Blast Injury Science and Engineering

A Guide for Clinicians and Researchers

Anthony M. J. Bull
Jon Clasper
Peter F. Mahoney
Editors

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Second Edition

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Foreword

We Will Remember Them.

A short walk from the central London campus of Imperial College lies Sloane Court East. This quiet, tree-lined residential street seems a world away from our busy university buildings and the work we do in its laboratories and classrooms to understand the complex effects of blast injury. Yet there is a very direct human connection. Anyone walking along Sloane Court East who cares to look can see two small memorial plaques, one inserted into the brickwork of a corner wall, and one set into the pavement. The plaques commemorate the deaths of 77 people who were killed at 07.47 am on 3rd July 1944, when a V1 rocket exploded at the north-western end of the street. Most of those who died were American service personnel. It remains the largest single loss of US military life on UK civilian soil in history. Despite this, the Sloane Court East explosion has been almost forgotten. Part of this process was deliberate. At the time, there was heavy press censorship in both Britain and the USA. The operation to liberate Europe, initiated on D-Day (6th June 1944), was less than a month old. The Allied advance was progressing, but at a slow pace. Neither government wanted their public to hear more bad news than absolutely necessary. Londoners were exhausted by sleepless nights spent dreading the onslaught of Nazi V1 flying bombs, whose destructive power was both brutal and inescapable.

The V1 rocket that appeared out of the morning haze over Sloane Court East on 3rd July carried 850 kg of high explosive in its warhead. Beneath its path were several six-by-six transport trucks loading up with US service personnel about to travel to their day's work at SHAEF offices in central London or to training camps outside the city. As it was designed to do, the rocket went silent a moment before gliding into its impact. These few seconds gave some of the soldiers and civilians just enough time to try to escape, but not enough for the men in the trucks to scramble out to join them. One young soldier remembered running around the corner of the street and finging himself to the ground at the moment of the explosion. He staggered to his feet, ears ringing, the air around him full of smoke and dust. Around him, other survivors were also able to stand and stumble back towards the scene of the blast. Although several of the buildings had been blown apart and there were fires breaking out in the rubble, many of the dead were lying in the open street, without crush wounds or other visible signs of injury. One witness remembered running through 20 or 30 bodies, desperately checking to see if any of them had a pulse, but their glassy eyes and dead weight told him that they were beyond his help.

At Sloane Court East, the worst killer had been completely invisible. V1 rocket explosive payloads produced huge blast waves that smashed out at least 400–600 yards in all directions. In the narrow street, this wave was confined and compressed by the solid stones of the buildings and tarmacadamed road surface, so its energy was exponentially increased, crashing through lighter, weaker structures. There was no blast crater where the bomb had landed, but bricks and masonry close by were pulverised into piles of dust. The sturdy US trucks and their loads of passengers were lifted into the air and slammed back on the ground, killing everyone inside. Pedestrians on the pavement were felled where they stood as the wave tore through their bodies. Fundamental blast physics explains the vacuum that always follows the initial blast wave that in turn generates a secondary wave. In Sloane Court East, one survivor described how it felt to directly experience such an effect where the building she was in "not only rocked, it did ground loops". The secondary wave caused at least three houses and two blocks of apartments to collapse, trapping many people inside. A first aid post was set up, and survivors pulled from the ruins of the devastated street, some taking days to be rescued. Later, the grim business of identifying the dead from the mangled body parts taken to a nearby morgue was assigned to two young US officers, Lieutenants Riley and Byleen.[1] All the human witnesses to the horror of Sloane Court East are now long gone. Only its trees remain, planted as saplings in the pavements of the street in the years before the war, and somehow able to survive the blast wave's effects. They can still be seen today, grown as high as the buildings and apartment blocks that were rebuilt when peace came in 1945.

Because of other wars in our own time, our understanding of blast effects and injury has been transformed. Blast injury studies have become an established and productive part of our academic infrastructure at Imperial College London, as well as providing a globally influential model for effective multidisciplinary co-operation across clinical, medical and bioengineering domains. We can model, predict and seek to mitigate the invisible wave and

[1]Thank you to Alex Schneider, of www.londonmemorial.org, who has preserved the history of the Sloane Court East explosion.

its effects. Few mysteries remain. Wherever this work is done, it should be in the remembrance that blast injury always begins with a blast. Every blast event comes with the primary, secondary, tertiary and quaternary effects detailed and analysed so effectively in this volume. We should also be mindful of the other, less quantifiable effects: the horror, agony and fear just before the darkness and the silence. What for some is the point of wounding is for many others the point of dying. As we work to secure better ways for human beings to survive blast injury, we should also honour those beyond our help but never beyond our memory.

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September 2022

Emily Mayhew

Preface

There have been many advances in the field of blast injury research since the publication of the first edition of *Blast Injury Science and Engineering* in 2016. As such, we believed that now is the right time to collate the second edition, allowing us to expand on the content of the book and update several chapters with important new research and knowledge.

Many of the chapters in this second edition are built on the foundation of the chapters in the first edition, incorporating additional research to bring them up to date. We have also added several new chapters and included an entirely new section on Rehabilitation. We believe that the addition of the Rehabilitation section is crucial to such a resource as this, given that understanding of blast injuries, and importantly their long-term impact, does not stop at the point of wounding. People's lives are changed by their injuries and acknowledgement and collation of the breadth of work being undertaken in the field of rehabilitation is an important consideration.

The editors would like to thank all those that contributed to the second edition of the book. This includes more than 60 authors who have given their time and who have brought together expertise in science, engineering and medicine, from both military and civilian perspectives. We would also like to thank the new section editors who worked on this second edition of the book—Dr Spyros Masouros, Dr Arul Ramasamy and Professor Alison McGregor.

Finally, we would like to thank all the funders that have enabled the research which is discussed in this book. In particular, the editors would like to thank the Royal British Legion who have funded a significant amount of the research that has come from the authors based at Imperial College London.

We hope you enjoy this second edition of the *Blast Injury Science and Engineering* book.

London, UK	Anthony M. J. Bull
London, UK	Jon Clasper
London, UK	Peter F. Mahoney

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Part IV Modelling and Mitigation

Part I

Basic Science and Engineering

1 Section Overview

Spyros D. Masouros

The multidisciplinary nature of blast injury, covering its pathophysiology, treatment, protection, and rehabilitation, requires an appreciation of a broad spectrum of topics in science and engineering. This section covers the science and engineering foundations required to follow the rest of the book. It introduces main principles, language, and techniques associated with the physics of blast, the behaviour of materials, biomechanics, computational modelling, and biology and physiology. The objective is to bring the reader, irrespective of background, up to speed with the clinical, scientific, and engineering aspects required to follow comfortably the contents of this book.

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2 The Fundamentals of Blast Physics

William G. Proud

Abstract

Blast injury is an explosion effect and generally comes from the use of manufactured explosives. In some relatively rare occasions, the rupture of a compressed gas bottle or the breakage of a high-pressure storage or reaction vessel will result in blast injuries. There are many books on explosives that start with a historical summary of the discoveries and personalities involved before developing the theories and the quantitative measures used in explosive chemistry, physics and engineering. In this chapter, however, the aim is to present the reader with a timeline of events relevant to blast injury working from the point of activation of the initiator to the arrival of the blast wave at the human body or target. The steps will be presented in a simple, largely non-mathematical, way, introducing the basic concepts and placing the materials within the wider context. For those who wish for a more mathematically developed understanding, a selection of references is indicated for further study.

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The Aims of this Chapter

1. Introduce some fundamental aspects of explosives and timescales
2. Provide an outline of how mines, blast and fragmentation work
3. Describe the energy release process and the efficiency of the process
4. Describe and distinguish between waves transmitting in solids, liquids and gases
5. Describe the range of fragment sizes and velocities
6. Provide an overview of shock and blast wave propagation
7. Outline how waves expand and interact with the surrounding environment

2.1 Explosives and Blast: A Kinetic Effect

Sound is a low magnitude stress wave. It propagates through air, causes a very minor change in the density of the air and moves at a fixed velocity. An **explosion** is a term used to describe the rapid expansion of gas, possibly from the rupture of a pressure vessel or a gas cylinder, the sudden vaporisation of a liquid, for example water exposed to hot metal, or by a rapid chemical reaction as seen in an explosive. The energy associated with a blast wave causes significant compression of the air through which it passes

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and the blast wave travels at a velocity faster than the speed of sound in air.

Explosives form part of a range of materials classified as '**energetic materials**'; other members of this class include propellants and pyrotechnics. The distinguishing feature of energetic materials compared to other materials is the very fast rate of energy release. This release rate and nature of the reaction products determine the use of the material.

The basic chemical components in an explosive are a fuel, an oxidiser and a material that allows a rapid ignition of reaction. In terms of total energy released, energetic materials are not particularly distinguished from other chemical reactions: petrol and butter release more energy per molecule when oxidised than tri-nitro toluene (TNT), probably the most famous explosive. This difference is that, while petrol needs to be mixed with air and then set alight, the explosive comes with the fuel and oxidiser intimately mixed, sometimes with both fuel and oxidiser present in the same molecule. The chemical or composition used gives information on intended application. Materials containing TNT, nitroglycerine, pentaerythritol tetranitrate (PETN), cyclotrimethylenetrinitramine (RDX) or tetrahexamine tetranitramine (HMX) are likely to be bulk explosives; these materials produce significant quantities of hot, high-pressure gas. Compositions using silver azide, lead azide, picric acid and the fulminates are sensitive to impact and flame, explode relatively easily, and so tend to be used in initiators, also called detonators; these materials are, however, very sensitive and so it would be unwise to use them as a main explosive charge.

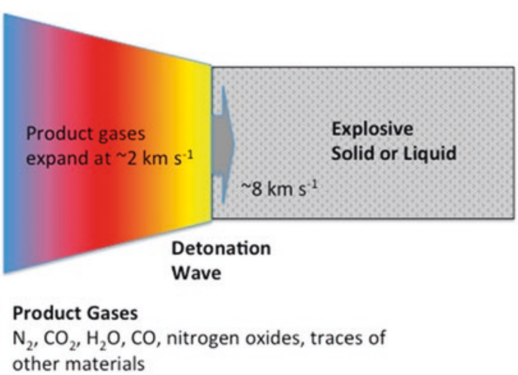

Fig. 2.1 Schematic of detonation process

High explosives, correctly called **high order explosives** include TNT, HMX, RDX and PETN often mixed with other chemicals to make the resulting explosive composition more stable, easier to process industrially or fit into cavities within a munition. When these materials detonate, they release their energy due to the passage of a reactive shock wave. A shock wave is a high-pressure stress and temperature pulse which moves through the material at supersonic speed. In the case of many compositions used in munitions, the shock wave consists of a thin, often sub-millimetre, region where the explosive turns from a solid or liquid into a hot, high-pressure gas (Fig. 2.1). The velocity of a detonation wave is of the order 5–8 km s^{-1} and the energy release rate is of the order of Gigawatts; that is the same output as a large electrical plant over the time of a few microseconds. The pressure associated with detonation waves is many hundreds of thousands of atmospheres. The product gases expand quickly; a rule of thumb gives that the gas expands ~a quarter of the detonation velocity; it is effectively a 7200 km h^{-1} wind. It is not surprising that such aggressive energy release can be used to shatter and push fragments at high velocities.

Within the class of high explosives, there is a sub-division between so-called **ideal** and **non-ideal** explosives. Ideal explosives have very thin detonation-reaction zones where the reaction takes place on a sub-microsecond basis. Ideal explosives have a more pronounced shattering effect, or brisance, on materials placed in contact with them; this class of materials includes TNT. Non-ideal explosives have much thicker reaction zones and give out their energy over slightly longer timescales, up to several microseconds. The most widely used explosive in the world, Ammonium Nitrate: Fuel Oil (ANFO), falls into this class. The chemical make-up of ANFO results in much lower pressures of detonation, about a quarter of the pressure seen in high explosives, producing less shattering effect. ANFO, however, generates a lot of gas, which gives the mixture a lot of 'heave', the ability to lift and move the surrounding material. The low detonation pressure means this mix produces cracks in rocks which are then pushed open, heaved, by the detonation gas. These materials tend to be used in the mining and quarrying industry.

Low Order Explosives are materials that give out their energy as a result of rapid burning, called deflagration. This class of materials includes much of the materials often called gun-powders, where the reaction is due to the convective motion of pressurised hot gas moving through the powder bed causing particles to ignite. Here, the energy reaction rate is much lower of the order of a thousandth of that seen in high explosives. Again, to emphasise, the total energy output is similar to that of high explosives but the lower rate produces much lower pressures.

Both high and low explosives generally need a degree of mechanical **confinement** in order to detonate. A gramme of gun-powder on a bench will react quickly with a flash of light, a small fireball, some heat, but little other effect. The same quantity placed inside a sealed metal can, however, where the confinement allows the hot gases to stay close to the powder and pressure and temperature to build-up accelerating the reaction until the confinement shatters, produces an explosion or propels a bullet down a barrel.

Propellants are a wide group of materials where the reaction pressures are of the order of thousands of atmospheres and the reaction timescales are measured in milliseconds. These materials are used in rocket motors and to drive bullets or shells. In some cases, propellants, if confined or impacted at high velocity, may detonate. Missile motors may detonate if they overheat or if the product gas cannot vent quickly enough. As a group of materials, they can present a significant fire hazard and are sometimes used in improvised explosive systems.

Pyrotechnics are a wide range of materials including flares and obscurants. They do not produce high pressures but can generate significant heat and can be used to ignite explosives. They can be very sensitive to electrostatic discharge, but badly affected by moisture. Magnesium-Teflon-Viton (MTV) flares have been used to protect aircraft from heat-seeking missiles as they emit very strongly in the infra-red region but have no visible flame. As a class, they can produce severe burns to casualties and often are not easily extinguished by water or conventional fire extinguishers as they possess significant quantities of oxygen and energy within their composition.

2.2 Explosive Systems: The Explosive Train

Energetic materials are extensively used in munitions, both the type of energetic material and its function cover a wide range of masses and outputs. This can be seen clearly if we consider the amounts contained in a number of common munitions; a small arms round often has less than 1 g of propellant to drive the bullet, while a hand grenade would contain something of the order of a few tens of grammes to shatter and throw the casing, and a large anti-tank mine would contain up to 25 kg of high explosive. Given this array of systems, it is easiest to think of the explosive content within the munition in terms of the explosive train (Fig. 2.2), which indicates the basic prerequisites of an explosive munition system.

A variety of stimuli can be used to put the explosive train in motion such as standing on a mine, closing an electrical switch, activating a magnetic action, etc. These actions input energy into an ini-

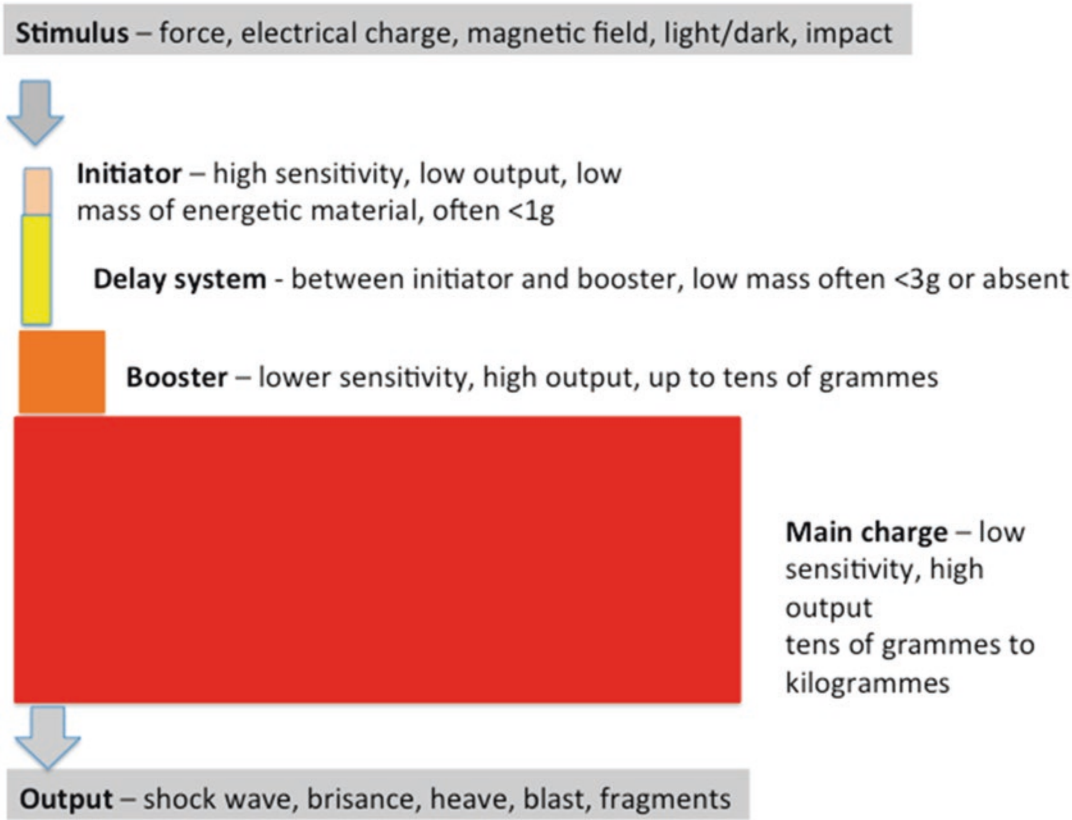

Fig. 2.2 The Explosive Train

tiator that contains a material that is highly sensitive, reacting when mildly crushed, impacted or heated. In anti-personnel mines, a simple design is based on a crush switch where applied pressure stabs a metal pin into a small metal thimble filled with a sensitive explosive. The main function of the initiator is to produce heat or a shock wave.

From the point of view of weapon-system design, it is useful to keep the amount of initiatory compound as low as practical, otherwise the weapon could become very sensitive to being dropped, shaken or transported. For many systems, a physical space or barrier exists between the initiator and the rest of the explosive system for the purposes of safety. This barrier is removed when the system is armed either manually or by an electrical/mechanical arming system. In the case of many bombs and missiles, the arming sequence occurs after the weapon has been launched to provide security to the user and the launch platform.

The heat or shock from the initiator is transmitted either along a delay system or directly into a booster charge. The function of the delay is to burn for a known period allowing time for other processes to occur such as throwing a grenade or allowing a small propellant charge to 'bounce' a mine into the air. Delays are important in quarrying and mining where exploding a set of explosive-filled boreholes in a sequence allows for more efficient material extraction.

The booster system consists of an explosive compound that is less sensitive than the initiatory compound but will detonate due to the heat of the shock wave from the initiator. The energy output of the booster is much higher than that of the detonator. The shock wave is then transmitted to the main charge.

The main explosive charge consists of a relatively low-sensitivity explosive composition that will detonate from the energy delivered by the booster. The mass of such main charges is extremely variable, however, the output from the main charge can be defined in terms of the outputs.

These outputs include: (1) shock waves transmitted into the immediate environment around the explosive; (2) the shattering of surrounding material, such as a metal case; (3) the expansion of hot product gases violently accelerating the surrounding shattered material opening a gas-filled void in a structure; (4) the coupling of some of the explosive energy into the air, producing a blast wave; (5) the aggressive acceleration of fragments, shrapnel or target vehicles and people.

2.3 Energy Levels and Energy Distribution

To allow some generalisation, it is useful to consider the amount of energy output and the physical characteristics of the materials involved. One gramme of propellant will produce 1 kJ of energy. The temperature of explosive product gases can easily exceed 3000 K.

As gases are about a thousand times less dense than solids and liquids, the result of this is that 1 g of solid turns into 1 L of gas if that gas is at room temperature. The gas, however, is ten times hotter than room temperature (using the absolute, Kelvin temperature scale). This means that the 1 g of material turns into a gas that, if allowed to expand freely, would occupy 10 L. Alternatively, it would take 10,000 atm to keep the gas from expanding. This order of magnitude calculation for the less aggressive propellant materials shows the levels of force that are produced.

Of the energy that is released by the propellant, the proportion of the energy that goes into different processes is revealing. Again, a simple order of magnitude estimate indicates that 20–30% of the energy will go into the kinetic energy of the fragments, 60% into the kinetic energy and temperature of the explosive product gases, while the remaining 10–20% will be spread over a number of other effects, such as the blast wave, the energy required to fracture shell casings, the motion of the ground, etc.

2.4 Formation and Velocity of Fragments

Many explosive systems have a casing, usually of metal or plastic, often with features or inserts to form specific fragment shapes (see Chap. 8). Some systems have very thin casings; this is generally done to reduce the metal content or mass of the munition. Reducing the metal content makes landmines harder to detect—'low-signature'—but also has the effect of slightly increasing the blast as less energy is used in fragmentation or acceleration of fragments. As many landmines are designed to maim rather than kill, reduction of metal content also assists in this.

The initial effect of the detonation wave is to send a shock wave into the casing which produces a violent acceleration of the casing. Figure 2.3 shows the velocity profile from the outer surface of a copper cylinder subject to loading from the explosive filling. On the right-hand side of this figure, a schematic of the process of acceleration is shown. This diagram considers the motion of the wall of a cylinder filled with explosive, as the explosive detonates: the horizontal axis represents distance and the vertical axis time. The detonation products send a shock wave into the cylinder from the inner surface to the outer surface; this is represented by the black line with an arrow on it. A wave reflection takes place at the outer surface, sending a release wave back into the casing, towards the inner surface. This in turn is reflected from the inner surface as a shock. As this back-and-forth wave reflection process takes place, each reflection represents the acceleration of the casing expanding in a series of steps. As the casing is accelerated outwards, it expands and becomes thinner. The diagram on the left also indicates the violence of this acceleration; the outer surface moves from rest to reach a velocity of 0.8 mm μs^{-1} (corresponding to 800 ms^{-1}) within 5 µs. More detailed discussion of how this can be calculated can be found in the specialised texts of the explosives engineering community, detonation symposia and shock physics literature. The total distance moved by the casing within this very narrow time window was only 0.6 mm. Thus, the metal casing is subject to very violent acceleration; this causes fracture and fragmentation within the casing producing metal splinters moving at high velocity.

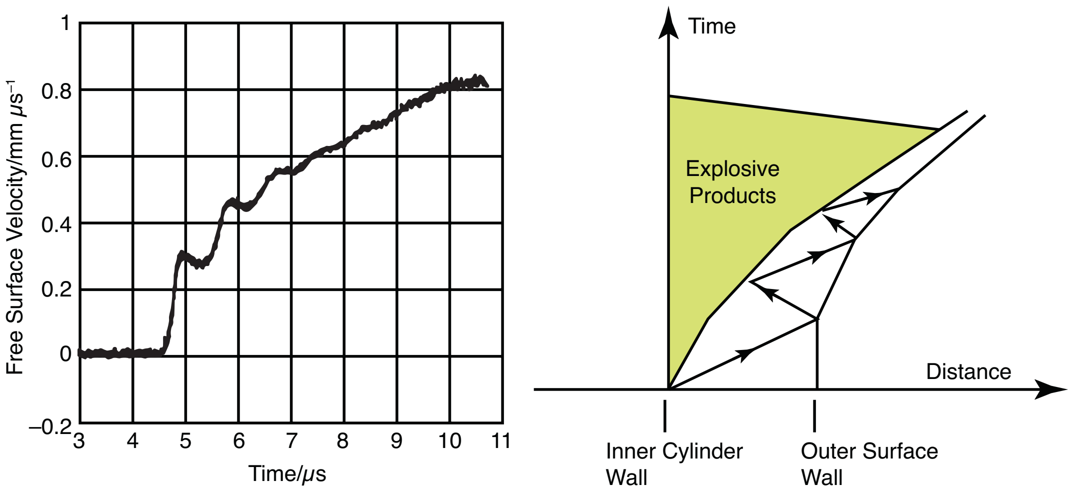

Fig. 2.3 Left: the velocity history of the outer surface of an explosively loaded metal cylinder. Right: a distance-time diagram of the wave processes taking place within the casing

The study of fragments and fragmentation took a major step forward during the Second World War (1939–1945) with significant scientific effort placed in the service of the militaries of industrialised nations. The two major studies are named after their main originators: Neville Mott and Ronald Gurney. The number and size of metal fragments from a casing were addressed by the Mott fragmentation criteria. This is a statistical model based on the idea of a rapidly expanding metal ring which contains imperfections. The imperfections form the basis of fractures which eventually break the ring into fragments. As the ring breaks, the stress inside the resulting fragment reduces and further fragmentation does not occur. This model was developed and populated in a semi-empirical fashion where the results of theory and experiment were compared.

The result of the modelling and experiments was to produce a series of curves giving fragment-size distribution. An example is shown below, in Fig. 2.4, where a Mott calculation is compared to experimental data of fragment distribution from an iron bomb.

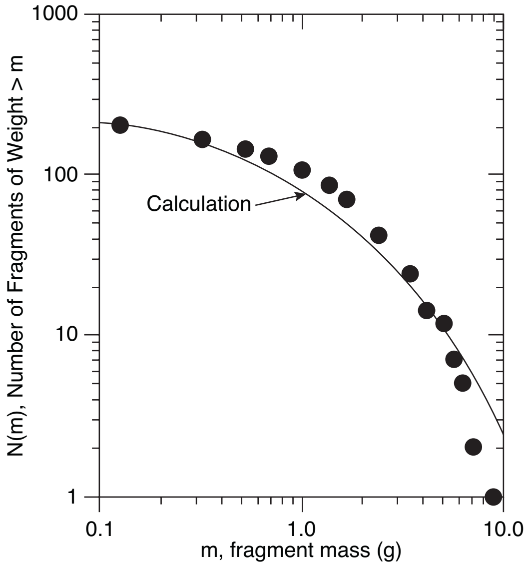

Fig. 2.4 Fragment distribution from an iron bomb [1]

The research of Gurney considered the balance of energy in the system; how much of the released energy would be captured by the metal casing, so predicting the expected fragment velocity. In the derivation of the velocity, the casing was assumed to remain intact during the acceleration process. The shape of the explosive charge and the casing has a major effect and this resulted in the production of a whole series of Gurney equations. In Eq. (2.1), below, the expression is for the velocities achieved by explosive loading of a cylinder

$$\frac{V}{\sqrt{2E}} = \left(\frac{M}{C} + \frac{1}{2}\right)^{-1/2} \tag{2.1}$$

where C is the mass of the explosive charge, M is the mass of the accelerated casing and V is the velocity of accelerated flyer after explosive detonation. The term $(2E)^{1/2}$ is the Gurney constant for the explosive; it has the same units as velocity and is sometimes called the Gurney velocity. Each explosive has a particular value of Gurney constant and accounts for the coupling between the energy of the detonation products and the energy deposited into the metal casing.

While the two approaches may seem contradictory, one assuming fragmentation and the other an accelerated but intact material, using both approaches is useful due to the timescales of the initial acceleration and fragmentation. The energy delivery by the detonating material occurs very quickly giving the initial impulse, while fracture takes time and expansion to develop. Once the fractures have started to open, the product gases escape, potentially causing a fireball, but only producing a limited change on fragment velocity post-fracture. Overall, the two approaches have provided a reasonable and predictive approach to fragmentation and fragment velocity.

From this section, it can be seen that explosives produce violent fragmentation and acceleration: a large number of small fragments and much fewer large fragments, moving at velocities that range from several hundred to a few thousand of metres per second. These fragments can produce significant, life-threatening injury in themselves, irrespective of the blast wave associated with the explosive charge.

2.5 Shock and Stress Transmission

At this point on the timeline, the munition has detonated and a shock wave has passed through the casing, which is starting to move and fragment. In total time, a few tens to hundreds of microseconds has passed since the main charge started to detonate.

The next effect that needs to be considered is **the transmission of the stress waves between materials**: how does the detonation pressure transmit from the product gases into the surrounding material, soil or rock? what are the properties involved? what is the resulting velocity of material when it is subject to a stress wave?

In all cases of wave transmission, it is important to consider what the type of the wave is, what the magnitude of the wave is, what are the properties of the material through which it is travelling and what is the change in material properties across the interface.

2.5.1 Wave Type

Wave motion can be broadly defined into three classes: compression, tension and shear. Compressive waves are associated with positive stresses and pressures, tensile waves with negative stresses and pressures, while shear waves produce motion lateral to their direction of propagation. To think of the action of a lateral wave, consider a pack of cards placed on a table; if you press down on the top card and move it sideways, then the cards underneath will move sideways as well, however, each card does not move quite so much as the one above it; the resulting spreading of the deck of cards is the result of shear. Shear waves produce the same kind of motion in materials as seen in the card deck.

The velocity at which the stress wave moves changes with stress level. For solids with strength, at stress levels below that of the elastic strength, the velocity of the wave is the same as the speed of sound in the material. For stress pulses that are above the strength of the material, the compression of the material results in the wave speed being higher than that in the uncompressed material; this is the definition of a shock wave.

Waves that reduce the stress level are generally called tensile waves and are most commonly seen when a material is pulled. If the material is dropping from a shock-compressed state, then this wave is more correctly called a release fan. In most common materials, the shock velocity increases with stress, while the stress-reducing release waves move at increasingly slower velocities as the stress drops; this is a result of the change between the very aggressive shock loading and the relatively slower and milder release process. Thus, the release region spreads, or fans out, over a much wider region than the thin shock front, and is most often associated with a continuous decrease in stress to the ambient pressure state.

2.5.2 Magnitude of the Wave

The magnitude of the stress wave is defined in terms of its stress level. One factor that often leads to confusion in the field of stress propagation is the relationship between stress and velocity. The important thing here is to remember one of Newton's laws of motion—**a body will continue in a state of rest or motion until acted upon by an external force**. It is, therefore, perfectly possible to have a material, fragment, etc. moving at a high velocity under no external force and also possible to have a material under high pressure but not moving.

The basic equation to be defined is the relationship between stress, the volume through which it moves and the acceleration it produces in the material. The law to be considered here is conservation of momentum; the product of mass multiplied by velocity.

The definition of stress (σ) is force (F) divided by the area (A) it acts over.

$$\sigma = F / A \tag{2.2}$$

The mass (m) of the material that is affected by the passage of the wave is going to be the volume the stress wave has moved through multiplied by its density.

The volume (V) that the stress has swept through will be the area (A) multiplied by the velocity at which the wave moves (U_σ) over the time window we are interested in (δt). The density of the material is represented by ρ.

So the mass that has been accelerated will be

$$m = V\rho \tag{2.3}$$

and from the argument above

$$V = AU_\sigma \delta t \tag{2.4}$$

So the mass is

$$m = \rho AU_\sigma \delta t \tag{2.5}$$

The final step is to consider the acceleration and the final velocity obtained. Here, we use one of the fundamental equations representing the acceleration (a) produced by a force (F),

$$F = ma \tag{2.6}$$

where the acceleration (a) is the change in velocity of the material (δu_p), sometimes called the particle velocity, over the time window we are considering (δt).

$$a = \delta u_p / \delta t \tag{2.7}$$

Combining these terms to relate stress to change in velocity, we arrive at the equation

$$\sigma A = \rho A U_\sigma \delta t (\delta u_p / \delta t) \tag{2.8}$$

which simplifies to

$$\sigma = \rho U_\sigma \delta u_p \tag{2.9}$$

This is one of the fundamental equations in shock physics, one of the so-called Rankine–Hugoniot relations. The full set of these Rankine–Hugoniot relations can be found in the introductory texts by Meyers or Forbes [see further reading section].

The fundamental relationship here is that the change in velocity produced by a stress wave is dependent on the density and the wave velocity of the transmitting material. The value of the density of the material multiplied by the wave speed is called the impedance (Z) of the material.

At low stress levels, the velocity at which a stress wave travels through a material is the same as the material's sound speed. At higher stresses, when a lot of force is applied as in an explosion, the velocity of the stress wave can be higher than the sound speed, a shock wave. As stated earlier, a detonation wave is a shock wave that is driven and supported by the energy release of the chemical reaction.

2.5.3 Impedance: A Property of the Material

In principle, the value of impedance for low stress levels is easy to calculate; it is the product of the density multiplied by the sound speed in the material. Table 2.1 contains the density, sound speeds and impedances of some common materials; air at 1 atm, water, iron, tungsten and Perspex®. These materials have been studied extensively and their properties are quite well known. The impedance of air changes strongly with pressure and is discussed in Sect. 2.6.1.

Table 2.1 Impedance, density and sound speed values for a range of materials

Material	Density/kg m^{-3}	Sound speed at 1 atm pressure/ms^{-1}	Impedance/kg m^{-2} s^{-1}
Tungsten	19,220	4030	77.4×10^6
Iron	7850	3570	28.0×10^6
Perspex®	1190	2600	3.09×10^6
Water	1000	1500	1.50×10^6
Air	1.292	343	4.43×10^2

As the stress level in the wave increases, however, then other properties such as strength and compressibility become important. As the stress level increases, so does the amount of energy that is being deposited in the material; some of this will result in increasing the kinetic energy of the material while another part of the energy will result in the material being compressed and becoming hot. The exact mathematics of this situation is complex and beyond the space available in this brief chapter, however, we can outline some simple conceptual guidelines.

The strength of a material is its ability to resist distortion; this strength will be different in compression, shear or tension. Materials with high strength tend to have high sound speeds as a result. Metals, in general, have similar strengths in tension and compression, while rocks and ceramics are strong in compression but weak in tension. Granular materials such as sand have no tensile strength, but can have significant compressive strength. Given the three-dimensional jigsaw-like nature of sand, the more you press down on the sand the harder it is to move it sideways, in other words shear it.

Compressibility is the ability of a material to deform and is the inverse of strength. Highly compressible materials such as foams are often used to protect objects from impact; they do this because the energy of the impact is absorbed in locally distorting and compressing the material in the region of impact and not into increasing globally the kinetic energy of the foam. Sands, soils and granular materials absorb energy in a number of ways by grains deforming, and grains fracturing, the particles moving together to fill the pores. These mechanisms absorb energy and act to mitigate the shock or blast wave.

As the material compresses, its density will change and its sound speed will tend to increase. At some point, the amount of compression in the foam will result in the removal of the majority of the voids at which point the material will behave like a stronger, solid mass; there is a limit to energy absorption. Similarly, all materials will have a yield point, where their yield strength is exceeded and they begin to deform and compress so there will be a change in how the energy in the stress pulse is deposited; more will go into temperature and into internal compression processes and less into velocity. In addition, time-dependent processes will also be occurring; the time for pores to collapse in foams and for particles to fracture in sands, so there will be a time dependence to the stress transmission.

The same issue of time dependence in the change of impedance, degradation of strength, and interplay between kinetic energy and compressibility occur in biological materials. The impedance of the material will change through the stress pulse and so the simple equations given should always be used with caution and to produce estimates.

2.5.4 Wave Transmission Across Interfaces

When a stress wave reaches an interface, it is the difference in the mechanical impedance of the materials which determines how much of the stress wave is transmitted and how much is reflected. By using conservation of momentum and the impedances of the materials, the amount of the stress transmitted and that reflected can be calculated and the **change in stress** in the materials calculated. The results of this is given in Eqs. (2.10) and (2.11)

$$T = 2Z_2 / \left(Z_1 + Z_2\right) \qquad (2.10)$$

$$R = \left(Z_1 - Z_2\right) / \left(Z_1 + Z_2\right) \qquad (2.11)$$

where T is the fraction of the stress transmitted, R is the fraction of the stress reflected, Z_1 is the impedance of the material through which the stress is originally transmitting and Z_2 is the material on the other side of the interface. Three situations are shown in Fig. 2.5b–d.

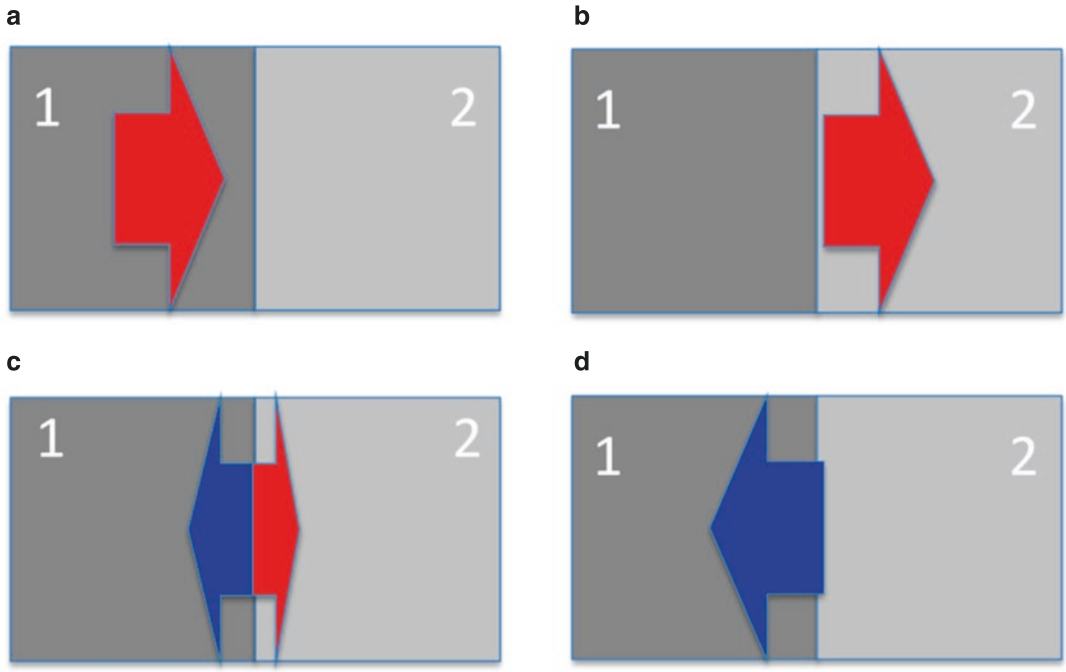

Fig. 2.5 Stress transmission across the interface between two materials. (**a**) A shock wave moves through material 1 towards an interface with material 2, (**b**) If the impedance of material 1 and 2 is the same, all the stress transmits, (**c**) If there is a difference in impedance, some of the stress transmits and some is reflected, and (**d**) If the impedance of material 2 is very low compared to material 1, then most of the wave is reflected back into material 1

Here, it is important to remember that **stress, force and velocity are all vectors**; they have a magnitude and a direction. By convention, we make an increase in velocity from left to right to be positive and right to left as negative. A wave which acts to compress a material will be regarded a positive stress, and a wave which puts the material into tension or releases will be regarded as negative change in stress.

In Fig. 2.5a, a stress wave travels through material 1 of impedance Z_1 towards an interface with material 2, impedance Z_2. Figure 2.5b shows what happens when the wave reaches the interface if the impedances of material 1 and 2 are the same; all the stress wave transmits through the interface. The value of the transmission coefficient is 1 and so the stress in material 2 is exactly the same as the stress level in material 1, i.e. both end up at the same stress. The amount reflected is 0 and so the stress in material 1 remains unchanged.

In Fig. 2.5c, the materials have impedances that are of a similar magnitude but different values; Z_1 being half the value of Z_2. Calculating the stress transmission and reflection coefficients results in $T = 4/3$ and $R = -1/3$. This means that that stress produced in material 2 is higher than that initial wave. This may seem odd, until it is remembered that impedance is related to density and sound speed, so material 2 in this case is denser and/or stronger than material 1; effectively material 2 slows down the motion of material 1; the result is higher stress and less particle velocity.

While the transmitted force is still going in the positive direction, it is higher by 4/3 over the initial stress in material 1. The reflected portion has a negative reflection coefficient, and the wave is propagating in a negative direction. Mathematically, two negative numbers multiplied together equal a positive, so what this implies is that the amount of reflection means we have a stress change of +1/3 of the initial pressure ***adding*** to the initial pressure. The stress in material 1 increases as the denser, stronger material 2 prevents it from moving forward, effectively swapping a change in velocity for higher stress.

This reveals an important point; if surfaces remain in contact, then the stress is the same in both sides of the interface. This is the basis of a technique called impedance matching used in shock physics to determine the stress and velocity of materials under the action of shock waves.

A material with a high density and high sound speed will have a higher impedance and as a result will exert a higher pressure at a given impact velocity than a low density or low speed material. This means that a much higher stress is produced by tungsten, a preferred material for anti-tank weapons, striking an object at a given velocity than the lower impedance iron, while Perspex will produce a much lower velocity.

In the case of Fig. 2.5d, the impedance of material 2 is very low compared to material 1. In this case, the change in the reflected stress is a value of 1 and it is going in the negative direction. Following the argument above, this means it changes the stress by −1. This implies that the stress change of the reflected wave cancels the stress of the original wave. In this case, the stress falls to zero and the energy of the stress pulse accelerates material 1 to a higher velocity, approximately doubling the particle velocity.

2.5.5 The Solid–Air Interface

After the stress wave from the detonation has transmitted through the casing, accelerating it and ultimately shattering it, it is then transmitted through the surrounding soil and sand compressing and fracturing it. When the stress pulse arrives at the solid/air interface from the transmission-reflection coefficient, it is clear that the vast amount of the stress is reflected from the solid/air interface. This reflection is like the case in Fig. 2.5d discussed above; the velocity of the sand/soil particles double as the stress drops to zero. Sand/soil has a very limited tensile strength so the result of this is to throw the sand and soil from the surface as a cloud of fast-moving debris.

While the stress wave compresses the material and ultimately results in a cloud of fast-moving debris, the product gases from the explosive devices are also pushing the soil and fragmenting the compacts formed by the stress wave. It is the expansion of the hot product gases that produces the largest effect of the blast wave.

2.6 Blast Waves

The product gases have a velocity of approximately 2000 ms^{-1}, considerably higher than the sound speed in air of 330 ms^{-1}. The resulting high-pressure pulse of air is pushed outwards by the hot explosive products. In order to allow comparison between charges of different sizes and compositions, explosive engineers conducted experiments using the simplest scenario possible—a bare explosive charge in an empty, flat field. This has resulted in a large body of blast wave literature based around the classic ‘Friedlander’ blast wave form.

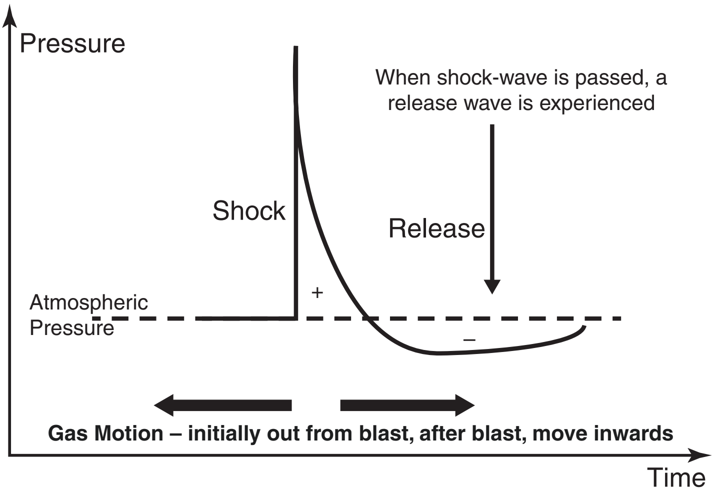

Fig. 2.6 The classic Friedlander form of a blast wave

Figure 2.6 shows the pressure-time profile of this classic blast wave. After the initial rapid rise of the blast wave, there is a region of positive pressure, accelerating outwards from the explosion. The speed of the gas moving behind the blast front, in the so-called blast wind, can be as high as 2000 km h^{-1}. This is followed by a release wave that drops the pressure below atmospheric pressure. The release occurs as both air and product gases have expanded outwards, away from the place where the explosion initiated. As a result, the explosive products have expanded and therefore performed 'work' on the surroundings and started to cool. This leaves a partial vacuum in the region of initial explosion, and this lower pressure now causes air and gases to flow backwards over a longer period to equalise the pressure. The resulting push-pull movement experienced in the blast wave can be especially damaging to structures and humans.

This simple waveform was often observed in much early research into the effects of blast on humans. These studies used data either from open explosive ranges or with well controlled blast reflections from single walls or barriers. It is important to note that in a cluttered urban or vehicle environment, a casualty will experience a number of waves—those directly arising from the charge and those reflected from a wide variety of surfaces and arriving from different directions. In general, the amount of reflected pressure waves can be equated to more damage and injury.

2.6.1 Change in Impedance of a Gas in a Blast Wave

In the discussion above, the energy deposited in a material can manifest itself in a number of ways; it can give the material kinetic energy, for instance, and it can compress the material increasing its temperature.

In many engineering applications, solids and liquid are often assumed to be incompressible. In shock wave studies, this is not the case and compressions which halve the volume of the material are relatively common. Gases, however, are very compressible. This means that blast waves are supersonic with respect to low pressure sound waves and they also are associated with a very large change in impedance and often with a large increase in temperature. Figure 2.7 shows the change in impedance of air with pressure. Atmospheric pressure is located at the point where the impedance increases sharply.

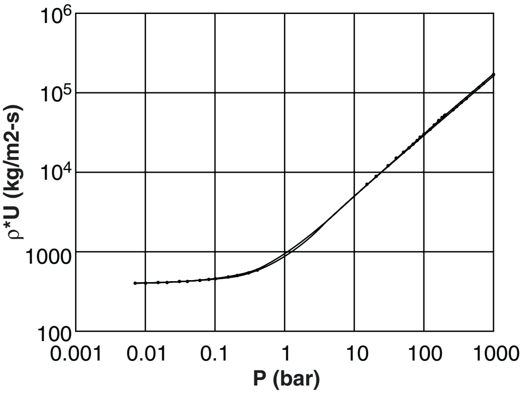

Fig. 2.7 The change in impedance of air with pressure

A major difference between shocks in solids and liquids and blasts is that blast waves from munitions tend to have durations measured in **milliseconds**, while shock waves in solids exist on timescales of **microseconds**. This thousand-fold difference in duration is why the relatively modest pressure seen in a blast wave can produce more movement and damage than the much higher pressures in shock waves.

2.6.2 Reflected Waves

While the compression of the air produces a large change in impedance, it is still the case that the impedance of the compressed gas is very much lower than that of any solid or liquid. When the wave hits against a solid barrier, a compressive pulse is transmitted and the stress in the gas also increases. Calculating the values of the reflected stress change indicates that the stress level almost doubles.

In the case of an explosion in an enclosed space, or in the partially confined space below a vehicle, there is time for multiple wave reflections; this leads to complex blast waveforms of widely differing stress histories. Within a vehicle, the reflections add to the injuries seen, over and above the push-pull effect seen in the relatively simple Friedlander waveform. Figure 2.8 shows an example of the type of waveform seen inside a vehicle, showing a longer lower level with additional pressure spikes.

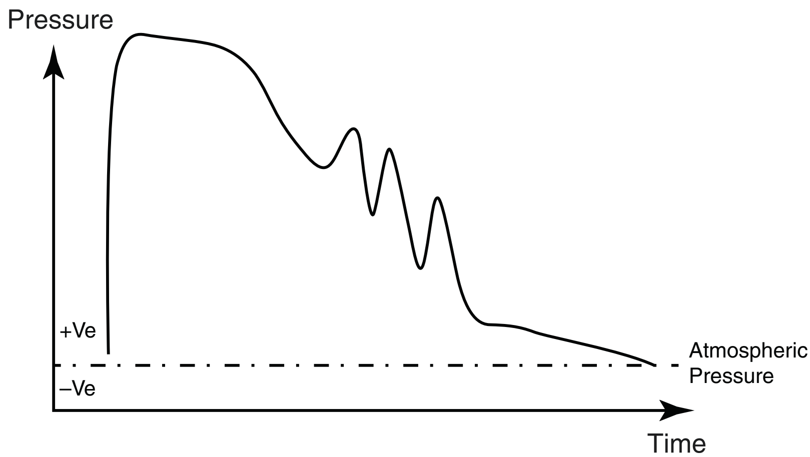

Fig. 2.8 Waveform showing a possible form of loading seen due to reflections internal to a vehicle

2.6.3 Temperature Rise

As well as the motion of the blast wave there is also a temperature rise associated with it. This can result in burns and combustion. Inside a vehicle, the heat deposited into a material can be divided almost evenly between the heat from the intense light flash associated with the detonation or gas compression while physical contact with the hot gas deposits the other half of the energy. The timescale for this energy transfer is of the order of milliseconds.

Shock waves also result in significant heating within metals; even after the metal has been dropped back to normal stresses, there may be a residual temperature increase in the fragments of over 100°C. The timescale for this heating is in the order of microseconds.

2.7 Comparing Explosive Scenarios: Scaled Distance and TNT Equivalence

Munitions and explosive charges come in a range of sizes and vary in terms of materials. In the technical literature, the term 'scaled distance' is often used to relate the effects of large explosive charges over tens of metres to those of small charges at close range. This is useful as it allows the effects of small-scale experiments to be extended to larger explosives charges.

The two terms used in virtually all scaling equations are (1) the explosive mass and (2) the mathematical cube of the distance between the charge and the target, i.e. the increase in volume in which the energy is deposited and ultimately dispersed.

There are often other terms involved; one of the easiest to conceptualise is the distance of the charge above the ground. If the charge is in mid-air, then the energy expands evenly in all directions, effectively spreading the energy through a sphere. If the charge is on the ground, however, then the effects of wave reflection occur. The gas has very low impedance compared to the ground and so virtually all the energy is reflected back into the expanding hemi-sphere above the ground. Within the hemisphere, the energy and pressure are approximately doubled compared to that of the mid-air charge.

Another simple comparative scale between explosive types is that of 'TNT equivalence'. Historically, TNT was a widely used material, which could be easily melted and cast into a variety of shapes, unlike many other explosive materials. Given its castable nature, many experiments were conducted, resulting in large databases. TNT equivalence is a simple factor that allows a well-defined reference point for the broad comparison of the effects on a non-TNT explosive charge to be estimated.

2.8 The Three-Dimensional World and the Physical Basis of Blast and Fragment Injury

The real world has a complex topography and is made of a wide variety of materials, many of which change their properties based on the accelerations produced in them by an explosion. This somewhat mundane statement also indicates the difficulty of understanding the precise effects in terms of the materials and the three-dimensional world. A well-founded method, however, is to take that complex situation and divide it into smaller, more tractable parts.

2.9 Summary/Conclusion

This chapter has given an overview of the basis of explosive technology and presented some of the basic processes relevant to the blast process. The importance of the detonation wave, stress transmission, fragmentation, the ejection of sand/soil, expansion of the product gases and flash heating has been introduced using simple first-order approximations.

The complexity of the mechanical processes and the resistance of the human body can result in injury patterns that show effects that are distant from the immediate blast or impact. It is increasingly possible, however, to adopt an interdisciplinary approach that can bridge the vital mechanical-biological gap in our knowledge. Following chapters in this book will address and expand on these issues.

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3 Biomechanics in Blast

Anthony M. J. Bull

Abstract

Blast injuries induce a mechanical insult to the human body at all levels. The study of the mechanics of biological systems is called biomechanics. This chapter starts at a foundational level to briefly summarise the fundamental physics of mechanics and then how to apply that to the human body through an understanding of functional anatomy and musculoskeletal dynamics.

3.1 Overview

Biomechanics is the study of biological systems from a mechanical perspective. In this discipline, the tools of mechanics are used in which the actions of force, motion, deformation and failure and their relationship to anatomy and functional aspects of living organisms are analysed. Traditionally, biomechanics has been considered in terms of either biofluid mechanics or the biomechanics of connective tissues ("solid biomechanics"), where the former is a key branch of science and engineering that is applied to the cardiovascular and respiratory sciences and the latter is applied to orthopaedics and musculoskeletal rehabilitation. Other branches of biomechanics investigate the *interactions* between solids and fluids and these are applied in ocular, cellular or the biomechanics of the brain, for example.

These branches of biomechanics are typically associated with specific anatomical regions and physiology; for example, biofluid mechanics might focus on the effect of flow in the arteries and its relationship to cardiovascular disease, whereas connective tissue biomechanics might focus on the mechanical effect of ligaments at a joint and deal with their repair post injury.

Biomechanics in blast is a key discipline in blast injury science and engineering that addresses the consequences of high forces, large deformations and extreme failure and thus relates closely to knowledge of materials science (Chap. 4) as applied to tissues and, at the smaller scale, cells and molecules (Chap. 6).

The aim of this chapter is to give the reader a basic understanding of biomechanics and its utility in the analysis of blast injuries. The specific objectives are to:

1. introduce fundamental terminology and concepts in biomechanics; and
2. describe how forces are transmitted through the human body at all loading rates, and hence provide an analytical framework to analyse forces on all relevant tissues and structures associated with blast injury.

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3.2 Terminology in Biomechanics

3.2.1 Biomechanics of Motion

Kinematics is the "Biomechanics of Motion"; in other words, it is the branch of biomechanics that deals with the description of motion. It is most commonly used in the analysis of activities of daily living and sporting activities. In blast, we consider the movement of the objects such as blast fragments (secondary blast—see Chap. 9) and displacement or movement of the person due to blast and their interaction with objects such as vehicles (tertiary blast—see Chap. 9), where their movement is considered as *whole body* movement, or also takes into account the *relative* movement of different body parts.

Mass is how much matter an object contains. Conceptually, this is the number of atoms in the object which remain constant regardless of location or gravitational conditions (for example Earth, Mars or gravity in outer space). Weight, however, would vary under these three conditions. The importance of mass in blast biomechanics is that it represents the resistance to a change of linear motion (a speeding up or slowing down). This is important in blast when considering smaller fragments that are energised (small mass—secondary blast effects) by the blast when compared to larger items (a person or a vehicle—tertiary blast effects) that require more energy to get them moving. The SI unit of mass is the kilogramme (kg).

A **Rigid Body** is an object that doesn't change shape. This is usually a major assumption in biomechanics, but serves to allow force analysis to be conducted (the subject of this chapter) prior to deformation analysis (the subject of later chapters). Formally, a rigid body is a collection of particles that do not move with respect to each other. In a force analysis, the assumption is that however large a force may be, the rigid body does not deform. Obviously, this is an approximation in every case because all known materials deform by some amount under the action of a force. In the biomechanics of blast, we tend to analyse the human body first as a series of rigid bodies (forearm, upper arm, head, trunk, shank, etc.) that can move relative to each other and then secondly consider their local deformation and failure.

Formulating problems in biomechanics in terms of the **Centre of Mass** (CoM) simplifies the problem to a single point. The CoM is the point on the rigid body that moves in the same way that a single particle containing all the mass (m) of the object and is subjected to the same external forces would move. Therefore, when analysing the body, we consider this as a series of rigid bodies, each with a constant CoM. Clearly, the whole person will have a varying overall CoM when independent movement of each of the joints is taken into account and this is where rigid body biomechanics moves into the realm of dynamics.

The **Centre of Gravity** (CoG) is the point at which the weight (W) of the body or system can be considered to act. In other words, it is the point at which the weight of the body, $W = mg$, should be applied to a rigid body or system to balance exactly the translational and rotational effects of gravitational forces acting on the components of the body or system. The CoG and the CoM are coincident when gravity (g) is constant and are thus frequently used interchangeably. These two points are important in biomechanics as they are the reference points for calculations.

The **Moment of Inertia** is the rotational equivalent of mass in its mechanical effect; it is the resistance to a change of state (a speeding up or slowing down) during rotation. This is dependent on the mass of the object and the way the mass is distributed

$$I = mr^2$$

where m is the mass and r is the distance of the CoM of the object from the axis of rotation.

The SI unit of inertia is the kilogramme metre squared (kg m^2).

The moment of inertia is especially important in high loading events where the forces that are applied to the person are not applied directly through the CoM. This means that the forces produce a turning effect and thus cause the person, or body segment, to rotate. A high moment of inertia will resist that rotation. For example, a coat hanger that supports two heavy bags hooked in the middle of the hanger will have a smaller resistance to rotation than a coat hanger with the same two heavy bags hooked on either side of the hanger.

A **Torque** or **Moment** is the turning effect of a force (Fig. 3.1). This is calculated in mechanics by multiplying the magnitude of the force by its **moment arm**, or **lever arm** which is the perpendicular distance from the point of application of a force to the axis of rotation. Therefore, a large force and a large moment arm will result in a large moment. As the forces in blast are extremely high, only a small moment arm will result in a large moment and thus cause angular (or rotational) acceleration of the person or body segment. In vector terms, the calculation is the vector (cross) product of force and distance. The SI unit of moment is the Newton metre (N m).

Scalar quantities in biomechanics have magnitude only. For example, mass, length, or kinetic energy (described later) are scalar quantities and can be manipulated with conventional arithmetic.

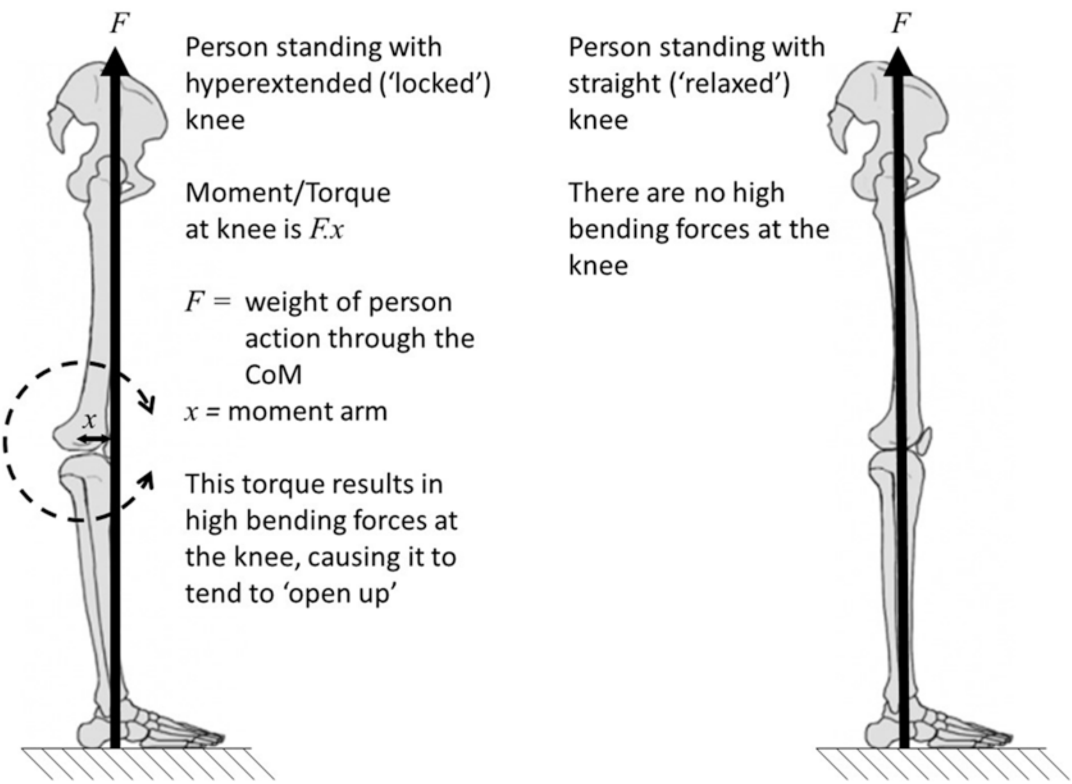


Fig. 3.1 Moment at the knee joint for different loading conditions

Vector quantities in biomechanics have both direction and magnitude (Fig. 3.2). A force, for example, is always described by its magnitude and by the direction in which it is acting. Velocity is also a vector quantity because it expresses the rate of change of position in a given direction. This is experienced when going round a corner; a racing driver going at a constant speed (scalar) round a corner has a changing velocity (vector), because the direction of travel is moving.

This means that when performing calculations with vectors, ordinary arithmetic will give the wrong answers; therefore, vector addition and other types of vector algebra must be used.

Acceleration is the rate of change of velocity with respect to time (the first time derivative of velocity or the second time derivative of displacement). Acceleration is a vector quantity. Taking the racing driver described above, the constant speed going round a corner reflects a change in velocity with respect to time and is, thus, an

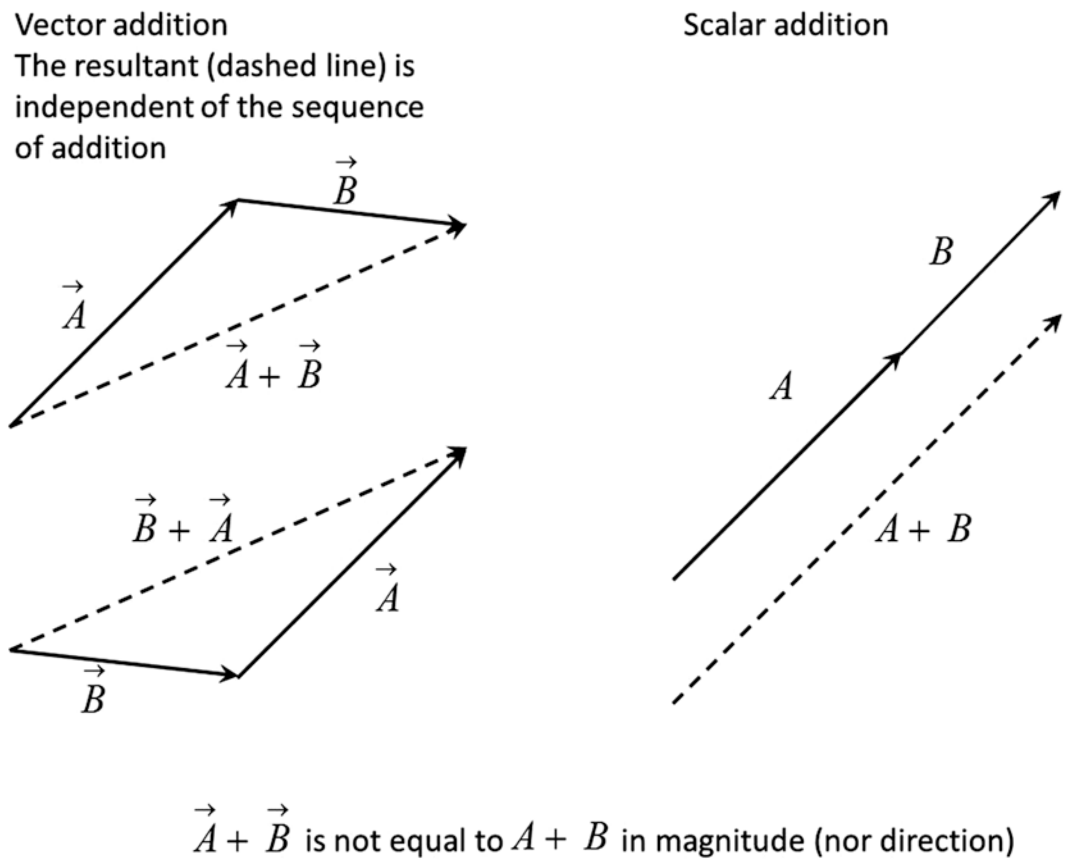


Fig. 3.2 Vector and Scalar addition

acceleration around the corner. A passenger in the car will experience no acceleration when going in a straight line (constant speed and constant velocity), but will experience the acceleration going around the corner as a tendency to be moved away from the corner sideways.

(Deceleration is simply negative acceleration and so is a term that is not normally used in biomechanics.) The SI unit of acceleration is the metre per second squared (ms^{-2} or m/s^2).

Angular Acceleration is the rate at which the angular velocity of a body changes with respect to time. The SI unit of angular acceleration is radians per second squared (radian s^{-2} or radian/s^2).

3.2.2 Forces

Force is a vector quantity that describes the action of one body on another. The action may be direct, such as the floor of a vehicle encroaching on the foot of an occupant, or it may be indirect, such as the gravitational attraction between the body and the Earth. Force can never be measured directly. It is always estimated, for example, by measuring the deflection of a spring under the action of a force. Measuring force, therefore, requires some knowledge of the deformation characteristics of materials (see Chap. 4). The SI unit of force is the Newton (N).

At any one time, many forces may be acting on a body. The **Resultant Force** is the result obtained when all the forces acting are added vectorially and expressed as a single force (Fig. 3.3).

Equilibrium is when the resultant force and moment acting on a body are zero. In Fig. 3.3, if the blast force, *B*, were removed, then the forces *P*, *W* and *GRF* would be vectorially added to come to zero. Therefore, there can be equilibrium when no forces are applied to a body as well as

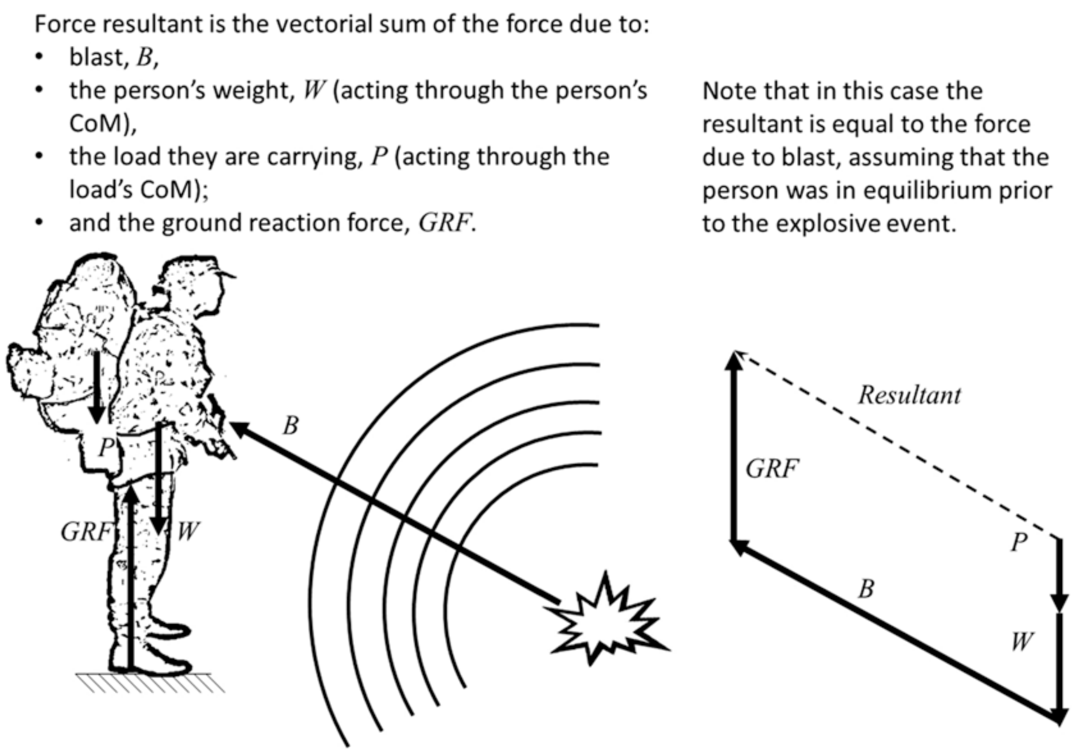


Fig. 3.3 Resultant force on a person under blast

where a combination of forces is applied to a body. Note that in blast the person, or object in proximity to the person, is rarely in equilibrium; the resultant force will be non-zero.

Weight is the force that results from the action of gravity on a mass; it acts through the CoG. Another term for this is **Gravitational Force**. When standing on weighing scales, weight is the force that the person applies on the scales when they are aligned perpendicular to the gravitational field, i.e. flat on the ground. This is the equal and opposite to the force the scales exert on the person (Newton's third law—see Sect. 3.2.3).

Contact forces are the forces between objects in physical contact. The description of the force between the person and the weighing scales is a type of contact force.

Friction is a type of contact force that is often forgotten, yet can be overpowering in its effect. It is the tangential force acting between two bodies in contact that opposes motion or the tendency to motion. If the two bodies are at rest (resisting *tendency* to motion), then the frictional forces are called **static friction**. If there is motion between the two bodies, then the forces acting between the surfaces are called **kinetic friction**. Often kinetic friction is less than static friction, so when static friction is overcome, then the friction level reduces and the object accelerates. Friction is affected by many parameters, for example the roughness of the surfaces or the presence or absence of fluid (and the type of fluid or pressured state of the fluid). Biomechanical analyses need to consider these different parameters; for example, there is friction without fluid (sometimes called *dry friction*) between your foot and the floor, and friction involving fluid (sometimes called *fluid friction*) acting within your knee joint when you move.

When fluid is not involved, the ratio of the magnitude of friction to the magnitude of the normal force is called the *coefficient of static friction* and shows the following relationship: $f = \mu P$, where f is the friction, μ the coefficient of friction and P the magnitude of the normal force.

Taking the example in Fig. 3.3, the effect of friction at the feet has been neglected. For completeness, there will be static friction between the feet and the ground that will act to resist the movement of the feet to the left (backwards) and thus the effect of the blast force is to produce a friction force at the feet to the right that, combined, will result in an anti-clockwise moment on the person (Fig. 3.4).

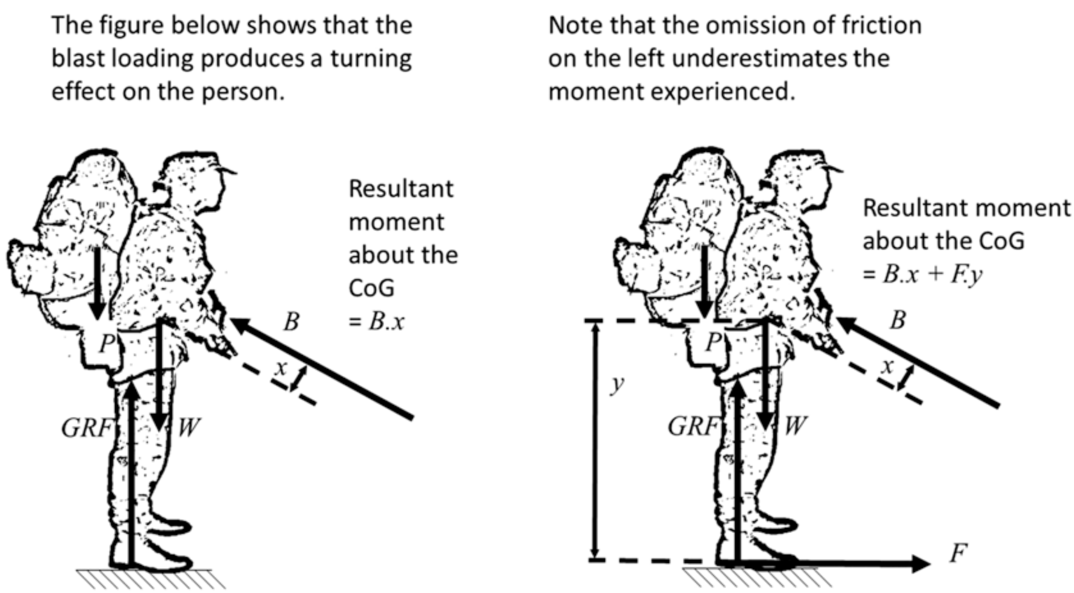


Fig. 3.4 Friction effects in blast loading. *GRF* = ground reaction force; *W* = weight of person and armour; *B* = force due to blast; *P* = weight of backpack; *F* = friction force

The **Joint Reaction Forces** exist between the articular surfaces of the joint, for example the forces between the surfaces of the tibia and femur at the knee joint. Joint reaction forces are the result of muscle forces, gravity and inertial forces (usually, muscle forces are responsible for the largest part when under normal, physiological, loading; blast is a very different case—see Sect. 3.3 and Chap. 46).

Other types of contact forces are **Ground Reaction Forces** (GRF) that specifically act on the body as a result of interaction with the ground. These are shown in Figs. 3.3 and 3.4. Newton's third law (see Sect. 3.2.3) implies that ground reaction forces are equal and opposite to those that the body is applying to the ground.

Pressure is the amount of force acting per unit area. Pressure is a scalar quantity. **Fluid pressure** is the pressure at some point within a fluid. There can be pressure within a fluid that is flowing in a pipe, and **fluid dynamics** is used to analyse this pressure and flow. There can also be pressure due to the height of fluid. For example, a diver will experience an increase in pressure when diving to greater depth; this pressure is called **hydrostatic pressure**. The SI unit of pressure is the Newton per metre squared (Nm^{-2} or N/m^2).

Centre of Pressure (CoP) describes the centroid of the pressure distribution. It can be thought of as the point of application of the resultant force, because if all the forces across a surface were summed and their resultant taken, the CoP would coincide with the position of the resultant force acting across that surface. In the more general case, the force is applied over an area (for example the plantar aspect of the foot or boot). In the example in Fig. 3.3, if we were to take away the blast loading, then the CoP at the boot would coincide with the line of action of the resultant force on the person, the GRF. Generally, when first conducting a rigid body analysis in blast, the CoP is considered as the point of force application. However, when deformations are being analysed, then it is important to also consider how the force is distributed. In biomechanics, the distribution of a contact force can be very important when considering conditions such as pressure sores, for example. Distribution of force is discussed in greater detail in Chap. 4.

3.2.3 Newton's Laws and Kinetics

Kinetics is the study of forces associated with motion. For example, kinematics and kinetics will be used to analyse the muscular contraction forces that are required to produce movement, or to analyse the sequence of movement for optimal performance. The key laws that form the basis of conventional mechanics are Newton's laws.

Newton's first law states that a body will maintain a state of rest or uniform motion unless acted on by a net force. If there is no resultant force acting on an object, then that object will maintain a constant velocity. In cases where the velocity is zero, then the object is not moving ("remains at rest"). If the resultant force is not zero, then the velocity will change because of the force. Newton's first law is often described as the Law of Inertia.

Whereas Newton's first law describes what happens when the resultant force is zero, **Newton's second law** describes what happens due to a non-zero resultant force and how the velocity of the object changes. This law defines the change in force as being equal to a change in momentum per unit time. Rearranging this definition results in the formal definition of **momentum** described below. The second law also states that the change in momentum will be in the direction of the resultant force, so you see why it is important when analysing rigid bodies to be able to quantify the resultant force. Newton's second law is often described as the Law of Momentum. The second law is frequently stated as, "Force equals mass times acceleration".

$$F = ma$$

This is rigorously derived from the formal statement of the law regarding momentum. We won't go into that here, but the mathematics works out to allow the second law to be restated as in the equation above. Newton's second law is the basis for formulating the equations of motion, the formulation of the impulse momentum relationship, and, more fundamentally, defines for us the units of force.

Newton's third law is the Law of Reaction; it states that action and reaction are equal and opposite and is shown by the earlier examples, for example, someone standing on weighing scales. The force the person applies to the scales is equal and opposite to the force that the scales apply to the person.

Impulse is the effect of a force acting over a period of time. Impulse is determined mathematically by the integral of the force-time curve, the area under the force-time curve. Impulse is important in blast as the forces due to blast vary significantly with time and often act for a very short period of time. The impulse (area under the curve) due to a Friedlander curve blast (Fig. 2.6) is very different from the impulse due to blast wave reflections inside a vehicle (Fig. 2.8). Therefore, impulse is a useful single measure that quantifies this difference without describing the full wave form. The SI unit of impulse is the Newton second (N s) that reduces to kg m/s.

Linear Momentum is the product of the mass of an object and its linear velocity. Its units are the same as those for impulse (kg m/s).

Newton's second law allows us to quantify the effect of a force on the velocity of an object. If we take $F = ma$ and multiply both sides by time, t, then we end up with $Ft = mat$. Now.

a = the change in velocity over time $= \frac{\Delta v}{t}$, so

we can rewrite our equation to be: $Ft = m\frac{\Delta v}{t}t$,

i.e. $Ft = m\Delta v$.

This is the **impulse–momentum relationship** and states that the change in momentum experienced by a body under the action of a force is equal to the impulse of the resultant force: Impulse = Change in Momentum. When calculating the effect of blast, we seek to know how the force is changing over time and integrate that over time to calculate the impulse. This then allows us to see how the momentum of the displaced person or object changes and allows a forensic examination of the blast scenario.

Angular Momentum (L) is the rotational equivalent of linear momentum; it is the "amount of motion" that the body possesses during rotation. Computationally, it is the product of the moment of inertia (I) and the angular velocity (ω):

$$\text{angular momentum } L = I\omega$$

Because angular momentum is a vector quantity, it can be resolved into components. It is possible to have angular momentum about one axis and none about another. Through algebraic manipulations, angular momentum can be transferred from one axis to another.

A projectile (for example a bullet or blast fragment) may have both angular and linear momentum. As it travels, it will have mass and linear velocity. However, it might also be spinning about its long axis (this is often desirable to reduce sideways turning of the projectile) which means that it will also have angular momentum about its long axis. In addition, the projectile may be tumbling end over end, meaning that it has angular momentum about an axis perpendicular to its long axis. These two components of angular momentum can be summed algebraically to give its total angular momentum. The SI units of angular momentum are kg m^2/s.

Work is done when a force moves an object. This is strictly defined as the integral of force with respect to distance:

$$W = \int F dx.$$

If the force is constant, then $W = Fd$, where d is the distance over which the force acts. Note that the definition of work is independent of time. Thus, the same amount of work is done in going up the stairs slowly or quickly. The power in these situations is not, however, the same. The SI unit of work is the Joule (J).

Power is the rate of doing work—the derivative of work as a function of time:

$$P = \frac{dW}{dt}.$$

Average power is equal to the work done divided by the time during which the work is being done: $P = \frac{W}{t}$. The SI unit of power is the Watt (W).

Energy is the capacity for doing work. In any system, this capacity cannot be destroyed, but energy can be transformed from one form to another (this is a statement of the Principle of Conservation of Energy). There are many different forms of energy, for example kinetic energy, potential energy, strain energy (all three are defined below) and heat. The units of work and energy are the same—the Joule (J)—because of the relationship of these two quantities through the work-energy principle.

Kinetic Energy is that component of the mechanical energy of a body resulting from its motion and can be split up into two constituent forms, just like acceleration (linear and rotational), forces (force and moment) and impulse (linear and angular momentum).

These are **kinetic energy of translation**: $KE_{xion} = ½mv^2$.

and **kinetic energy of rotation** $KE_{rot} = ½I\omega^2$.

Heat (or heat energy) is a form of kinetic energy in which particles within a material, substance or system transfer their kinetic energy to each other. It is always in defined in terms of the transfer of energy from one system to another, not in terms of the energy contained within systems.

Potential Energy is the energy of a body resulting from its position. Clearly the reference point for this is important and therefore potential energy is always quantified according to an arbitrary datum and can therefore assume any value depending on the choice of said reference point. Note that the *change* in potential energy is important in biomechanical analyses and this is independent of the choice of a reference point. Potential Energy, $PE = mgh$ where m is the mass, g is the acceleration resulting from gravity and h is the distance above the datum.

Strain Energy is the energy stored by a system that is being deformed. The energy is released when the load that is causing the deformation is removed. Although this chapter does not go into deformation in any great amount (see Chap. 4), it is worth noting that strain energy is important in some loading conditions, for example the energy contained within a bungee cord when fully stretched, or the energy stored in a trampoline with someone standing on it.

The **Work-Energy Principle** states that the work done on a body is equal to the change in the energy level of the body. This principle is used widely in forensic blast analysis where the work done is visible and can be analysed through the structure or person's deformation or gross movement and thus the change in energy can be quantified. $W = \Delta KE + \Delta PE = (KE_f - KE_i) + (PE_f - PE_i)$, where f represents the final state and i the initial state.

Fluid Mechanics is the study of forces that develop when an object moves through a fluid medium. This medium can be a fluid like blood, but also very importantly in blast, this can be air (if free field blast, for example), or water (torpedo strikes, for example). In many cases of loading fluid forces have very little effect on kinetics and kinematics, yet in other cases, these can be significant. For example, a shot put will not experience significant forces due to the air, yet a shuttlecock in badminton will be significantly affected by the air through which it travels.

Drag is one of the most important forces produced by a fluid. It is the resistive force acting on a body moving through the fluid. The **surface drag** depends mainly on smoothness of surface of the object moving through the fluid. Practical examples of this include shaving the body in swimming or wearing racing suits in skiing and speedskating to reduce the surface drag. **Form drag** depends mainly on the cross-sectional area of the body presented to the fluid; this is why cyclists have a crouched position rather than sitting upright when trying to go fast.

Because fluids can flow, the influence of the fluid on a body moving through it depends not only on the body's velocity but also on the velocity of the fluid. Walking headlong into a stiff wind requires more force than standing still facing the same stiff wind. Walking with the wind reduces the force required. This is amplified to an extreme level in primary blast where the fluid (air) travels at extremely high speeds.

Of course, the human body is filled with different fluids and therefore fluid mechanics effects are apparent internally as well as externally.

3.2.4 Functional Anatomy

In biomechanics, the term **functional anatomy** is used to describe the physical function of the biological structure of interest. This means that this requires knowledge of the loading on the structure, the constituent materials of the structure and their shape. In the human body, the constituent materials are complex. For example, biological fluids include protoplasm, mucus, synovial fluid and blood; these are described in more detail in Chap. 4. Solid material constituents include actin, elastin and collagen. How these combine to give material properties are described in Chap. 4.

An example of functional anatomy is that of articular cartilage which describes the interrelationship between its shape, constituent materials, loading and motion environment, wear, deformation and pathology.

Articular Congruency is the description of how the two surfaces of a joint overlap one another. A hip joint, for example, is highly congruent in that the full surface of the spherical femoral head is in contact with the full surface of the acetabulum. This congruency allows efficient transfer of load from one articular surface to the other. The glenohumeral joint of the shoulder is incongruent in that the spherical humeral head has a very small contact area with the virtually flat glenoid. Other terms used are "conforming" and "non-conforming". (Note that there is a physiological reason for this difference in congruency; the hip joint has a smaller range of motion than the glenohumeral joint.)

3.3 Biomechanics of Force Transmission

There are multiple sources of force and deformation in the human body. Within blast, the main sources are gravity (weight), posture (deformation of tissues due to the seated/standing position), pressure (through the shock wave) and impulse (through contact with an external agent). These *external* sources result in *internal* forces, stresses and deformations.

Newtonian mechanics are used to understand these internal forces, and then fluid mechanics and stress analysis are used to understand stresses, deformation and flow. The construct used to understand these forces is a **Free Body Diagram**. In this, the region of interest is outlined, and all the external forces acting on that region are identified and quantified (Fig. 3.5). Frequently such analyses are conducted where the regions of interest are in equilibrium. As described previously, in blast, the person is rarely in equilibrium and is moving and therefore a free body diagram analysis must be conducted at different time points.

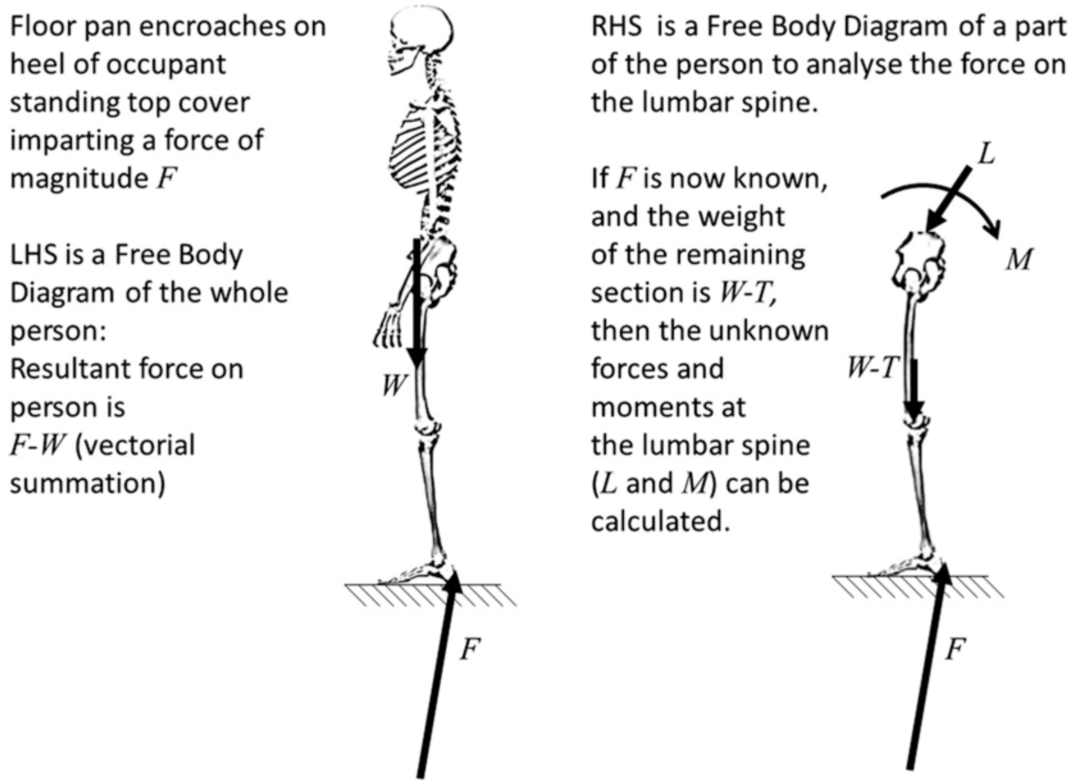


Fig. 3.5 Different free body diagrams under the same blast loading situation at time $t = 0$. These demonstrate that the appropriate choice of a free body diagram can allow internal forces to be quantified

3.3.1 Muscle Forces

The main purpose of a muscle is to rotate a joint against a force. This rotation is produced by a *torque*, or *moment*, therefore, the joint torque is the key mechanical parameter that muscles need to produce. Muscles attach very close to the joint centres of rotation and therefore their *lever arm* is small. Normally the external force is applied further away from the joint centre of rotation and therefore it has a large *lever arm.* Using Newtonian mechanics, we find, therefore, that forces in muscles during normal movement are orders of magnitude greater than the externally applied force.

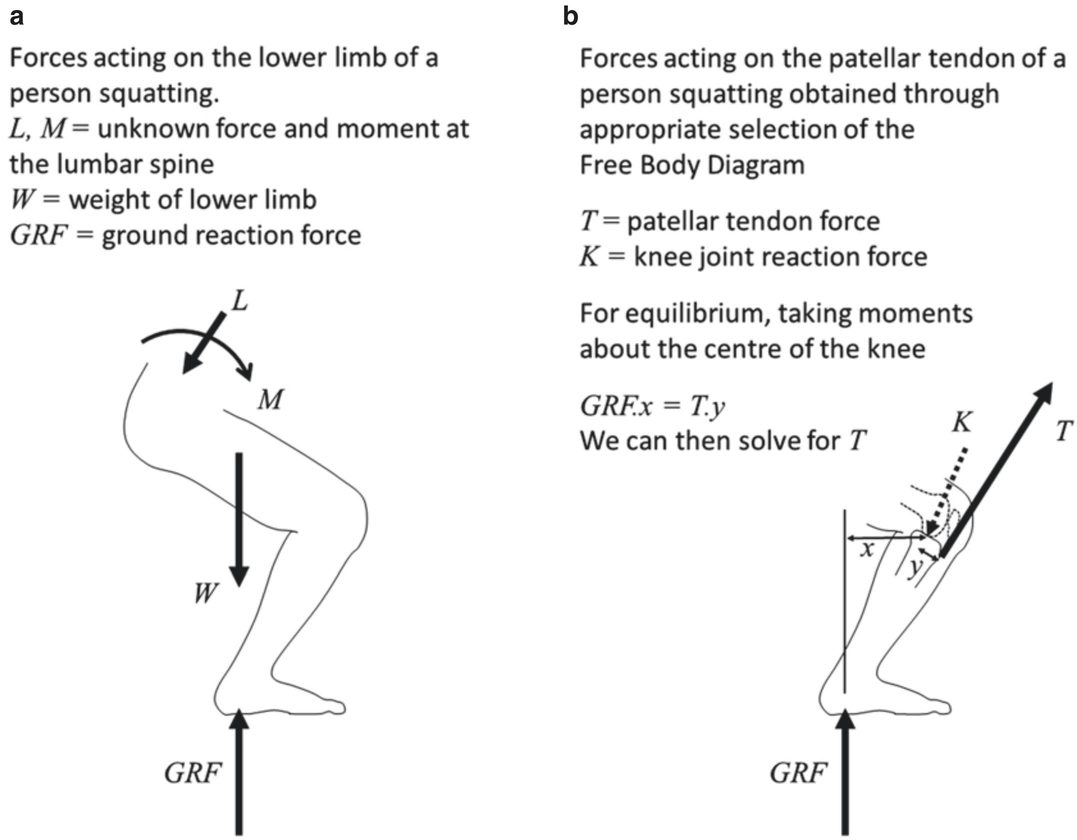
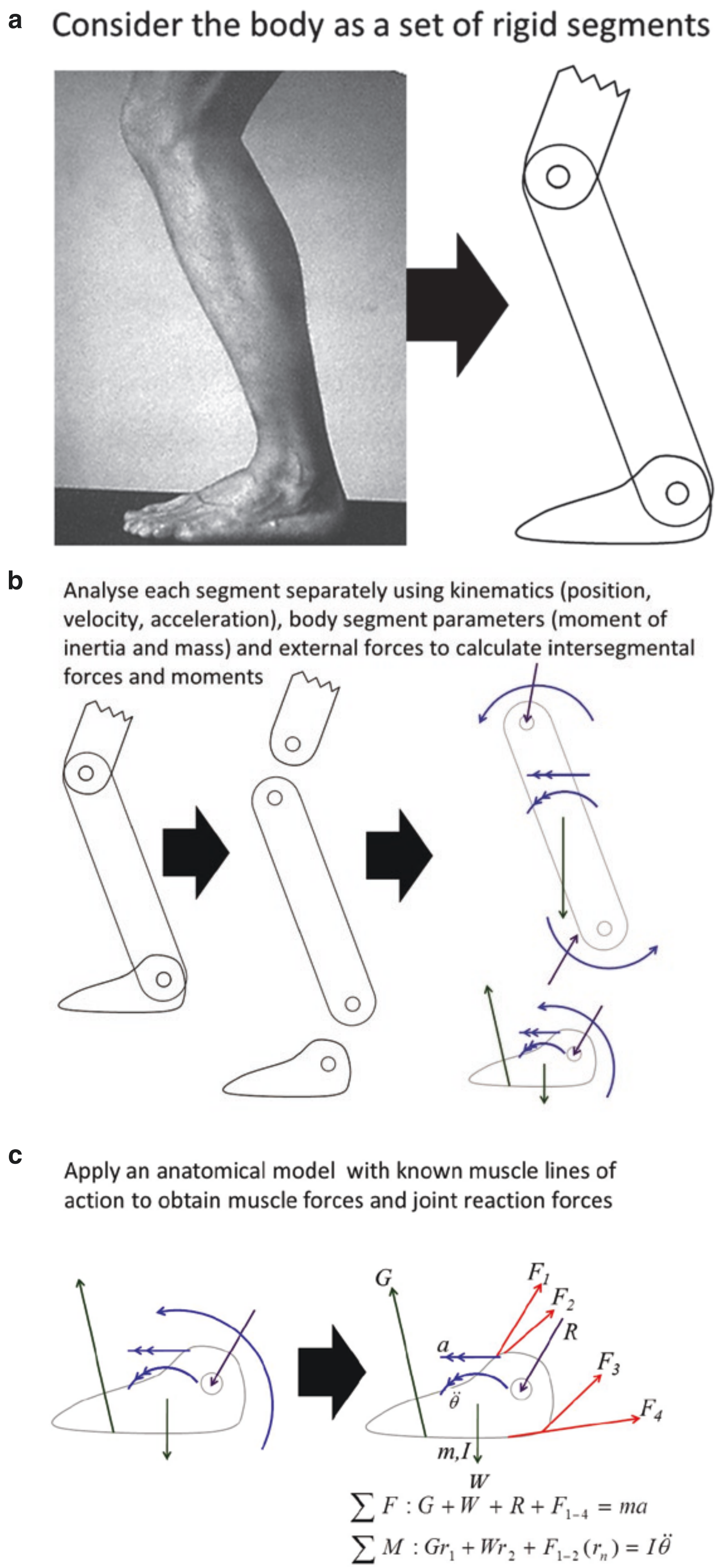


Fig. 3.6 (**a**) Forces acting on the lower limb when squatting. (**b**) Force in the patellar tendon in a simple squat. The force is much higher than the external load due to the lever arm effect ($x > y$)

Musculoskeletal Dynamics is the engineering tool that is used to quantify muscle forces during activities. Because very few loading scenarios can be simplified to that in Fig. 3.6 in which only one muscle force acts and an analytical solution to the problem can be found, other approaches need to be taken to quantify muscle forces in more complex situations. Musculoskeletal dynamics uses in biomechanics are described in this chapter and outlined in the flow chart in Fig. 3.7.

Fig. 3.7 Musculoskeletal dynamics: from motion to muscle forces. (**a**) Rigid body segments; (**b**) intersegmental analysis; (**c**) muscle force distribution



Initially kinematics is used to quantify the motion of the person. The motion of each part of the person, each segment, is analysed separately. The data input for this can be video analysis or the use of optical motion tracking such as is used in the computer graphics industry. The output of this is the position, velocity and acceleration (linear and rotational) of each body segment.

This information is combined with estimated knowledge of the mass and moment of inertia of each part of the person (each "segment") as well as estimates of the external forces acting on each segment. For example, knowledge of the deformation of the floor pan of a vehicle can give an estimate of the force that such a deformation would apply to the foot segment. All of this information is brought together using inverse dynamics in which knowledge of the kinematics and forces on the body are applied to quantify the moments and forces between each body segment. These are called "intersegmental forces and moments".

Finally, the intersegmental forces and moments are combined with an anatomical model of the physiological joint (the ankle, in this case) in which the articular geometry, ligament geometry and muscle lines of action are known. The equations of motion are then solved whereby the sum of all the muscle and ligament forces gives the true joint reaction force and then the sum of all the moments due to the muscles (the product of the muscle force times its moment arm about the joint centre of rotation) gives the intersegmental moment which is known from the previous step in this technology. What is apparent is that there are many more unknowns than equations to solve the unknowns as there are many muscles crossing each joint. Numerical methods, termed "optimisation techniques", are used to solve this set of equations and the output of this is the muscle forces, ligament forces and joint reaction forces at the joints of interest [1]. There are three leading software technologies available to conduct musculoskeletal dynamics analysis (OpenSim [2]; AnyBody [www.anybodytech.com/]; Freebody [3, 4]).

3.3.2 Forces in Joints

Once the muscle forces during a loading activity are known (as shown above), then the loading on all the other tissues of the joint can be characterised using additional free body diagrams. The example below (Fig. 3.8) shows how understanding of the muscle forces at the knee allows quantification of the loading in the anterior cruciate ligament (based on its geometry) and the articular cartilage.

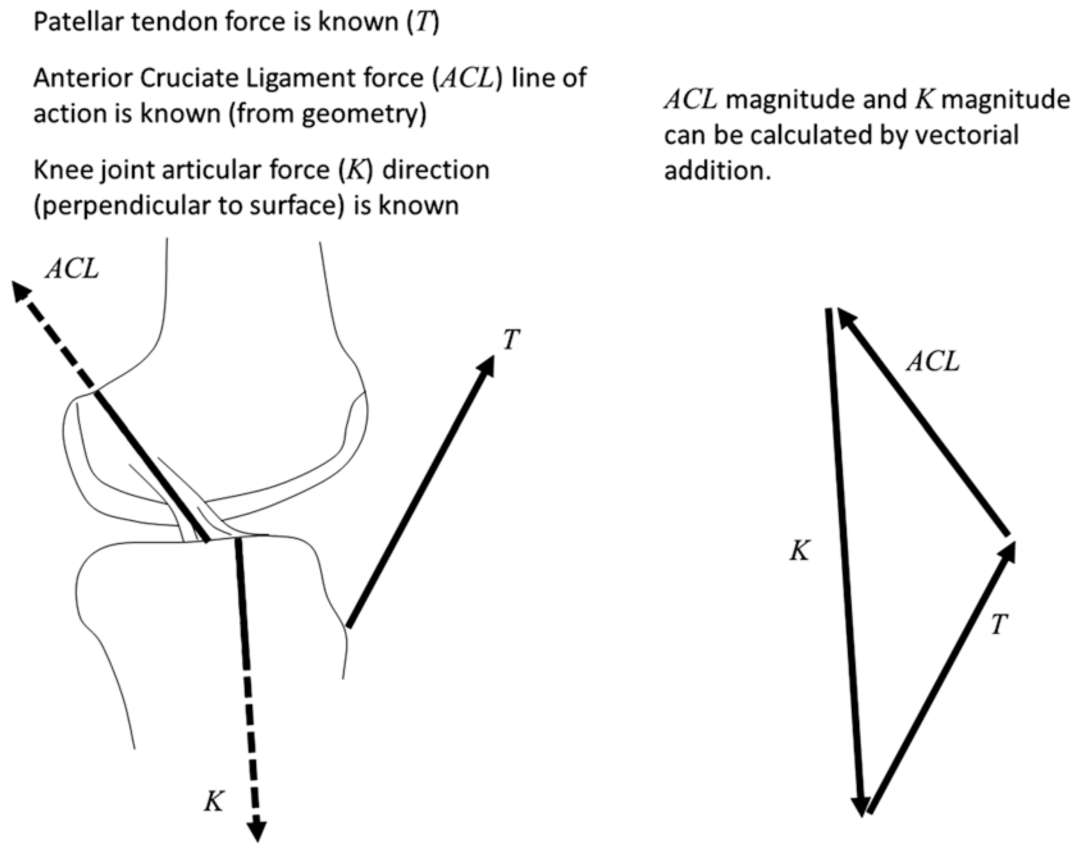


Fig. 3.8 Forces in tissues of the knee joint

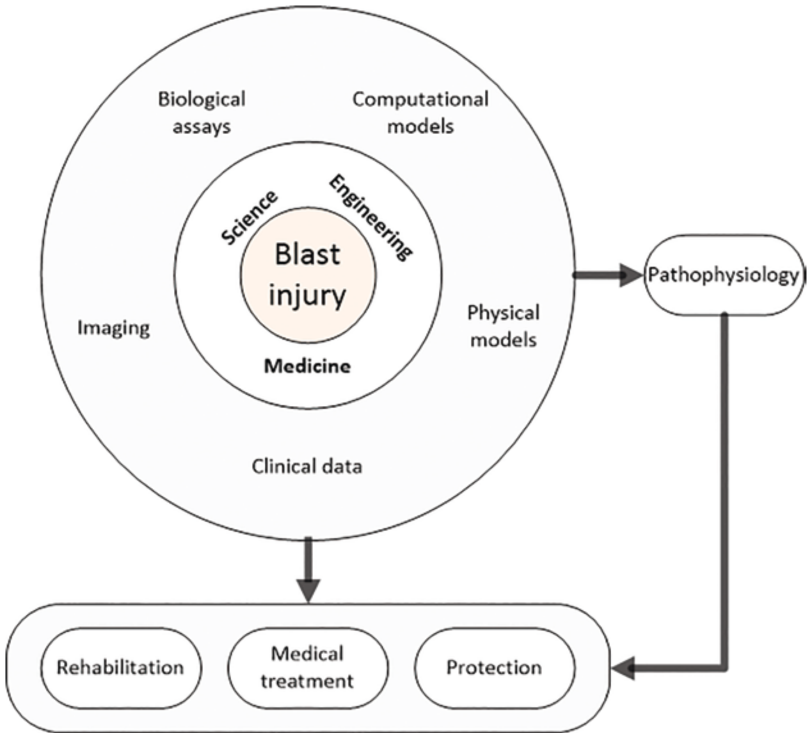
The cartilage loading at the knee is then distributed over an area. A congruent joint will distribute that over the full articulating area; however, an incongruent joint will distribute that over a very small area. The analysis of how forces are distributed within a tissue or structure is called **stress analysis** and is described in Chap. 4.

What will become apparent in this simple analysis is that, because of very complex three-dimensional structures, it becomes impossible at some point to analyse internal forces in the human body without some computational help. In addition, as the structures deform, so the loading on them changes due to the geometrical change. At this point, computational biomechanics takes over. In computational biomechanics, the same methods are used as described here, but numerical techniques are used to solve the free body diagrams for all tissues and all *parts* of tissues. At this point, loading analysis and stress analysis (that accounts for deformation as well) become the same and this is the subject matter described in Chap. 5.

3.4 Bringing it All Together: Forensic Biomechanics of Blast

The basic biomechanics described in this chapter allows for a simple analytical framework to be devised for analysing loading on the human body due to blast. Although the framework is simple, the application of this is not so straightforward due to lack of information about key aspects of the blast situation. Despite these deficiencies in data, the literature has countless examples [5] of the use of biomechanics to analyse blast loading and many of these are presented in this book. Generalising the approaches taken in the literature to a single framework is nigh on impossible and therefore the tactic taken in this book follows closely that presented in Ramasamy et al. [6] in which incident and clinical data is combined with biomechanics utilising imaging, computational models and physical models in order to understand pathophysiology of blast injuries to then devise better protection, mitigation, medical treatment and rehabilitation (Fig. 3.9).

Fig. 3.9 Research approach for analysing blast injuries with biomechanics as the core discipline for much of the work



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4 Behaviour of Materials

Spyros D. Masouros and Daniel J. Pope

Abstract

This chapter presents the fundamental principles surrounding materials and their behaviour under load as relevant to blast. It begins by describing engineering and biological materials. Basic principles of the analysis of material behaviour follow, introducing the reader to terms such as stress and strain and the types of behaviour that different types of material might exhibit. The mechanical response of materials under static and dynamic loading is presented and basic tools for their analysis are introduced, including for estimating failure.

4.1 Introduction

The aim of this chapter is to introduce the reader to the mechanical behaviour of materials. The term mechanical behaviour refers to the response of materials to load; under load the material will deform and possibly break. The behaviour of a material should not be confused to that of a structure. The behaviour of a structure depends not only on the materials of which it is made but also on its dimensions. In order for one to understand or predict the behaviour of a structure, one needs to understand first the behaviour of the materials of which it is made. This will be the focus of the chapter.

4.2 Materials

Advancements in material science and engineering have been the vehicle of success in industry for solving human-centred problems and improving the quality of life. A traditional classification of types of materials is the following:

- Metals
- Non-metallic inorganic materials (for example ceramics and glass)
- Organic materials (polymers)
- Composite materials (combinations of the above)
- Biological materials (living tissue)

The first four categories comprise a selection of materials used in industry. An integral part of the design process for products, equipment and infrastructure is the selection of such materials as appropriate for the application. The study of

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these materials has been extensive with composite materials being the latest addition to the list in the twentieth century (Fig. 4.1). In the second half of the twentieth century, human injury and disease and the interaction of the body with industrial materials have spawned the comprehensive study of biological materials and their behaviour.

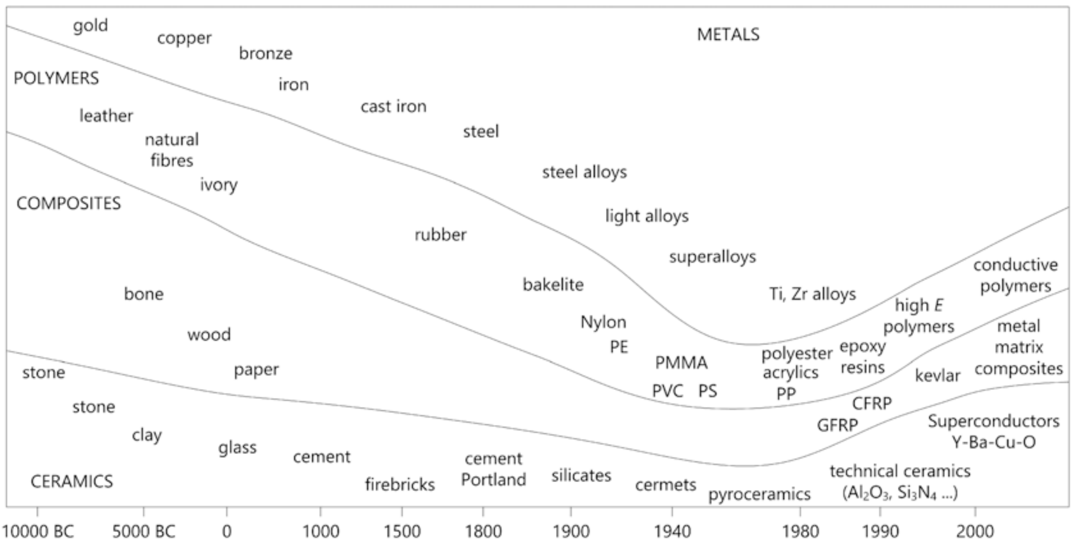


Fig. 4.1 Relative use of technical materials through the years. Adapted from Παντελής ΔΙ, 1997; *Μη μεταλλικά τεχνικά υλικά*

4.2.1 Metals

Metals are inorganic substances that consist of a highly ordered microstructure. All metals have similar physical and material properties, exactly because of their 'metallic' microstructure. Some of these properties are summarised below:

- They are solids at room temperature (except for mercury; Hg).
- They have high density.
- They are good conductors of heat and electricity.
- They reflect all visible wavelengths of light, and so appear white (except for copper, Cu and gold, Au).
- Most of them are magnetic to some degree.
- They are ductile.
- They have good machinability.

4.2.1.1 Microstructure

The metallic microstructure is formed of *crystals*. These are periodic arrangements of atoms closely positioned next to one another in a particular way to form a lattice.

The potential geometric arrangements of atoms in crystals and of crystals in the bulk metal depend primarily on the thermodynamics of the forming process as the atoms will seek the conformation with the lowest energy. Fourteen arrangements have been documented, but most metals commonly form in one, or combinations of, the following three; bcc (body-centred cubic), fcc (face-centred cubic) and hcp (hexagonal close packed). In the bcc arrangement, the unit cell has atoms at each corner of a cube plus one at the centre of the cube; in the fcc arrangement, the unit cell has atoms at each corner of a cube and at the centres of all cubic faces; the hcp arrangement is similar to the fcc but in a hexagonal formation rather than in a cubic.

4.2.1.2 Imperfections

The crystallic structure is never perfect; it contains imperfections that dictate some of the material's behaviour, especially in relation to plasticity, failure, corrosion, electric conductivity and alloying. These imperfections could be point, line, plane or complex (3D) defects. Point defects are of intra-atomic dimensions; they could be vacant atomic sites or extra (usually foreign) atoms positioned between atomic sites; they both result in distortion of the planes and the lattice arrangement. Line defects are also termed dislocations. They split the structure into two perfect crystals. They form during the solidification or during plastic deformation of the metal. This type of defect is also present in ceramics and polymers. The deformation mechanisms and therefore the plasticity and strength of materials are directly related to the formation of dislocations, as slip between crystal planes result when dislocations move. The greater the ability of a dislocation to move, the more ductile the material is. The density of dislocations (dislocation length per unit volume) in a metal that contains very few of them is at the order of 10^3 cm/cm^3, and when the metal is deforming plastically, it could be up to 10^{12} cm/cm^3. A plane defect present in all (polycrystalline) metals is the boundary of grains; grains (crystals) form during the solidification process and position themselves one next to the other to form a continuous, solid material. Crystallographic orientation, size and relative placement of grains all constitute some form of imperfection that affect the macroscopic mechanical and physical properties of the solid.

4.2.1.3 Hardening

The number and type of imperfections can be controlled in part by appropriate mechanical and heat treatments. It is usually desirable to increase the strength and toughness of a metal to make it withstand heavy loads for long times; but this is sometimes associated with reduction in ductility which is usually undesirable. The most common methods used to manipulate the imperfections (usually reduce the ability of dislocations to move) and, therefore, the macroscopic mechanical properties of metals are the following:

- Grain size reduction
- Cold working/strain hardening
- Solid solution strengthening and alloying
- Precipitation hardening and ageing
- Transformation hardening (for steels only)

4.2.1.4 Main Industrial Alloys

An alloy is a material that consists of two or more substances of which one is a metal. The most commonly used industrial alloys are ferrous alloys (steel and cast iron), copper alloys (brass and bronze), alloys of light metals such as aluminium (Al), titanium (Ti) and magnesium (Mg), superalloys such as alloys of nickel (Ni) and cobalt (Co), and alloys of zinc (Zn) and of lead (Pb).

Steel

Steel is used today widely in the construction industry as it has excellent mechanical properties such as high strength and toughness. It is an alloy of iron (Fe) with carbon (C) at consistencies less than 2.0% by weight. Other metals may be added in the alloying process to alter as required the physical and/or mechanical properties of the end product. Mild (or carbon) steels are Fe–C alloys with no other elements for alloying. In alloyed steels, C is no more than 1% by weight and the most common alloying elements are Ni, manganese (Mn), chromium (Cr), silicon (Si) and molybdenum (Mo). With regard to their use, steels can be classified as structural steels, tool steels, stainless steels, and steels for electromagnetic applications. In stainless steel, chromium (Cr) is present at consistencies greater than 12% by weight and it is primarily responsible for their good corrosion resistance. Tool steels are alloyed with elements that easily form carbides (such as Cr, vanadium (V), tungsten (W), Mo, Co, Ni, Si); these carbides do not allow the formation of large grains thus resulting in a hard alloy.

Cast Iron

Cast irons are alloys Fe–C–Si where C is at 2–4.5% by weight and Si at 0.5–3% by weight. These alloys are relatively cheap to make and are manufactured exclusively by casting. They are typically not as strong as any of the steels.

Copper Alloys

Copper (and its alloys) was the first metal used by humans. It has excellent electrical and thermal conductivity, and so half of its global production is used to produce electrical goods. It is very malleable, ductile and corrosion resistive. The main alloys are brass (Cu-Zn with Zn even up to 50% by weight) and bronze (Cu-tin, Sn). Other Cu alloys are with Al, Sn, Ni, Zn & Ni, beryllium (Be), Si.

Alloys of Light Metals

Al, Mg and Ti are light metals as their density is relatively low. The importance is that their specific strength (max stress/density) is higher compared to other metals and their alloys. They also have good resistance to corrosion.

4.2.2 Ceramics

Historically, the term ceramic meant objects made of clay and other raw materials subjected to heat; ceramics in the form of pottery and bricks are the first man-made objects. We class as ceramics today the inorganic, non-metallic materials that are fabricated using heat and cover a wide range of chemical compositions and physical and mechanical properties. They offer advantages compared to metals such as the relatively low density, high melting point, high modulus of elasticity, low thermal conductivity, good resistance to compression, high hardness and wear and heat resistant behaviour. Disadvantages include low resistance to tension, shear, fatigue, buckling and impact, brittleness, high production costs for some, low resistance to crack propagation, and sensitivity of their microstructure and of pores on their physical properties and strength.

Ceramics consist of elements that form strong ionic or covalent bonds. In terms of structure, we can classify ceramics into ionic, that consist of metals and non-metal elements, and covalent, that consist of two non-metal elements.

In ionic ceramics, the two elements have different electric charges resulting in attracting forces that contribute to forming the bond. The most stable microstructure is seen when the cations are closer to the anions resulting in high attracting forces that form stable crystalline shapes. The most common are the face-centred cubic shape of MgO and ZrO_2 and maximum density hexagonal shape of Al_2O_3.

In covalent ceramics, every atom that belongs to a covalent bond 'shares' the electrons of the outer shell with neighbouring atoms. The resulting shape is usually cubic either crystalline with formation of chains, 2D or 3D lattices, or amorphous.

Ceramics can be classified based on their main non-metal constituent (B, C, N, O, F and Si) into 6 categories: oxides (Al_2O_3, ZrO_2 UO_2), carbides (SiC, B_4C, WC, TiC), nitrides (Si_3N_4, AlN, BN), borides (ZrB_2, TiB_2), silicides ($MoSi_2$, $TiSi_2$) and fluorides (CaF_2, LiF).

Glasses are ceramics that are worthy of special mention. All commercial glasses are amorphous 3D lattices with main constituent the stable silica (SiO_2). The glass structure is a result of rapid cooling of melted oxides. The high values of viscosity and the strong bonds that form between the silicate tetrahedrons do not allow for a crystallisation process to commence during solidification. Solidification of a glass occurs due to the gradual thickening of the liquid as a result of an increase of its viscosity due to the cooling. Other oxides may be added that will transform the silica lattice and so affect the properties of the final product depending on the intended use.

4.2.3 Polymers

Polymers consist of large molecules (molecules of large molecular weight), the macromolecules; hence their name (poly = many; meros = part). The building blocks of a polymer are chemical units of small molecular weight called monomers that bond to each other to form the characteristic long chains of the polymer; the monomer quantity may vary from 100 to 100,000 per chain.

A big advantage of polymers is that they can be formed relatively easily compared to other materials. Their production is of low cost, they can form into products of complex geometry, they can be transparent (and therefore substitute the more expensive glass), they have low density, and they have good mechanical properties. Disadvantages include pollution (not easily recyclable), inferior mechanical properties to other materials, especially metals, and that they don't work at high temperatures.

In terms of microstructure, the polymers can be crystalline or amorphous; this is dependent primarily on the rate of cooling of the melted polymer. When cooling is gradual, the chains have time to align to one another and form a crystalline-like structure (not to be confused with the crystallinity of metals). Conversely, in amorphous polymers, the chains don't have time to align due to the rapid cooling; this structure can be loosely compared with that of liquids as it is characterised by a lack of order. Under heat, the long chains of the polymers can easily slip over one another. During cooling, this movement between chains reduces and the polymer transforms gradually from a liquid to an amorphous solid state. The temperature at which this transition occurs is termed the glass transition temperature, T_g; above this temperature, the polymer behaves elastically and below it the polymer is brittle and behaves similar to glass.

In terms of their physical properties, there are three types of polymers: thermoplastics, thermosets and elastomers or rubbers.

4.2.3.1 Thermoplastics

They consist of primarily linear chains that soften and flow with heat due to the relaxation of the molecular bonds and so can be formed, they solidify after cooling. This process is reversible, which means that they may be reshaped by heating them up. Widely used thermoplastics include polyethylene (PE), polyvinyl chloride (PVC), polypropylene (PP), polystyrene (PS) and the polyamides (Nylon).

4.2.3.2 Thermosets

They consist of relatively short chains in 2D and 3D networks. They are usually amorphous. They are formed by a curing process of a resin with a hardening agent with or without heat; the process is irreversible. Main representatives of this family of polymers include phenolic polymers (bakelite), epoxy resins, aminoplasts (melamine or urea resins with formaldehyde) and polyesters.

4.2.3.3 Elastomers or Rubbers

These are usually linear polymers with branching chains. They can deform a lot under load and yet return back to their original dimensions when the load is removed; they are hyperelastic. Main representatives are synthetic and natural rubber, synthetic polyisoprene, polybutadiene, polychloroprene and the silicones. Natural rubber vulcanises when heated with sulphur. The vulcanisation process entails the formation of cross-linking between molecules that result in reinforcing the structure of the material, making it tougher, more durable and less sensitive to temperature changes.

4.2.4 Composites

Composite materials are those that consist of a combination of two or more of the abovementioned material types. The intention is to fabricate a material that has special properties that none of the aforementioned material types can achieve on its own. Of the constituent traditional materials in a composite, one is termed as the matrix and the other as the reinforcement. Depending on the shape of the reinforcing constituent material, they can be classed as fibre-reinforced composites, particulate composites and laminar composites. The strength of a composite depends on the strength of its constituents, but also on the compatibility between the two when put together.

In fibre-reinforced composites, the reinforcement could be through long, continuous fibres or through short, discontinuous fibres. The strength of these composites depends on the directionality of the fibres. In unidirectional composites, the fibres are orientated parallel to one another, adding strength in that particular direction to the composite. In multidirectional composites, the fibres can be laid in random orientations, in a woven motif, or in layers perpendicular in orientation to one another. The intention in adding fibres to a matrix is to increase the strength of the matrix material. This is why the material of the fibre usually has a high modulus of elasticity, high strength and low density. The most common materials for fibre reinforcement are glass, carbon, polymers (such as nylon, polyethylene and aramid (aromatic polyamide)), metal (boron) or other raw/ceramic materials (such as mica).

The matrix secures the reinforcement from the fibres. Under load, the stresses are transferred from the matrix to the fibres. Importantly, the matrix can interrupt the propagation of cracks that might form in the fibres whence the load is too high. The material for the matrix is usually selected to be tough, ductile, and with a high melting point, higher than the maximum intended operating temperature of the composite. The most common type of matrix is organic, either thermoplastic or, more often, thermoset, such as epoxy resins, phenolic resins and polyester resins. At high intended operating temperatures, metallic materials are necessary for the matrix. Ceramic matrices are not that common; but a special mention is due to cement. Cement is used to form concrete (by mixing it with sand and stone)—which is a particulate composite—that can then be reinforced with steel rods; most infrastructure is at least partly built out of this composite.

Particulate composites are not that common as they don’t offer the superior mechanical properties that fibre-reinforced composites do, but they are cheaper to make than fibre-reinforced composites and tend to be more wear resistant.

Laminar composites can be categorised into coatings, bimetallics, multilayers and sandwich materials. The intention in coatings is mostly to improve the wear resistant properties of the surface whilst maintaining the superior material properties of the main material. The other two types have limited uses in industry.

4.2.5 Biological Materials

The constituents of the human body from the perspective of mechanical response are almost without exception a mixture of fluid and solid phases, either organic or inorganic. We will limit ourselves here to a quick overview of constituent biological materials that might be associated with blast injury; the response of tissues and organs in blast-related loading is discussed in Part IV.

4.2.5.1 Biological Fluids

In contrast to water, most fluids in our body are non-Newtonian, which means that they vary their viscosity dependent on the force they experience and have a substantial elastic (solid) component. Disturbance in the material and physical properties—primarily viscosity—of these fluids due to disease or injury may have adverse consequences on the function of tissues. Treatment, restoration or replacement of such tissues aims at recovering the unique material behaviour of its constituents in order to enable normal function.

Protoplasm

Protoplasm (*protos* = first; *plasma* = formed object) is the collection of a cell's contents that are encapsulated within the plasma membrane. It consists of the cytoplasm and various particles suspended in it. Its viscosity is several times greater than water.

Mucus

Mucus consists of glycoproteins and water. Its job is to protect epithelial cells in vital systems of the body, including the respiratory, gastrointestinal and urogenital, from infection. It does so by trapping foreign material and so its material behaviour is affected by the properties of the foreign material. The material behaviour of the mucus produced by the sex glands in both sexes—cervical mucus and semen—is appropriate for fertilisation. The properties of the cervical mucus are affected by hormones and so they vary during the menstrual cycle. The efficiency of the swimming of spermatozoa in semen and in the cervical mucus is key for enabling reproduction and so any change in the properties of either media may affect reproducing capabilities.

Synovial Fluid

Human joints undergo cyclical loading and yet can remain clinically asymptomatic for many decades. This is due to the virtually frictionless articulation provided by the articular cartilage present at the articulating surface and the lubrication from the synovial fluid. Synovial fluid is present in cavities of synovial joints. It contains hyaluronic acid and interstitial fluid. In addition to reducing friction in the joint, it acts as a shock absorber and as a medium of transportation for waste and nutrients.

4.2.5.2 Biological Solids

Actin and Elastin

Actin is a protein present in muscle and many types of cells including leukocytes and endothelial cells. It is approximately 7–20 nm in diameter, and its tensile strength has been measured to be approximately 2 MPa. It plays a role in many important cellular processes, including remodelling, primarily via its interactions with the cell's membrane.

Elastin is a protein present in connective tissue. It is responsible primarily for resuming the original shape of the tissue after deformation and for storing elastic energy when loaded. It exhibits an almost perfectly linear elastic behaviour—the only biological tissue to do so.

Collagen

Collagen is a key load transferring compound for many tissues in the body. It is present in various combinations and forms in different tissues and may act beneficially in wound healing. Twelve types of collagen have been identified to date.

Collagen is a protein. A collagen molecule (tropocollagen) (1.5 nm in diameter) is made up of three polypeptide strands (left-handed helices) that are twisted to form a right-handed triple helix. Collagen molecules combine to make up fibrils (20–40 nm in diameter) and bundles of fibrils combine to make up fibres.

Collagen is the main structural protein in connective tissue; it combines with actin, elastin, other proteins and a ground substance (a hydrophilic gel) to form fibrils and fibres that in turn combine to form bone, cartilage, skin, muscle, ligament, blood vessels, etc. The mechanical behaviour of a tissue is directly related to its microstructure and therefore the relative arrangement of fibres, cells and ground substance.

4.3 Stress Analysis

4.3.1 Introduction: General Terms

The response of materials to loading is termed mechanical behaviour. Quantifying this behaviour allows the engineer to design a product that is fit for purpose and to understand or predict what is going to happen to an existing structure under load.

When a material deforms under a small load, the deformation may be elastic. In this case, when the load is removed, the material will revert to its original shape. Most of the elastic deformation will recover immediately. There may be, however, some time-dependent shape recovery; this time-dependent behaviour is called anelasticity or viscoelasticity.

A larger stress may cause permanent, often called plastic, deformation. After a material undergoes plastic deformation, it will not revert to its original shape when the load is removed. Usually, a high resistance to deformation is desirable so that a part will maintain its shape in service when loaded. On the other hand, it is desirable to have materials deform easily when forming them into useful parts or when conforming on adjacent surfaces to distribute loading.

Failure is the discontinuation in the integrity of a structure rendering it unable to operate as intended. Failure usually occurs as soon as a critical amount of loading is reached; repeated application of lower loading may also cause failure; this is called fatigue.

4.3.2 Stress and Strain Tensors

The effect of external loading on to a body can be quantified through internal reaction loads and deformation. We use the concepts of stress and strain in order to normalise for cross-sectional size and shape that allows us to quantify material rather than structural behaviour. Furthermore, the stress–strain behaviour is unique for a material; we call that the constitutive law from which we can define material properties, unique to that material.

4.3.2.1 Stress

Stress is a normalised measure of force. Consider a body subjected to a static external force, F (Fig. 4.2). For the body to be in static equilibrium, every part of the body needs to be in equilibrium. If we make a virtual cut somewhere along the length of the body through the cross-section, then the remaining part should be considered to be in equilibrium. The internal reaction force at the cross-section can be considered as made up of a collection of infinitely small amounts of force dF_i acting over infinitely small areas dA_i. In order to maintain equilibrium $\Sigma dF_i = F$, whilst $\Sigma dA_i = A$. We define stress (at a point) as the internal force per unit area.

$$\sigma = \frac{dF_i}{dA_i}$$

The stress on a surface is defined as the intensity of internal distributed forces on an imaginary cut surface of the body. Stress acting perpendicular to a plane is termed direct or normal stress, whereas stress acting parallel or tangential to a plane is termed shear stress; we normally use the symbol τ for shear stress (Fig. 4.3).

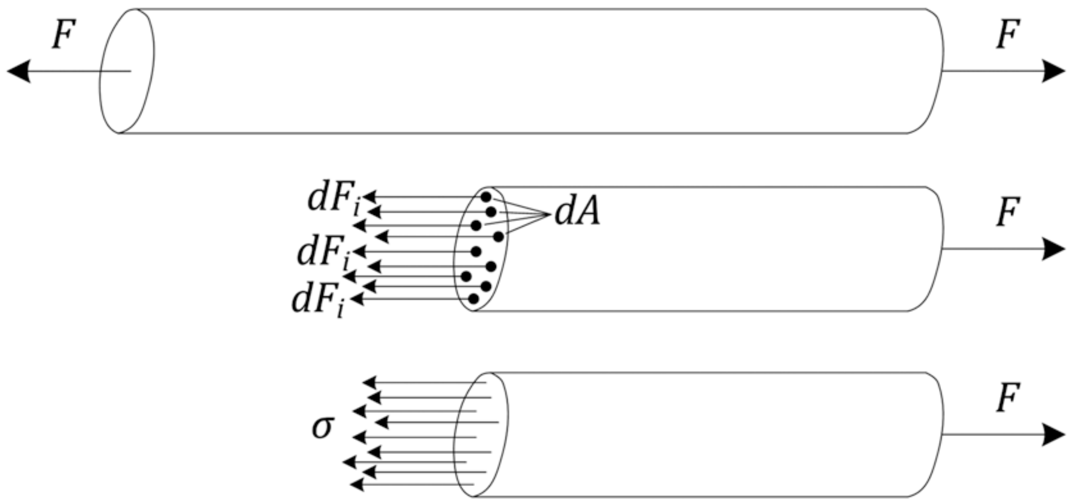


Fig. 4.2 A body with external force F acting on it with a virtual cut along its length showing a collection of infinitely small amounts of force acting over infinitely small areas the sum of which represents the stress at the cut

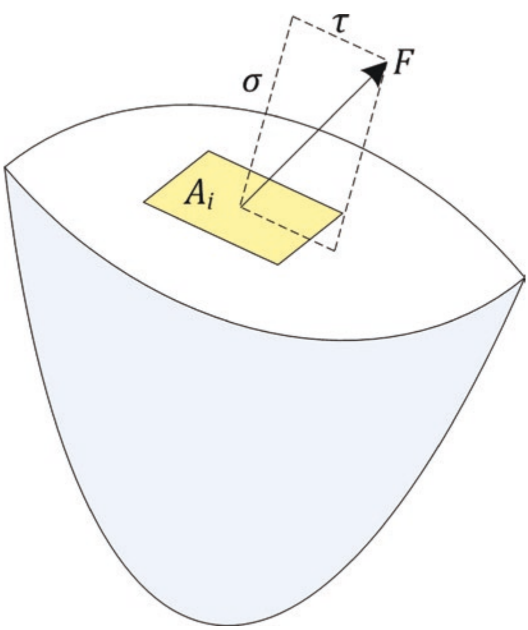


Fig. 4.3 Definitions of direct, σ and shear, τ stress on a cross-section of a body in equilibrium

$$\text{Direct stress}: \sigma = \frac{dF_{\text{direct}}}{dA}$$

$$\text{Shear stress}: \tau = \frac{dF_{\text{parallel}}}{dA}$$

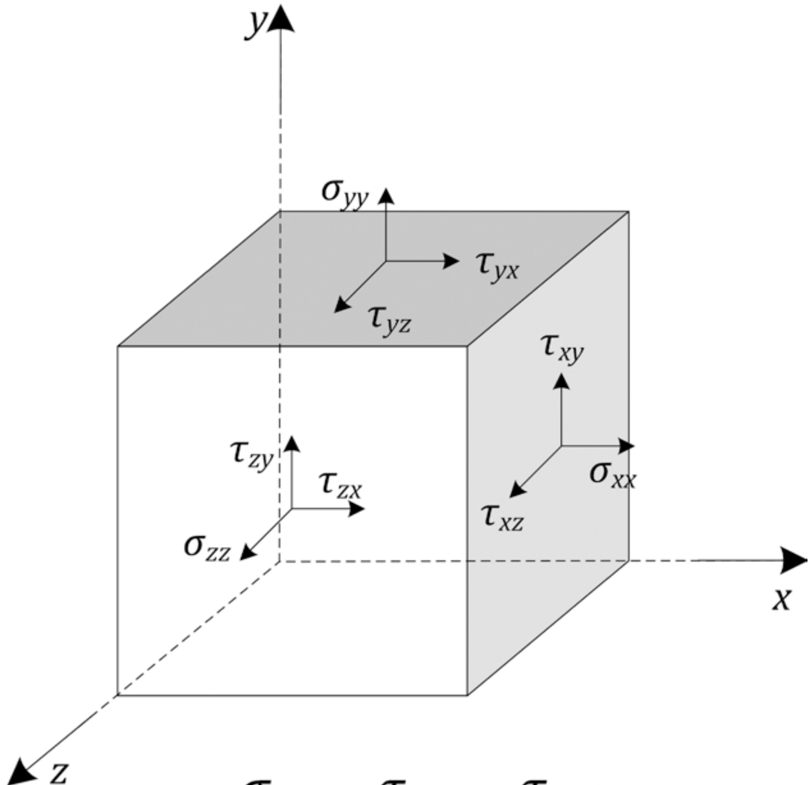
Stress is a second order tensor; this means that magnitude, direction and plane at which it acts are required to define it fully (Fig. 4.4). A vector (e.g. force), for comparison, is a first order tensor.

$$\sigma_{ij} = \frac{dF_j}{dA_i}$$

where i is the direction of the outward normal to the plane and j is the direction of the internal force component.

Therefore, stress can be represented by a 3 × 3 matrix with nine components. The diagonal elements of the matrix are the normal stresses and the rest are the shear stresses.

Fig. 4.4 The stress tensor. A unit cube representing a material point showing the components of stress on the 3 planes



$$\boldsymbol{\sigma} = \begin{bmatrix} \sigma_{xx} & \tau_{xy} & \tau_{xz} \\ \tau_{yx} & \sigma_{yy} & \tau_{yz} \\ \tau_{zx} & \tau_{zy} & \sigma_{zz} \end{bmatrix}$$

In order to maintain rotational equilibrium, the shear forces along the sides of a material point need to be equal; this means that shear stresses on perpendicular planes need to have the same sign and magnitude; we can write $\tau_{ij} = \tau_{ji}$; we call these complementary shear stresses. Therefore, the stress tensor is a 3 × 3 symmetric matrix and has only six independent elements. It is convenient to write the matrix as a column vector with the six independent elements.

A convenient way of representing a triaxial stress state, which is used often in stress analysis calculations, is by using the so-called principal stresses. There are three orthogonal planes where the shear stress is zero; these are termed *principal planes*. The magnitude of the traction (stress perpendicular to a plane) on each principal plane is a *principal stress*. These are the eigenvalues of the stress tensor. The characteristic equation is $(\boldsymbol{\sigma} - \sigma\mathbf{I})\mathbf{n} = \mathbf{0}$ ($\mathbf{n}$ the 3 unit normals).

For nontrivial solutions, $\det(\boldsymbol{\sigma} - \sigma\mathbf{I}) = 0 \Rightarrow$
$\sigma^3 - I_1\sigma^2 + I_2\sigma - I_3 = 0$

where I_i are the *stress invariants* and are independent of (invariant to) the coordinate system. The roots of the cubic equation are the principal stresses σ_{I}, σ_{II}, σ_{III} ($\sigma_{\mathrm{I}} > \sigma_{\mathrm{II}} > \sigma_{\mathrm{III}}$), and the invariants can be defined as

$$I_1 = \mathrm{tr}(\sigma) = \sum \sigma_{ii} = \sigma_{\mathrm{I}} + \sigma_{\mathrm{II}} + \sigma_{\mathrm{III}}$$

$$I_2 = \frac{1}{2}\left[\left(\mathrm{tr}(\sigma)\right)^2 - \mathrm{tr}\left(\sigma^2\right)\right] = \sigma_{\mathrm{I}}\sigma_{\mathrm{II}} + \sigma_{\mathrm{II}}\sigma_{\mathrm{III}} + \sigma_{\mathrm{III}}\sigma_{\mathrm{I}}$$

$$I_3 = \det(\sigma) = \sigma_{\mathrm{I}}\sigma_{\mathrm{II}}\sigma_{\mathrm{III}}$$

4.3.2.2 Strain

Strain is a normalised measure of deformation. As with stress, strain is a second order tensor and can be split in normal or direct and shear components in perpendicular and tangential directions, respectively. Let's consider an infinitesimal material plane element, dx, dy in dimensions, that displaces and deforms (Fig. 4.5). If u and v are the displacements in x and y respectively, then the deformations are $\Delta u = u + \frac{\partial u}{\partial x} dx$ and $\Delta v = v + \frac{\partial v}{\partial y} dy$. We define the direct strains in x and y as the normalised deformation along each direction; $\varepsilon_x = \frac{\partial u}{\partial x}$ *and* $\varepsilon_y = \frac{\partial v}{\partial y}$, respectively. We define shear strain on the plane as the change in angle; the initial right angle in Fig. 4.5 has changed by $\gamma_x + \gamma_y$, where $\gamma_x = \frac{\partial u}{\partial y}$ and $\gamma_y = \frac{\partial v}{\partial x}$.

Then shear strain in the x-y plane is defined as

$$\varepsilon_{xy} = \varepsilon_{yx} = \frac{1}{2}\left(\gamma_x + \gamma_y\right) = \frac{1}{2}\gamma_{xy} = \frac{1}{2}\gamma_{yx} = \frac{1}{2}\left(\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x}\right)$$

Using the tensorial representation of strain in three dimensions, we can write

$$\varepsilon = \begin{bmatrix} \varepsilon_{xx} & \varepsilon_{xy} & \varepsilon_{xz} \\ \varepsilon_{yx} & \varepsilon_{yy} & \varepsilon_{yz} \\ \varepsilon_{zx} & \varepsilon_{zy} & \varepsilon_{zz} \end{bmatrix} = \begin{bmatrix} \varepsilon_{xx} & \frac{1}{2}\gamma_{xy} & \frac{1}{2}\gamma_{xz} \\ \frac{1}{2}\gamma_{yx} & \varepsilon_{yy} & \frac{1}{2}\gamma_{yz} \\ \frac{1}{2}\gamma_{zx} & \frac{1}{2}\gamma_{zy} & \varepsilon_{zz} \end{bmatrix}$$

Similar to the stress tensor, the strain tensor has only six independent elements, has invariants, and principal values and planes.

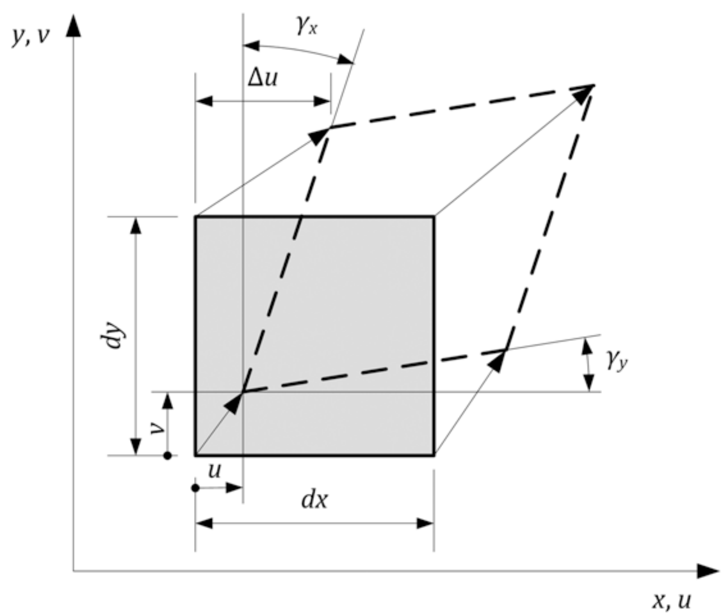


Fig. 4.5 2D strained infinitesimal element in undeformed (solid line) and deformed (dashed line) configurations

4.3.3 Stress States

There are four basic stress states under which a material could be (Fig. 4.6). Complex loading that results in complex stress states can by analysed as a combination of these four basic states.

1. *Simple tension or simple compression*. In these cases, the stresses are direct and uniaxial; there are no shear stresses.
2. *Biaxial tension*. Stresses act over two directions on every material point of the structure. A sheet being pulled equally from all directions or a closed spherical shell under gas pressure is under this stress state.
3. *Hydrostatic stress*. In solid mechanics, we use the term hydrostatic stress, σ_H instead of pressure to refer to a stress state whereby stress is equal and compressive in all directions; $\sigma_x = \sigma_y = \sigma_z = \sigma_H$. For example, an object submerged in liquid would be under hydrostatic stress.
4. *Pure shear*. A stress state whereby there are no direct stresses. When we apply torsion on a rod (moment about its longitudinal axis), then the rod is in pure shear.

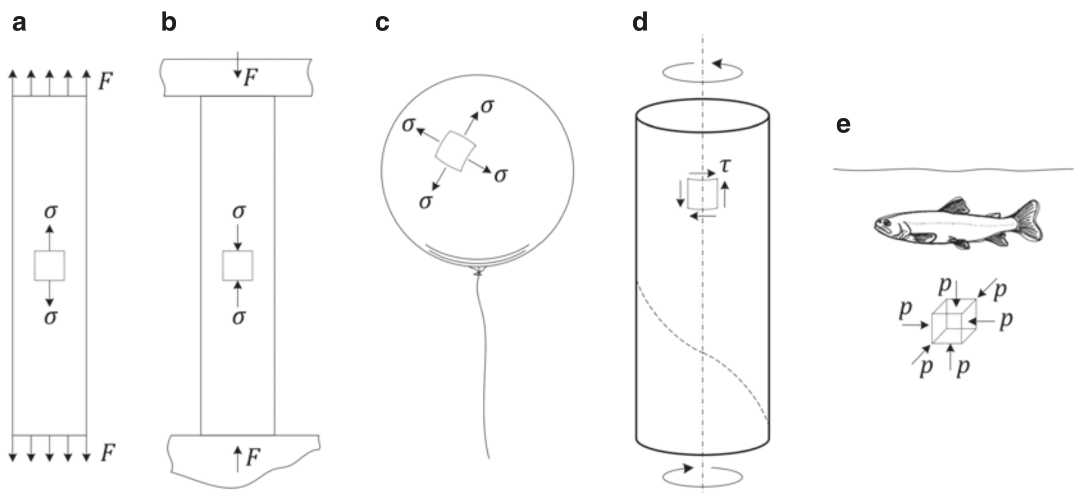


Fig. 4.6 The main stress states. (**a**) simple tension, (**b**) simple compression, (**c**) biaxial tension, (**d**) pure shear, (**e**) hydrostatic stress/pressure

4.3.4 Engineering Properties of Materials

A constitutive law is a relation between physical quantities that characterise the material behaviour in full. Each material is governed by its own constitutive law. Usually this is a form of a stress–strain relationship (Fig. 4.7). For materials undergoing small deformations (infinitesimal strains), this can be written in matrix notation as $\boldsymbol{\sigma} = \mathbf{D}\boldsymbol{\varepsilon}$, whereby $\mathbf{D}$ is a 6 × 6 matrix containing the necessary material parameters.

The first constitutive law was developed by Hooke and is known as Hooke's law; it was developed for linearly elastic solids. Every structure has a unique, unloaded state. Elasticity is the tendency of the structure to return to that unique, unloaded state when external loads are removed. Such behaviour stems from different physical properties for each type of material. For example, in metals it is the atomic lattice that changes in shape and size when load is applied and then returns to its original state of minimum potential energy when the load is removed. In most polymers, it is the polymer chains that deform under load and return to their original length when the load is removed. The energy stored and subsequently released by the structure when the load is removed is often called strain energy; this is the area under the stress–strain curve, termed strain energy density, integrated over the volume of the structure.

The stress–strain curve of linearly elastic and isotropic materials is a straight line through the origin. The slope of that line is called modulus. In a direct stress–direct strain curve, the modulus is termed Young's modulus, E and $\sigma = E\varepsilon$, whereas in a shear stress–shear strain curve, the modulus is termed shear modulus, G and $\tau = G\gamma$. In a linearly elastic and isotropic material, two material parameters are sufficient to characterise the material fully; the $\mathbf{D}$ matrix has two independent components.

Another material constant used in linear elasticity is the Poisson's ratio, v. The Poisson's ratio is defined as the ratio of transverse to axial strain; for example if an isotropic material is loaded in one direction, let's say x, then there is strain in the transverse plane; $\varepsilon_y = \varepsilon_z = -v\varepsilon_x$. There is no Poisson's effect in shear. Due to the Poisson's effect, the strain state of a linearly elastic and isotropic material when subjected to a triaxial stress state is the following and is termed the generalised Hooke's law.

$$\varepsilon_x = \frac{1}{E}\left[\sigma_x - v\left(\sigma_y + \sigma_z\right)\right] \quad \gamma_{xy} = \frac{\tau_{xy}}{G} = \frac{2\left(1+v\right)}{E}\tau_{xy}$$

$$\varepsilon_y = \frac{1}{E}\left[\sigma_y - v\left(\sigma_z + \sigma_x\right)\right] \quad \gamma_{yz} = \frac{\tau_{yz}}{G} = \frac{2\left(1+v\right)}{E}\tau_{yz}$$

$$\varepsilon_z = \frac{1}{E}\left[\sigma_z - v\left(\sigma_x + \sigma_y\right)\right] \quad \gamma_{zx} = \frac{\tau_{zx}}{G} = \frac{2\left(1+v\right)}{E}\tau_{zx}$$

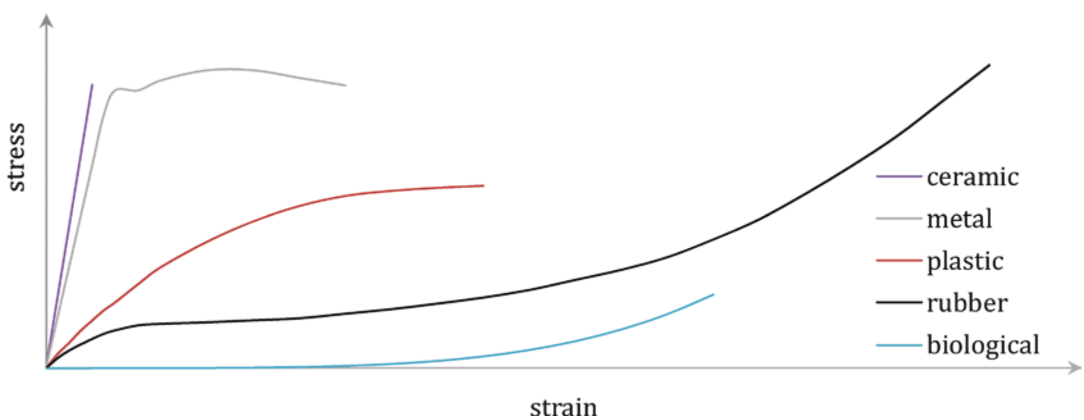


Fig. 4.7 Stress–strain curves in tension for various materials. The axes are not to scale. Note that only metals and ceramics exhibit a linearly elastic response and that the response of biological tissue is very variable depending on the type of tissue/constituent

4.4 Beyond Linear Elasticity

The most commonly used materials in the construction of useful objects have a linearly elastic and isotropic behaviour for most of their service life. There are certain materials and circumstances, however, whereby different behaviours are observed.

4.4.1 Hyperelasticity

The elastic behaviour of materials that undergo finite (as opposed to infinitesimal) deformations are called *hyperelastic* (or Green-elastic) materials. Examples of such materials are rubbers (elastomers in general) and biological tissues. The constitutive law (stress–strain relationship) for hyperelastic materials is nonlinear. For example, the 1D behaviour of a collagenous tissue may be expressed as $\sigma = A(e^{B\varepsilon} - 1)$, where A, B are material parameters that can be determined by fitting the mathematical model to experimental data.

The constitutive law for hyperelastic materials is usually expressed with a strain energy density function, W; this is the strain energy per unit volume and can be estimated as the area under the stress–strain curve. Common representations of the strain energy function are with strain invariants. The Neo-Hookean model, for example which appeared in approximately 1940, is a simple nonlinear model: $W = c_{10}(I_1 - 3)$, where c_{10} is the single material parameter to be defined experimentally and I_1 the first strain invariant. Tension and compression responses are different. It captures isotropic rubber well enough up to approximately 30% engineering strain (Fig. 4.8).

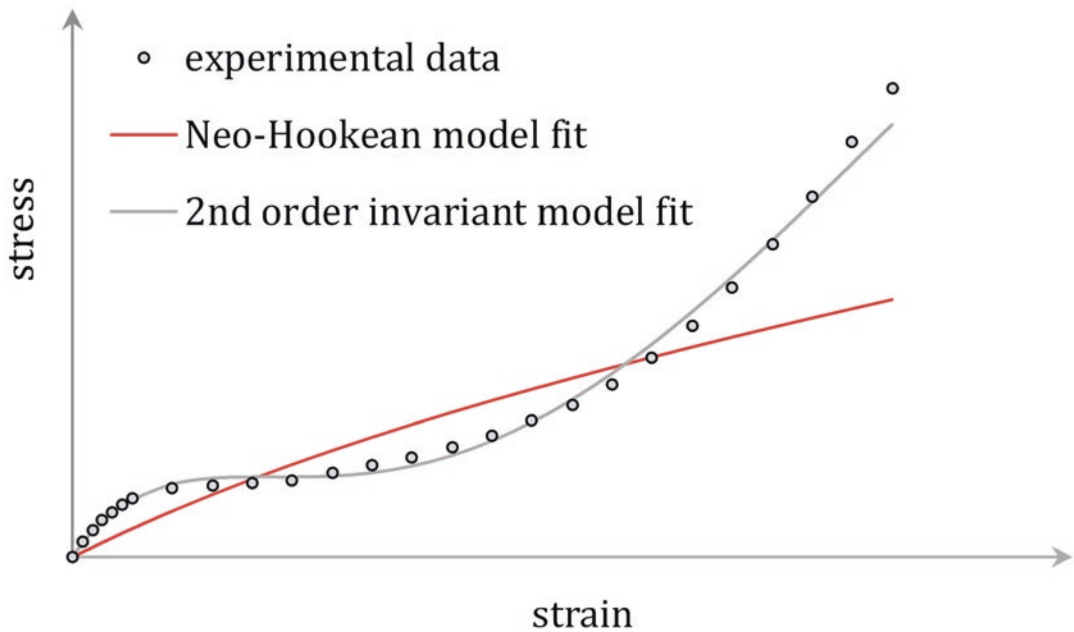


Fig. 4.8 A typical rubbery stress–strain curve with a Neo-Hookean and a second order invariant material model fit

4.4.2 Viscoelasticity

All materials have some element of time-dependent behaviour in them that can be brought out under certain loading conditions. Material behaviour which combines that of an elastic solid and a Newtonian, viscous liquid is termed viscoelastic. The resulting stress in these materials depends both on strain and on strain rate alike. Rubbers, most polymers and biological tissues exhibit viscoelastic behaviour under normal loading conditions and temperatures. A manifestation of viscoelastic behaviour is the hysteresis loop; the loading and unloading pathways for a viscoelastic material are not the same, thus forming a loop in the stress–strain material response (Fig. 4.9a). Viscoelastic materials exhibit creep (Fig. 4.9b) and stress relaxation (Fig. 4.9c). Creep is the phenomenon whereby when stress is kept constant then strain increases with time. Stress relaxation is the phenomenon whereby when strain is kept constant then the stress decreases (relaxes) with time.

The simplest material models for viscoelastic materials are the Maxwell and Kelvin–Voigt models, which represent the material as a combination of a spring and a damper; the spring represents the solid part and the damper represents the viscous part. In the Maxwell model, these are connected in series whereas in the Kelvin–Voigt model in parallel.

$$\text{Kelvin – Voigt model} \quad \sigma(t) = E\varepsilon(t) + \eta\dot{\varepsilon}(t)$$

$$\text{Maxwell model} \quad \dot{\varepsilon}(t) = \frac{\dot{\sigma}(t)}{E} + \frac{\sigma(t)}{\eta}$$

where E is the Young's modulus and η the viscosity.

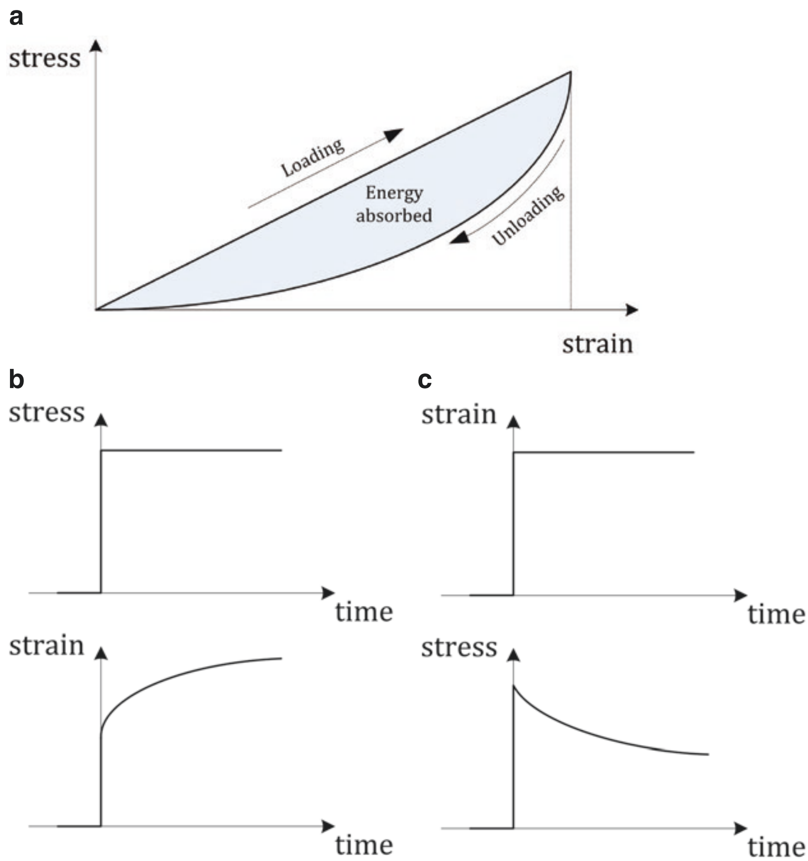


Fig. 4.9 Characteristic viscoelastic behaviours. (**a**) Hysteresis loop; the unloading path is not the same as the loading path. (**b**) Creep. When the load is held constant, the strain increases with time. (**c**) Stress relaxation. When the strain is held constant, then the stress required to hold that strain reduces with time

The most commonly used material model to capture time dependency in computational models of biological tissue is the quasi-linear viscoelastic (QLV) model. For a three-dimensional stress state and infinitesimal strains, the QLV relates stress with strain through a convolution integral.

$$\sigma(x,t) = \int_{-\infty}^{t} G(x,t-\tau)\frac{\partial\sigma}{\partial\varepsilon}\frac{\partial\varepsilon}{\partial\tau}(x,t)\,d\tau$$

where G is a relaxation function, $\frac{\partial\sigma}{\partial\varepsilon}$ represents the instantaneous elastic response and $\frac{\partial\varepsilon}{\partial\tau}$ represents the strain history.

4.4.3 Plasticity and Failure

4.4.3.1 Plasticity

When a material experiences strain beyond some limit (usually referred to as the elastic limit) then, upon unloading, it does not return to its original shape or/and size; rather, there is residual strain. The material behaviour beyond the elastic limit and prior to failure is termed plasticity for metals as the material ‘flows’, i.e. offers less resistance to further deformation with further application of load. The term plasticity is being used loosely to refer to the behaviour of all materials beyond the elastic limit. Plastic behaviour is rather complex and manifests itself in different ways for different materials.

A material that encounters large plastic deformations prior to failure is termed ductile, whereas one that doesn’t is termed brittle (Fig. 4.10a). Toughness is a property associated with how ‘large’ the plastic region is as it is a measure of the energy absorbed by the material per unit volume prior to failure; it is, therefore, a measure of resistance to failure. In most cases in construction, a tough material is favourable as it guarantees the limitations of sudden, unexpected failures. Steel, for example, is a very tough material whereas concrete is brittle, as are most ceramics.

Let’s consider loading of a sample of an isotropic material with negligible viscous response in simple tension (Fig. 4.10b). The material will elongate as the load increases. Upon load release, the material will follow the loading path back to zero elongation. If loaded beyond the elastic limit of the material, then the sample will deform plastically and, upon load release, it will exhibit permanent, residual strain. The transition from elasticity to plasticity in some materials is not smooth and depends on several factors, most important ones being the microstructure and the loading rate. For convenience in calculations, engineers have introduced the notion of yield stress; this is a value of stress beyond which the material can be considered to behave plastically. Identification of the yield stress varies among different material types; for example, in metals one can observe upper and lower yield stresses. A widely used method, however, to identify yield in metals and plastics is the offset rule; yield stress is the intersection between the stress–strain curve and a line parallel to the linear part of the stress–strain curve that crosses the x-axis (strain) at 0.2% (Fig. 4.10c).

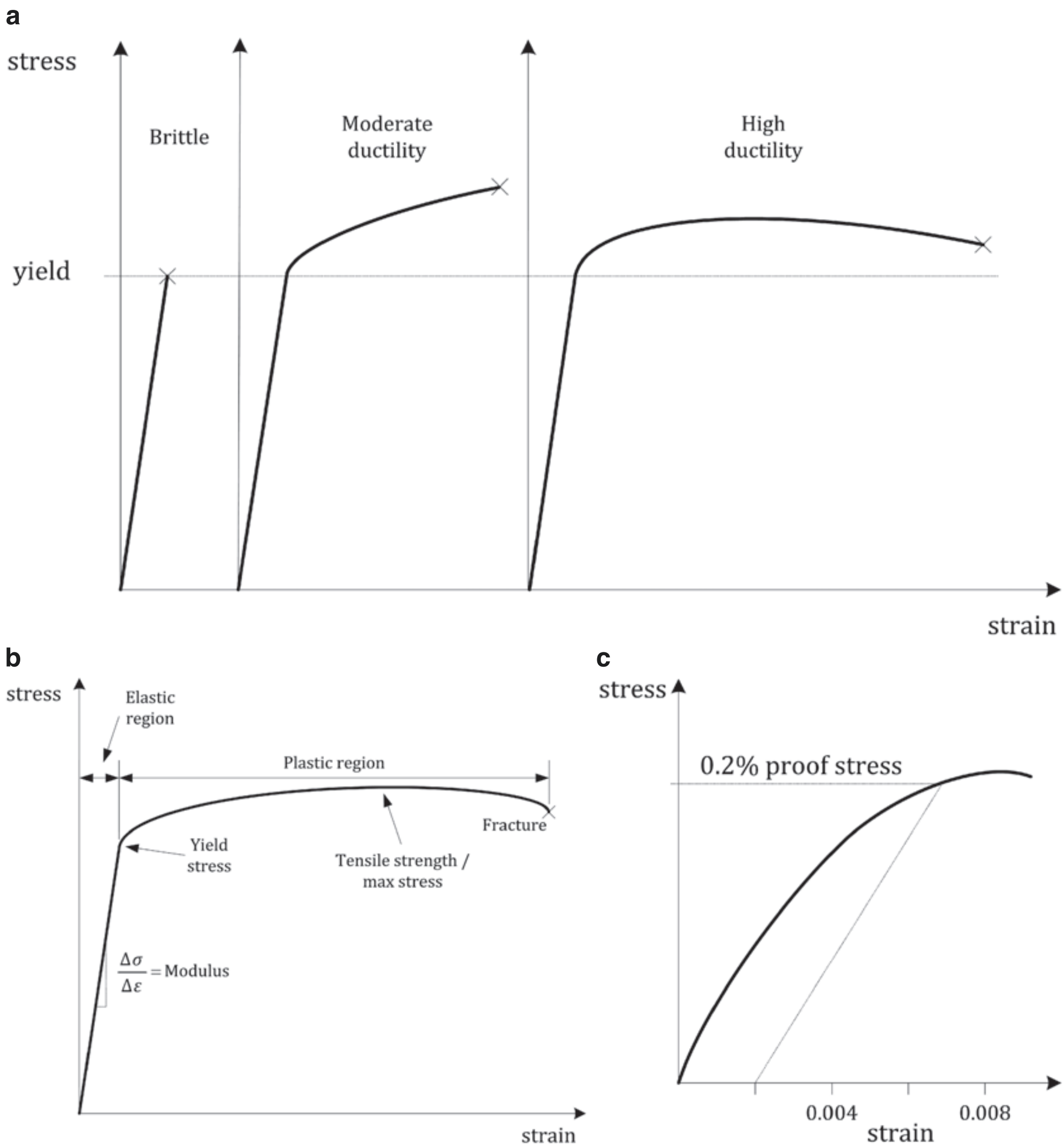


Fig. 4.10 (**a**) Stress–strain curves of brittle and ductile materials. (**b**) The stress–strain curve for a metal alloy. (**c**) Definition of 0.2% proof strength used as the yield stress for plastics and metals. Adapted from Askeland DR, 1998; *The science and engineering of materials*

4.4.3.2 Failure

Failure theories were firstly developed for metals and are usually expressed in terms of principal stresses as they represent conveniently the stress state of the material. All other types of material do not fail in similar ways to metals and therefore the theories described below are not necessarily valid for them.

The failure theory that has been proven to predict yielding and failure in ductile materials through extensive experimentation over the years is the shear (or distortion) strain energy theory. Also referred to as the von Mises theory, it postulates that the shear/distortion/deviatoric strain energy is the reason yielding will occur. A convenient measure of stress for isotropic ductile materials is the von Mises equivalent stress, which is a scalar. In an isotropic, ductile material, that is the stress one can compare against the yield or ultimate stress in order to ascertain the integrity of the structure. A convenient way of expressing the von Mises criterion is that yielding can occur when the root mean square of the difference between the principal stresses is equal to the yield of the material established by a simple tension test.

$$\sigma_{eq} = \sqrt{\frac{1}{3}\left(\left(\sigma_I - \sigma_{II}\right)^2 + \left(\sigma_{II} - \sigma_{III}\right)^2 + \left(\sigma_{III} - \sigma_I\right)^2\right)}$$

$$= \sqrt{\frac{1}{3} 2\sigma_{y,t}^2}$$

$$\text{for 2 D } \sigma_{eq}^2 = \sigma_I^2 + \sigma_{II}^2 - \sigma_I \sigma_{II} = \sigma_{y,t}^2$$

where $\sigma_{y,t}$ the yield stress in simple tension and σ_{eq} the equivalent von Mises stress.

4.4.3.3 Equations of State

At conditions of high temperature and high pressure, the behaviour of solids deviates substantially from what has been discussed up to now. Appropriate models of behaviour at such extreme conditions are the equations of state (EoS); these are relationships between state variables, most often describing how the density (or volumetric strain) and temperature (or internal energy) vary as a function of applied pressure.

In circumstances where deformations are small and the behaviour is independent of temperature, then a simple bulk modulus, K, can be used to relate hydrostatic stress (or pressure), p with volume changes; $\sigma_H = -K\varepsilon\upsilon$ where $\varepsilon\upsilon$ is the volumetric strain as it represents the volume change over the original volume; $\varepsilon_\upsilon = \varepsilon_x + \varepsilon_y + \varepsilon_z = \dfrac{\Delta V}{V}$.

For harsh loading environments where linear elasticity fails, mathematical expressions have been developed to express the EoS for different materials. Depending on type of material, these nonlinear behaviours may take the form of shock-wave generation, crushing of pores, or compaction of a granular material, to name but a few. Such behaviours may be fitted conveniently to a polynomial (power series) EoS in line with the principles established by Mie & Gruneisen;

$$p = a_0 + a_1\varepsilon_\upsilon + a_2\varepsilon_\upsilon^2 + a_3\varepsilon_\upsilon^3 + \left(b_1 + b_2\varepsilon_\upsilon + b_3\varepsilon_\upsilon^2\right)E$$

where a third order polynomial with constants a_i is used to reflect how the pressure varies with the volumetric change in the material, and a second order polynomial with constants b_i is used to account for the influence of temperature change, inherent in the internal energy per unit volume, E of the material. Note that use of the second term alone in this relationship is equivalent to the bulk modulus expression used above. Although discussion so far has been limited to materials under compression, similar general expressions can be developed for materials undergoing expansion.

4.4.3.4 Shock Loading

Under loading conditions such as high velocity impact or contact detonation of explosives, the pressures experienced by the material are far greater in magnitude than its strength; this results in a hydrodynamic behaviour, similar to a fluid that has negligible shear strength, and the generation of a shock. For convenience, in line with the typical measurements made during testing, such as plate impact, deformation can be expressed as a relationship between the velocity of the shock in the material, u_s, and the particle velocity in the material, u_p; for example in quadratic form

$$u_s = c_0 + s_1 u_p + s_2 u_p^2$$

where c_0 is the bulk speed of sound in the material and s_i are parameters obtained by fitting the experimental data. This relationship is usually termed a Rankine–Hugoniot, which has been discussed in more detail earlier, in the blast-physics chapter (Chap. 2). Such a relationship can be exploited to attain the full Mie–Gruneisen EoS.

4.4.3.5 Compaction and Unloading

With granular materials or materials exhibiting significant porosity, it is often convenient to define their nonlinear behaviour using a compaction EoS, based on experimental observation, which simply expresses the pressure as a function of volumetric strain in a piecewise fashion. Upon loading, these materials can undergo irreversible deformation (for example as a consequence of particle rearrangement or pore collapse) and so relationships for unloading have to be determined. A simple way to do this is to define a series of unloading bulk moduli, also as functions of volumetric strain, which are 'stiffer' (steeper) than the loading path for the material at this deformation point.

4.5 Dynamic Loading

All loading in nature is inherently dynamic, as the loading itself or the behaviour of the structure due to the loading changes with time. In some cases, time can be neglected in the analysis of a structure without loss of accuracy; we call this a static analysis. Time cannot be neglected, however, in loading scenarios such as impact, shock or repeated/oscillatory loading. Impact can be defined as the collision of two objects/masses with initial relative velocity. The term shock is used to describe any loading applied suddenly to a structure. These types of loading are associated with short, typically sub-second, durations.

Depending on the application and the amount of detail required in the analysis, impact and shock are studied either by considering the bodies involved as rigid or as deformable. When the bodies are considered rigid, then a system of differential equations can be formulated that describes the coupled motion of the objects as a function of the disturbance/loading. This system is usually solved computationally. The equation of motion each rigid body can be expressed by a system of differential equations of the form $m\,x + c\dot{x} + kx = F(t)$ (for linear motion; a similar equation can be written to account for rotational motion), where x is the displacement, m is the mass, c is the damping and k is the stiffness of the body; $F(t)$ is the external force applied to the system. When the bodies are considered deformable, then the behaviour of the materials from which the objects are made is considered. This system is also solved computationally, employing advanced computational methods such as the finite-element method. These computational techniques are described in more detail in the next chapter.

4.5.1 Elastodynamics: The Wave Equation

Transmission of stress waves through a solid is important when impact and shock loading is expected. For an elastic bar that is loaded axially suddenly, and assuming that plane cross-sections remain plane, then the displacement u, along the axis of the bar x, will cause strain of $\varepsilon_x = \frac{\partial u}{\partial x}$ and thus axial stress of $\sigma_x = \mathrm{E}\varepsilon = \mathrm{E}\frac{\partial u}{\partial x}$, where E is the Young's modulus of the bar. If we apply Newton's law of motion ($F = ma$) on an infinitesimal section of the bar dx, then the change in axial force through it should be equal to its mass (ρAdx) times acceleration.

$$
\begin{aligned}
Ad\sigma = \rho \mathrm{A}dx\frac{\partial^2 u}{\partial t^2} A\frac{\partial \sigma}{\partial x}dx = \rho \mathrm{A}dx\frac{\partial^2 u}{\partial t^2} A\frac{\partial \sigma}{\partial x}dx \\
= \rho \mathrm{A}dx\frac{\partial^2 u}{\partial t^2} A\mathrm{E}\frac{\partial^2 u}{\partial x^2}dx = \rho \mathrm{A}dx\frac{\partial^2 u}{\partial t^2}
\end{aligned}
$$

Therefore, the wave equation is the following second order partial differential equation

$$
\frac{\partial^2 u}{\partial t^2} = c^2\frac{\partial^2 u}{\partial x^2}
$$

where $c = \sqrt{\frac{E}{\rho}}$ is the speed of sound through the bar.

Solution of the differential equation results in a displacement $u = f(x - ct) + g(x + ct)$; functions f and g are arbitrary; the g function represents a wave travelling in the opposite direction to the applied load. Solving for the particle velocity, υ we get $\upsilon = \frac{\partial u}{\partial t} = \pm c\frac{\partial u}{\partial x}$ and using the stress–strain relationship yields $\sigma_x = \pm\frac{E}{c}\upsilon = \pm\rho c\upsilon$. The importance of this relationship has been discussed extensively earlier, in the blast-physics chapter (Chap. 2).

4.5.2 Design for Strength and Endurance: Fatigue Strength

4.5.2.1 Design of Parts and Structures

Designing parts and structures to serve a specific purpose is an important part of the engineering profession. The designer needs to take into account multiple factors in order to meet the specification and produce an object that is fit for purpose. From a stress analysis point of view, the main considerations usually are strength and endurance. No one can guarantee absolutely that a device will not fail; there is always a finite probability of failure. One way of addressing this is regular inspection of critical areas of a part. Another way is to provide redundant load paths in case a primary load path fails. No matter what though, the engineer needs to ensure s/he has done their utmost to ensure reliability of the design and predict lifespan in advance. In order to deal with variations and uncertainty, and to guarantee maximum safety, we utilise a safety factor. The value of the safety factor comes from experience and intention. Depending on the application, common and previous practice and lessons learned should define the safety factors. A common process in ensuring safety from a strength perspective is more-or-less (depending on application) the following.

- Choose the safety factor.
- Estimate maximum loading in operation (this is no easy task usually).
- Conduct static (and dynamic, if appropriate) stress analysis.
  - Hand calculations (for simple features and loadings and for most machine elements and connections).
  - Computational calculations (such as finite-element analysis) of parts or assemblies.
- Estimate fatigue life.
- Other (such as wear or corrosion resistance).

Dynamic stress analysis is conducted far less in the industry than static stress analysis as dynamics are associated with uncertainties in estimating the loading and in complexities with the calculations. Even for estimating strength in dynamic events, the tendency is to calculate an equivalent static loading scenario with an appropriate safety factor. If the structure is going to be subject to impact or blast, however, a dynamic analysis is usually necessary and is almost exclusively conducted computationally.

4.5.2.2 Fatigue

Fatigue is associated with repeated application of loading for long periods of time. It is the most common reason for failure of metal components in structures, and therefore needs to be considered in the design process. Such failures tend to initiate from design features (including connections) that result in stress concentrations. It is a process that starts with the movement of dislocations that in turn form short cracks that propagate with continuous loading (cumulative damage).

Estimating the fatigue life of a component is based primarily on past experience and lots of testing, as the process of fatigue itself is fairly stochastic. Multiple experiments with coupons have made available fatigue (*S*-*N* or Wöhler) curves for various materials and geometries (for example sheet versus bar) (Fig. 4.11). The loading to produce the curve is oscillatory with specific amplitude about a mean stress. Most metals have an endurance limit, which is the mean stress below which fatigue failure would never occur.

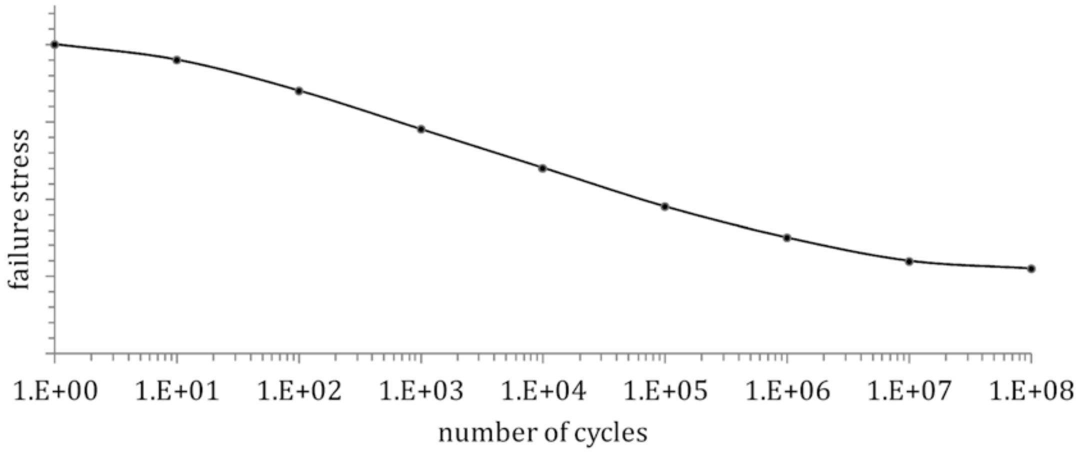


Fig. 4.11 A typical S-N (fatigue) curve for a metallic coupon. The x-axis represents the number of cycles to failure. One can look up fatigue life by finding how many cycles are required to failure for the mean operating stress that is expected due to the estimated loading environment

A simple way to estimate fatigue life is the following.

- Calculate expected maximum or equivalent stress. This will depend on the application and available experimental data. The expected loading is usually not periodical, but stochastic. One, therefore, usually needs to estimate an average stress level and its variation (alternating stress).
- Consider a fatigue level factor, K_f (this is $\frac{2}{3}$ UTS for most metals; then $K_f = 0.667$).
- Consider stress concentration, K_t based on the design of your component and available fatigue curves.
- Calculate the effective stress (that is, reduced for K_f and K_t).
- Look up life on an appropriate fatigue curve.

4.6 Summary

This chapter described fundamental principles related to the mechanical behaviour of materials and introduced analytical tools to study them. Some of these tools are expanded upon in later chapters. Often in engineering we reduce complex problems, such as blast loading on a structure or on the human body, with appropriate assumptions to problems that we can analyse with the tools we have at our disposal. The reduction of the problem always depends on the question we want to answer. Sometimes, even the simplest of tools as presented in this chapter may give a very good first approximation of the behaviour of a structure to complex loading. For further reading on the mechanical behaviour of materials, the reader is referred to the following textbooks:

Further Reading

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5 Fundamentals of Computational Modelling

Spyros D. Masouros and Daniel J. Pope

Abstract

This chapter presents an introduction to the main formulations used in computational modelling. It presents the commonly used ways of discretising the continuum of space and time, and then focusses mainly on describing the finite element method, which is the most commonly used computational tool to analyse the deformation of structures, including the human body. It concludes with introducing the important terms of sensitivity, verification and validation of computational models, which are used to quantify the integrity of a model and its range of utility to predict behaviour.

5.1 Introduction

The aim of this chapter is to introduce the reader to the main formulations used in computational modelling. Computational modelling is being used extensively in science and engineering to analyse the behaviour of systems. Analytical calculations allow us to quantify the behaviour of some simple systems or to find a first, approximation of the behaviour of complex systems, but their utility is quite limited, especially in blast science and engineering. With the advent of computational techniques, computational modelling has become an increasingly powerful tool to obtain solutions in science and engineering problems.

The choice of computational model will depend on a number of factors, including:

- The nature of the required output (for example whether a kinematic response or levels of deformation and stress development are needed)
- The degree of imponderability (for example the extent to which populating parameters or boundary conditions can be reliably defined)
- Natural variation (for example consideration of the fact that humans exhibit inherent anatomical differences)

The requirement could call for either a deterministic or probabilistic approach.

Simple, highly idealised techniques can take the form of a single equation, such as a Single Degree of Freedom (SDOF) method, which can be based on physical principles or derived empirically from a comprehensive dataset. Others may

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require the simulation of a myriad of complex non-linear physical processes and involve a huge number of degrees of freedom. These approaches are often dependent on highly detailed parameterisation and must be solved via an appropriate numerical method, often requiring significant computational effort. Examples of such approaches include:

- Multibody dynamics (MBD), suitable for modelling kinematics assuming rigid-body behaviour
- Computational Fluid Dynamics (CFD) and hydrocodes suitable for modelling scenarios involving fluids such as blast wave development
- Finite Element Analysis (FEA) suitable for modelling the deformation of structures such as the human body or structural components of a building under loading

5.2 Computational Continuum Mechanics: Overview

A continuum is a mathematical representation of a real material such as a solid, liquid or gas. By examining such media, we disregard the molecular structure of matter and assume the material is continuous. Continuum mechanics is concerned with the behaviour of such materials and is based on fundamental physical laws.

A wide range of numerical procedures have been developed, including the Finite Element (FE), Finite Volume (FV), Boundary Integral and Finite Difference methods. The FE method has been the most widely used for the analysis of solid-mechanics problems, while the FV method is the most popular method for the analysis of fluid-mechanics problems. Algorithms that combine the two methods have also been developed recently in order to address problems where deformable solids interact with fluids.

The numerical methods used to solve the problem involve the transformation of the mathematical model into a system of algebraic equations. In order to achieve this, the equations have to be discretised in time and space. In doing this, a number of approximations are made; the continuum is replaced by a 'finite' set of computational points (or nodes), areas or volumes (or elements) (Fig. 5.1); the notions of nodes and elements are discussed in more detail later in the text. The constitutive laws, which describe the material behaviour, are simplified, and continuous functions representing the exact solution to the mathematical model are generally approximated by polynomials of a finite order. The resulting set of algebraic equations is then solved either by direct or approximate, iterative methods. Finally, post-processing facilities are required in order to interpret the resulting numerical solutions and present them in a graphical way, for example stress contours.

The FE method is a computer based, approximate method to solve engineering problems. The robust mathematical framework means that the method is now being employed to solve problems for static and dynamic stress analysis in solid mechanics (including biomechanics), thermal and thermal/structural coupled analysis, electromagnetic analysis and other multi-physics problems including some fluid-mechanics problems.

In the next sections, the two commonly used mathematical formulation of FEA will be presented for structural static and dynamic problems in solid mechanics. Common in these methods is the necessity to discretise the problem so it can be solved computationally.

5.3 Material, Spatial and Other Descriptions

The essence of the FE method revolves around the spatial discretisation of the region of interest. This representation may be defined in different ways, depending on the problem at hand. As an example, consider a typical pre- and post- impact scenario as shown in Fig. 5.1a.

5.3.1 Lagrangian Representation

When quantities are phrased in terms of their initial position, x, the description is known as Lagrangian and the position itself is tracked with time (Fig. 5.1b). When the motion or deformation is described using the current configuration this is known as the updated Lagrangian description. The Lagrangian and updated Lagrangian descriptions are also referred to jointly as a material description as a material particle is followed in time.

5.3.2 Eulerian Representation

An alternative formulation is the Eulerian method. It is commonly used in fluid mechanics or when solid material deformations would be large enough to cause unfavourable element distortions or mesh entanglement within a Lagrangian framework. (The approach has some similarities with the updated Lagrangian approach and in some solid-mechanics textbooks updated Lagrangian is referred to as Eulerian.) The Eulerian formulation describes the motion at a given spatial point of different particles which occupy that point at different times. Thus we examine a fixed region of space that does not change throughout the simulation (the 'control volume'), as opposed to following a changing configuration, and particles pass through this fixed region (Fig. 5.1c). The position vector x is then used to denote the point in space which can be occupied by different particles at different instants of time.

Using an Eulerian mesh typically requires the definition of more elements than when using a Lagrangian approach to model substantively the same problem. Additional nodes are required to track the constituent materials as they move away from their initial position and, for this reason, an approximate idea of the expected response is also useful to the analyst. Perhaps one of the main disadvantages when using an Eulerian framework is the inherent difficulty in keeping track of the 'local axis' associated with materials that exhibit general anisotropy or fracture in a 'directional' manner.

5.3.3 Other Forms of Spatial Integration

Additional schemes have been developed that attempt to hybridise the virtues of the Lagrangian and Eulerian approaches, such as the Arbitrary Lagrange Euler (ALE) method (Fig. 5.1d). Within this framework, material can flow between elements whilst the mesh also concurrently deforms. The manner in which this occurs can be dictated by imposing particular arbitrarily-defined constraints on the global and local elemental behaviour within the mesh. The method can, for example, be exploited when attempting to achieve temporarily finer resolution in a particular part of a mesh which contains highly transient or localised behaviour, such as the development and subsequent expansion of a blast wave. The mesh may contract appropriately (Fig. 5.1e) providing potentially greater accuracy, when resolving the thin, high-pressure zone that constitutes the front of the wave. Once the wave has passed through this part of the model, and less transient behaviour ensues, the mesh automatically coarsens again. Adaptive Mesh Refinement (AMR) schemes have also been developed that are based on a similar principle of providing finer resolution where required; an Euler-based example of this is shown in Fig. 5.1e. In this case, although all elements within the mesh maintain a rectilinear shape, during the simulation, temporary local subdivision occurs where greater resolution is desired.

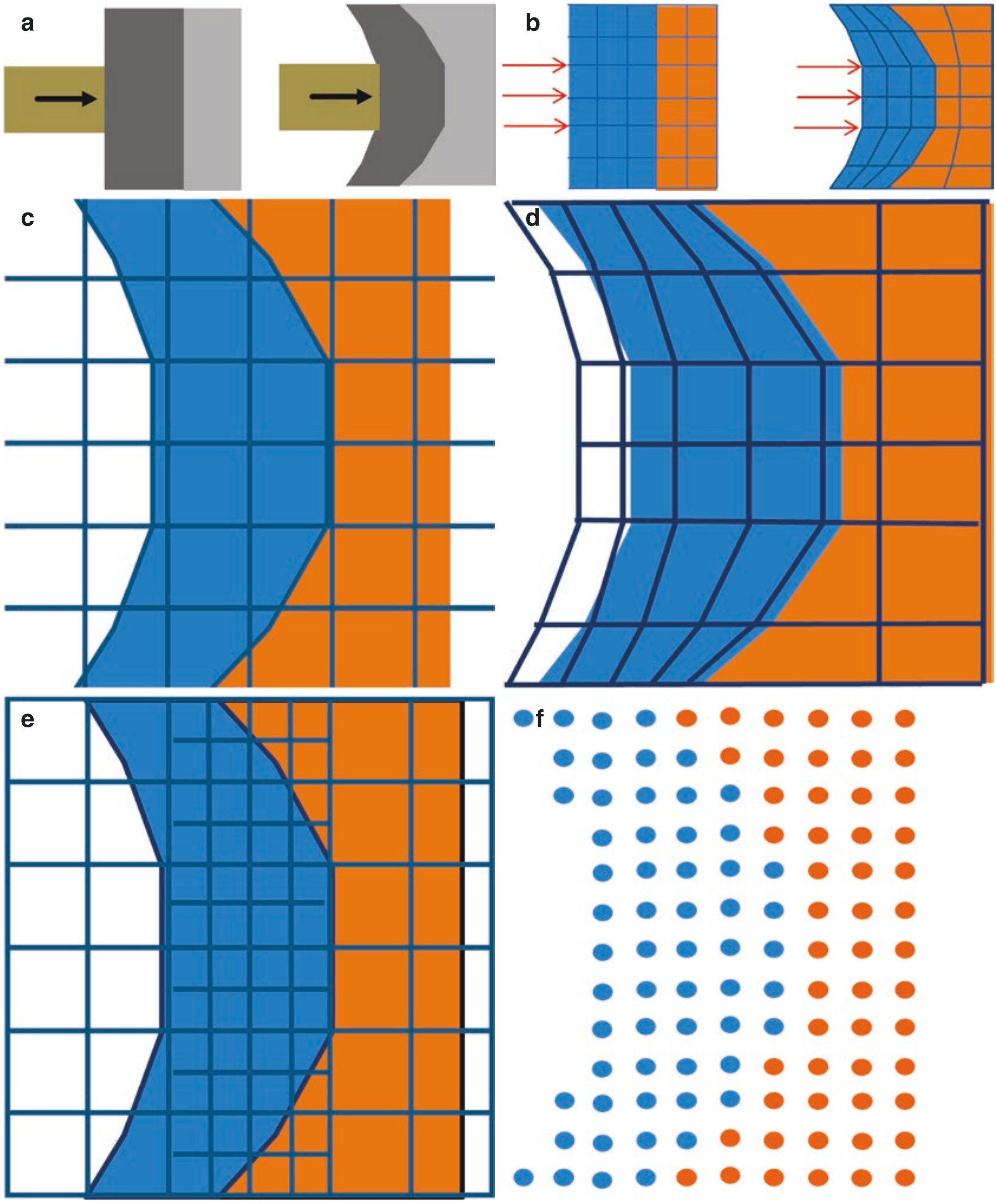

Fig. 5.1 Various meshing techniques. Each finite element is represented diagrammatically by a rectangle. (**a**) Schematic showing impact of a bar on a bi-material structure. (**b**) Lagrangian. (**c**) Eulerian (generic mesh from which (**d**) and (**e**) stem). (**d**) Adaptive Eulerian. (**e**) Arbitrary Lagrange–Euler (ALE). (**f**) Smooth Particle Hydrodynamics (SPH)

Within aggressive, dynamic loading regimes, failure, fracturing and subsequent fragment production is a common occurrence; in addition to providing adequate material models, representing these phenomena spatially, however, can also be a significant challenge. Whilst a Lagrangian scheme can be tailored to deal with relatively complex, directional material behaviour, problems with element distortion or mesh entanglement can limit its use when simulating structures undergoing large deformations. In other situations, a particular stress or strain state should lead to material fracture and so a way to allow portions of material to 'detach' within the solution framework has to be found. One expedient way of dealing with this within a Lagrangian system is to allow unstable or highly distorted elements to be deleted or 'eroded' from the simulation. The criterion for erosion can be based, for example on a particular stress or strain state within the element, or a combination of both. Erosion can also be enforced if element distortions lead to a very small controlling time step. Although effective in many situations, the use of different erosion criteria can lead to very different model outputs and it must be remembered that the process largely violates physical laws; for example, mass from eroded elements is either ignored or somehow distributed amongst their surrounding elements.

Alternative schemes, such as particle-based approaches have been developed, to cope partly with the issues discussed above. One example is the Smooth Particle Hydrodynamics (SPH) method (Fig. 5.1f). Here, the solution space is populated with particles, rather than elements. A 'kernel' is defined that effectively determines the radius of influence that each particle has on its neighbouring particles. As with the classic Lagrangian method, this provides the interdependence within the model to allow quantities such as stress to be transferred from one zone to another. In contrast to the Lagrangian approach, however, if the distance between particles exceeds the radius of influence at any point, they can simply detach from one another.

5.4 Implicit Finite Element Analysis

The FE method was developed by engineers in order to carry out stress analysis of objects with complex geometry. Practically, it stems from the structural analysis of large frames with the advent of computers (1930–1950) by implementing the direct stiffness method that employs matrix algebra; it was pioneered by the aeronautical industry in the 1950s by taking the idea of the 'discretised' frame consisting of individual trusses and applying it to any solid by splitting it into finite regions. Mathematically, the method was documented in the late 1960s and is based on the Galerkin numerical methods.

In a nutshell, in order to run a finite element analysis of a part or an assembly using commercial FE software, one needs to carry out the following steps:

- Acquire the geometry of the part(s) (usually from Computer Aided Design (CAD))
- Mesh it (split it into small regions, the finite elements)
- Assign material properties
- Define initial and boundary conditions (including loading), where boundary conditions are the constraints that are necessary to solve the problem and can include force, pressure and temperature
- Run the simulation (solve/stress recovery)
- Plot and interpret the results

The objective of the finite element code is to calculate the distribution of the variable in question (the displacement field in the structural case) using functions defined over the whole structure and, secondly, satisfy the boundary conditions. The objective in a structural FE analysis is to calculate the displacement field $\mathbf{u}(\boldsymbol{x})$ in the structure and from that calculate the strain and therefore, through the constitutive laws, the stress distribution. Hence, the FE method converts the problem of calculating a variable over the continuum—of an infinite number of degrees of freedom (DOFs)—to a surrogate problem of calculating the variable over a finite number of DOFs.

We will develop the formulation here for a static FE analysis of a linearly elastic material and small strains, and generalise later in the text.

5.4.1 Meshing/Discretisation

The essence of the FE method is the division of the body in discrete regions, the finite elements. The finite elements are of ordinary shapes (depending on the dimension of the simulation) such as lines, triangles, rectangles, cubes, tetrahedra, etc., which are connected with one another at the corners of each edge with *nodes*. A set number of DOFs is associated with every node. Elements and nodes are collectively termed the finite element *mesh*.

5.4.2 Shape Functions

The shape (or basis) functions are interpolation functions—unique to each element type—that relate the displacement across the element $\mathbf{u}(\boldsymbol{x})$ to that at its nodes, $\mathbf{U}$, where $\mathbf{U}$ is a vector of dimension equal to the DOFs of the element, and

$$\mathbf{u}(\boldsymbol{x}) = \begin{Bmatrix} u(x,y,z) \\ v(x,y,z) \\ w(x,y,z) \end{Bmatrix}$$

The interpolation functions are polynomials, usually linear or quadratic. The number of finite elements in which the body is divided and the degree of the polynomial of the interpolation functions are in most cases directly related to the accuracy of the solution.

Let's consider a one-dimensional truss element (Fig. 5.2). A truss is a structural member that can only resist tension and compression, i.e. axial loading; it cannot take bending moments. Using the general form of a polynomial to represent the interpolation function, the displacement at a point x along the length of the element is

$$u(x) = a_1 + a_2 x + a_3 x^2 + \ldots + a_{m-1} x^m + \ldots$$

$$u(x) = \begin{bmatrix} 1 & x & x^2 & \cdots & x^m & \cdots \end{bmatrix} \begin{Bmatrix} a_1 \\ a_2 \\ \vdots \\ a_m \\ \vdots \end{Bmatrix}$$

$$u(x) = \mathbf{M}(x)\mathbf{a}.$$

The terms a_i are constants, which depend on the element type and are associated with the displacements (but could also be associated with their derivatives, for example in bending; a type of loading that results in curving the structure). Their total number is equal to the total number of nodes, NNODES. At node N_i, i = 1, 2, …, NNODES of the element, the displacement will be

$$u_{N_i}(x_i) = U_i = \mathbf{M}(x_i)$$

where x_i are the coordinates of node i. Applying this to all the nodes we get

$$\mathbf{U} = \mathbf{Aa}$$

Combining we get

$$u(x) = \mathbf{M}(x)\mathbf{a} = \mathbf{M}(x)\mathbf{A}^{-1}\mathbf{U}$$

$$u(x) = \mathbf{N}(x)\mathbf{U}$$

We call the matrix $\mathbf{N}$ the *shape function* of the element.

For example, let's consider the simplest possible interpolation

$$u(x) = a_1 + a_2 x = \begin{bmatrix} 1 & x \end{bmatrix} \begin{Bmatrix} a_1 \\ a_2 \end{Bmatrix}$$

$$u(x) = \mathbf{M}(x)\mathbf{a}$$

Apply this to the displacements at the two nodes

$$\begin{Bmatrix} U_1 \\ U_2 \end{Bmatrix} = \begin{bmatrix} 1 & 0 \\ 1 & L \end{bmatrix} \begin{Bmatrix} a_1 \\ a_2 \end{Bmatrix}$$

$$\mathbf{U} = \mathbf{Aa}$$

Solve and substitute

$$u(x) = \left(1 - \frac{x}{L}\right)U_1 + \frac{x}{L}U_2$$

$$u(x) = \begin{bmatrix} 1 - \frac{x}{L} & \frac{x}{L} \end{bmatrix} \begin{Bmatrix} U_1 \\ U_2 \end{Bmatrix} = \begin{bmatrix} N_1(x) & N_2(x) \end{bmatrix} \begin{Bmatrix} U_1 \\ U_2 \end{Bmatrix}$$

$$u(x) = \mathbf{N}(x)\mathbf{U}$$

There is a computationally more elegant way to represent shape functions, which is what commercial FE codes utilise; that is the 'parent' element on the dimensionless s-space. No matter what the shape of the element is, the FE code would always calculate at the simple, unit, parent element and map the result to the actual element (Fig. 5.3). The parent line element lies between $s = -1$ (node 1) and $s = 1$ (node 2).

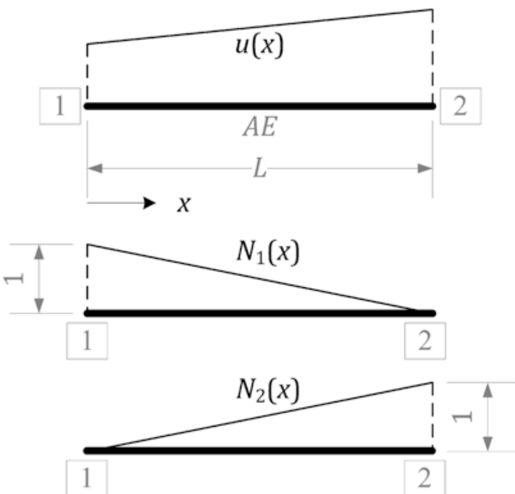

Fig. 5.2 A line element with two nodes of length L, cross-sectional area A and Young's modulus E

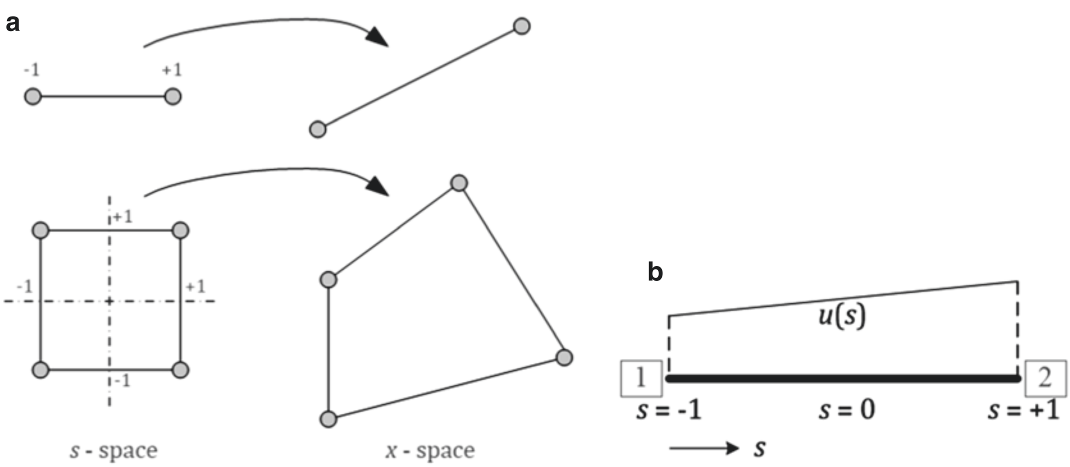

Fig. 5.3 (**a**) Schematic showing mapping from the s-space to the x-space for a line and a square parent element. (**b**) The parent line element in the dimensionless s-space

5.4.3 Strains and Stresses / Constitutive Laws (See Chap. 4 for More Details)

Strain in each finite element is related to the displacement field:

$$\begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \varepsilon_z \\ \gamma_{xy} \\ \gamma_{yz} \\ \gamma_{zx} \end{Bmatrix} = \begin{bmatrix} \frac{\partial}{\partial x} & 0 & 0 \\ 0 & \frac{\partial}{\partial y} & 0 \\ 0 & 0 & \frac{\partial}{\partial z} \\ \frac{\partial}{\partial y} & \frac{\partial}{\partial x} & 0 \\ 0 & \frac{\partial}{\partial z} & \frac{\partial}{\partial y} \\ \frac{\partial}{\partial z} & 0 & \frac{\partial}{\partial x} \end{bmatrix} \begin{Bmatrix} u(x,y,z) \\ v(x,y,z) \\ w(x,y,z) \end{Bmatrix}$$

$$\boldsymbol{\varepsilon} = L\mathbf{u}$$

$$\boldsymbol{\varepsilon} = L(\mathbf{NU}) = L(\mathbf{N})\mathbf{U} = \mathbf{BU}$$

where L is the differential operator.

Stress is related to strain through the constitutive law.

$$\boldsymbol{\sigma} = \mathbf{D}\boldsymbol{\mu} = \mathbf{DBU}$$

5.4.4 Formulation

5.4.4.1 Internal and External Energy: Principle of Virtual Work

Consider a finite element of volume V^e with boundary S^e as part of the mesh. The following forces may be contributing to the overall external loading of that element:

- Traction forces (normal stresses), $\mathbf{p} = \boldsymbol{\tau}\hat{\mathbf{n}} = \lvert\, \mathbf{p}_x, \mathbf{p}_y, \mathbf{p}_z \sim$ due to neighbouring elements
- Global, body forces, $\mathbf{b} = \{b_x, b_y, b_z\}$; and
- Concentrated nodal forces, $\mathbf{P} = \{P_x, P_y, P_z\}$

Then the work done by external forces (force times displacement) on the element, Ψ^e, is

$$\Psi^e = \int_{V^e} \{\mathbf{u}^e\}^{\mathrm{T}} \mathbf{b}^e dV^e + \int_{S^e} \{\mathbf{u}^e\}^{\mathrm{T}} \mathbf{p}^e dS^e + \{\mathbf{u}^e\}^{\mathrm{T}} \mathbf{P}^e$$

$$\Psi^e = \{\mathbf{U}^e\}^{\mathrm{T}} \mathbf{F}_b^e + \{\mathbf{U}^e\}^{\mathrm{T}} \mathbf{F}_p^e + \{\mathbf{U}^e\}^{\mathrm{T}} \mathbf{F}_P^e$$

$$\Psi^e = \{\mathbf{U}^e\}^{\mathrm{T}} \mathbf{F}^e$$

where

$\mathbf{F}^e$ the element force vector	$\mathbf{F}^e = \mathbf{F}_b^e + \mathbf{F}_p^e + \mathbf{F}_P^e$
$\mathbf{F}_b^e$ the body force vector	$\mathbf{F}_b^e = \int_{V^e} \{\mathbf{N}^e\}^{\mathrm{T}} \mathbf{b}^e dV^e$
$\mathbf{F}_p^e$ the surface traction vector	$\mathbf{F}_p^e = \int_{V^e} \{\mathbf{N}^e\}^{\mathrm{T}} \mathbf{p}^e dV^e$
$\mathbf{F}_P^e$ the nodal force vector	$\mathbf{F}_P^e = \mathbf{P}^e$
$\mathbf{F}^e$ the element force vector	$\mathbf{F}^e = \mathbf{F}_b^e + \mathbf{F}_p^e + \mathbf{F}_P^e$

Using the stress–strain and strain–displacement relationships above, we can evaluate the strain (also referred to as internal or stored) energy of the element, W^e

$$\left.\begin{aligned} W^e &= \int_{V^e} \{\boldsymbol{\varepsilon}^e\}^{\mathrm{T}} \boldsymbol{\sigma}^e dV^e \\ \boldsymbol{\sigma}^e &= \mathbf{D}^e \boldsymbol{\varepsilon}^e \\ \boldsymbol{\varepsilon}^e &= \mathbf{B}^e \mathbf{U}^e \end{aligned}\right\} \Rightarrow W^e = \{\mathbf{U}^e\}^{\mathrm{T}} \left(\int_{V^e} \{\mathbf{B}^e\}^{\mathrm{T}} \mathbf{D}^e \mathbf{B}^e dV^e \right) \mathbf{U}^e = \{\mathbf{U}^e\}^{\mathrm{T}} \mathbf{k}^e \mathbf{U}^e$$

where $\mathbf{k}^e = \int_{V^e} \{\mathbf{B}^e\}^{\mathrm{T}} \mathbf{D}^e \mathbf{B}^e dV^e$ is the stiffness matrix of the element.

The principle of virtual work may be applied in the FE context. Virtual work is the 'weak' formulation of the balance of linear momentum and states that if a system of forces acts on a body that is in static equilibrium and the body is given any virtual displacement then the net work done by the forces is zero and so the virtual work is equal to zero. *Virtual work* is the work done by real loads on a solid body when virtual displacements are applied. *Virtual displacement* is an imaginary, small, arbitrary displacement that must be geometrically possible. An important note is that stresses do not change due to the virtual change in displacement, but forces do work due to the virtual change in displacement. In essence, the principle of virtual work suggests that in order for a deformable body to be in equilibrium then the work done by external forces when a virtual displacement is applied is equal to the virtual strain energy; that is the energy stored within the material due to it deforming under load (Fig. 5.4).

Stiffness of bar, $k = EA/L$
Apply virtual displacement, δu
External virtual work done, $\delta W_e = F\delta u$
Internal virtual work done, $\delta W_i = ku\delta u$
Total virtual work done, $\delta W = \delta W_e - \delta W_i$
Principle of virtual work states that $\delta W = 0$
Therefore $\delta W_e - \delta W_i = 0$
and so $F = ku$

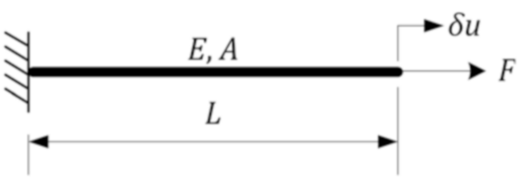

Fig. 5.4 A simple example of the principle of virtual work using a bar in tension. Try calculating the internal virtual work done by calculating the internal strain energy (integral of stress times virtual strain ($\sigma\delta\varepsilon$) over the volume) instead of just using the 'spring' reaction force ku

In the context of FE, virtual displacement and virtual strain may be associated with virtual nodal displacements using the shape functions and the matrix **B**.

$$\delta\mathbf{U}^e = \mathbf{N}\delta\mathbf{u}^e$$

$$\delta\boldsymbol{\varepsilon}^e = \mathbf{B}\delta\mathbf{U}^e$$

When we apply a virtual displacement $\delta\mathbf{u}$ to the element we were considering above, then the virtual external and internal energies, $\delta\Psi^e$ and δW^e, respectively, would be

$$\delta\Psi^e = \{\delta\mathbf{U}^e\}^{\mathrm{T}} \mathbf{F}^e$$

$$\delta W^e = \{\delta\mathbf{U}^e\}^{\mathrm{T}} \mathbf{k}^e\mathbf{U}^e$$

and by applying the principle of virtual work, the total work done should be equal to zero.

$$\delta\Psi^e - \delta W^e = 0$$

$$\{\delta\mathbf{U}^e\}^{\mathrm{T}} \left(\mathbf{F}^e - \mathbf{k}^e\mathbf{U}^e\right) = 0$$

$$\mathbf{F}^e - \mathbf{k}^e\mathbf{U}^e = \mathbf{0}$$

5.4.4.2 Assemble

In order to account for all elements in the mesh, NEL,

$$\sum_{e=1}^{\mathrm{NEL}} \left(\mathbf{F}^e - \mathbf{k}^e\mathbf{U}^e\right) = 0$$

$$\mathbf{KU} = \mathbf{F}$$

where **K** is the global stiffness matrix, **F** the global force matrix, and **U** the global displacement vector.

Some remarks about the stiffness matrix.

- The matrix is symmetrical ($K_{ij} = K_{ji}$).
- The coefficients of the matrix K_{ij} are the force F_i required to achieve a unit displacement U_j and zero other displacements.

- The sum of the coefficients in each column is zero as each column represents global forces/loads that are in equilibrium.
- The matrix is singular (i.e. det**K** = 0); since the sum of the coefficients in each column is zero, the lines are linearly dependent. This means that one cannot inverse the matrix (and so cannot solve for nodal displacements, **U**) unless one includes appropriate boundary conditions.

5.4.4.3 Solve

In order to solve for the unknown nodal displacements, an appropriate numerical method needs to be implemented that can calculate the inverse of the stiffness matrix.

$$\mathbf{U} = \mathbf{K}^{-1}\mathbf{F}$$

Several numerical techniques may be used to solve this problem (Gauss elimination, Cholesky decomposition, etc.). The size of the stiffness matrix of the system will depend on the number of degrees of freedom present in the system.

5.4.5 Evaluation of the Stiffness Matrix: Numerical Quadrature

The stiffness matrix for each element is an integral over the element's volume. In order to carry out the integration numerically, FE codes use Gaussian quadrature. Consider the one-dimensional integral

$$I = \int_{-1}^{1} f(s)\,ds$$

This integral can be approximated by a sum

$$I = \int_{-1}^{1} f(s)\,ds \approx \sum_{n=1}^{\mathrm{NQ}} w_n f(\xi_n)$$

where ξ_n the position of the Gauss points; w_n the 'weight' assigned to each Gauss point; NQ the order of the quadrature; for a linear quadrate NQ = 1; for a quadratic quadrature NQ = 2 and so on.

If we have NQ quadrature points, then we can integrate exactly a polynomial of order 2NQ − 1; e.g. with 1 Gauss point we can integrate a linear function exactly (polynomial of order 1 has two parameters: $p(x) = (a_0 + a_1x)$. Coordinates and weights of Gauss points that correspond to a particular order of integration can be evaluated easily by integrating simple expressions with known results (see Table 5.1).

Table 5.1 Gaussian quadrature. Coordinates of Gauss points and weight assigned to each of them for up to third order quadrature

Order of quadrature, NQ	Position of Gauss points, ξ	Weight assigned to each Gauss point, w
1	0	2
2	$-\frac{1}{\sqrt{3}}$ $\frac{1}{\sqrt{3}}$	1 1
3	$-\frac{\sqrt{3}}{\sqrt{5}}$ 0 $\frac{\sqrt{3}}{\sqrt{5}}$	$\mathbf{\frac{5}{9}}$ $\mathbf{\frac{8}{9}}$ $\mathbf{\frac{5}{9}}$

If we consider a double integral, then it will be approximated by a double sum

$$I = \int_{-1}^{1}\int_{-1}^{1} f(s_1, s_2)\, ds_1 ds_2 = \sum_{m=1}^{\mathrm{NQ}}\sum_{n=1}^{\mathrm{NQ}} w_m w_n f(\xi_m, \xi_n)$$

and if we consider a triple integral, then it will be approximated by a triple sum.

One can apply the Gaussian quadrature in order to integrate the element matrix over the element's volume. For example, if we assume 2D space, that the material properties do not depend on position, and that parent element and actual element coincide, then

$$\mathbf{k} = \int_V \mathbf{B}^{\mathrm{T}}\mathbf{D}\mathbf{B}\, dV = \sum_{m=1}^{\mathrm{NQ}}\sum_{n=1}^{\mathrm{NQ}} w_m w_n \mathbf{B}^{\mathrm{T}}(\xi_m, \xi_n)\mathbf{D}\mathbf{B}(\xi_m, \xi_n)$$

which means use of four Gauss points, marked with × in Fig. 5.5.

5.4.5.1 Mapping of Elements from the s- to the x-Space: The Jacobian

It is convenient numerically to calculate matrix **B** always at the parent element. Therefore, in the calculation of the stiffness matrix the parent element needs to be mapped to the actual element (Fig. 5.6). If we consider integration of a function $f(x)$ in one dimension, then

$$\int_x f(x)\, dx = \int_x f(x)\frac{dx}{ds}\, ds = \int_x f(x) J(s)\, ds$$

where $\dfrac{dx}{ds}$ is known as the Jacobian, J of the mapping.

Mapping insofar as elements are concerned happens through the shape functions (Fig. 5.6).

$$\mathbf{x}_i(s) = \sum_{n=1}^{\mathrm{NEL}} \mathbf{N}^e(s)\mathbf{X}_i^e$$

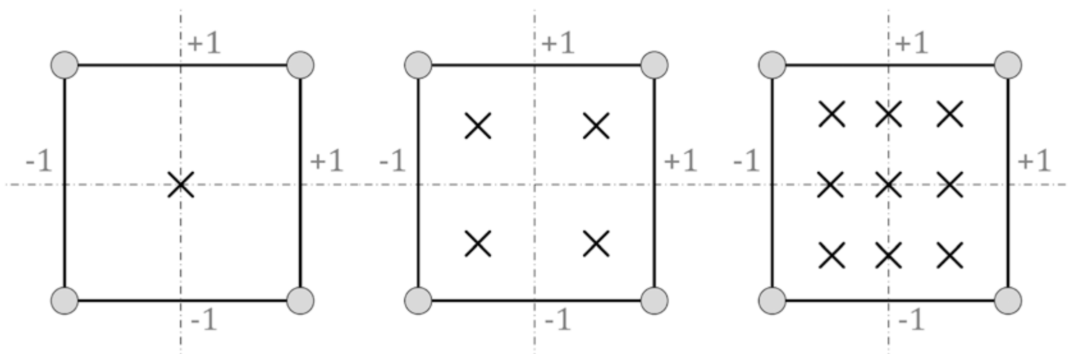

Fig. 5.5 A 2D quadrilateral parent element with 1 Gauss point (reduced integration), 4 Gauss points, and 9 Gauss points that are required for full integration of the stiffness matrix over the element volume; the coordinates in x and y of the Gauss points are shown in Table 5.1

where $\mathbf{x}_i$ the coordinates of a point on the mapped element and $\mathbf{X}_i$ the coordinates of the nodes of the mapped element.

Elements that use the same shape functions to map the nodal coordinates and the nodal displacements are known as isoparametric elements. The vast majority of elements do so.

In 2D, the Jacobian of the mapping, J is the determinant of the Jacobian matrix

$$J = \det\left(\frac{\partial x_i}{\partial s_j}\right) = \det\left(\frac{\partial}{\partial s_j}\left(\sum_{n=1}^{\mathrm{NEL}} \mathbf{N}^e(s)\mathbf{X}_i^e\right)\right)$$

In 2D $\dfrac{\partial x_i}{\partial s_j} = \begin{bmatrix} \dfrac{\partial x_1}{\partial s_1} & \dfrac{\partial x_1}{\partial s_2} \\ \dfrac{\partial x_2}{\partial s_1} & \dfrac{\partial x_2}{\partial s_2} \end{bmatrix}$. This can be easily generalised to 3D.

Then the element stiffness matrix of the actual element, for example a quadrilateral element (Fig. 5.6), can be evaluated as

$$\mathbf{k} = \int_V \mathbf{B}^{\mathrm{T}}\mathbf{D}\mathbf{B}dV = \sum_{m=1}^{\mathrm{NQ}}\sum_{n=1}^{\mathrm{NQ}} w_m w_n \mathbf{B}^{\mathrm{T}}(\xi_m,\xi_n)\mathbf{D}\mathbf{B}(\xi_m,\xi_n)J(\xi_m,\xi_n)$$

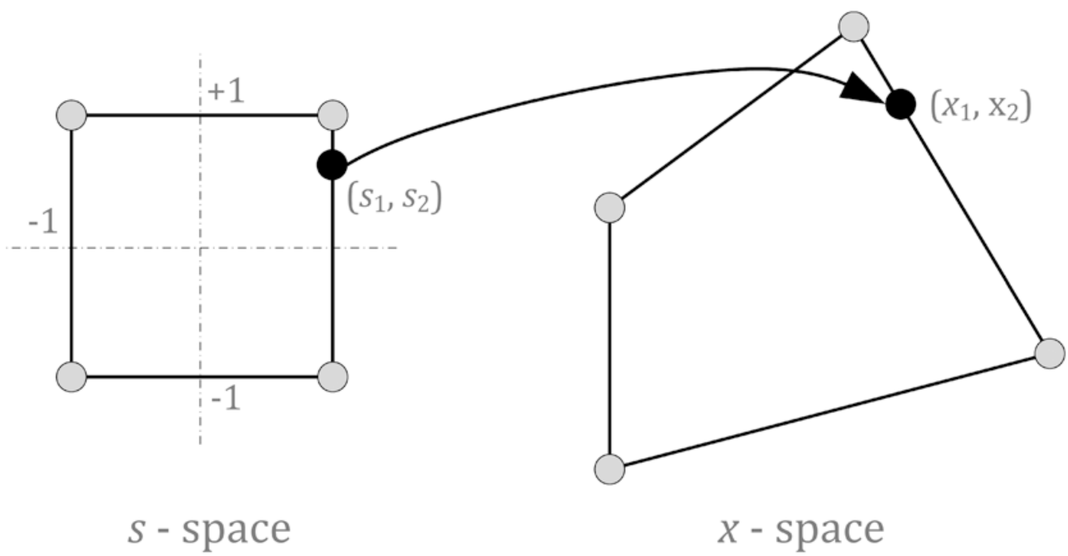

Fig. 5.6 A 2D quadrilateral parent element (s-space) is being mapped onto the actual element in the x-space

5.4.6 Recover Strain and Stress

As the **B** matrix is evaluated at the Gauss points, it is straightforward to evaluate strain and then stress, using the constitutive law, also at the Gauss points. Then strain and stress output may be extrapolated to the nodes or averaged over the element to give a single value for each element. The single value of stress / strain per element is a source of numerical error. Therefore, one needs to carry out a mesh sensitivity study (discussed in detail in the last section of this chapter) and refine the mesh appropriately in order to eliminate unacceptably large variations of stress and strain between adjacent elements.

5.4.7 Overview of the Linear Static FE Method

Epigrammatically, below are the steps taken to run a linear static FE analysis. Steps 4–8 are carried out by the FE code, whereas 1–3 and 9 by the analyst.

1. Set up FE mesh and define element type

$$\mathbf{u} = \mathbf{N}\mathbf{U}$$

2. Assign material properties
3. Apply boundary (including loading) conditions
4. Set up element matrices and assemble global stiffness matrix

$$\mathbf{k}^e = \int_{V^e} \left\{\mathbf{B}^e\right\}^{\mathrm{T}} \mathbf{D}^e \mathbf{B}^e dV^e \text{ and } \mathbf{K} = \sum_{e=1}^{\mathrm{NEL}} \mathbf{k}^e;$$

5. Set up force matrix

$$\mathbf{F} = \sum_{e=1}^{\mathrm{NEL}} \mathbf{F}^e$$

6. Apply displacement boundary conditions and therefore eliminate known DoFs from the system of equations
7. Invert stiffness matrix and solve for displacements

$$\mathbf{U} = \mathbf{K}^{-1}\mathbf{F}$$

8. Recover strain and stress

$$\varepsilon = \mathbf{B}\mathbf{U} \quad \text{and} \quad \sigma = \mathbf{D}\varepsilon$$

9. Carry out post-processing.

5.4.8 Non-linear Finite Element Formulation

The above formulation is valid for a linear analysis, whereby the geometry is linear (small displacements and rotations) and the material response is linear. The formulation that accounts for non-linear responses is similar, but the stiffness matrix and the external force will be a function of the displacement and therefore an iterative numerical schema (such as the Newton–Raphson) needs to be implemented.

$$\mathbf{K}(\mathbf{U})\mathbf{U} = \mathbf{F}(\mathbf{U})$$

As most non-linear problems—and definitely those with non-linear material models—are associated with large deformations, the finite-strain (as opposed to infinitesimal strain) theory is implemented. Although a presentation of the finite-strain theory is beyond the scope of this book, it needs to be noted that the stress and strain values that are evaluated at recovery from commercial FE codes are either the second Piola–Kirchhoff stress with Green-Lagrange strain (when using a total Lagrange schema) or Cauchy stress with logarithmic strain (when using an updated Lagrange schema).

If we consider dynamic effects, the equation to be solved in FE will be of the form

$$\mathbf{M}\ddot{\mathbf{U}} + \mathbf{C}\dot{\mathbf{U}} + \mathbf{K}\mathbf{U} = \mathbf{F}$$

where **M** is the FE mass matrix, **C** the damping matrix, $\ddot{\mathbf{U}}$ the nodal accelerations and $\dot{\mathbf{U}}$ the nodal velocities (see also Sect. 4.5).

$$\mathbf{M} = \int_V \mathbf{N}^{\mathrm{T}} \rho \mathbf{N} dV$$

To carry out a *modal analysis* of a structure, one needs to solve the undamped free vibration problem

$$\mathbf{M}\ddot{\mathbf{U}} + \mathbf{K}\mathbf{U} = \mathbf{0}$$

from which natural frequencies and the corresponding (modal) shapes of the structural system can be evaluated.

5.5 Explicit FEA for Dynamic Systems

5.5.1 The Single Degree of Freedom System (SDOF)

5.5.1.1 Formulation

Perhaps the simplest form of numerical analysis of a dynamic system would be the solution of an SDOF system via an explicit, time-stepping analysis. Consider a spring-mass system, as discussed in Sect. 4.5 (Fig. 5.7a). The equation of the (undamped) motion is

$$m\ddot{y} + ky = F(x,t,\ldots)$$

This equation can be used to represent a simple structural element, such as a beam, that is loaded by a force per unit length, p. For each of the terms of the equation, at every point during transient response, the energies associated with the SDOF must be equal to those in the real system. A real system has spatially varying properties (in this case in the x direction) whilst the properties within an SDOF act at a single point. Consequently, simple integration techniques must be used to 'transform' the 'distributed' mass and loading into 'effective' quantities as follows:

$$m_e = \frac{m}{L}\int_0^L \phi(x)^2\, dx$$

$$F_e = p\int_0^L \phi(x)\, dx$$

where φ is a dimensionless 'shape function' and in this case describes the spatial deflection of the beam as a function of x. The deformed shape will depend on the support conditions of the beam as well as the loading distribution upon it and its material characteristics. For example, the simply supported, elastic beam exposed to a uniform spatial load, as shown in Fig. 5.7a would produce the following shape function:

$$\phi(x) = \frac{16}{5L^4}\left(L^3x - 2Lx^3 + x^4\right)$$

5.5.1.2 Closed-Form Solution

For scenarios in which strong approximations or idealisations are acceptable, the functions representing the individual terms in the ordinary differential equation above can be contrived to allow solution in a closed-form manner. For such cases, it must also be assumed that the shape function remains constant throughout the entire motion of the structure. For example, the equation of motion of a purely elastically deforming, simply supported beam (of effective stiffness, k) exposed to an idealised blast wave, exponentially decaying as a function of T (Fig. 5.7b), is

$$m_e\ddot{y} + k_e y = F_{e0}e^{-(t/T)}$$

This ordinary differential equation (ODE) can be solved in a closed-form manner to yield a harmonic displacement response (Fig. 5.7c)

$$y(t) = y_{stat}A\left(\frac{\sin(\omega t)}{\omega T} - \cos(\omega t) + Be^{-\left(\frac{\omega t}{\omega T}\right)}\right)$$

where y_{stat} is the deflection that the structure would attain if the load was applied statically and A is given by

$$A = \frac{(\omega T)^2}{1+(\omega T)^2}$$

with $\omega = \sqrt{\frac{k_e}{m_e}}$ being the angular frequency of the system.

Whilst solution in this case is extremely quick, its scope of usage is highly limited. When a structure possesses a complex geometry, is exposed to a highly variable temporal loading, and exhibits non-linear material behaviour or experiences a change in mode during its transient deformation, then an analytical approach is not possible and a numerical approach must be implemented.

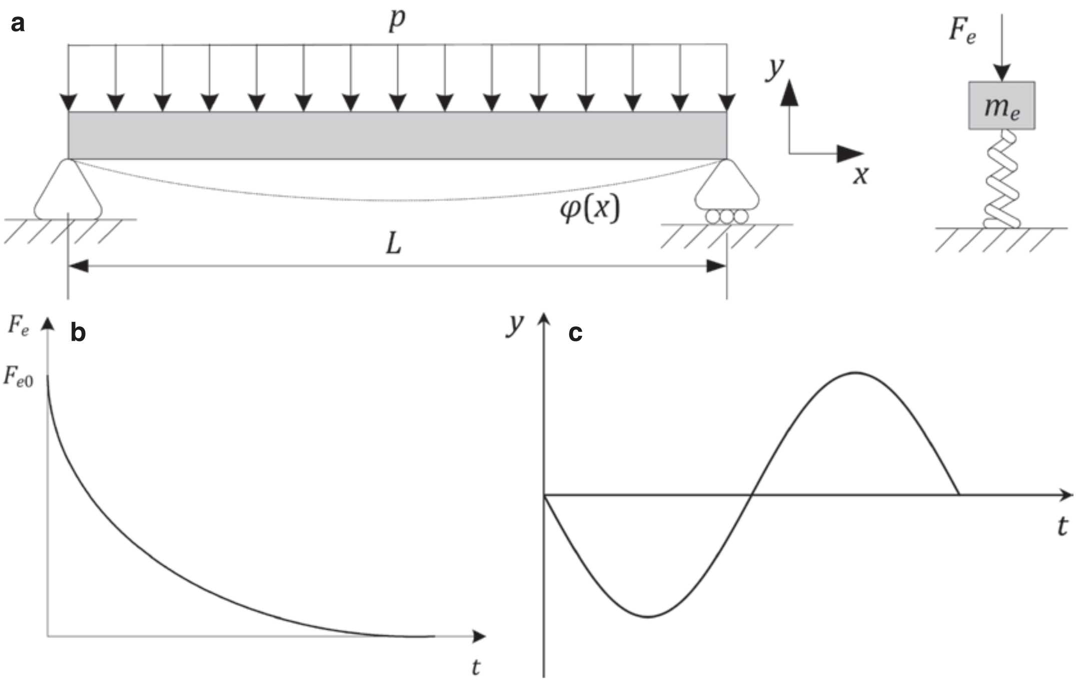

Fig. 5.7 (**a**) Simply supported beam under a distributed load and its SDOF equivalent representation. (**b**) Exponentially decaying blast loading. (**c**) Harmonic solution to the resulting ordinary differential equation (ODE)

5.5.1.3 Numerical Solution with Explicit Time-Stepping: The Linear Acceleration Method

The linear acceleration method represents one of the most simple and effective ways of attaining a solution for an equation of motion that cannot be solved analytically. Its application within a SDOF framework is shown in the flow diagram in Fig. 5.8a. Instead of providing a continuous solution, the response of the system here, similar to that shown in Fig. 5.7b for example, is predicted over a series of discrete time steps with linear changes in acceleration being assumed between them (Fig. 5.8b). By utilising the Simpson's rule, the velocity at a given instant, t_i can be estimated as

$$\dot{y}_i = \dot{y}_{i-1} + \frac{\Delta t}{2}\left(\ddot{y}_i + \ddot{y}_{i-1}\right)$$

whilst displacement at the next time step can be estimated as

$$y_{i+1} = y_i + \dot{y}_i \Delta t + \frac{(\Delta t)^2}{6}\left(2\,\ddot{y}_i + \ddot{y}_{i+1}\right)$$

For an equation of motion of the type described above the acceleration at any given instant of time and as such at t_{i+1} can be expressed by

$$\ddot{y}_{i+1} = \frac{F_e\left(t_{i+1}\right) - k_e y_{i+1}}{m_e}$$

and this expression can be substituted into the one above and rearranged to yield

$$y_{i+1} = \frac{y_i + \dot{y}_i \Delta t + \frac{(\Delta t)^2}{3}\ddot{y}_i + \frac{(\Delta t)^2}{6}\frac{F_e\left(t_{i+1}\right)}{m_e}}{1 + \frac{(\Delta t)^2}{6}\frac{k_e}{m_e}}$$

In order to instigate the calculation process, initial conditions must be assumed such as velocity ($\dot{y}_0$) and displacement (y_0). In this case, the force term in the equation of motion can be exclusively equated to the inertial term to yield an initial acceleration as follows:

$$\ddot{y}_0 = \frac{F_e(t_0)}{m_e}$$

This can then be substituted into the equation directly above it to determine the displacement for the next time step and the output, in turn, substituted into the velocity equation to establish the velocity for the next time step.

As equilibrium must only be satisfied at an instant in time, non-linear functions can be readily accommodated for all terms of the equation of motion, but the time step must be small enough to avoid instability and inaccuracy.

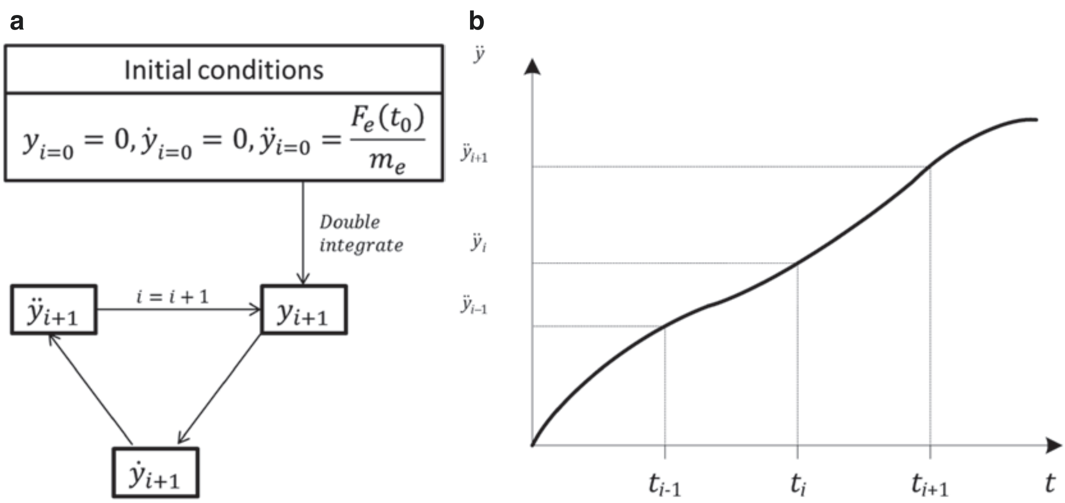

Fig. 5.8 Linear acceleration method when applied within a SDOF framework. (**a**) Calculation framework. (**b**) Temporal integration technique

5.5.2 Explicit FE and Hydrocode Techniques

In cases where levels of spatial variation (in terms of geometry, loading distribution and material characteristics) are too great to be represented via the coarse 'lumping' approach as described above, explicit FE and Hydrocode (HC) techniques can be applied. These are often used to tackle dynamic problems where, at times, the strength of the loaded material is substantially lower than the pressures exerted during the simulation. There is a requirement, therefore, for information on material behaviour at extreme regimes; this is provided by equations of state (EoS) (see Sect. 4.4.3.3). The EoS and the constitutive relationship of a material usually are termed collectively as a material model.

Despite the relative spatial complexity, the transient deformations within the model can be predicted using a similar explicit, time-stepping scheme to that describe above for an SDOF. In terms of the spatial treatment, when operating within a Lagrangian numerical framework, the application of the external load (during the initial time step) causes deformation of the nodes which bound the elements in immediate proximity to the loading. Deformations, displacements and rotations in this scheme are calculated in a similar manner to what was discussed in the previous section, whereby the stiffness matrix is evaluated at every time step. In contrast to an implicit (Newton–Raphson type) iterative, dynamic scheme, an explicit method does not require equilibrium between the internal structural forces and the externally applied loads to be satisfied at each time step.

For the stability to be maintained during the calculation, a 'disturbance' originating in one particular element must not travel beyond its immediate neighbours. The time step can be determined based on the Courant criterion for the element.

$$\Delta t = \frac{l_{\min}}{2c}$$

where $l_{\min}$ is the minimum element length dimension and c the factored wave speed (a factor of two is applied here).

When loading rates produce velocities that exceed the material wave speeds then it is these which may dictate the time step. The mesh (or series of interacting meshes) that constitute a numerical model may contain elements of different type and size; for numerical calculations using an explicit temporal solution scheme it is the minimum element dimension within the whole model that often dictates the absolute calculation time step size. Generally speaking, smaller elements (higher mesh resolution) produce results of greater accuracy, but care must be taken when developing a model that appropriate element sizes are used to avoid prohibitively long calculation times.

When considering the general stress state affecting a material under load, it is convenient in numerical analysis to separate the stresses that cause deviatoric response (change in deformation shape or 'shearing behaviour' but no volume change) from those that result in hydrostatic response (volume change but no change in shape; see Chap. 4). When generating a solution, it is worth considering that the strength-related behaviour (including aspects of failure and damage) are usually controlled by the deviatoric response whilst volumetric changes, pressure dependency and temperature (internal energy) changes are dictated by the hydrostatic response.

The time step dependence of explicit methods can lead to long run times compared to equivalent implicit approaches but the lack of dependency on convergence often allows the representation of more complex material interactions (such as those associated with component contact) and material damage. In scenarios involving highly transient phenomena, explicit methods can also better facilitate the coupling of two different numerical schemes. For example, Fluid Structural Interaction (FSI) can be readily implemented to couple a Lagrangian structural mesh to an Eulerian mesh which contains a fluid. The approach is often used when modelling the effect of a blast loading on a structural entity. As discussed earlier, the Eulerian scheme is better suited to tracking the rapidly expanding blast wave whilst the Lagrangian scheme is often better suited to tracking complex material responses when excessive deformations are not involved. An example of explicit FSI is described in Chap. 28 for analysis of human response to fragment penetration. In relation to the discussion on stability earlier in this section, the consideration of interactions between different materials or different schemes can result in yet further time step factoring.

5.6 Verification, Validation and Sensitivity Studies in FEA

A computational model is an approximation of a physical phenomenon; as such, it is inherently inaccurate to some extent. It is important, therefore, to ascertain the utility, credibility, predictive ability and interpretation of each computational model.

5.6.1 Verification

Verification of a computational model according to the American Society of Mechanical Engineers (ASME) is 'the process of determining that a computational model accurately represents the underlying mathematical model and its solution'.

Verification is associated with the writing of the mathematical code itself. A verified code will have been tested against benchmark problems for which analytical solutions exist; this means that the underlying physics, numerical discretisation, solution algorithms and convergence criteria of the code are correctly implemented. This process has been conducted for all commercially available FEA software.

What an FE code cannot guarantee is the so-called calculation verification, which is associated with the user. Calculation verification in FEA is usually conducted by assessing the adequacy of the discretisation in space and time. Usually FE model predictions are 'stiffer' compared to analytical solutions and so a mesh refinement will make the model more 'compliant'. At the same time, if the problem is dynamic, the time step has to be decreased to account for the smaller element sizes as per the discussion in the previous section. Refinement is a process of diminished returns, as there exists a level of discretisation beyond which the change in the resulting values of variable(s) the analyst is interested in (such as stress and strain) is adequately small. Refinement comes with computational cost; therefore, the modeller needs to decide how much discretisation is 'enough' and communicate it appropriately. Terms such as mesh convergence or mesh validation are used to refer to verification of the spatial discretisation in FEA.

Verification of a model does not guarantee adequate predictive ability of a physical phenomenon, but only consistency in prediction. If the model is used with substantially different loading or boundary conditions to those that the verification process was carried out initially, then a consequent mesh verification study might be necessary. Experience and good engineering judgement are critical in this process.

5.6.2 Validation

A model can be considered validated when its predictions fall within an acceptable range of the corresponding experimental results. The range over which an FE model has been validated depends on the case and should be communicated appropriately. It is usually good practice to design validation experiments that are well controlled and test specific predictive abilities of the FE model; the intention is not to carry out the complex experiment that the FE model is trying to simulate (else what do you need to model for?), but to build confidence in the predictions of the FE model at specific key locations for specific variables, such as strain. After all, an FE model is usually developed in order to explore parameters and behaviours that are impossible or very expensive to quantify experimentally. It may be that model predictions are not similar to the corresponding experimental data, in which case the model cannot be deemed fit for its intended use; a reassessment of FE model parameters and assumptions, consequently, is essential.

5.6.3 Sensitivity / Uncertainty Quantification

Sensitivity studies involve altering key input parameters of the FE model over a sensible/expected/physiological range in order to assess the dependency of key outputs, such as strain, on them. It is common for a number of input parameters to be associated with uncertainty. It is important to assess the dependency of the prediction on especially uncertain input data; essentially, this forms part of the validation process as it will dictate the range of input/loading parameters over which the model is valid. Sensitivity studies may be conducted using statistical (such as Monte Carlo or Taguchi methods) or one-at-a-time approaches, depending on the application.

5.7 Conclusion

Computational modelling is a powerful tool to analyse deformation and flow. There are many different formulations of computational models; this chapter has focused on computational continuum mechanics, where there are different approaches to discretisation and two main approaches to solving the problems: implicit FEA and explicit FEA. In all cases, model formulations must be suitably verified, and validated for the relevant use case, using techniques such as sensitivity analyses.

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Fundamentals of Blast Biology and Physiology

6

Sarah Stewart and Claire Higgins

Abstract

Understanding the biological and physiological effects of blast on living organisms is key to developing strategies to mitigate the deleterious effects that blast confers on individuals. Blast affects organisms from nano to macro scale - by impacting DNA structure and epigentic marks, through to complete organ systems. These changes may be advantageous to the cell, through increased proliferation or viability, or deleterious to the cell causing increased cell death through damage to integral structures; this is dependent on factors related to both blast characteristics and organism characteristics. Many of the non-deleterious effects are mediated through complex mechanotransductive pathways that commence at a cellular level following exposure of the cell to a blast wave. The mechanical stimulus conferred by a blast wave activates biochemical signalling pathways within the cell. This results in fundamental changes which can ultimately alter the cell's phenotype and also generate adaptive transformation enabling the cell to respond to its environment in a homeostatic manner. Identifying the threshold whereby the effect of blast is sufficient to stimulate these mechanotransductive pathways without having a deleterious impact, as well as how these pathways can be modulated to alter cell behaviour, forms the cornerstone of much of cellular blast research. In order to better understand these effects, it is necessary to consider the effect of the blast wave at a number of biological hierarchical levels. This chapter aims to explain the fundamental biological and physiological effects that a blast wave and its associated energy confer on the structure and function of cells, which make up tissues, organs and organisms.

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6.1 Basic Cellular Biology

The cell is the basic unit of life and found in all living organisms. Although cells vary widely in their physical characteristics, they have the unifying denominator of possessing liquid cytoplasm containing a solid nucleus, all encapsulated by a cell membrane. In order to understand the relationship between blast waves and cells, a basic explanation of each of these components is given here (Fig. 6.1).

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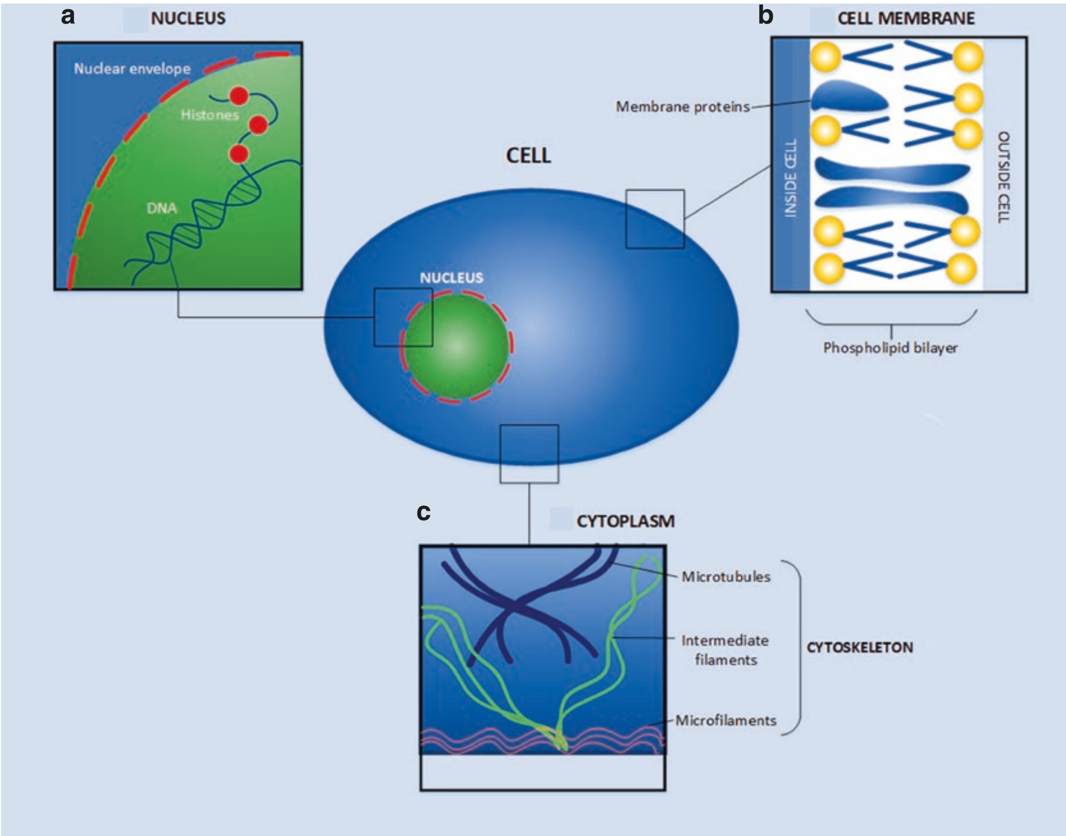

Fig. 6.1 The cell is the smallest structural and functional unit of an organism, with the unifying characteristics of a nucleus, cell membrane and cytoplasm. (**a**) The nucleus contains DNA linked to histones (nuclear proteins), enclosed within the porous nuclear envelope. (**b**) The cell membrane comprised of a phospholipid bilayer, interspersed with membrane proteins which can serve several functions, including ion channels, attachment sites and molecular sensors. (**c**) The cytoskeleton confers integrity to the cell cytoplasm and conveys messages to the nucleus through structural deformation. It is comprised of microtubules, intermediate filaments and microfilaments

6.1.1 Nucleus

The nucleus in a plant or animal cell is typically a spherical structure, consisting of nucleoplasm within a nuclear envelope. Within the nucleoplasm are found dense bodies called chromatins, consisting of deoxyribonucleic acid (DNA) and proteins referred to as histones. Genetic information encoded in the DNA is transcribed into proteins via messenger ribonucleic acid (mRNA). The mRNA can therefore be considered as an instruction released from the nucleus to drive protein synthesis in the cytoplasm. mRNA passes into the cytoplasm through pores in the nuclear envelope, which selectively regulates the passage of molecules into and out of the nucleus. The nuclear envelope therefore serves not only to contain and segregate DNA from the rest of the cell, but also to control the process of protein synthesis by release of mRNA.

6.1.2 Cytoplasm

The cytoplasm forms the bulk of the cell's mass, enclosing the nucleus and surrounded by the plasma membrane. The characteristics of the cytoplasm will differ significantly between cell types. However, common features include an endoplasmic reticulum (ER) to synthesise cellular material, golgi apparatus to sort the materials synthesised, and mitochondria which supply energy to the cell. Interspersed between these entities are the cross-linkages of the cytoskeleton, comprising microfilaments, intermediate filaments and tubules enabling a cell to both maintain and change its structure, according to what function it has been assigned to fulfil. The cytoskeleton also acts as a highway to transmit messages (in the form of forces) from the cell membrane to the nuclear membrane.

6.1.3 Cell Membrane

The cell membrane is a dynamic structure forming a boundary around the cell, comprising of two principal entities: a phospholipid bilayer and proteins. The phospholipids that form the bilayer are amphiphilic, having both hydrophobic and hydrophilic components. This results in a bilayer naturally forming: the hydrophobic tails line up to form the bilayer interior, and the hydrophilic heads of the phospholipid molecules are exposed to the aqueous exterior. Although small, non-polarised ions can diffuse passively across the phospholipid bilayer, the majority of molecules and biological substances cannot and require active transport utilising membrane proteins.

Membrane proteins make up approximately 50% of the plasma membrane and serve a number of vital cellular roles. They can be either embedded in the plasma membrane ('intrinsic'), associated with the membrane on either its interior or exterior ('extrinsic') or can span across the entire width of the phospholipid bilayer ('membrane spanning'). As well as actively transporting molecules across the cell membrane, they also function as receptors for molecular messengers such as hormones, act as anchors to provide attachment points both within the cytoskeleton and extracellular matrix, serve as sensors to guide a cells behaviour within its microenvironment or operate as markers to aid in cellular self-recognition.

6.2 The Biological Hierarchy

Tissues are collections of specialised cells whose individual role is orchestrated to fulfil the overall function of the tissue. Four different tissue types exist in mammals depending on their embryonic germ cell origin: nervous tissue (derived from ectoderm), connective and muscle tissue (both derived from mesoderm) and epithelial tissue (derived from ectoderm, mesoderm and endoderm depending on location within the body). The infrastructure of connective tissue is maintained by an extracellular matrix (ECM) between the cells, which acts as a bulk compound to provide structural support and integrity to the tissue. It is composed of both structural and adhesive matrix proteins, including collagen, elastin and fibronectin. Cell to cell communication within the ECM is further enhanced by the presence of cell adhesion molecules known as integrins which connect to, and interact with, ECM, thus linking cells together. These connections between cells have been demonstrated to be integral in embryonic development, homeostasis, immune responses and malignant transformation. In contrast to connective tissue, epithelial tissue is characterised by a lack of ECM, and cells are in direct contact with one another. The presence of gap junctions between cells allows direct transfer of molecules between cytoplasms, whilst cell adhesion molecules including cadherins and integrins enable anchoring of cells together.

Organs are the functional units of an organism, and represent a number of different tissue types that together combine to form a structure capable of fulfilling a specific functional role within that organism. Organs rarely function in isolation and are grouped together to form 'organ systems', such as the respiratory system and the central nervous system. Together, these organ systems form a living organism.

6.3 Understanding the Cellular Effect of Blast

Post traumatic conditions such as blast traumatic brain injury (bTBI) and heterotopic ossification (HO) following blast wave exposure have led to increased interest in understanding the biological and physiological effects of blast waves on cells and tissues, as well as a clearer understanding of the mechanotransductive properties that blast waves can confer on cells (Fig. 6.2a).

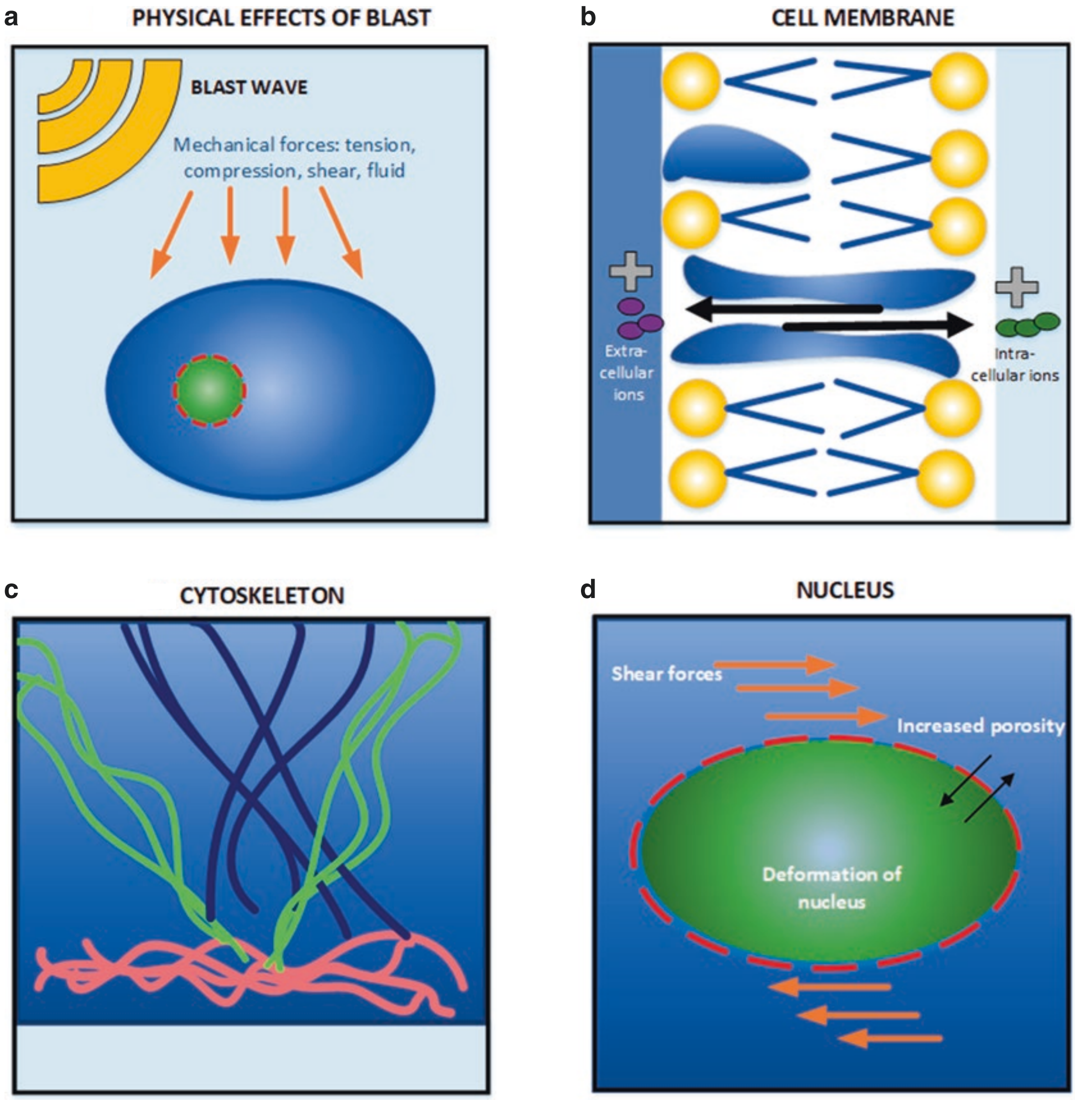

Fig. 6.2 The physical effect of a blast wave on a cell. (**a**) A blast wave can exert a number of mechanical forces on a cell, creating a complex loading environment of deformation and strain. This serves to alter the structure and function of the components of the cell. (**b**) The cell membrane increases its permeability in a bidirectional manner, allowing increased passage of molecules across the cell membrane. (**c**) The cellular cytoskeleton becomes disorganised, with microfilaments becoming shortened and thickened. This can be a reversible process. (**d**) In response to the shear forces conveyed by a blast wave, morphological changes to the nucleus include enlargement and elongation. The nuclear envelope also becomes more porous, allowing increased influx and efflux of molecules

As described in Chap. 2, shock waves are a major component of blast, and therefore experimental data utilising shock waves are included in this section.

6.3.1 The Effect of Blast on the Cell Membrane

The cell membrane has been reported to be the most sensitive part of a cell to shock wave exposure [1], reflecting its vulnerability at the periphery of the cell. In vitro work by Arun et al. applied a blast wave generated by a shock tube to human neuroblastoma cells to better understand the cellular response following traumatic brain injury in injured military personnel exposed to blast [2]. They demonstrated that the cell membrane was compromised in a bidirectional manner, with increased permeability causing both increased release and increased uptake of molecules across the cell membrane. This finding has been corroborated with other studies assessing different cell types in vitro, such as dorsal root ganglion cells which are more permeable to dye–protein complexes immediately after exposure to blast [3]. Damage to the cell membrane following blast exposure resulting in increased permeability may lead to the influx of unwanted extracellular ions, such as calcium, and the efflux of cytosolic components resulting in a deleterious effect to the cell's viability (Fig. 6.2b).

Regarding the physical process through which a cell membrane is compromised by blast wave exposure, work by Ohl and Wolfrum demonstrated two mechanisms behind shock wave-induced membrane damage in an epithelial cell line [4]. Firstly, the generation of cavitation bubbles caused by the trailing negative wave pressure after the initial positive pressure front. Upon collapse of the cavitation bubbles, the membrane increases in permeability leading to molecular uptake. This process is referred to as *sonoporation*. Secondly, the shock wave generates shear stresses, which are sufficiently strong to break the anchoring junctions between cells in its vicinity leading to membrane damage.

6.3.2 The Effect of Blast on Cytoplasm

The cytoskeleton of the cytoplasm plays a vital role in strengthening and maintaining the shape of the cell, as well as interacting with the cell membrane to control cellular functions including cell motility, morphology and intracellular transportation. Moosavi-Nejad and colleagues investigated the effect of high energy shock waves on the cellular ultra-structure conferred by the cytoskeleton [5]. Using immunofluorescence staining, they demonstrated that morphological changes to the cells were a result of the disorganisation of the intracellular cytoskeleton filaments, with actin (a major cytoskeletal protein) becoming disorganised, shortened and thickened (Fig. 6.2c). This disorganisation occurred immediately as a direct physical effect of the blast, as three hours later deformed cells reorganised their cytoskeletal network, demonstrating the reparative and regenerative capability of the cytoskeletal framework.

Beyond the cytoskeleton, a number of signalling pathways exist in the cytoplasm to transmit messages to the cell nucleus or the ECM. The molecules that make up these signalling pathways can also be augmented by blast waves. Kinase pathways are comprised of enzymes that modulate the actions of proteins through phosphorylation and form important intracellular signalling pathways. When the cell is not physically compromised, pathways augmented by mitogen-activated protein kinase (MAPK) and extracellular signal-related kinases (ERK) have been shown to be activated by shock waves. This activation led to enhanced proliferation of three different cell types in vitro (mesenchymal progenitor cells, adipose tissue-derived stem cells and T-cells), demonstrating the beneficial effect that shock waves can confer on a variety of cells [6]. Subsequent development of a rodent ischaemic wound model by the authors showed that this proliferative effect on cells by shock waves improved wound healing, providing scope for its use in the clinical sphere.

6.3.3 The Effect of Blast on the Nucleus

Early work by Steinbach et al. looking at the threshold sensitivities at which cell components were damaged following exposure to high energy shock waves demonstrated that nuclei required the highest threshold of energy to incur damage [1]. Energy levels of 0.55 mJ mm^{-2} (compared with 0.12 mJ mm^{-2} for the cell membrane) were required to induce morphological changes in the nuclei of cells following shock wave exposure. Similarly to the cell membrane, the nucleus is also subject to shear stresses in the direction of the wave front. The shear forces are generated by the velocity gradient between the nucleus surface and the other cell organelles, which results in elongated deformation of the nucleus along the shock direction [7] (Fig. 6.2d). This results in compromise of the nuclear membrane, illustrated in work by Arun et al. who demonstrated increased uptake of a nucleic acid-binding dye after blast wave exposure [2]. Evidence for enlargement of the size of the nucleus following blast wave exposure has also been observed. Logan et al. found increased nuclear area in mesenchymal cells, starting at six hours following blast wave exposure. This increase in nuclear size was not evident immediately after the blast event, indicating possible restructuring of the nucleus as a key tenet of mechanotransductive pathways following blast, rather than a direct physical effect of the blast itself.

6.4 Whole Cell Response

6.4.1 Cell Viability

So far, the response of individual components of the cell to blast exposure has been discussed. It is also important, however, to understand how the cell as a single entity responds to the energy conferred by a blast wave in terms of autonomous function and its interaction with other cells.

The effect of blast on cell viability is an area that has been researched extensively. A US military research lab investigated the effect of an explosive blast on dissociated neurones. They demonstrated that although cell death was minimal after a single insult (of approximately 32 psi), cell viability decreased significantly following repeated blast exposure. They attributed this to the repeated mechanical damage to the cell membranes and increased ion transport [8]. The amplitude of the blast can also play a role; Logan et al. demonstrated that rat skin fibroblasts had a significant decrease in viability 24 h after shock wave exposure with a peak blast overpressure of 127 kiloPascal (kPa), but not at 72 kPa [9]. Conversely, however, external physical forces are also capable of increasing cell viability. Vetrano et al. showed that tenocytes (differentiated cells derived from tendons) are metabolically 'activated' by shock waves and significantly induced to proliferate at multiple timepoints post-shock [10]. Osteoblasts display a dose-dependent increase in proliferation following exposure to shock waves in vitro [11]. Evidently, a balance exists between the capacity of a blast wave to have a deleterious effect on a cell or an effect that confers the opposite: an increase in proliferation or viability. Finally, it appears in some cases that a shock wave can confer both effects: fibroblasts demonstrated a decrease in viability one hour following low intensity shock wave exposure, with a subsequent increase in proliferation from day 6 to 9 compared with controls [12].

In addition to amplitude, the nature of the forces applied to the whole cell following blast can also influence cell viability. The presence or absence of shear forces following application of blast may be a key determinant in post-blast cell survival. Ravin et al. found little cell death in human brain cells at pressures as high as 220 psi in the absence of shear [13], correlating with in vitro and in vivo work demonstrating that shear and tension strains are significantly more damaging to tissue than compressive strains [14]. Shear forces causing stretching and microdamage to neuro-axonal structures underpin the aetiology of bTBI [15]. Minimising the shear forces that blast waves confer on cells may therefore prove to be a novel way through which morbidity and mortality following blast exposure can be reduced.

6.4.2 Mechanotransductive Pathways Within the Cell

The ability for cells to sense and respond to external physical forces is vital in development and physiological tissue function. Cells experience numerous mechanical stimuli in vivo, from the shear stresses that are conferred to an endothelial cell membrane from blood flow through a vessel, to the gravitational loading of osteocytes on weight bearing. These external forces act to activate intracellular biochemical signals, altering the behaviour of that cell and fundamentally changing the function of the surrounding tissue. This process is termed *mechanotransduction.* The adaptation of a cell to its surrounding physical environment through mechanotransductive pathways forms the cornerstone of homeostasis (the ability for a living system to maintain a dynamic state of equilibrium), as well as proliferation, differentiation, apoptosis and organ development. Consequently, aberrations in these pathways are implicated in a wide spectrum of disease states including cancer, muscular dystrophies and hearing loss [16]. In terms of blast, a cell in one location of the body may respond quite differently to the same cell in another location, due to differences in the mechanics of the surrounding tissue which regulate both cell function and response.

The ability of cells to alter both their own function and the function of surrounding tissue in response to external physical forces has been a major area of research interest in recent years. Application of in vitro physical forces to cells to modulate their function and phenotype has been demonstrated through utilisation of shear stress, electromagnetism, gravity, stretch, hydrostatic pressure and piezoelectricity [17–19]. Given that blast waves also confer a physical force, they too have also become a focus for mechanotransductive research in recent years (Fig. 6.3).

Logan et al. evaluated the mechanotransductive effects on whole cell function of human mesenchymal stem cells exposed to a blast wave in vitro [20]. They selected a blast wave peak pressure which did not cause cell death, but instead enhanced osteogenic differentiation, and evaluated the effect of the blast wave on both genetic and epigenetic changes prior to onset of differentiation. Looking at DNA methylation profiles to identify which signalling pathways were responsible for these cellular changes in response to the blast wave, they identified a transmembrane receptor integrin ITGAV as pivotal; the ITGAV gene was demethylated which led to an increase in ITGAV mRNA expression in response to shockwave exposure. The authors postulated that ITGAV orchestrates the determination of whether a human Mesenchymal Stem Cell (hMSC) commits to an osteogenic fate, and showed that chemical perturbation of ITGAV expression suppressed the differentiation response of MSCs to the shockwave. This ability for a shock wave to alter the differentiation potential of a cell through mechanotransductive pathways has been observed by others, with additional mediators including Ras (a receptor anchored to the cell membrane) and P2X7 receptors (an ATP-gated cation channel in the cell membrane) also being implicated [21, 22]. The work by Logan et al. was the first to demonstrate the phenomenon utilising a shock wave that recreated a blast wave experienced in an open environment.

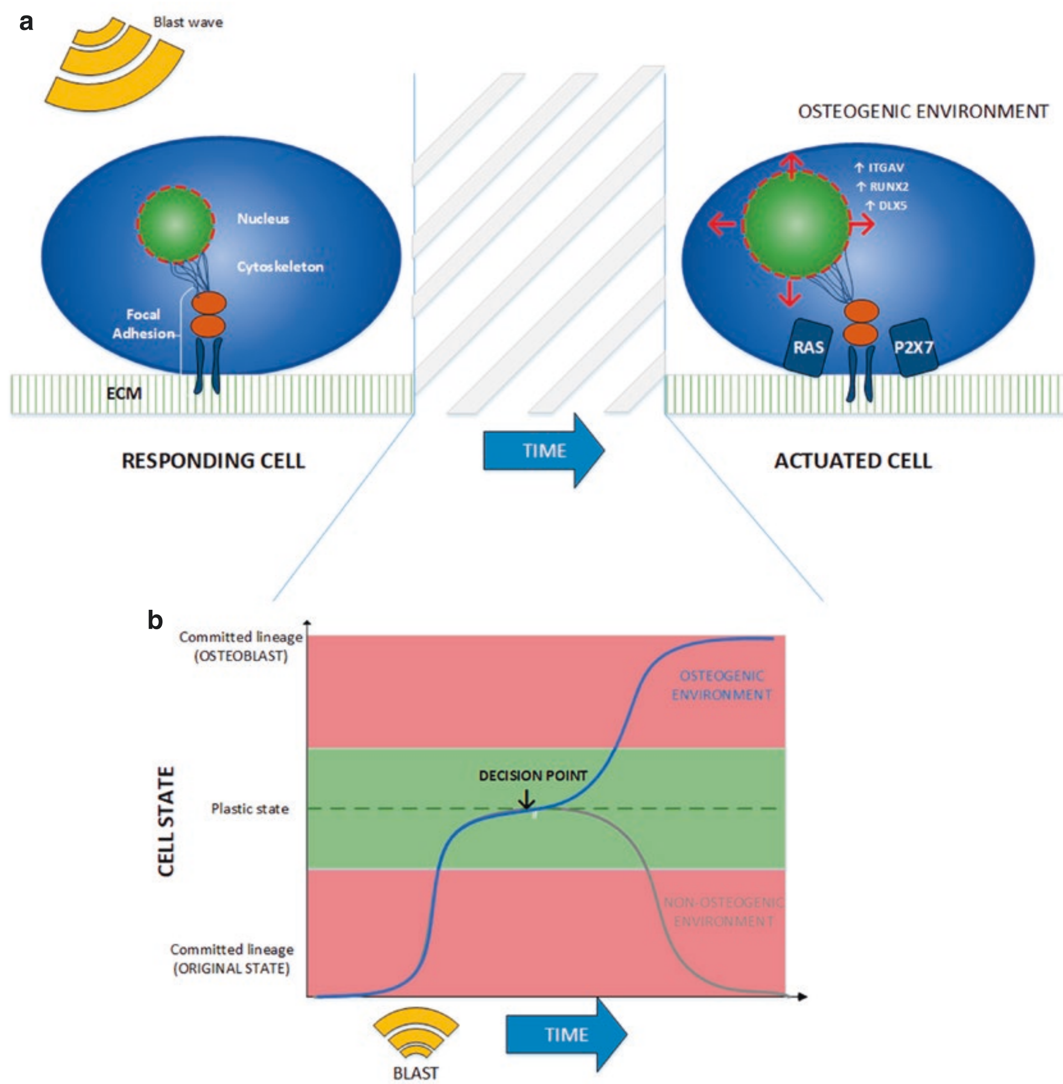

Fig. 6.3 Mechanotransductive changes following blast. (**a**) Following exposure to a blast wave, several events are set in motion. Transmembrane proteins termed 'integrins' link extracellularly with the ECM and intracellularly with proteins, together forming a focal adhesion complex (FAC). FACs serve to convert information about the external environment into intracellular biochemical signals. This is referred to as 'mechanotransduction'. As a result of the blast wave, the cell undergoes structural and functional reconfiguration. Activation of mechanotransductive pathways leads to reorganisation of the cytoskeleton, nucleus enlargement and alteration of transcription factors. (**b**) Following blast wave exposure, mechanotransductive processes within the cell may elevate it into a highly plastic state. If environmental conditions have not changed, the cell will revert to its original state over time. In the appropriate conditions, however, such as in an osteogenic environment, the increased plasticity results in commitment of the cell to a different lineage

6.4.3 Cell Interactions with Its Environment: The Inside-Out—Outside-In Concept

Once a mechanotransductive pathway has been altered by an external force such as a blast wave, a cell can fundamentally undergo a change in its phenotype as discussed above. How does this change get communicated to adjacent cells in order that an entire tissue undergoes alteration? Much of this two-way conversation between the external environment and the intracellular environment is orchestrated by integrins.

Integrins are transmembrane proteins which have been identified as being part of a critical, two-way mechanotransductive pathway termed the 'ligand-integrin-cytoskeleton' linkage. The integrins act as a link between extracellular ligands, transmitting forces to the intracellular actin cytoskeleton (*'outside-in'*), whilst also propagating signals from intracellular domains to the extra-cellular matrix to affect the binding affinity of extra-cellular molecules (*'inside-out'*). The integrin complexes are made up of two heterodimers, termed the alpha and beta subunit, which together form the transmembrane protein.

Mechanotransduction via the 'outside-in' manner is predominantly controlled by integrins acting through focal adhesion complexes. Focal adhesions (FA) are macromolecular protein complexes formed by the binding of integrins to intracellular 'linker proteins'. The linker proteins include talin, vinculin, paxillin, p130Cas and focal adhesion kinase (FAK) [23–25] and, once adhered to the integrins, the FAs form a connection between the extracellular matrix and the cytoskeleton. These FAs serve to transmit forces from 'outside in' through their connections to the F-actin cytoskeleton via their attachments to linker proteins, which subsequently activate mechanotransduction signalling cascades. The phenotypic destiny of MSCs has been shown to be augmented by the rigidity of the surface on which they are placed. Seminal work by Engler et al. demonstrated that MSCs placed in a soft matrix adopt a neuronal cell lineage, whereas those placed on a stiff substance differentiate into bone-like cells [26]. These changes were determined by mechanosensitive channels interacting with FAs, subsequently stimulating signalling channels downstream to augment transcription factors which commit the cell to a specific lineage.

'Inside-out' signalling is coordinated through integrins connected to the extra-cellular matrix. Intracellular events are transmitted through the integrins and into the ECM. This can lead to alteration of the binding affinity of the cell and changes in the architecture of the ECM (including alterations in its stiffness and rigidity). This in turn will have an effect on surrounding cells, which on detection of changes in the ECM can alter their own behaviour (including alteration in their phenotype).

The 'outside in-inside out' concept may explain the behaviour of cells in response to a blast wave, particularly the ability for mesenchymal stem cells to preferentially differentiate into a particular cell lineage. Clearly, there exists a tipping point between the detrimental impact on cellular structures that a blast wave can confer, and the positive, non-detrimental effect that a blast wave can have. It is yet to be determined whether a non-deleterious blast wave can put a cell into a 'primed', highly plastic state, with the eventual differentiation response of the cell to the blast dictated by the cellular environment. It is these mechanotransductive, non-detrimental effects whereby cell lineages and entire tissues can be fundamentally altered by blast waves that require further research, as their clinical applicability and translatability may be paramount to help with conditions such as heterotopic ossification and bone non-union.

6.5 Summary

Exposure to blast has a profound impact at a number of biological levels, from microscopic cell organelles to whole organisms. Whilst the physiological effect of blast waves on cells can be deleterious, it is evident that blast waves and their analogues also possess the ability to modulate cells and surrounding tissues in a non-fatal manner. Although the pathways are complex and there remains much to be elucidated in the field of blast wave physiology, research to date has shown promising evidence of how blast waves can modulate the behaviour of cells through mechanotransductive techniques. Improving our understanding of mechanotransductive pathways, as well as mechanisms through which they can be influenced, may prove to be key in improving the outcomes of blast-induced conditions such as bTBI and heterotopic ossification.

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Part II

Weapons Effects and Forensics

7 Section Overview

Peter F. Mahoney and Jon Clasper

The aim of this section is to give the researcher, investigator or clinician the tool kit to analyse blast incidents and explosive injury in a methodical way.

Chapter 8, Weapon construction and function, explains the basics of industrially manufactured and improvised systems. A key lesson to take away is that weapons are designed with the target in mind.

When the targets are people, or vehicles and buildings containing people, different types of injuries may be produced. These are discussed and explained in Chap. 9, Blast Injury Mechanism.

The learning from these two chapters is brought together in Chap. 10, Analysis of explosive events.

Explosions leave evidence traces and trails. Section 10.1 considers the physical evidence at the scene of an explosion and how this should be collected and managed. Sections 10.2 and 10.3 describe how documents, reports and images can be analysed to draw conclusions and produce statements to assist court proceedings.

When undertaking casualty analysis, the question of survivability often arises. This can be addressed by the methods as described in Chap. 10. Injury scoring, both anatomical and physiological, may form part of this analysis—but has limitations. These limitations are discussed in Chap. 11.

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8 Weapon Construction and Function

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Abstract

The term 'weapon' covers a wide range of articles which can be used in combat to injure or to kill, it includes bayonets, grenades and chemicals. Military forces use weapons for various reasons including neutralising, illuminating, screening or suppressing a target. When considering those weapons used to neutralise, there are a number of intended targets including people, vehicles, structures, equipment and aircraft. This chapter will focus on those designed to neutralise the human target, i.e. weapons with an anti-personnel effect which are designed to transfer lethal force to a body.

8.1 Introduction

The main means of delivering an effect to personnel is through the use of shrapnel, fragments and blast or, more often, a combination of all three. Despite commonly being used interchangeably, shrapnel and fragmentation are technically two different phenomena. Shrapnel is pre-formed, contained within a munition, generally homogenous in size and shape and dispersed rapidly on initiation. Fragments are the small metal pieces of the munition casing when it is broken up following initiation of the munition and are much more random in size and shape.

When an explosive is initiated, there is a sudden release of energy which results in pressure waves being propagated away from the site of the explosion. This is a blast wave and can cause significant injuries even when not accompanied by fragmentation. Other by-products of an explosive initiating are light, heat and sound which can cause injuries in their own right if experienced in close proximity.

Military weapons are manufactured to a high standard and undergo considerable testing and evaluation before they are brought into use and, as a consequence, the results when they function can be predicted and quantified. The opposite is true of Improvised Explosive Devices (IEDs) as they are often manufactured to lower standards, with lower quality materials and the result can be much less predictable.

8.2 Small Arms Ammunition

Small arms ammunition, an example of which is shown at Fig. 8.1, refers to rounds up to 20 mm in diameter which are fired from a weapon system. The initiation of the explosive component, the propellant contained in the cartridge case, occurs in the weapon system to propel the bullet or

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projectile towards the target. As such, there is no blast affect at the target but the effect on the human is caused by energy transfer. The maximum effect is achieved if the bullet does not leave the body meaning all energy is transferred to the target. The damage inflicted is dependent on a number of variables including the calibre and velocity of the bullet on impact as well as the organs which are affected either by the bullet itself or by the shock wave imparted to tissues, organs and structures.

Fig. 8.1 A sectioned small arms round showing the bullet, the cartridge case and the percussion cap

8.3 Mortars

A mortar system incorporates a launch tube which directs the force of launch, or recoil, into the ground through a base plate. Mortars have been around for several hundreds of years and it is a very simple system whereby the fin-stabilised mortar bomb, which has its own integral and augmenting propelling charges, is inserted base down into the tube and fired at low velocity and high trajectory towards the target. The most commonly used mortar systems, up to 81 mm in calibre, are man portable. A typical mortar bomb can be seen in Fig. 8.2.

Fig. 8.2 A 60 mm mortar bomb showing the fuze, body and augmenting propelling charges around the tail unit

8.4 Grenades

These are small explosive munitions which can be hand thrown, fired from a rifle or discharged from a launcher and which can deliver a number of effects from smoke to noise and chemicals. A typical high-explosive hand grenade comprises the body, the explosive filling, fragmentation (which can be either a separate component or from the body itself) and a fuze as illustrated in Fig. 8.3. The fuze generally incorporates a safety pin which holds the fly-off lever in place and which the firer has to physically remove, a pyrotechnic delay to allow the grenade to travel a safe distance from the firer before initiating and a detonator. It is essential that the ergonomic design and weight of the grenade enable it to be thrown further than its lethal radius.

Fig. 8.3 A hand grenade showing safety pin, fly-off lever, detonator, explosive fill and pre-notched grenade body for fragmentation

8.5 Artillery Shells

Large calibre guns mounted on vehicles can fire munitions with a sizeable payload facilitating the delivery of a target effect out to great distance. Figure 8.4 illustrates the shells which need to be robust enough to withstand high firing forces within the barrel and can be used to deliver a wide range of materials including high explosives, armour piercing darts, sub munitions, chemical agents and propaganda leaflets. When used in an anti-personnel role, the kinetic energy (KE) of the fragments is significantly lower than the KE of a bullet but hundreds of sharp sided, hot fragments are delivered to the target as a result of the initiation of the explosives increasing the probability of inflicting damage to the target. This is illustrated in Fig. 8.5.

Fig. 8.4 Artillery shells showing the fuze cavity at the top and the differing thickness of the shell walls which on initiation of the explosive fill become fragments

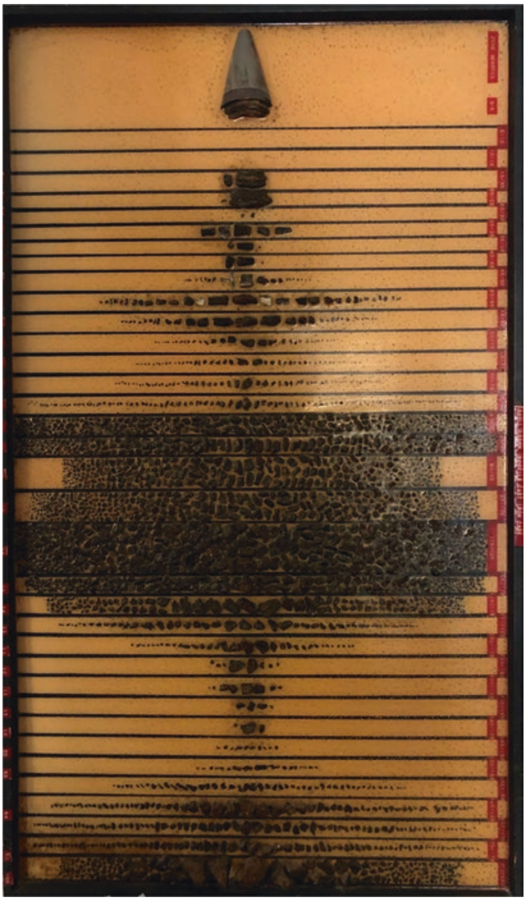

Fig. 8.5 The quantity and size of fragments resulting from initiation of a 105 mm artillery shell

8.6 Fuzes

A fuze is the component designed to initiate the munition at the correct time or place. Fuzes have varying degrees of complexity and can be broken down into:

- Percussion fuzes require a degree of physical force or impact in order to function.
- Time fuzes use either combustion, mechanical or electronic means to delay function.
- Proximity fuzes use electronic circuitry to function when the munition is within a predetermined distance of the target.
- Multi-role fuzes are designed to utilise a combination of the methods listed above.

8.7 Munition Components

Regardless of whether describing a conventional military munition or an improvised device, all require some basic components in order to function. The more complex the item the more components, but the basics are container, main charge and initiator (Table 8.1).

Table 8.1 Examples of munition components

	Container	Main charge	Initiator
Round 5.56 mm Ball L2A2	Brass cartridge case	1.52 g propellant	Percussion cap
L4A1 Mortar bomb 60 mm HE	Cast iron body	300 g TNT	Fuze nose percussion direct action M9815C1
L109A2 Hand Grenade	Pre-fragmented cast steel spherical body	155 g RDX/TNT high-explosive filling	Percussion fuze which functions when the fly-off lever is released
Shell 155 mm HE L20A1	High carbon steel alloy	49 × M85 dual purpose (shaped charge and fragmentation) bomblets	Fuze nose electronic time L132A1
Petrol bomb	Glass milk bottle	Petrol	Petrol-soaked rag
Pipe bomb	Scaffolding pipe	Black powder	Electric detonator

The majority of military munitions form part of a weapon system so could have a separate launcher like a mortar, an artillery shell or a shoulder launched guided weapons but when looking at IEDs they are often self-contained to allow them to be placed and allow the individual who placed them to move away.

8.8 What is an IED?

An IED is a device which is put together in an improvised manner and incorporates a payload which is designed to destroy, incapacitate or distract. It may include military stores, but normally comprises non-military components. The IED has increasingly become a weapon of choice in use against military forces during conflict and also against civilian targets around the world.

8.9 IED Components

There are five basic components to all IEDs as shown in Fig. 8.6 but they may also include enhancements such as fragmentation or fuel:

- Main charge
- Container
- Switch
- Power source
- Initiator
- (Enhancements)

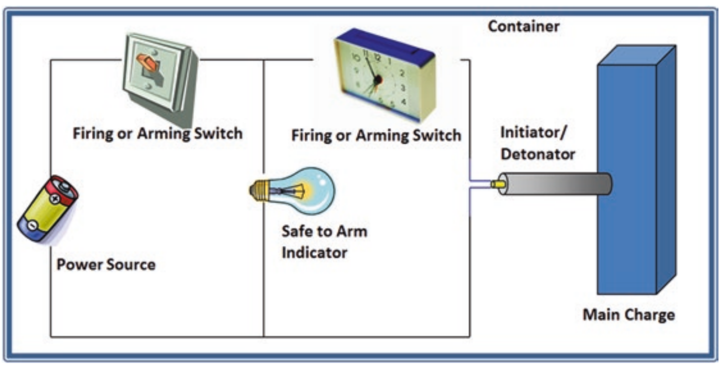

Fig. 8.6 Diagram showing the basic components of an IED

8.9.1 Main Charge

There are a number of options for this:

1. Homemade Explosive (HME). This can be easily made from a combination of readily available materials including ammonium nitrate fertiliser, aluminium powder, fuel oil and sugar. The reliability and performance characteristics of HME depend on many variables including the mix ratio, the quality of ingredients, how well the ingredients are mixed, their age, storage conditions and how long they have been mixed. All this leads to uncertainties as to if the HME will initiate, how much will initiate and how powerful the initiation will be.
2. Military Grade Explosives. This can be harvested from illegally acquired conventional military munitions and as such is of a high quality; however, its performance is reliant on how it has been stored and if it has been contaminated.
3. Commercial Explosives. These are commercially available high energy materials which could be used in the firework, mining or demolition trades.

8.9.2 Container

Containers can be anything that can hold explosives and other components. They vary in size according to the target and the desired effect. Military munitions come with an integral container but there are infinite container options for IEDs ranging from envelopes to cargo vehicles. Plastic containers which are often used prove very difficult to detect and are highly water resistant. Pipes make a good container for explosives and various groups have used pipe bombs made from scaffolding poles or heavy duty plumbing pipes to great effect. Pressure cookers have also been used; these are readily available and are tightly sealed kitchen pots which have the effect of increasing the force of the blast—they are made of relatively thin metal allowing the power of even low-grade explosives to rupture them.

8.9.3 Switch

There are different options used for switches depending on many things including available components, the skill of the device builder and how the device is intended to be initiated.

1. Command. This is where an individual at the firing point keeps control of an IED until they choose to fire it and generally falls into one of two categories, Radio Controlled (RC) and command wire. In Iraq, as in Northern Ireland, there was a continual development and progressive expansion through the Electro Magnetic Spectrum (EMS). In the early days, simple radio-controlled servos from toys along with short range key fob car alarms were used. Electronic Counter Measure (ECM) developments then saw insurgents moving to personal mobile radios (walkie-talkies) and long-range cordless telephones which were also encountered in Afghanistan. In late 2005, Iraq saw the development of its digital cellular phone network infrastructure which was then utilised by the insurgent. Further developments saw the use of frequency hopping technology being used in Iraq but there has been continued use of extraordinarily simple but effective command pull devices.
2. Victim operated (VO). These switches were regularly encountered in Afghanistan at the low technology end of the spectrum with the use of pressure bars, pressure plates and pressure release switches. These are all simple to manufacture and quick to emplace. At the higher tech end are the Passive Infra-Red (PIR) switches. These were often encountered in Iraq, especially by US forces and are associated with Explosively Formed Projectiles (EFPs). A normal household PIR switch (for example from a motion sensor light) can have the sensor window masked to make it smaller, further reducing the detection area, thus making it a very effective VO switch against vehicles.
3. Time. Incense sticks have been used in the past by animal liberation type organisations to

provide a time delay to incendiary devices used to destroy property. Acid contained within a couple of condoms has also been used as a simple but highly effective delay switch.

8.9.4 Power Source

As long as there is enough power left in a battery to heat the bridge wire in a detonator, then it is a useful power source. An issue on recent operations was troops discarding spent batteries from items like torches, or radios, which were then picked up by insurgents to use in their IEDs.

8.9.5 Initiator

The initiator or detonator contains a highly sensitive primary explosive that starts the explosive train. These can be initiated by electrical or igniferous means. Lead azide is widely used as an initiating explosive in high-explosive detonators as it does not react with the aluminium container. The manufacture of improvised detonators takes both education and skill and can also be very hazardous depending on the composition used. Improvisation is an indication that the supply of commercial or military detonators has been interrupted.

8.9.6 Enhancements

IEDs are commonly found with an enhancement to achieve a number of effects. Fuel added to an explosive device produces a blast incendiary effect which spreads the fuel across a wide area before igniting it, while if an anti-personnel effect is desired, then IEDs will typically incorporate scrap metal or ball bearings. On 15 April 2013, two IEDs exploded near the finish line of the Boston marathon with devastating consequences. The devices comprised explosive packed pressure cookers containing shards of metal, ball bearings and nails. Likewise, the fatal Manchester Arena bomb was packed with a large number of small metal objects when it was detonated on 22 May 2017.

8.10 Using IEDs to Attack Personnel

There are a number of methods of delivering the IED to the human target:

- Vehicle Borne IEDs (VBIEDs) where the vehicle acts as the container and the IED is initiated either remotely, by timer delay or by victim operation
- Suicide Vehicle Borne IEDs—similar to VBIEDs except the ‘bomber’ remains with the vehicle and detonates the IED
- Under Vehicle IEDs, these are attached to the underside of the target’s vehicle and usually designed to initiate due to movement of the vehicle
- Person Borne IEDs or ‘Suicide vests’ where the ‘bomber’ has explosive materials strapped to their person and detonates them when close to the target
- Postal IEDs or ‘letter bombs’ which are sent through the postal system and are designed to initiate once opened by the recipient.

8.11 Summary

Weapons, conventional or improvised, are merely ways of delivering a target effect and they come in many different forms and sizes and deliver many different target effects. All military personnel are trained in the use of one or more anti-personnel weapon for offensive or defensive actions, even if that is just their rifle or pistol. The aim of these anti-personnel weapons in combat is to incapacitate or kill the enemy using blast, fragmentation or a combination of the two.

9 Blast Injury Mechanism

Jon Clasper and Dafydd Edwards

Abstract

This chapter considers the mechanism of blast injury, describing the traditional classification and relating that to the type of injuries seen. It emphasises that although the different mechanisms of injury can be separated from a scientific point of view, clinically, it is rare to see an injury from a single mechanism. The cause of death after explosions, as opposed to the mechanism of the injury, is also detailed, continuing the theme that this is more than just a scientific classification of injury.

A brief description of the weapons involved is presented, concentrating on the different clinical effects from the weapons, and how the way in which the weapon has been deployed can determine the injury pattern. The importance of the environment in which the explosion occurs is also highlighted, looking at how both patterns of injury and fatalities rates are related, and how the environment is as important as the weapon itself in determining injury.

Finally, suicide bombings will be discussed, highlighting their increased use and the devastating effect this method of deployment can have on fatality rates.

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9.1 Blast Injury Mechanisms

9.1.1 Overview

Traditionally, the injurious effects from explosions have been divided into primary to quinary mechanisms, with the initial descriptions credited to Zuckerman during the Second World War [1]; a more detailed description was subsequently produced by the US Department of Defense in 2008. Whilst the classification is considered by many as the definitive one, it has limitations, particularly as the initial work related to free-field blast explosions in the open environment. It was acknowledged in the early literature that different environments will result in different injury patterns, but this seems less well appreciated in recent literature. This will be discussed in greater detail following a description of the traditional classification.

9.1.2 Primary Blast Injury

A primary blast injury results from the blast overpressure which can result in direct transmission of the wave through the tissue, as well as compression and acceleration; differential acceleration can occur at the interface of tissues of different densities and impedance. This can result in compression, shearing forces (see Chap. 2, Sect. 2.5.1) and spallation (fragment ejection due

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to impact or stress). In addition, it has been suggested that a more subtle biochemical injury mechanism can occur; given the complex nature of the body it is likely that there are multiple injury mechanisms, which almost certainly create and increase the effects of one another.

Previously, it was believed that only the gas containing structures, the lungs, ears and the hollow abdominal organs were affected; however, it is now appreciated that the blast overpressure also affects the brain, solid abdominal organs, the musculoskeletal tissues as well as other tissues. The exact extent and mechanism of injury is still not fully understood.

Blast-related lung injury is probably the most important effect of the blast overpressure in air. From a pathophysiological basis, alveoli septal rupture occurs with pulmonary haemorrhage and oedema, resulting in impaired gas exchange and hypoxia which may be fatal. Pneumothoraces and evolving pulmonary contusions further exacerbate the effects. The exact cause of the lung damage is not fully understood, and the microscopic and metabolic effects have also not been described in detail. Lung injury is further complicated in survivors, particularly the critically ill, as resuscitation, transfusion and ventilation strategies will all have effects and complicate the clinical picture.

In addition, it is believed that significant air emboli can occur, leading to cardiac and central nervous system insults; this may be the mechanism of the fatalities with no external signs of injury. It has been proposed that emboli result from the re-expansion of compressed gases in the lungs, which can access the circulation via the damaged microvasculature.

Blast bowel and tympanic membrane rupture are influenced by the environment, blast bowel being more common when the victim is in water, particularly when half-submerged such as when wearing a life vest. The impulse travels faster and further in the almost incompressible water compared to air, presumably resulting in relative protection to the chest which is above water. The concept that different injuries occurred from transmission of the blast wave through the different states of matter, gas, liquid and solid was recognised soon after the original classification [2]. Solid blast will be considered in more detail below.

Although tympanic membrane rupture is often considered a primary blast injury, it seems to be related to the position of the head and may also be related to any head protection that is being worn. It is associated with head injury rather than other primary blast injuries; tympanic membrane rupture has been shown to be unrelated to significant lung problems, the most serious consequence of the blast overpressure [3].

It has also been reported that the blast overpressure can affect the skeletal system resulting in fracture of long bones, and this was thought to be the initial event with traumatic amputations following explosions. Stress concentration at the diaphyseal/metaphyseal junction and a shattering effect has been proposed as a possible mechanism [4–7]. However, this is incompletely understood and as will be described later, there is likely to be a flail element as well as a solid blast element, re-enforcing the concept that the specific environment is at least as important as the proposed blast mechanism.

Most survivors of explosions have sustained secondary or tertiary injuries and, in general, there are few survivors with significant primary blast injuries, as casualties with the necessary blast loading will have usually been killed immediately from a combination of all effects. This is as a direct result of the fact that the energy carried by the blast wave rapidly diminishes as the energy is subjected to the inverse cube rule.

$E \propto \frac{1}{r^3}$ (where E is energy and r is distance from the explosion).

However, this also appears to be related to the environment, with a higher incidence of primary blast injuries in confined spaces. Leibovici et al. [8, 9] reported that all confined space casualties who died in hospital succumbed to respiratory failure secondary to blast injuries, and Katz et al. [10] noted a higher than expected incidence of primary blast injuries following a bus explosion when all the windows were closed. It is understood that the reason for this is due to the pressure reflections that maintain an overpressure for a longer duration.

9.1.3 Secondary Blast Injury

Secondary effects are due to fragments accelerated by the blast wind; these might be from the device itself (confusingly referred to as primary fragments by some authors, despite being a secondary effect), or other environmental objects such as stones or soil, particularly when the explosive device is buried. These objects are sometimes referred to as secondary fragments. Shrapnel is a specific rather than a collective term, and refers to a fragment containing artillery, designed as an anti-personnel device to increase its injurious effects (see also Chap. 8, Sect. 8.1). It is the fragments that are the most lethal mechanism following explosions, with a greater radius of effect than that of the blast overpressure. This is because, unlike the blast wave, the energy of a fragment is subjected to the inverse square rule of dissipation.

$E \propto \frac{1}{r^2}$ (where E is energy and r is distance from the explosion).

9.1.4 Tertiary Blast Injury

Tertiary effects relate to the displacement of the body, or the displacement of solid objects which come in contact with the body, by the blast wind and are often similar to the effects of civilian blunt trauma, although usually at greater injury levels. Head injuries are common (and may be fatal), as are fractures. Crush injuries and injuries from buildings collapsing are also included in this group.

Recently, the concept of "solid blast" has been re-described, having been ignored in the literature since first described as "deck slap" during the Second World War [11]. As this is related to transmission through a solid structure such as a vehicle floor or a ship's hull from an underwater blast, it cannot be adequately classified using a free-field blast classification; however it results in injury patterns predominately to the musculoskeletal system, similar to those seen from the tertiary effects of blast. In addition, flailing can also cause injury, either from relative restraint or protection of one part of the body compared to another, usually a limb. It appears that it can also result from solid blast. At its most extreme, it seems to result in traumatic amputation of the limb, again illustrating the need to consider the specific situation when considering blast injury mechanisms.

The concept of behind armour blunt trauma (BABT) is also placed in the tertiary blast injury category. BABT is defined as a non-penetrating injury due the deformation of body armour that lies in close proximity to the body, such as a helmet or Kevlar© chest plate, and the injury mechanism similar to that of the "solid blast" in that the armour transfers energy to the body [12]. Whilst the injurious mechanism is that of the tertiary category, the energy can arise due to the primary or secondary effects of the blast.

9.1.5 Quaternary Blast Effects

Quaternary effects are essentially a miscellaneous group of injuries not specifically associated with one of the other groups. The category was added after the original classification. Burns, inhalation injuries and other toxic effects would also be included in this category. The extent and distribution of burns have been reported to fall into two distinct patterns with one group sustaining burns to the exposed areas of the hands and face, often relatively superficial flash burns from the initial detonation [13]. The second group have more extensive, deeper burns from fires that break out after the explosion; in this group the clothing offers much less protection.

More recently, it has been proposed that quinary effects of blast should be included. These are effects from specific non-explosion related effects such as radiation, bacteria or viral infections. They have been referred to as "dirty bombs" [14]. This has been a concern with suicide bombers who may deliberately infect themselves with the hope that the infection will be transmitted from biological fragments. This is considered in more detail below.

9.1.6 Cause of Death After Explosions

Human casualty research suffers from the inability to control the environment, and lack of specific injury details, particularly fatality data. In many papers, the most common causes of death were head injury, both blunt and penetrating, and haemorrhage usually from penetrating fragment injury. With the use of Computerised Tomography in Post-Mortem (CTPM) analysis, further evidence will be produced which may also help understand the phenomenon of victims who died without any external evidence of injury; said to be one of the initial stimuli to blast research in the early twentieth century [15].

CTPM findings have been reported from British fatalities from Afghanistan [16]. These reported different patterns of injuries when comparing service personnel who were on foot to those within a vehicle, again emphasising the need to consider the environment. Those on foot were most likely to die from haemorrhage commonly associated with severe lower limb injuries, whilst those in vehicles were more likely to die of head injuries, possibly from a tertiary mechanism. It is worth repeating that this may just be relevant to a specific scenario, that of a buried improvised explosive device, and the effects of the soil on modifying the injury mechanism must be considered. In particular, the victims may have been protected from the blast overpressure, and most of these weapons were not designed to injure by fragments. As will be described in the next section, the patterns of injuries in survivors suggest that a solid blast element may be responsible, a mechanism not usually considered in the classical description of blast injuries.

9.2 Weapons

Broadly speaking, explosive weapon systems are manufactured to produce their effect by two distinct mechanisms—blast or fragmentation [17]. However, the mechanisms of injury are not mutually exclusive. Casualties can be injured by the blast or fragmentation or both.

Table 9.1 Blast pressure effects on personnel (blast injury science and engineering, 2016)

Pressure (atmosphere)	Injury
0.34	Tympanic membrane injury
1	50% chance tympanic rupture
5.44	50% chance lung injury
9.14–12.66	50% chance of death
14.06–17.58	Death

9.2.1 Blast Weapons

Conventionally, the use of the blast wave as a primary means of causing injury to a casualty is rare.

Injuries sustained due to blast overpressure are well documented in the literature. Work in the 1970s details the pressures at which the hierarchy of injury is likely to be sustained [14, 18]. Focussing on air containing compartments, the most common injury seen in blast weapons is that of tympanic membrane injury, from simple contusions through to rupture (Table 9.1).

9.2.2 Fragmentation Weapons

Weapons whose primary purpose is to cause damage due to fragmentation generally share a design that have a high casing to explosive ratio, the casing being the source of fragmentation. Whilst the lethality zone is greater than that of a conventional blast weapon, the probability of death relies upon a direct strike to vital organs. The most common type of injury seen in this scenario is that of extremity penetration. The coverage of an area by fragmentation is almost a random event and, due to the diverging effect of their trajectory and an increase in volume of the lethal zone, the further a fragment carries, the probability of coverage decreases with distance.

The mechanism of delivery of the fragment to the target is by the utilisation of the blast wave from the detonation of the High Explosive (HE). The quantity of HEs is determined by the intended target, mass of fragmentation and size of the munition.

The sources of fragmentation are multiple. The simplest and most well-known form of fragmentation device is the hand grenade. The word is thought to be derived from its comparison to the pomegranate fruit from the French and Spanish languages. However, the use of grenade type devises extends back to the Byzantine period where "Greek fire" was contained in ceramic jars. In contemporaneous times it is William Mills, and the bomb that shares his name, that is credited with the modern well-known design of the "pineapple" hand grenade. Its obvious limitations are those of accuracy and range of the thrower. With respect to artillery munitions, the fragmentation may be derived from the casing, formally called fragmentation, or by pre-formed fragments, or shrapnel within the casing. Improvised shrapnel is seen in Improvised Explosive Device (IED) and suicide scenarios where everyday objects, such as ball bearings, nuts and bolts, are utilised.

In addition, objects in the environment near the field of detonation may themselves become part of the fragmentation load. This can be intentional or unintentional. Masonry, rubble or building material may be considered in a similar manner to formal fragments, however, agricultural material, sewage or soil, which again may contaminate the injured person intentionally or unintentionally must carry special consideration during the medical management of the casualty. The possibility of biological implantation either from agriculture, other injured persons or the suicide bomber must be established from the scene.

On the modern battlefield, fragmentation injuries predominate.

9.2.3 Blast and Fragmentation Effects

In reality, blast and fragmentation mechanisms are not mutually exclusive of each other. The degree and ratio to which an individual is subjected to either modality is proportional to the distance they are from the centre of detonation (Fig. 9.1) [14]. As previously stated, it is fragmentation injury that usually predominates as casualties subjected to significantly high-pressure wave fronts to cause blast-related injury experience unsurvivable levels of trauma. Signs of blast injury itself may not become apparent for 48 h. Therefore, other blast-related injuries noted should alert the clinician to intensive observations of the effects of hidden primary blast injury. British and American experience in recent conflicts in Iraq and Afghanistan is that over one-third of casualties demonstrated blast injuries [19].

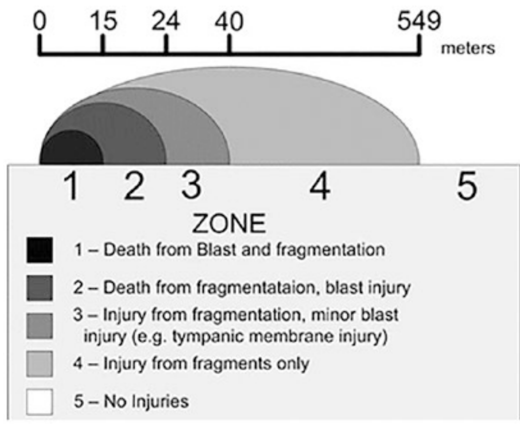

Fig. 9.1 Zone of injury from blast and fragmentation munitions

9.2.4 Mines

In contemporary history, the principle use of mines has been as a potent defensive force multiplier [19]: protecting the military assets and/or boundaries with the added advantage of freeing valuable personnel. Blast land mines exploit the simple physical properties of blast weapons, that of supersonic detonation of a high explosive. However, it is the unique interaction with the local environment that leads to specific mechanism of injury or vehicle disablement. Three distinct processes determine and quantify the degree of kinetic energy transfer that occurs once detonated:

Interaction with soil—Following total consumption of the explosive by the detonation wave heat is transferred to the adjacent soil whilst the wave front passes in the surrounding of the explosive. The heated soil is subsequently compressed creating a “cap”. On interaction of the compressed soil with the air–soil interface, two processes occur. Either the compressed cap is reflected back towards the explosive, fracturing the soil cap as it does so, or the wave is transmitted into air but with relatively little kinetic energy.

Gas expansion—All high explosives produce large quantities of expanding gas which expands at supersonic speeds. The high-pressure gas expansion escapes the detonating area by travelling through the fracture lines caused by the compression soil cap. In direct correlation with the Venturi effect, the gas pressure decreases as it seeks out the fractures lines but as it does so its velocity increases. The velocity reaches supersonic speeds. Together with the escaping jets of compressed gas and the ejected soil plug due to gas expansion, the combined effect of transfer of energy is enough to cause vehicle floor deformation or significant injury to personnel.

Soil ejecta—Soil disturbance occurs with the initial radial compression wave but as detailed before often poses little threat to individuals or vehicles. Towards the end of gas expansion, shear forces at the boundary area of the explosion crater cause the upward mass movement of the soil. Together with the detonation gases, this may cause the vertical displacement or movement of a vehicle or individual.

9.2.5 Anti-Personnel Devices

Anti-personnel devices (AP) or mines can be defined as:

> *a mine designed to be exploded by the presence, proximity, or contact of a person and that will incapacitate, injure or kill ….*

In general, they are designed to maim or injure, thereby causing not only devastating injury to the casualty but also damaging the logistical supply chain due to the mechanism required to treat and evacuate an injured soldier [20]. AP mines are classified into blast or fragmentation types which denotes the primary mechanism of injury, but in reality, a significant overlap exists.

Table 9.2 International committee of the red cross classification of anti-personnel mine injuries [21]

Type 1	Injuries from standing on a mine	Traumatic amputations or devastating soft-tissue injuries
Type 2	Injuries from a mine triggered in the vicinity	Randomly located fragmentation type injuries
Type 3	Injuries sustained from handling a mine	Severe upper limb and facial injuries

Blast AP mines exploit the use of small amounts of HEs and the subsequent formation of a blast front. They are often triggered by short-trip wire or direct pressure and, consequently, the intended casualty is rarely more than a foot from the centre of detonation. Upon detonation, a shock front is delivered directly upwards through the soil and into the lower limb. Longitudinal stress is applied through the foot and tibia and related soft tissues and traumatic amputations ensue. This is compounded by the secondary effects of the mine by the carriage of soil and casing and the dynamic overpressure after the front. The injury patterns are relatively reproducible and have been classified into three distinct groups by the International Committee of the Red Cross [21] (Table 9.2).

9.2.6 Anti-Vehicle Devices

Anti-vehicle mines (AV) were primarily manufactured in response to tanks. Initially manufactured purely as large AP mines, subsequent modifications have made them vehicle or tank specific. In comparison to AP mines, AV mines are often in excess of 5 kg in weight, including casing and detonator, with HE charges of typically 3.5–7 kg [19]. The logistic demand of carrying and placing large numbers of landmines being therefore greater than that of AV mines. The triggering mechanism is traditionally set at 100 kg to ensure target specific detonation and so therefore not waste mine strikes on smaller targets.

Whilst modern shaped AV mines can penetrate vehicles, this is a departure from the normal mechanism of targeting the track or wheels of vehicles which are seen as their weak points. The gas expansion phase of mine detonation is enough to cause damage to vehicle chassis or tank tracks to immobilise the vehicle in question. The floor pan may deform but rarely penetrate the vehicle sufficiently to cause bodily harm. However, it is the shape of the underbelly of the vehicle that dictates further energy transfer to the vehicle during the "soil ejecta" phase. A flat underbelly surface will create high concentration fields of the soil ejecta and detonation products, whereas a "V" shaped underbelly, designed to mitigate against this effect, will allow the detonation products to flow in a linear pattern horizontal to the vehicle surface and not produce increased field pressure concentration [19].

9.2.7 Improvised Explosive Device

The IED has become the explosive weapon of choice of terrorist organisations worldwide. The term IED has almost become synonymous with the conflicts in Iraq and Afghanistan due to the large number of casualties as a direct result of their use. Their design has been to ultimately not only cause devastating injuries, but their magnitude is such that it will also overwhelm logistical elements of the force to which they are directed. Whilst the road-side bomb has been used effectively, an IED is any device that uses modified conventional, or unconventional, munitions to exert their effect. They have been used to great effect in the form of pipe bombs, car bombs (Vehicle Borne - VBIED), letter bombs and indeed suicide bombs (see later in chapter). According to the United States Department of Defense the definition of IED is any:

"*devices placed or fabricated in an improvised manner incorporating destructive, lethal, noxious, pyrotechnic or incendiary chemicals, designed to destroy, disfigure, distract or harass and often incorporate military stores*".

9.3 Environmental Factors

The majority of the research into blast effects and injuries dates from during and after the Second World War, particularly in relation to the effects of nuclear weapons, hence the focus on free-field blast. However, as bombs have always been a favoured weapon of terrorists, it is this aspect rather than conventional warfare that has provided the injury data. This experience has led to the understanding that the fatality rate, and the pattern of injury is related to the environment in which the explosion occurs. In terms of air blast (as opposed to liquid or solid blast), these can be considered as:

Explosions in the open air
Explosions in confined spaces
Explosions associated with structural (building) collapse

Arnold et al. [22] reviewed the outcome of 29 incidents that collectively produced 8364 casualties and 903 immediate deaths. They analysed the relative mortality for the three types of bombing and noted that this differed depending on the environment. One in four victims died in bombings involving structural collapse, one in twelve in confined space explosions and one in twenty-five in open air bombings.

9.3.1 Open

Fragments are by far the most common cause of injury in the open environment. Blast lung and other conditions associated with primary blast injury do occur in survivors but are much less common than in confined space explosions. Fractures can occur from both the fragments and the casualty been thrown by the blast wind.

9.3.2 Confined Spaces

This generates an environment where large pressures are created for extended periods of time allowing for further energy transfer to a casualty, increasing the lethality of an explosion (Table 9.2). This has been demonstrated in literature emerging from suicide bus bombings in Israel and the Underground Train bombing in London in 2005.

Bus and train bombings can be considered as a separate "ultra-confined" space based on injury profile, with particular reference to primary blast injuries.

Kosashvili et al. [23] in a review of 12 separate incidents from Israel noted that explosions occurring in buses had the highest mortality rates (21.2%) both as a result of crowding and reflection of the blast in the confined space. Similar findings were reported from the Madrid train bombings, with Turégano-Fuentes et al. [24] noting that more deaths occurred in the two trains that had their doors closed, compared with the two that had their doors open in the station.

When compared to the other two environments, primary blast injuries such as blast lung are far more common in confined space explosions. Burns are also far more common, and fractures and head injuries are also more common than the other two groups. This is likely to be due to the close proximity to the blast, causing flash burns, and throwing the casualty against the sides of the structure. The energy involved in throwing the casualty can cause severe head injuries, as well as fractures. There is also a higher rate of abdominal injury to the liver and spleen which may cause bleeding into the abdomen, and if severe could result in a casualty bleeding to death. However, this is still less common than penetrating injuries from fragments and much less common than blast lung.

As a result of these different injury patterns, a casualty who survives the initial explosion and appears responsive but dies soon afterwards, is more likely to be in an enclosed space environment than the other two environments. A casualty who is initially conscious and talking is more likely to have died of blast injuries to the lung, or bleed to death from internal injury, as casualties with severe head injuries are usually deeply unconscious from the time of the explosion.

9.3.3 Structural Collapse

This is the most lethal environment. As well as exposure to the complex blast waves and/or fragmentation of the other two environments, there is the added risk of crush and other severe tertiary blast trauma. Fragment injures are less common that with open air explosions, but fractures are more common, due to crushing of the casualty when the structure collapses. Head injuries are also more common for the same reason. Modern buildings, particularly apartment complexes or business centres, are often designed to prevent building collapse in the event of an explosion.

9.3.4 Deck Slap

An increased IED threat during the Iraq and Afghanistan conflicts resulted in the change of tactics by coalition forces. Troops patrolling on foot were replaced by patrols in vehicles. Strategies developed to avoid or mitigate death or serious injury succeeded but a new cohort of lower limb casualties was borne. These injuries have subsequently been named the "modern deck-slap" injury.

First described in World War 2, the deck-slap injuries comprise fractures to the foot and lower limb in response to a rapidly accelerating floor, or deck, of a vehicle or ship. In the instance of the original deck-slap injuries, cases were described when ships were targeted by sea-mines. It was noted that the "*deck rose suddenly beneath the feet of those injured, and the force transmitted upward through the skeleton produced a series of injuries including fractures of the Os Calcis, tibia and knee*" [25]. These injuries are created by the direct transfer of energy from the blast wave through the ship via solid blast mechanism where the victims are not subjected to primary blast effects. It was hypothesised by Barr et al. [25] that ancient wooden ships would better protect their crews from sea-mines than modern steel hulled vessels.

An almost identical pattern of injuries has been noted in passengers of military vehicles following IED under-vehicle explosions [11]. Of note is that this casualty cohort experienced multiple segment injury to their limbs of a severity that resulted in nearly half requiring amputation. The field of injury mitigation in the "deck-slap" scenario is an area of significant research in the military and civilian environment.

9.4 Suicide Bombings

Suicide bombings have significantly increased in number in the last 50 years. Their history dates back 2000 years to Roman occupied Judaea and the establishment of the Sicarii ("daggers") Jewish sect. The following millennia saw the rise of the Muslim Hashashin Group, which lends itself to the contemporary term "assassin".

Whilst the vast majority of contemporary attacks have been seen in the middle East and Asian subcontinent, notable incidents in the developed world, e.g. New York, (09/11/01), London (07/07/05) and Manchester (22/05/17), have raised its profile in the eyes of emergency services globally. The attraction of suicide bombing by terrorist organisations is multifactorial. The combination of low cost, high lethality and pinpoint target accuracy is devastating. Bloom showed that suicide attacks cause 6 times more fatalities, 12 times more casualty figures and 8 times more media coverage [26]. The effects of a suicide bomb extend further than the physical injuries caused by the index event but also the on-going psychological impact on those that witness the incident and the local community and population.

The simplest and most effective method of delivery of an explosive is that of the individual suicide bomber. The first 10 years of this millennium witnessed the devastating effect of the individual suicide bombers using an explosive belt in Israel [27]. The clear advantage of this method of deployment is that the bomber himself or herself may choose the optimal temporal and geographical environment to detonate the explosive to cause maximum damage. Other modes particularly used by the IRA in London, although not suicide, is the use of the car bomb; the advantage being to deploy large quantities of explosives whilst the clear disadvantage being a static delivery method of the parked vehicle or final access to a target. The 11th September 2001 saw the use of multiple aircrafts in a combined attack in New York, Washington and Pennsylvania resulting in 2986 deaths. The advantage of this type of attack to the terrorist organisation involved the large quantity of casualties, logistical consumption of the emergency services and obvious media coverage. However, this form of attack requires large-scale planning, many individuals being party to the plans (19 attackers used in the hijacking of 4 airplanes) and significant funding requirements. The mechanism of injury differs between these three modes. The former two primarily resulted in a blast injury mechanism, and possible secondary additional environment factors such as building collapse, whilst the aeroplane method causes death by crash landing, such as in Pennsylvania or as seen in New York, deliberate crashing into high-rise buildings causing building collapse.

The effectiveness of suicide bombing depends on mode of delivery, accuracy, concentration of target population and volume of explosive used. A sample of 89% (135) human suicide attacks in Israel between 2000 and 2005 resulted in a mean of 3.7 fatalities and a mean of 24.2 injured personnel [27] Benmelch and Berrebi continued to explore the specifics of human capital of suicide bombers and their subsequent effectiveness [28]. Statistically significant results demonstrate a suicide bomber with an increased age, increased level of education and the detonation in a city environment results in higher fatality and casualty figures.

As noted above, the impact of suicide bombings extend beyond that of those injured during the incidents. Psychological evaluation of a cross-sectional group of the population of New York and London after their respective suicide bombings revealed that approximately 30% had significantly increased stress levels 1–2 weeks after the respective incidents despite no direct exposure [23]. Dissemination of information regarding terrorist attacks by the media has been shown to increase stress levels in a population via an indirect means.

A scenario which is particular to suicide bombing is that of biological implantation. Whilst biologically implantation amongst casualties from non-suicide explosions is possible, deliberate self-contraction of blood-borne viruses such as hepatitis B, hepatitis C and HIV poses additional medical concerns. Biological implantation can be in the form of bony and soft-tissue contamination and also the contamination of metal from the environment, or indeed fragmentation from the explosive, contaminated by the perpetrator's tissue or blood [29]. Environmental contaminants resulting in infection should be considered after an explosion within a market or agricultural environment. This has been demonstrated by high rates of candidaemia seen in casualties from marketplace suicide bombings [30]. The literature suggests routine prophylactic treatment with intravenous antibiotics, anti-tetanus and vaccinations against hepatitis B [31].

In addition to the casualties, the protection of first responders to the scene handling potentially infected material must also be considered, and personal protection equipment and prophylactic immunisation should be considered in that cohort of medical caregivers.

9.5 Summary

- A primary blast injury results from the blast overpressure which can result in direct transmission of the wave through the tissue and subsequent transfer of energy.
- Secondary blast injuries are due to fragments accelerated by the blast wind.
- Tertiary effects relate to the displacement of the body, or the displacement of solid objects which come in contact with the body, by the blast wind.
- In practice, most casualties have sustained injuries due to more than one mechanism.
- The environment in which the explosion occurs is also important, in relation to the pattern of injuries sustained.
- Broadly speaking, explosive weapon systems are manufactured to produce their effect by two distinct mechanisms—blast or fragmentation, however, these are not mutually exclusive of each other.
- IEDs can be classified into 3 groups—(1) Conventional explosive formed from munitions, (2) Explosive-Formed Projectiles (EFP), and (3) Suicide or vehicle delivered devices.
- Suicide bombings have significantly increased in number in the last 50 years and result in 6 times more fatalities, 12 times more casualties and 8 times more media coverage per event.

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10 Analysis of Explosive Events

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Abstract

This chapter considers the forensic investigation of explosions with three sections each outlining methods to assist the Court.

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In Sect. 10.1, we describe the immediate aftermath and evidence collection indicating how clinical staff caring for victims may help or hinder this process. Thereafter follow two case studies of expert panel review.

Section 10.2 reports on the 2005 ‘7/7’ attacks in London. Individuals had been reported alive after the explosions but died before reaching hospital. Had any of these individuals had died from potentially survivable injury? The absence of post-mortem CT imaging made this a complex task, so weapon effects and blast over pressure for each environment were modelled and correlated to probability of survival.

In Sect. 10.3, we report on the 1974 Birmingham Pub Bombings on behalf of the Coroner for the Inquests into the bombings. This required examination of reports and images that were over 40 years old. Hospitals had closed and relocated several times during the intervening period. Clinical notes other than autopsy reports could not be located. The reports available including scene photographs were used to assess injury mechanisms and correlate individual’s distance from the seat of the explosions. Contemporary publications in the open literature were matched to individual patients and used to give clinical detail.

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10.1 The Examination of Post-Blast Scenes

Karl Harrison and Mike Harris

10.1.1 Introduction

Post-blast scene examination has traditionally formed a component of the general training and awareness undertaken by Crime Scene Investigators (CSIs). While the environments of operation (potentially widely dispersed fields of disrupted or detonated debris), the nature of the examination (the prospect of large numbers of casualties) and the surrounding investigative concerns of a high-profile investigation with wide-ranging political ramifications all conspire to distance the post-blast scene from the general experience of most CSIs, the application of their core technical disciplines remains as important throughout the scene examination as with more routine examinations. There will undoubtedly be pressure to identify the cause of the blast in order to commence subsequent investigations. It is therefore critical that an early indication is given as to whether the explosion[1] at the scene was caused by an accident, such as a gas leak, or explosives placed with criminal intent. Indeed, for the CSI, the requirement to provide exhaustive photographic and locational documentation is even greater, given the chaotic nature of such scenes and the importance of reconstructing the distribution of debris at a later date for the courtroom, for understanding the relative position of affected individuals, or for modelling the nature and placement of the charge. As a consequence, it is crucial to understand the 'standard' model of training and approach to scenes adopted by CSIs in order to understand how adaptations to post-blast scenes might be managed. For this chapter, the focus will be on crime scenes resulting from the use of explosives that results in a chemical explosion rather than a mechanical, nuclear or electrical explosion.

[1] An explosion is "a violent expansion of gas at high pressure". Akhavean et al. (2009: 4)"Introduction to Explosives"Jan 2009. Cranfield University Page 4.

10.1.2 Coordination of the Post-Blast Scene

CSIs working for UK police forces are now almost entirely a body of civilian specialists operating in a niche role. The shift away from warranted police officers engaging in crime scene investigation began as early as the late 1960s in some police forces, but this small number greatly expanded following the publication of the recommendations of the Touche Ross Report in 1987 [1]. A further expansion of civilian specialists followed as a consequence of the growing importance of DNA evidence, as the required level of technical knowledge increased beyond the general forensic awareness of most warrant-holding police officers. By contrast, Bomb Scene Managers (BSMs) who supersede the role of the Crime Scene Manager (CSM) on post-blast scenes are much more likely to be warranted police officers who do not engage in core CSI activities, but rather gain their training and experience through specialised roles within Counter-Terrorism units.

Of vital importance to the initial management of all major crimes scenes, post-blast scenes included are the first actions undertaken by uniformed police response teams, who in relation to this role are referred to as the First Officers Attending (FOA). The role of the FOA entails not only the confirmation of the suspected major offence but also the initial identification of obvious foci of forensic attention (the presence of a body or weapon, for example), the administering of emergency first aid, the identification of obvious risks to health and safety and the recording of details relating to witnesses still present at the scene. The fulfilment of these duties should ideally be completed in a non-invasive manner that does not jeopardise the forensic potential offered by the scene—the preservation of life is recognised as the one FOA responsibility that takes precedence over scene preservation, but clearly in relation to any wide-ranging disruption such as the aftermath of a blast, this would be an impossible task, and initial disturbance of elements of the scene is an inescapable fact. On occasions, a suspect explosive device may be identified prior to it functioning either due to a warning from the terrorist group or by a vigilant member of the public. In these instances, the FOA, potentially aided by local contingency plans, will commence cordon and evacuation procedures based upon the 4Cs framework (Confirm, Clear, Cordon, Control). Thus, the CSI may attend a scene which is already prepared.

Any intervention a FOA is forced to undertake in the commission of their duties (such as forcing a door to reach the body of a victim thought to still be alive) should be recorded in detail at the earliest opportunity and that record be made available to the incident room (the advent of body-worn cameras and the saturation of CCTV systems have made this process far more comprehensive). In the example of a post-blast scene of magnitude, this is likely to comprise the actions of numerous first responders including the police, ambulance and fire and rescue assets.

Initial attendance at the major scene and ongoing examination would generally be completed by CSIs. Any CSIs deployed to a major scene would be managed directly by a CSM or BSM who has a responsibility to ensure that a forensic strategy is complied with, and that findings from the crime scene are communicated back to the Incident Room (see Fig. 10.1). Whilst the CSM is deployed to the scene with CSIs, the Crime Scene Coordinator (CSC) has overall responsibility for deploying staff to scenes; coordinates the examination strategies of numerous CSMs and ensures integration between the forensic strategy and the overall investigation directed by the Senior Investigating Officer (SIO). The role of CSC might be filled by any suitably trained individual within the Scientific Support Department, from Senior CSI to Head of Scenes of Crime, depending on the size of the police force, the complexity of the forensic investigation and the wider public impact of the offence.

Because of the close relationship between the SIO and CSC, there is an expectation that crime scene coordination should be managed from the incident room. As such, there is generally no requirement for CSCs to deploy to crime scenes, as this would compromise their pivotal management role. Post-blast scenes are somewhat different; the scale of disruption and the level of attention focused at the scene are more likely to result in a permanent coordination team being deployed to the scene.

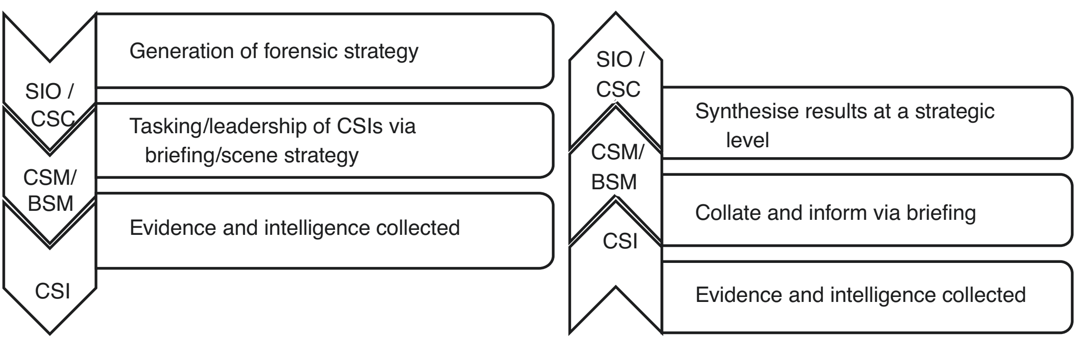

Fig. 10.1 Directions of tasking and information flow

10.1.3 Optimal Capture of Forensic Evidence

While the methods of scene examination can be adapted depending on the requirements of the investigation, the general commanding concept is that of unrepeatability; a crime scene can be revisited, but it can be examined in its entirety only once, hence there is an onus on the role of the CSC and CSM to ensure an optimal capture of potential forensic evidence. The notion of 'optimal' rather than 'total' is crucial; any one scene examined in its entirety might contain hundreds of items suitable for some form of recovery or analysis, which in turn might generate thousands of fragments of forensic data (trace evidence, fingerprints, DNA profiles, for instance). Consequently, while it is important that a forensic examination maintains a degree of independence from the investigation, it must remain driven by an investigative strategy if it is to retain any sort of focus that can bring meaning to the results of the examination.

The concept of unrepeatability of examination and the requirement to optimise evidence gathering puts great emphasis on the sequence of examination. Generally speaking, whatever techniques of examination are required at the scene, they are undertaken in a sequence that begins with the least invasive and disruptive of methods and ends with the most potentially disruptive.

All major scenes are likely to see some adaptations from the generalised approach that will form part of the written forensic strategy; such adaptations might be required by limitations of access to a scene, or environmental variations. Post-blast scenes are more likely than standard large major scenes to see the need to make considerable changes to otherwise standard scene approaches; initial scene and safety assessments must include a consideration of potential threats, such as the presence of secondary hazards, including explosive devices that have yet to function or have been designed to catch first responders, CBRN (Chemical, biological, radiological and nuclear) materials, or other risks associated with extensive structural damage to buildings—all of which can cause considerable delay to the forensic examination commencing.

10.1.4 Access Control and Cordoning

While perimeters need to be established for all crime scenes, control of access through extensive double cordons is frequently required for post-blast scenes together with large numbers of scene guards, and these might be located within highly populated urban areas with residences located within cordoned areas. The inner cordon encompasses the explosion area and has a radius of approximately one and a half times the distance from the seat of the explosion or centre to the furthest identifiable piece of evidence. Only the BSM and their team can enter the inner cordoned area until the examination and evidence retrieval is complete. The outer cordon marks a perimeter which ensures public safety while preventing those who are not associated with the investigation from disturbing the scene, destroying or moving evidence, observing examinations too closely or overhearing conversations pertinent to it. It also provides a safe working area within which members of the police and other emergency services can operate [2].

10.1.5 Explosives, Seat of Explosion, Device Identification

The explosive used will have an effect on the dispersal of debris and evidence surviving at the scene. Factors such as whether a low explosive or high explosive was used, as well as the integrity of the composition, its confinement and placement will need to be considered. Indeed, observation of debris at the scene could provide an initial indication of the type of explosive involved; debris being spread over a wide area will lead to complexes of material preserving multiple instances of forensic opportunities that would require the imposing of a sequence.

Explosive Ordnance Disposal (EOD) or Bomb Disposal Operators should be present to assist when the use of explosives is suspected. In the UK, EOD support (outside of the Metropolitan Police area) is provided by military personnel through Military Aid to the Civil Authorities (MACA) and is requested by the relevant Police Force to the Joint Service EOD Operations Centre (JSEODOC). EOD support for the Metropolitan Police is provided by their own Explosive Officers (EXPOs) department. The role of the EOD Operator is primarily to ensure explosive safety in order to allow the scene investigation to commence. Therefore, the EOD Operator may have undertaken various actions on the scene before the CSI is allowed to enter. This means the EOD Operator can be a source of information as a witness, in addition to providing a set of technical functions. Once satisfied that the immediate scene is explosively safe, the EOD Operator will 'hand over' to the CSI. At this point, it is critical for the CSI to understand what the EOD Operator has done, and more importantly, not done. 'Explosively safe' does not necessarily mean the removal of all explosive material nor does it mean that the entire area has been searched for other explosive devices or hazards. In complex post-blast cases, the BSM may require EOD support to remain on site for safety as well as in an advisory capacity.

Just as 'standard' major investigations require the identification of a range key scenes (a murder investigation might feature an attack site and a body deposition site, in addition to suspect addresses and vehicles), post-blast examination has similar specific challenges. The identification of the focus of the blast is crucial for both the sampling of material that might retain chemical traces of the explosive used [3, 4], but also to facilitate a reconstruction of items that might relate directly to the placement and nature of an explosive device. In terms of reconstructing events around the blast, the BSM must consider a strategy of examination that seeks to identify material traces that assist in building a picture of events prior to the placement of the device, the complex of activity around the blast itself, and the events that immediately followed activation.

In addition, the search of debris directly associated with the centre of the blast may reveal components of the device (timers, switches and batteries) that both assist with understanding the nature of the device's operation (and hence potentially providing intelligence regarding the technical capability of the maker of the device), as well as providing forensic opportunities related directly to the identification of the makers or placers of the device. However, as highlighted above, the spread of debris may be such that device componentry can be found hundreds of metres away from the seat of the explosion.

10.1.6 Zoning and Detailed Recording

The activities that follow a blast are almost certain to include the action of first responders discussed above, and the associated evacuation of casualties or the movement of walking wounded. The disturbance of debris associated with their activities might result in the contamination of items later found to be of forensic importance.

One of the key challenges facing the BSM is that of identifying exhibits that might prove to be of significance, forensic or otherwise, amongst a vast quantity of scattered and disordered debris. The standard means by which this is achieved is by the division of the scene into zones that then defines the aggregation of recovered debris. This enables the rapid clearance of material, while still allowing the tracing of a recovered item back to a generalised location on the scene. While the nature of the zones created depends upon the topography of the scene and the extent of the debris field, this long-standing technique can now be supplemented with three-dimensional scanning capability that assists with the reconstruction of the scene and the more specific location of items within the zones. In the UK, liaison with the Forensic Explosives Laboratory (FEL), and if deemed necessary, attendance by FEL scientists themselves at the scene may also benefit the decision-making process regarding evidence location, retrieval and best practice.

The identification of potential items of evidential importance requires a teamwork approach and, following confirmation of explosive safety, is initiated with a walk-through of the scene, during which (as with other crime scenes), evidence marking, photography and recording are constant tasks. Each evidence item is collected into an appropriate sterile container (e.g. metal cans, glass containers, paper, plastic or nylon bags) on which details describing the item and its location would be recorded. In post-blast examination, clearing a zone requires that all loose debris and material are swept up and either sieved at the scene or placed into bags or containers for further examination. The purpose of such a search is the singling out of component pieces of the device. A combination of coarse and fine sieve meshes can reveal very small items such as pieces of switching mechanisms, circuit boards, power sources and wires [5].

The meticulous examination of the seat of the explosion is usually one of the most painstaking tasks, requiring the swabbing of the area for trace explosive residues, the measurement of crater dimensions, the removal of loose debris (which may be treated as a single evidential exhibit), and further excavation of the crater in order to locate embedded components.

In addition to searches of the ground and crater, the examination of any secondary craters (if present) can also be forensically lucrative. Furthermore, items in the vicinity of the central explosion area which are positioned perpendicular to the ground—such as signposts, building walls or nearby car doors if outside; furniture or walls inside—may harbour pertinent forensic evidence whether they exhibit signs of blast damage or not. Consideration should also be given to the possible route of the blast wave given the surrounding environment as items may have been taken some distance due to pressure channelling. In addition, seeing how structures, vehicles, people and other objects have been affected will assist in the assessment of the explosive quantity and potentially identifying supplementary sources of evidence.

Fragmented remains of a device and associated explosive residues can also become embedded within skin and tissue; intended and unintended victims of the incident are therefore also potential sources of evidence. The BSM must ensure that if casualties are involved, then investigating personnel are dispatched to hospitals to recover any evidence either with emergency room staff or pathologists.

10.1.7 Forensic Intelligence and Evidence

There is an implicit challenge for the BSM and investigating police in the recognition of important intelligence gathered from blast scenes. This recognition touches on the conflation that persists between concepts of forensic intelligence and evidence, and the tendency to regard only specific forensic evidence types as being suitable providers of intelligence (most specifically those derived from DNA profiling [6]). By contrast, the experience of security services and military over many years of gathering weapons intelligence from Improvised Explosive Devices (IEDs) and other weapon systems is that devices, their placement locations and understanding of their ‘tactical design’ represent rich loci of potential intelligence. Whereas some complex enquiries that might be led in some parts by forensic intelligence in its broadest sense can be hamstrung by a syndrome of tunnel vision that equates the term ‘intelligence’ directly with ‘biometric identification’.

Alongside the role of developing and delivering strategies to conduct a full methodological forensic examination, it is the responsibility of the BSM to ensure the welfare and safety of the forensic team. All must be suitably equipped with appropriate materials to perform their tasks effectively; be provided with sufficient refreshments and breaks during lengthy investigations and the required personal protective equipment, which may include hard hats (to protect from falling debris and head strikes) and face masks to protect from noxious gases and dust which may be present in confined areas. Welfare considerations should also extend to the psychological impacts when potentially dealing with multiple casualties, body parts and other significant hazards in a high profile and therefore pressurised situation. This could also have implications for individuals after the investigation in complete.

It is also the role of the BSM to consider the use of tents and screens which can be used to guard the investigation from prevailing weather conditions or to provide a degree of privacy to the investigators. In addition, the BSM needs to consider the use of lighting to lengthen the period of examination and protect the scene, or to halt the scene examination when lighting levels are especially poor (the use of flood lights can cause deep shadows that can mask potential evidence, and it may not be best to work actively through the nights). A further consideration is the impact of certain devices on the safety of the scene. For example, transmitting devices (including radios, mobile phones and wearable devices) should not be used where there remains the potential for electrical initiation of an explosive device.

One role of particular importance for the BSM is to maintain consultation and liaison with relevant parties throughout the investigation. If there are disruptions to the investigation, zone clearance might take many days, and throughout this time it is the duty of the BSM to regularly update the SIO as well as facilitate contact with the media in order to ensure the community and other interested parties remain suitably informed about progress.

10.1.8 Conclusion

As with any major crime scene, no two bomb scenes are the same as one another, each varying substantially in size, impact and associated scene variables. The roles, responsibilities and considerations outlined above are relevant to all scenes but particular investigative tactics will vary on the unique set of challenges each post-blast scene presents. Moreover, the general examples above are predominantly applicable to civilian scenarios which are only time-gated by the pressure of closure of urban areas. By contrast, post-blast investigation in a military context may have significant limits to available time to complete an examination and potentially a lack of resources. In either care, specialist systems of operation and the associated skillsets of personnel assist in distinguishing post-blast scenes from other major incidents. Despite this, the fundamental reliance on the core skills of scene examination is clearly present throughout the investigation process, as well as in the mindset of those involved.

10.2 Case Study 1: Modelling the Blast Environment and Relating this to Clinical Injury: Experience From the 7/7 Inquest

Alan E. Hepper, Daniel J. Pope, Maria Bishop, Andrew J. Sedman, Robert Russell, Peter F. Mahoney and Jon Clasper

10.2.1 Introduction

On the 2nd August 2010, the United Kingdom Surgeon General was instructed by Her Majesty's Assistant Deputy Coroner for Inner West London (Rt Hon Lady Justice Hallett DBE) to provide Expert Witness Reports relating to the terrorist events of 7 July 2005 on the London Public Transport Network. These Reports were required to review the evidence that had been gathered during the investigations into the event surrounding the bombings. Her Majesty's Coroner asked a series of specific questions relating to the survivability and preventability (with respect to the medical interventions and care) of the deaths of many of the victims, and these had to be answered on an individual basis with a review of all of the relevant information. It was appreciated that the most appropriate and current experience of dealing with personnel injured in this type of event came from the UK Ministry of Defence Surgeon General's Department who are experienced in dealing with combat-related injuries, particularly in the context of the current operations. This was also assisted by the fact that the UK Military Medical community already had a proven technique for the regular review of operational mortality and medical response [7, 8].

There had also been concerns about the nature of the events, criticism about the initial response, and one review in particular was highly critical of the communication systems of the emergency services which led to delays in understanding what was happening during the first few hours of the events of 7 July 2005 [9]. Survivors had also raised concern at the response of the emergency services [10].

10.2.2 Approach

In order to answer all of the questions posed by Her Majesty's Coroner, a multidisciplinary team was essential. This would take expertise from the Royal Centre for Defence Medicine (RCDM), Birmingham and Defence Science and Technology Laboratory (Dstl) Porton Down.

Her Majesty's Coroner was particularly concerned with the victims who were not killed immediately by the explosions, but died prior to reaching hospital. Of interest was what happened to them; what attention and/or treatment they received, whether there were any failings in the way that they were treated, the circumstances of their eventual death, and whether any failings in the emergency response contributed to or were causative of their death.

The decision was made at an early stage that a single report covering all personnel would be inappropriate and unique reports for each of the people in question would be written. There were two reasons for this:

- The victims were all individuals and should be regarded on an individual basis.
- The reports may be released to the families of the deceased and the reports would need to be redacted to ensure what was released was only relevant to their relative. There was a risk that such redaction would leave the feeling that some vital information had been removed, and this would simply amplify any conspiracy theory or any feeling that the Government (or in particular, the Ministry of Defence or Ministry of Justice) wanted to hide something of relevance.

This increased the workload substantially, resulting in multiple unique reports.

10.2.2.1 Work Strands

The broad ranging and complex nature of these questions required a substantial investment of time to address these questions. A three-phase approach was adopted as the only practical way to answer the questions within the challenging timescale (3 months start to delivery). These three phases were conducted in series; however, any hypotheses, assumptions or conclusions from either of the analysis phases were not allowed to affect or influence the other, in order to keep all options open.

The first phase required an engineering expert in blast effects on structures and injury modelling to review photographs of the damaged carriages and bus to give a view on the likely physical effects on people close to the explosions. This was coupled with a review of the forensic evidence relating to the explosions. This provided one strand of opinion on the nature of the injuries (the blast effects and injury mechanism) that was used in the final comparison.

The second phase was a clinical review of the evidence by military clinicians to assess blast injury in the casualties. This used techniques developed both in the deployed environment and at regular morbidity and mortality reviews over a number of years [7, 8] to review mechanisms of blast injury and likely cause of death. This method has shown significant benefit in demonstrating the survivability and preventability of the deaths of personnel and to provide a robust evidence base to guide the changes in medical care and response to the critically injured patient. This was coupled with a review of the nature of injuries from other terrorist incidents to provide a baseline comparison of injury mechanisms, as well as a review in the progression of pre-hospital care to advise the Court of changes in treatment strategies that may assist in survival rates.

In the third phase, the blast environment was modelled by the structural dynamics experts [11] to assess likely blast loading on victims. This loading information was then assessed by physiology experts with access to data from experimental studies that provided a correlation of precisely measured blast data with injury, focusing principally on blast lung [12] since this is one of the most difficult aspects to evaluate from post-mortem reports. Simple modelling was also undertaken in isolation of the complex structural dynamics modelling to provide simple predictions of the risk of blast lung and other injury mechanisms.

The relationship of these phases is shown in Fig. 10.2.

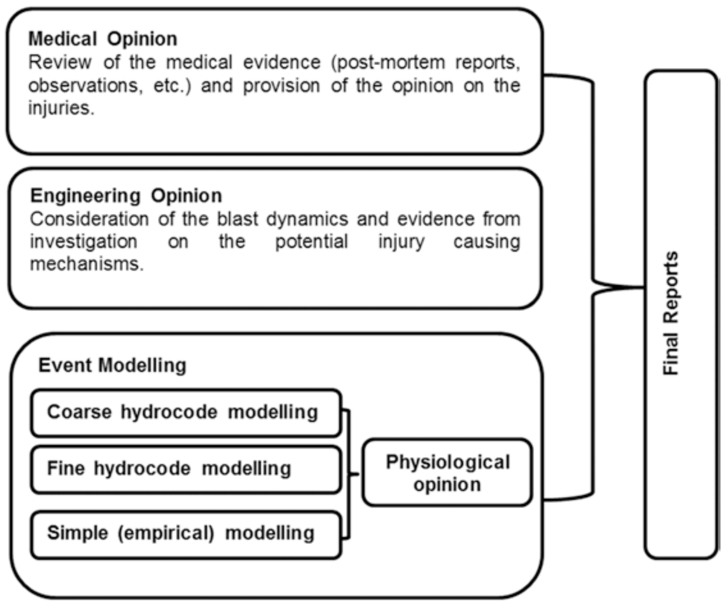

Fig. 10.2 Relationship of three phase work strands

The outputs from these three phases were combined into a joint report and a single opinion on the nature of the injuries and the survivability of personnel as described in the transcripts from the Inquest [13–15]. Each report was formatted to provide a main section written by the principal author and summarising the work that was undertaken.

10.2.2.2 Model Design and Risk Reduction

Substantial risks were inherent in the mathematical models of the blast environment because of the model complexity and the degree of uncertainty (exact charge size, exact charge dynamics, exact charge location, location and orientation of victims, etc). As a result, three different levels of model were run for each of the events in the trains:

- A coarse hydrocode model (see Chap. 5) was used to:
  - Study the mechanisms of blast load development and provide broad levels of peak overpressure and specific impulse.
  - Establish 'zones of blast wave intensity'.
  - Determine the extent to which the fireball extended within the carriage during the event.
- A fine hydrocode model to quantify the probable pressure time history loading sustained by occupants within each carriage. This model also produced images and videos of the effects of the blast that showed the blast propagation (see Fig. 10.3). These images were useful for the team, the Court and families to understand the nature of the blast environment.
- A simple (uniform blast wave model) to give an empirical relationship of blast pressure from idealised explosives and compare the results to simple estimates of lethality from blast lung.

10.2.2.3 Resources

The team had access to a combination of scene photographs, post-mortem photographs, external post-mortem reports and witness statements to form an opinion of the internal and external injuries received by the victims and for how long they showed signs of life after the bombing (if at all).

The team looked particularly at witness statements to understand if the victims were noted to be breathing and have a pulse after the bombing, whether or not they were conscious and the likely time course over which they died from their injuries.

Information provided by the court to support this activity was stored on encrypted memory drives, secured at Dstl and at RCDM, where they could be examined in a secure environment.

The scene reports included seating plans for the underground carriages and the bus indicating positions of individuals pre- and post-explosion (where this information was known) and during recovery of the deceased.

As some deceased and live casualties had to be moved at some of the bombing locations after the attacks to allow access to other casualties, the position of a victim post-explosion does not always indicate where that person was prior to the explosion or if that position was the location where they died. This meant that the team needed to use a number of methods to try and work out how close a victim was to the seat of the explosion and from this offer a view on likely internal injuries, as well as providing a review of relevant related information to inform a final opinion on the probable nature of injuries.

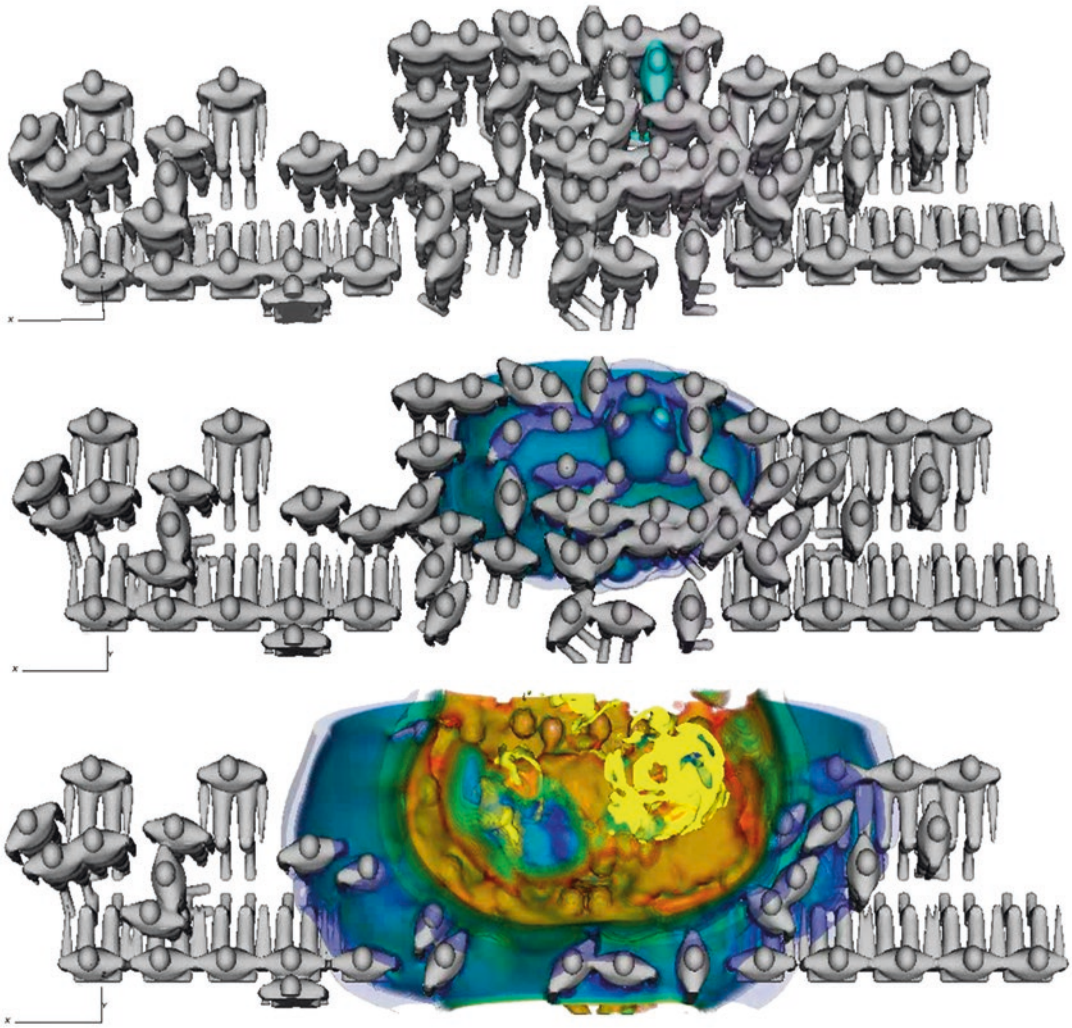

Fig. 10.3 Sample blast propagation from fine hydrocode model

10.2.2.4 Challenges: Quality of Information

Usually when conducting such a review, the clinicians and scientists looking at the information would have a complete list of the victim's injuries derived from a combination of a full post-mortem examination plus X-ray imaging. This in turn would be used to calculate mathematical trauma and injury scores which help in assessing whether or not a particular combination of injuries would or would not be expected to be survivable. On this occasion, the information from internal post-mortem examination was not available and the X-ray imaging information was limited to fluoroscopy. The fluoroscopic examination was used to identify some fractures and foreign materials present in the victims' bodies.

The team, therefore, relied upon a number of sources of information and scientific methods to come to a considered opinion for each of the victims; however, in an ideal world, more structured observations, measurements and opinions would have been available for the team to consider.

The amount of information missing from a simple external post-mortem was a significant challenge in this work. If anything can be stressed from this work, the importance of a detailed post-mortem examination must be one element.

10.2.3 Conclusion

We believe that this detailed understanding of the nature of injury from blast and fragmentation threats, and the modelling and understanding of the physical interaction of combat-related threats can only come from a multidisciplinary grouping such as the group formed to address the events of 7 July 2005 and the applicability of this form of analysis should be considered in the event of other terrorist events.

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This section reproduces the J R Army Med Corps material, with permission.

10.3 Case Study 2: Injury Mechanism and Potential Survivability Following the 1974 Birmingham Pub Bombings

Jon Clasper, Alan E. Hepper, Ruth McGuire, Anthony M. J. Bull and Peter F. Mahoney

10.3.1 Introduction

On the 21st November 1974, two bombs exploded in Birmingham. There were 21 fatalities, and it is reported that 220 people were injured [16].

On the 28th May 1975, Inquest hearings were held and evidence relating to the identification of the fatalities was given. As a criminal investigation was taking place, the Inquest was then formally adjourned. Following an application from some of the families of the victims of the bombings, the Inquest's proceedings were resumed on 1st June 2016.

In view of the complexity of explosive injuries and in particular, the length of time that had passed since the initial Inquest, expert evidence was required in order to explore any issues with causation and time of death. Required expertise included explanations of the consequences of an explosion, understanding of the variables that can impact the blast load, and an analysis of the injuries suffered by the deceased.

Potential survivability of the injuries with current advanced medical treatment was highlighted, as a number of the victims had survived long enough to reach hospital. An approach was made to the Centre for Blast Injury Studies (CBIS) at Imperial College to assemble and co-ordinate a team to provide this, and the authors were formally instructed on 5th May 2017 by His Honour Sir Peter Thornton QC, Coroner for the Birmingham Inquests (1974).

10.3.2 Overview

The two bombs exploded in different Public Houses in Birmingham: the Mulberry Bush and the Tavern in the Town. Both establishments were busy, resulting in multiple casualties. It was reported that of the 21 killed, 18 'were killed outright' and 3 died later in hospital [17].

Although 6 people were convicted of the murder of the 21 casualties, following a Court of Appeal ruling, these convictions were quashed as being 'unsafe and unsatisfactory' on 14th March 1991 [18]. As noted above, the Inquest's proceedings were resumed on 1st June 2016.

Unlike most civilian trauma, victims of explosions are subjected to multiple mechanisms of injury. As described in Chap. 9 blast injuries fall into five main categories Primary, Secondary, Tertiary, Quaternary and Quinary injury.

As a result of the multiple casualties, multiple injuries and multiple mechanisms of injury, the Coroner sought expert evidence to explain the consequences of an explosion and the variables that can impact the blast load, and an analysis of the injuries suffered by the deceased in order to explore any issues with causation and time of death. This analysis required an approach that could understand and articulate the physics of the blast, the injury causing mechanisms and consequences to the victims. This was required to assist Her Majesty's Coroner, the families and any other interested parties of the events of 21st November 1974. As well as a narrative of the blast events and possible injury modes, the potential survivability of the victims needed to be understood, especially for those who were reported as having signs of life at the scene.

10.3.3 Approach

10.3.3.1 Multidisciplinary Team Approach

Three of the authors (JC, AH, PM) had been involved in providing evidence for the 7/7 bombing Inquest, and so a similar multidisciplinary team approach was used. The team comprised personnel from CBIS, the Defence Science and Technology Laboratory (Dstl) and the Royal Centre for Defence Medicine (RCDM). All are authors of this chapter (JC, AH, PM, AB). The academic disciplines and experience of the group are summarised in Table 10.1. The involvement of personnel from a range of organisations and backgrounds also brought independent learning and a range of opinions—this was seen as important in exploring as many eventualities as possible.

Table 10.1 Academic disciplines and experience of the CBIS, Dstl and RCDM team

Clinical management of multiply injured blast and ballistic casualties on military operations
Civilian pre-hospital care
Bioengineering and trauma biomechanics
Forensic investigation
Blast physics
Explosives chemistry
Specialist engineering knowledge of human vulnerability
Injury scoring and wound mapping

The team was provided with contemporaneous witness statements, post-mortem reports and photographs and scene photographs and sketches. Additional information was available in the open literature [17]; this was obtained, reviewed and considered in the context of the other provided information.

The panel members held a series of review meetings at which information provided was analysed and other experts were recruited to provide additional analysis. The witness statements and reports for each victim were reviewed in turn to build up an understanding of where they were at the time of the explosion, the injuries they suffered and any treatment they received. An analysis of the two incidents was conducted; this included consideration of survivability for each of the victims.

Injuries were collated onto external injury mapping software by Dstl [19] and Abbreviated Injury Scale Scores [20] assigned (see Fig. 10.4), which could then be calculated into Injury Severity Scores [21] (ISS) and New Injury Severity Scores [22] (NISS). The NISS is implied to provide a better calculation of overall injury severity than ISS [23], yet there are known caveats in the use of ISS and NISS with blast (see Chap. 11 and [24]).

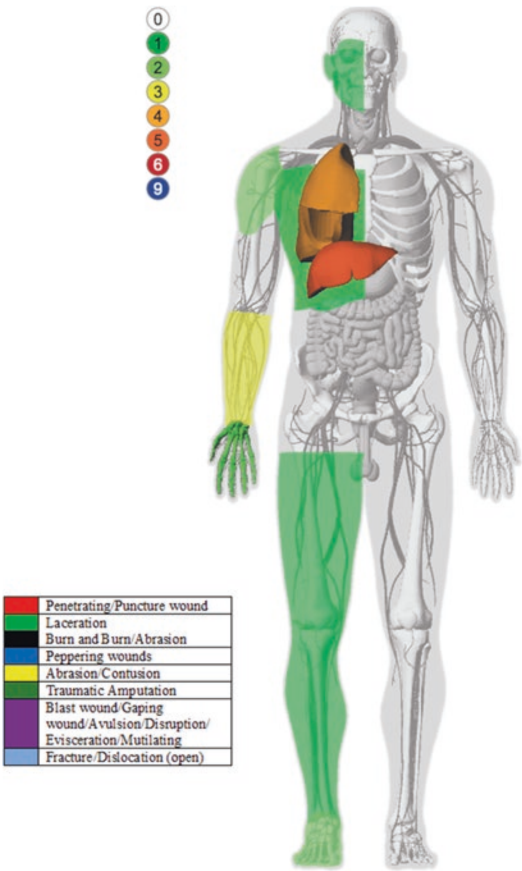

Fig. 10.4 Example of injury map and wound scoring for a fictitious casualty

In each case evidence of primary, secondary, tertiary and quaternary blast injuries was sought, together with the injury (or injuries) most likely to have caused death. Following detailed discussion, a consensus on potential survivability was given.

10.3.3.2 Challenges: Missing Clinical Information

Due to the length of time, a series of hospital closures and reorganisations in Birmingham since the bombings, contemporaneous clinical records were not available for the casualties admitted to hospital. There were, however, several peer reviewed clinical publications in relation to the bombings [17, 25]. In conjunction with witness statements and the post-mortem reports, a forensic analysis was possible. One particular publication allowed a sufficiently detailed analysis of the in-hospital deaths to be made [26].

All the collated data was reviewed and re-reviewed iteratively to ensure that any lessons identified during the first stages of the review were applied consistently to all cases.

10.3.4 Findings

Of the 21 fatalities, 17 were found to be dead at the scene, 2 were declared dead soon after admission to hospital, 1 died 6 days, and 1 died 18 days after the explosion. The age range was 17–56, and of the 21, 14 (66.7%) were male.

All 21 fatalities had multiple injuries, and died as a result of significant head injury, severe blast lung, or major haemorrhage. Fourteen of the 21 (66.7%) had more than 1 of these 3 modes of death.

Based on the analysis, of the 21 deaths, 13 people (71.4%) had evidence of all 4 blast injury mechanisms. Six people (19.0%) had evidence of three blast injury mechanisms. One person had evidence of two (4.8%) mechanisms, and one (4.8%) had evidence of only one mechanism of injury; both these casualties had a significant secondary blast (penetrating) injury to a vital structure and died rapidly after the explosion.

Despite initial reservations, it was possible, based on a knowledge of the injuries at post-mortem together with a relevant publication [26], to state with confidence that the four deaths that occurred after arrival at hospital were not preventable. We were therefore able to confirm at the inquest that, on balance, all 21 casualties sustained non-survivable injuries.

Although we were not tasked to provide an opinion on the location of individuals in relation to the seat of the explosion, we were able to confirm that the injuries were consistent with the location proposed in the witness statements. For the three fatalities who could not be placed accurately, it was possible to give an opinion that they were close to the seat of the explosion. Two individuals could be placed outside a building when the bomb detonated.

The absence of detailed records presented a challenge but, as noted above, this was mitigated by the availability of additional sources of expert information. The variability in the quality of available information also presented a challenge and there needed to be an understanding that the standards of 1974 are not the same as today, and much of the information was recorded without any belief that this would need to be critically reviewed in legal proceedings more than 40 years later. The methods and techniques used in this work had been previously used, and the use of injury scoring was found valuable in determining a common way of describing the injury types and severities. Despite all this, some of the scoring methods, notably ISS, were seen as under-predicting the total burden of injury. NISS, in this limited set, was better, but still not ideal and clinical expertise, experience and judgement were seen key ways in which injury mechanisms and the overall survivability could be provided.

10.3.5 Conclusions

As with our previous Inquest work [7/7], the value of a multidisciplinary team approach was confirmed. All conclusions were based on evidence or expert opinion, and all team members confined their comments to their area of expertise. The multidisciplinary and multiorganisational approach also provided internal group review that ensured that narratives were widely understandable.

The lack of contemporaneous hospital records was a potential problem but managed by cross-referencing an open access peer reviewed publication with the witness statements and post-mortem reports. This was declared to the Coroner and caveated.

There was considerable variety in the narrative description and detail of injuries in the post-mortem reports but the images provided allowed us to interpret blast mechanism where written detail was lacking.

In this analysis, ISS did not appear particularly helpful, as it appeared to under-score some injuries and injury complexes. This may be due to the fact that blast injury can result from different mechanisms (primary to quaternary), and some injuries, notably lung trauma, may occur from several mechanisms occurring at the same time, when the explosion occurs. The effect of this may be cumulative, even though the individual components may appear less severe. In addition, severe head injuries and blast lung evolve over time, due to the physiological response. Thus, significant injuries, such as diffuse brain injury may appear less severe at post-mortem if death is rapid.

The effect on different mechanisms of injury has been reported previously [27], but to our knowledge no scoring system has considered blast detail, particularly in relation to lung injury from the different blast mechanisms, which occurred in most of the casualties we reviewed in this work. Whilst NISS appeared in this limited cohort to be better than ISS, there was no replacement for training, experience and judgement of the medical experts in the group.

Based on this Inquest work, a number of recommendations were made to the Coroner. These are included at Annex 1.

Annex 1

The panel's recommendations to the Coroner, Sir Peter Thornton QC were as follows. The Coroner and the families of those killed have approved the inclusion of these recommendations in this chapter.

1. The case review approach taken by the expert panel has resulted in a detailed understanding of the 1974 Birmingham bombings. This would not have been possible if the approach had been to produce separate reports without meeting in person and conducting a joint review.

   We recommend that similar inquests follow the suitably formed expert panel approach.

   Recommendation owners: HM Coroners.
2. The lessons learned by the expert panel are important and should be captured for the

benefit of others and to inform future inquisitions.

We recommend that the lessons learned by the expert panel are written up as a report to be published in the open literature.

Recommendation owners: Expert Panel.

Birmingham Bombings—1974 inquest Coroner.

3. We acknowledge the complexity of the review we undertook in the context of the elapsed time since 1974 as well as lack of detailed record keeping in a form which facilitates such an understanding. This lack of detail made the panel's work difficult and hampered the total progress that could be made. Detailed record keeping of all key facts would facilitate timely analysis of any future incidents.

   We recommend that detailed record keeping of the following facts is made following any explosive event by the Coroners/Police and separate digitised copies of these be held, ensuring that these are future-proofed to any technological changes:

   1.1 location of all individuals involved (fatalities, survivors and uninjured);
   1.2 detailed and consistent injury recording of fatalities and survivors, post-mortem records, photographs;
   1.3 medical notes of all individuals involved;
   1.4 detailed plans of the location;
   1.5 details of explosives, device construction, detonation method, fragmentation;
   1.6 structural damage records; and
   1.7 building records and construction techniques.

4. In particular, we would highlight the necessity to obtain information not only on fatalities, but also on injured survivors, as the ability to learn from other incidents in order to effect changes that could increase survivorship requires knowledge of current survivors, not only fatalities.

   We recommend the keeping of detailed records for all those injured in a blast event.

   Recommendation owners: HM Coroners/ Police.

5. The injury 'scoring' systems currently may not adequately capture the complex injuries associated with blast. Such systems need to be consistent and improved. A key point in this is that the pathologist reports did not have a standard form and, as such, the articulation of injuries were affected by individual pathologist reports.

   We recommend that research is conducted to ensure a consistent method for injury recording that is appropriate for the review and analysis of blast injuries.

   Recommendation owners: Academic community. Research funders.

6. We were unable to conduct a detailed computer model analysis of the blast in the 1974 Birmingham bombings. Validated computer models could be used pre-emptively to design new facilitates to mitigate the effects of potential explosions. Whilst we are aware these are being developed, validation is limited. Detailed analyses of incidents such as the 1974 Birmingham bombings could be used retrospectively to improve computer model analysis.

   We recommend that the information collated as part of this inquest be released to conduct a detailed computer analysis of the blasts in the 1974 Birmingham bombings.

   Recommendation owners: Birmingham Bombings – 1974 inquest Coroner.

   Competent computer modellers.

   We recommend that such computer analysis be considered and commissioned for any future incidents.

   Recommendation owners: HM Coroners.

DSTL/PUB120062.

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Injury Scoring Systems

11

Phill Pearce

Abstract

Explosions are chaotic, and the resulting injuries can be varied in nature and severity. Comparison of injury burden between individuals and cohorts requires accurate quantification of injury. This chapter discusses the rationale for injury scoring and the importance of these systems for trauma care. The different types of injury score (including physiological, anatomical, and combined) are described along with commonly used examples of each. The scientific and clinical merits and limitations of these systems are highlighted, particularly regarding evaluation of blast injury. Specifically, current systems are shown to be inadequate in determining the true limits of survivability from severe multisystem injury. The chapter concludes with a discussion of the "perfect" blast injury score, which is highly dependent upon the intended use of the system.

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11.1 Introduction

Blast injury is now the most common mechanism of wounding in modern warfare. High velocity military rifles can cause great tissue injury, but explosive weapons carry with them a greater propensity for multi-mechanism, multisystem, multi-region, multi-casualty trauma than the more localised action of a bullet. The severe nature of blast injury is reflected in both the clinical manifestation of these injuries (such as dismounted complex blast injury) and greater burden of injury within the military injured casualty compared to a civilian cohort [1–3]. The quantification of this injury burden is important, and this chapter will explore the degree to which current systems may adequately describe injury and potential survivability due to battlefield and blast injury.

11.2 Why Score Injury?

Accurate quantification of injury is a crucial element of several aspects of trauma care:

Clinical decision making: Scoring systems (particularly simpler systems) may be used as point of care decision support tools. A quantitative assessment of injury may be used to triage patients to an appropriate pathway such as referral to a specialist centre or pre-notification to a centre of severe injury.

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Clinical governance: The "denominator" of any trauma system is the volume and severity of injury for which it must care. Ensuring that any trauma system cares appropriately for the patient relies upon comparing inter- and intra-system outcomes against trauma severity.

Ensuring high standards of care is of concern within military health systems. War is often reported to advance medical practice, particularly for the care of the injured but the retention of the skills and knowledge gained during conflict may be difficult during periods of relative military quiescence [4, 5]. Military health systems must undergo constant evaluation [6]. This burden of injury is measured with the use of combat casualty statistics to determine effectiveness of the system and to facilitate benchmarking against other military and civilian trauma systems [7]. Detailed analysis of injury patterns, severity, and outcomes enables ongoing assurance of standards despite changing threats and environments.

Research: Injury burden may be used not only as a comparator of trauma systems but also to compare the outcomes of interventions for clinical research. Similarly, injury severity is often used as inclusion or exclusion criteria for studies addressing those with specifically mild or severe injury.

Injury prevention: Of relevance to this book is the use of trauma scoring within biomechanical studies. Anatomically derived scores, particularly those that apply to individual organ or organ systems may be used as the dependent measure of trauma with various biomechanical parameters used to quantify the independent measure of injury.

11.3 Types of Injury Scoring Systems

The broad range of uses of injury scoring systems requires a similarly broad range of scoring tools and a thorough understanding of the appropriate use of each. These systems can be classified, based upon the parameters which they measure, into physiological, anatomical, and combined.

11.3.1 Physiological Scores

Physiological scores are based upon clinical findings and the response of the patient's physiology to the injury. Physiological tools may assess injury to a particular body system, the most notable being the Glasgow Coma Scale (GCS). First described in 1974, the GCS predicts outcome from head injury based upon eye opening, movement, and speech of a patient at a particular moment [8]. Physiological scores may also combine measures from different organ systems and include the Revised Trauma Score (RTS) [9] which is based upon neurological function (as described by the GCS), respiratory rate, and systolic blood pressure.

Physiological scores are used for both outcome prediction and triage within the acute setting and often form part of the clinical description of trauma patients. These scores are dependent upon the patient being (a) alive and (b) undergoing some form of real time measurement of the relevant physiology. These scores therefore have no place in assessing those injuries which resulted in immediate death. Although they may be used for assessing and stratifying patients for research interventions, they are not typically used as outcomes for biomechanical studies which frequently use cadaveric or component level testing.

11.3.2 Anatomical

Anatomical systems are based upon anatomical injuries without physiological measures. They are an objective measurement of the degree of tissue injury. The most commonly used scoring tool is the Abbreviated Injury Scale (AIS). The tool was initially developed in 1969 [10] and has subsequently been maintained and updated by the Association for the Advancement of Automotive Medicine [11]. AIS is an anatomically based, consensus derived, global severity scoring system that classifies an individual injury by body region according to its relative severity on a 6-point scale (1 = minor and 6 = maximal). Given the consensus-based nature of the AIS system, this number is based upon an agreed degree of severity rather than derived from outcome data.

Specific injuries are graded according to an extensive AIS dictionary which is regularly updated and lists specific injuries by affected organ. Importantly, the scores between different organs are not necessarily comparable (for instance, AIS 5 head and AIS 5 liver injuries are both described as "critical" but are not necessarily associated with the same outcome).

For many injuries, there are individual grading systems which may be based upon a true anatomical degree of injury (such as % volume of organ injured or depth of laceration [12, 13]). In most cases, the AIS scores of such injuries are mapped to their individual grading systems.

AIS codes are used only for describing and quantifying individual injuries. The burden of trauma caused by multiple injuries may be described by combining individual scores.

The most commonly used score is the Injury Severity Score (ISS), first described by Baker and O'Neil [14]. The ISS is made up from the three most severe injuries from three separate anatomical regions. The severity score (calculated using the AIS scale) of each of these injuries is squared and these are summed.

$$\mathrm{ISS} = \max \mathrm{AIS}_a^{\,2} + \max \mathrm{AIS}_b^{\,2} + \max \mathrm{AIS}_c^{\,2}$$

A major criticism of ISS has been in the use only of injuries from different regions [15]. Additional severe injuries from the same regions do not form part of the score even if the severity of these additional scores is higher than the most severe score from another region. This criticism led to the development of the New Injury Severity Score (NISS) [16]. In contrast to ISS, the three most severe injuries are used to make up the total score, irrespective of the region in which they occur. A further criticism of the ISS is on the weighting of injuries in that the same effect on survivability is to be expected from injuries of the same AIS severity from different regions and this patently cannot be true. The Anatomic Profile (AP) tool was developed by Champion with the aim of overcoming some of these limitations [17]. The AP tool divides up injuries into four body components (Head/Spinal cord, Thorax, Abdomen and Pelvis, and all other injuries). In contrast to ISS and NISS, the AP uses all injuries within each region to define the component score which is the square root of the sum of squares of all the AIS codes within that component. Each component score is therefore most affected by the highest AIS score within that region and decreasingly affected by subsequent lower scores. The score effectively measures the distance that each injury component moves the casualty away from non-injury. A modification has been made to the AP which uses the maximum overall AIS injury as the fourth component in place of "any other injuries". This modification was made on the basis that the single worst injury may be a more accurate predictor of death [18].

Comparison of the different AIS utilising scoring systems has shown that modified AP (mAP) is the best predictor of survivability (with a higher Area under the Receiver Operating Curve (AUROC) compared to ISS and NISS). Despite this, mAP has been criticised for its relative complexity and difficulty in application given that it is made up of four different component scores and is rarely used as a control score for either research or governance.

The relatively recent International Classification of Diseases (ICD)-based Injury Severity Score is based upon ICD diagnostic codes [19]. These codes are routinely used for resourcing and billing purposes. Within a sufficiently large dataset, survival rates of each ICD code can be compiled to create an outcome prediction score which utilises the population-based survival rates for all injuries as coded. This score has been shown to predict survival risk ratio as well as ISS within selected populations but has not been widely validated [20, 21].

11.3.3 Combined Scores

The Trauma and Injury Severity Score (TRISS) is a combined score which includes the ISS in addition to physiological data (the Revised Trauma Score—RTS) and age. The score uses a logistic model to predict probability of survival (between 0 and 1) [22].

$$P_S = \frac{1}{1+e^{-b}}$$

where $b = b_0 + b_1(\text{RTS}) + b_2(\text{ISS}) + b_3(\text{Age})$

$b_{0..3}$ are coefficients based upon the population. Numerous revisions and modifications of the TRISS methodology have improved and refined performance. More recent models take account of mechanism, vary the physiological component (including only the GCS in place of the RTS), and include population specific coefficients [23, 24]. It is important to emphasise that the anatomical component of these scores (the measure of trauma) retains the same limitations of ISS and therefore of AIS.

11.4 Limitations of Contemporary Current Scores

As noted above, physiological scores may be of use for triaging patients, and for evaluating systems of care for those patients who are alive at presentation to medical treatment facilities. Therefore, these tools are unsuitable for assessing the injury burden in those casualties who die in the battlefield or pre-hospital settings.

The anatomical systems are also limited in their ability to assess survivability of such severe injuries. Both ISS and NISS are calculated retrospectively based upon an extensive "AIS dictionary". Neither score is therefore of any utility within the acute setting. Other features limit the ability of the systems for describing survivability. Two injuries with the same score within the same region may have different outcomes. Given that the severity scores of each code are not based upon a quantifiable metric, they are non-linear in distribution. The difference in severity between a score of 1 and a score of 2 is not necessarily analogous to the difference between a score of 2 and a score of 3. This non-linearity is compounded when scores are squared for ISS or NISS.

The use of the scores as a predictor of survivability is further hampered by statistical features of the systems. Both ISS and NISS are formed by the combinations of between 1 and 3 squares, dependent upon the number of injuries. This method provides a range of scores between 1 and 75 for all injuries of 5 or less severity. Some scores are then formed more frequently as they can be formed from a greater number of score combinations. In contrast, some scores are impossible to achieve. The resultant distribution contains characteristic peaks and troughs and is not transformable to a Gaussian distribution by any conventional method [25]. Despite this, ISS and NISS are frequently used as a continuous variable despite neither being a continuous distribution nor is there a quantifiable relationship between score and tissue injury. Further evidence of a lack of established relationship between score and injury is in the discrepancy in outcomes for casualties with the same score formed from different injury combinations [26, 27].

The use of these scores is particularly problematic for analysis of survivability. Peaks of death or morbidity are associated with particular values due to the higher likelihood of these scores occurring. AIS 6 scores are given to those injuries that are deemed unsurvivable (by consensus). As a result, any casualty who sustains a grade 6 injury is automatically scored 75 by ISS/NISS. An ISS/NISS or 75 (achieved from either 3 × AIS-5 or 1 or more AIS-6) is the maximum achievable score.

As an example, death from mounted blast may be a consequence of one or more severe injuries. Many of these injuries would thus be graded as AIS 6 resulting in a total ISS or NISS of 75. This distribution of NISS injury scores across survivors and non-survivors is bimodal with peaks both of either low of maximum scores (Fig. 11.1) [28]. Consequently, there is very little granularity in describing the potential survivability of those who died.

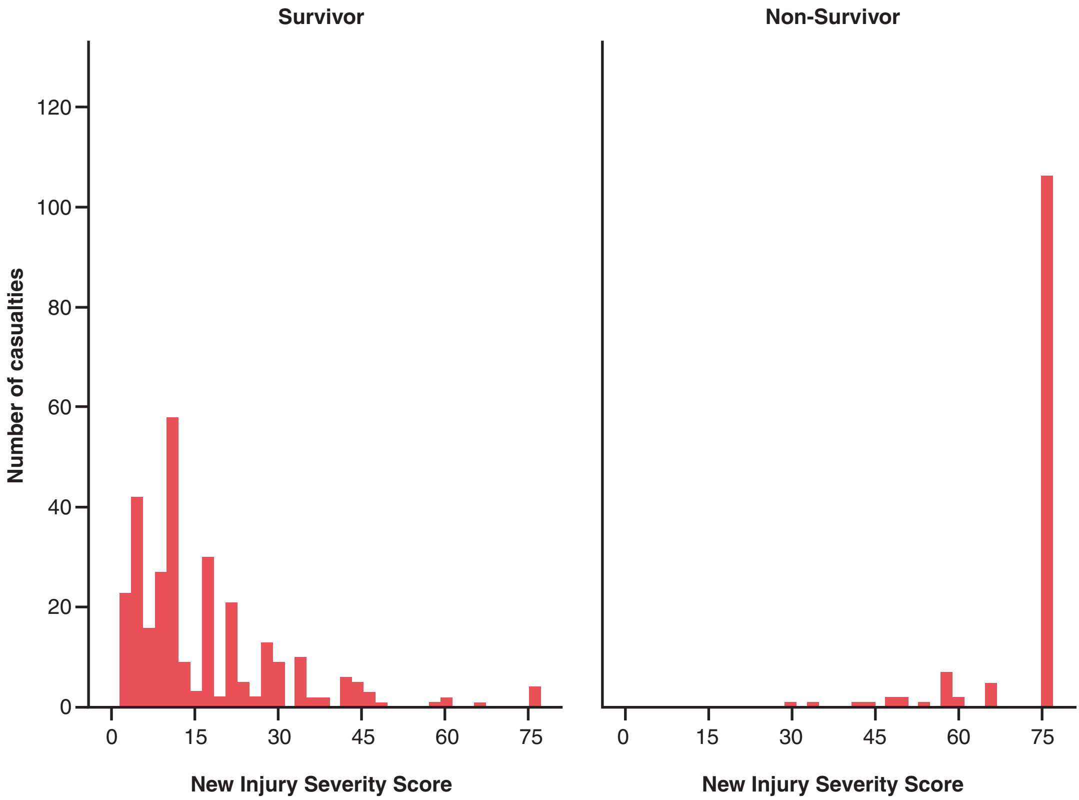

Fig. 11.1 NISS of survivors and non-survivors from mounted blast showing bimodal distribution of scores

Paradoxically, the unsuitability of extant systems to define the "spectrum of unsurvivability" has been emphasised by improved survival in response to injury.

Advances in trauma healthcare have led to an increasing number of casualties surviving despite an ISS 75 injury burden. An examination of over 2800 ISS 75 US trauma admissions found that 48.6% were discharged alive [29]. In a military setting, Penn-Barwell et al. (2015) have described sequential improvement in survival for a given battlefield injury burden. Although a testament to exceptional care and dedication to quality improvement, this data shows the degree to which a conventional scoring system inadequately predicts death in the face of advanced or improving clinical care. More than 40 "unexpected survivors" (NISS > 75) were treated during UK combat operations in Afghanistan [30].

Although this data shows a laudable increase in the average probability for survival for a given injury burden as a result of clinical and systems improvement, it highlights the fallibility of injury scoring for assessing true survivability. If survivability scores are influenced by clinical ability, then no score will be truly accurate until the absolute limit of medical advancement has been reached. The value of scoring systems may be in stratifying the needs of different casualty cohorts and identifying the future medical advances required.

11.4.1 Military Specific Scores

There has been some recognition of the limitations of AIS and ISS to adequately describe military injuries. As noted above, ISS was first developed in order to describe injuries resulting from motor collisions [31]. Although there have been several iterations of the AIS dictionary with subsequent major expansion in the number of the codes, injuries particular to combat are not well described [32]. Similarly, the relative severity of apparently similar injuries may differ between combat and non-combat environments due to the higher energy of insult. Indeed, even injuries with similar degrees of anatomical disruption may result in worse outcomes within a deployed environment with extended timelines from injury to treatment and lack of resources. This recognition led to the development of the military Injury Severity Score (mISS) [33]. The mISS uses a separate AIS (2005-Military) dictionary in which the codes were derived by identifying each combat injury that is more severe than the corresponding civilian injury and increasing the AIS code by one increment of severity. As with the standard AIS system, changes were made following consensus opinion by an expert panel of military surgeons and physicians. A few injury descriptions were increased by two AIS severity increments to reflect the increased risk of death or morbidity in a military setting. AIS 2005-Military 6 injuries are those which were determined by the panel to be untreatable within the combat setting, but which might be survivable if evacuation to an out of theatre facility was possible. The new dictionary included 356 changes with 326 injury codes upgraded by 1 increment and 30 codes upgraded by 30. The most common upgrade was for an AIS 3–4. Head injuries were those most likely to be upgraded.

The use of these military specific codes to describe combat injury does increase the predictive ability of the score for mortality. A retrospective cohort study compared use of ISS and mISS across a large cohort of combat casualties from the US Joint Theatre Trauma Registry [33]. This study demonstrated that 18% of patients had discordant ISS and mISS codes with a median difference in scores of 9. Greater discordance between scores was shown in the more severely injured group. mISS was shown to have a significantly greater predictive ability for mortality than ISS using both receiver operating curve and multivariate logistic regression analysis. Examination of these discrepancies between mISS and ISS showed them to be especially common in body region 1 (head/neck, 23.4%) and body region 5 (extremities/shoulder and pelvic girdle, 17.5%). These regions are of relevance for battle injury since they are poorly protected by body armour and commonly the injured region from blast.

Although representing an improvement on conventional ISS, mISS is still prone to the same statistical limitations as ISS and still reliant upon discrete, consensus derived injury codes. Some effort has been made to couple these methods with objective description of wounds. Surface Wound Mapping™ and the Surface Wound Analysis Tool™ have been developed within the US Department of Defense in order to correlate the external wounds of (predominantly penetrating) injured service personnel with internal injuries, protective equipment, and mechanism of injury (Fig. 11.2) [32]. Even for this tool, the internal injury data associated with the wounds are in the form of AIS codes.

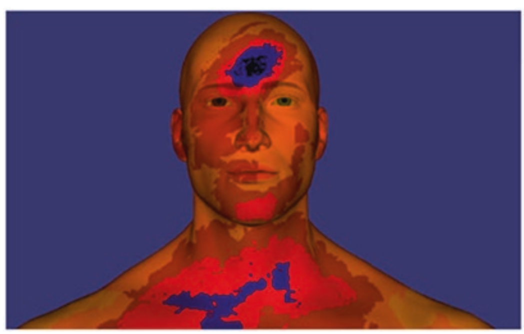

Fig. 11.2 Surface Wound Mapping TM tool. Patterns of injury (along with incident data, equipment details) may be mapped visually and coupled with AIS codes. (Image used with permission from SimQuest, Annapolis, MD, USA)

High prevalence of explosions as a mechanism of injury was part of the rationale for devising mISS. The issue of primary blast lung injury has been addressed by the inclusion of novel codes which acknowledge that blast lung in comparison to conventional pulmonary contusion is more likely to be bilateral and associated with air embolism. Despite this specific set of codes, the overall effect of blast upon the cumulative injury burden is not assessed by the score.

In addition, certain injury patterns associated with explosions may be more severe than the individual scores suggest, particularly when death is rapid, and therefore secondary physiological changes do not develop. Examples include superficial facial burns, which in combination with blast lung is indicative of a severe multi-mechanism injury. Similar patterns can occur in the head when a basal skull fracture is associated with a diffuse axonal injury, or even a relatively minor penetrating injury.

The TRISS method, as discussed above, utilises population level coefficients to predict mortality based on both anatomic and physiological parameters. The conventionally used TRISS model does differentiate between penetrating and blunt mechanisms of injury. Recent work has sought to devise new coefficients for combat injury with either explosive or gunshot wounds [34]. Use of explosive specific coefficients increased the accuracy of the tool for predicting mortality to around 98%. These results must be interpreted with caution however, as accuracy was determined by whether the tool had predicted either less than or greater than 50% chance of survival in those who died or survived, respectively. As with other physiology-based tools, even the revised TRISS has little utility in those patients killed in action. The anatomic input to the revised TRISS is still ISS.

11.4.2 The Ideal Scoring System?

Despite attempts to revise and improve upon scoring systems for explosive injury and combat casualty care, difficulties remain in trying to reconcile scoring systems which remain clinically relevant and scientifically useful. A recent review has recommended that an ideal scoring system should incorporate the severity of injuries with the mechanism of blast injury [35]. The new system may include refinement of codes, additional codes being added, or even severity scores being added to the Abbreviated Injury Scale for blast-specific injuries. A weighting factor for body regions associated with a higher risk for death and blast-specific trauma coefficients was also proposed as well as the removal of the maximum value, which would enable the classification of more severe constellations of injury.

It is important to recognise that any single system is unlikely to be suitable for all purposes. Evaluation and comparison of deployed and domestic trauma systems may not require a novel quantitative assessment of anatomical disruption. The use of a combined system (such as TRISS) which includes a physiological component and coefficients based upon combat injury mechanisms may be suitable assuming that the populations, injury mechanisms, timelines, and resources across the systems of comparison are similar. As noted above, there would be issues in assessing fatalities.

The use of military AIS codes for such comparisons is logical. Some caution should perhaps be employed in the interpretation of these scores regarding the treatment environment. Military AIS codes may have been upgraded based on the austere environment but outcomes from severe trauma (such as bilateral above knee amputation) may be better for a given injury severity given the concentration of trauma expertise and experience within this environment.

It is in the accurate and objective description of anatomical disruption that our current scoring systems are most flawed. AIS scores are discrete, non-linear, and opinion based. While the use of such scores to describe injury as mild, moderate, or severe may be useful for governance and comparison purposes, these scores are frequently used as numerical outcomes. Injury biomechanics research may use these or similar scores to denote the clinical significance of such injuries but derivation of injury criteria and relationships between insult and injury should be based upon robust, reproducible, and objective measure of tissue disruption. Measures based on volumetric imaging have been employed in both biomechanical and clinical research but are not commonly used [36–38]. There is currently little evidence that correlates these purely quantitative measures of tissue disruption with clinical outcome. It is logical, however, that such a measure may be of more benefit in non-survivors where physiological data is irrelevant and unavailable, and currently used methods are saturated at a pre-set point.

11.5 Conclusions

There is an ongoing need for robust description and quantification of injury. AIS, ISS, and NISS are intuitive and ubiquitous and certainly of important use for description of injury. These tools are perhaps not suited for quantification of injury since they are underpinned by fundamental issues of subjectivity, non-linearity, and statistical misinterpretation. Combined scores such as TRISS are of use in the evaluation of trauma systems but rely upon accurate physiological data and are unsuitable for assessment of potential survivability of severe trauma.

Biomechanical and survivability assessments should be based upon objectively quantified parameters of anatomical disruption which may provide an estimate of the "survivability" gap of severe injury.

The scoring of the blast injured and killed should include specific measures of the insult itself and estimate the global effect on survivability of the blast.

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Part III

Clinical Problems

12 Section Overview

Arul Ramasamy

Until recently, the predominant focus of medical battlefield research has been on increasing survivability from ballistic trauma. Recent advances in torso protection, enhanced pre-hospital care and rapid aero evacuation to medical facilities capable of providing optimised resuscitation and damage control surgery, have led to an unprecedented improvement in survivability. This increase in survivability of those with life-changing injuries has pushed medical practice to unprecedented levels. Ensuring that survivability is further improved and injury severity is reduced requires a concerted effort on mitigation.

The key to mitigation lies in understanding the multi-modal mechanisms with which blast interacts with the body. Prior to the recent conflicts in the Middle East, detailed information on combat injury and injury mechanisms was often limited to sporadic case series, other than the Weapons Data and Munitions Effectiveness Team (WDMET) from the Vietnam war [1]. The development of combat injury registries in the UK (Joint Theatre Trauma Registry, JTTR), US and other countries has enabled current researchers to gain an invaluable insight into injury mechanisms. These have allowed clinicians and engineers to link injury profiles to specific events. Furthermore, these registries have provided the basis to allow us to understand the long-term effects of blast injury and to aid the development of better protection, as well as to focus research towards reducing or preventing death and disability.

In this section, we provide a comprehensive overview on the effects of blast on the body, from the cellular effects on inflammation (Chap. 13) to the mechanism of traumatic amputation (Chap. 15). Building on from the first edition, we also investigate the challenges that blast injury presents to physicians looking after casualties, both military and civilian. The long-term effects of hearing loss (Chap. 20) and brain injury (Chap. 22) have had a devastating effect on the quality of life of combat casualties. In addition, the sequelae of combat extremity injury should not be forgotten. Blast injury results in complex soft tissue and bone defects that are on the extreme of the trauma spectrum. The management of heterotopic ossification (Chaps. 23 and 24), failure of bone healing and managing deep-seated infection, challenges the most experienced trauma surgeon, and continues to blight the lives of blast injured casualties. In these chapters, we outline current management of these injuries, as well as offering insights into future research foci and treatments.

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13

The Immune and Inflammatory Response to Major Traumatic Injury

Jon Hazeldine and Mark Foster

Abstract

Two opposing clinical syndromes characterise the immune and inflammatory response to major trauma. Defined by immune activation and raised circulating levels of pro-inflammatory cytokines, a systemic inflammatory response syndrome (SIRS) is accompanied by a counteracting compensatory anti-inflammatory response syndrome (CARS), which is characterised by peripheral immune suppression and elevated circulating concentrations of anti-inflammatory cytokines. With it becoming increasingly recognised that it is the magnitude and/or duration of the SIRS:CARS response that influences a patient's clinical course, the field of trauma immunology is starting to focus upon understanding the mechanisms that underpin both the initiation and maintenance of the post-trauma immune response in an effort to improve patient outcome. Thus, the purpose of this chapter is to provide a summary of these emerging areas of research that are combining the results of both basic science and clinical-based studies to (1) further our understanding at the molecular level of how traumatic injury promotes systemic inflammation, (2) determine whether novel therapeutic strategies aimed at restoring immune homeostasis can reduce the incidence of secondary complications amongst hospitalised trauma patients and (3) establish whether features of the SIRS:CARS responses can serve as prognostic indicators of patient outcome.

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Table 13.1 Effect of traumatic injury on the human innate and adaptive immune system

Cell	Frequency/number	Surface phenotype	Function
Neutrophil	Neutrophilia [17, 18, 33] Raised immature neutrophil counts [17, 18] Increased frequency of suppressive neutrophil subsets [17, 34]	Decreased CD62L, CD88, CXCR1 and CXCR2 expression [17, 21–23, 35] Increased CD11b expression [17, 24, 36]	Reduced phagocytosis [18, 25] Impaired NET production [19, 20] Reduced ROS production [17, 18, 20] Impaired chemotaxis [20, 26] Reduced responsiveness to fMLP stimulation [17, 21, 27]
Monocyte	Monocytosis [17, 28, 29] Increased frequency and number of $CD14^{+}$ 16^{-} and $CD14^{+}$ 16^{+} monocytes [17] Increased frequency of $CD14^{+}$ $HLA\text{-}DR^{-/low}$ monocytes [17]	Increased TLR2, TLR4 and CD86 expression [17] Reduced HLA-DR expression [17, 29, 30]	Impaired cytokine production [17, 29, 37] Increased proliferation [29] Maintained phagocytic activity [29] Increased ROS production [29]
NK and NKT cells	Increased number of NK cells, NKT cells, $CD56^{DIM}$ and $CD56^{BRIGHT}$ NK cells (<2H post-injury) [17, 33] Reduced number of NK cells, $CD56^{DIM}$ and $CD56^{BRIGHT}$ NK cells (>4H post-injury) [17, 31]	Increased CD39 expression on NK cells [28]	Impaired NK cell cytotoxicity [38, 39]
T cells	Immediate (<1H post-injury) lymphocytosis [17] Lymphopenia (>4H post-injury) [17] Expansion in T_H1, T_H17 and Treg $CD4^{+}$ T cells [29, 32]		Increased IL-17 production upon CD3/28 stimulation [29] Elevated frequencies of IFN-γ producing $CD4^{+}$ T cells [16]

fMLP N-Formylmethionyl-leucyl-phenylalanine, *IFN-γ* Interferon gamma, *IL-1β* Interleukin-1 beta, *IL-17* Interleukin-17, *LPS* Lipopolysaccharide, *NET* Neutrophil extracellular trap, *NK* cell Natural killer cell; *ROS* Reactive oxygen species, *TLR*, Toll-like receptor, *TNF-α* Tumour necrosis factor-alpha, *Treg* T regulatory cell

13.1 Introduction

During military operations in Afghanistan between October 2001 and December 2014, the U.S. saw the largest proportion of coalition forces wounded in action (WIA), with 20,000 personnel injured and 1833 killed in action [1]. In the same conflict, of the 2188 UK military and civilian personnel WIA, 404 battle related deaths were recorded [2]. The year-on-year improvement in survival rates for combat casualties of any given injury severity [3] produced a cohort of survivors that were expected to die from their injuries [4]. Attributable in part to medical advancements in military trauma care, particularly the management of haemorrhage [5, 6], these statistics suggest that if the likelihood of patients surviving severe catastrophic injury are increasing [4], then it is the onset of secondary complications that will pose the greatest risk to the long-term outcome of survivors. With complication rates ranging between 25 and 75%, hospitalised trauma patients represent a high-risk cohort for the development of multiple organ dysfunction syndrome (MODS), multiple organ failure (MOF) and sepsis [7–12]. The onset of these secondary complications is associated with an increased risk of in-hospital mortality, increased lengths of intensive care unit (ICU) and hospital stay and a greater requirement for post-discharge care and rehabilitation [10, 13]. As these conditions have elements of immune activation and suppression underpinning their pathophysiology, their development indicates that trauma must have a profound impact upon the immune system. Indeed, data from military [14–16] and civilian based studies [17–32] has revealed both immediate and sustained changes in the composition, phenotype and/or function of the innate and adaptive arms of immune system post-injury (Table 13.1). With the results of these studies summarised across several review articles [40–44], the purpose of this chapter is to introduce emerging areas of research within the field of post-trauma immunity that are using the results of both basic science and clinical-based studies to (1) further our understanding of the mechanisms that underpin the immune response to injury, (2) determine whether novel therapeutic strategies aimed at restoring immune homeostasis can reduce the incidence of secondary complications and (3) develop prognostic indicators of patient outcome.

13.2 Redefining the SIRS:CARS Paradigm

Severe traumatic injury is associated with two inflammatory syndromes: a systemic inflammatory response syndrome (SIRS) and a compensatory anti-inflammatory response syndrome (CARS). Characterised by raised circulating levels of pro-inflammatory cytokines (e.g. Interleukin (IL)-6, IL-1β, tumour necrosis factor-alpha (TNF-α)) and activation of innate immune cells, the SIRS response is triggered in the immediate aftermath of injury, with a robust and heightened response a significant risk factor for the early onset of organ dysfunction/failure [45, 46]. Counteracting the SIRS response is the CARS response. Defined by elevated circulating concentrations of anti-inflammatory cytokines (e.g. Transforming Growth Factor-β, IL-10), cytokine antagonists (e.g. IL1-Receptor antagonist, soluble TNF Receptor-1) and immunoparesis, the CARS response is initiated in order to restore homeostasis [47]. However, if exaggerated or prolonged, the CARS response can leave the hospitalised trauma patient at risk of infection and the late onset of organ dysfunction/MOF [45–47].

When conceptualised in the early to mid 1990s, the SIRS:CARS model of trauma-induced inflammation proposed a biphasic response, theorising that the SIRS response dominated the immediate and early post-injury phase before the counterbalancing CARS response emerged in the ensuing days and weeks to restore homeostasis (Fig. 13.1a). Nearly 30 years later, whilst the notion of opposing SIRS and CARS responses remains an integral feature of the immune and inflammatory response to injury, their sequential nature has been challenged, with researchers now favouring a model in which the SIRS and CARS responses occur simultaneously (Fig. 13.1b).

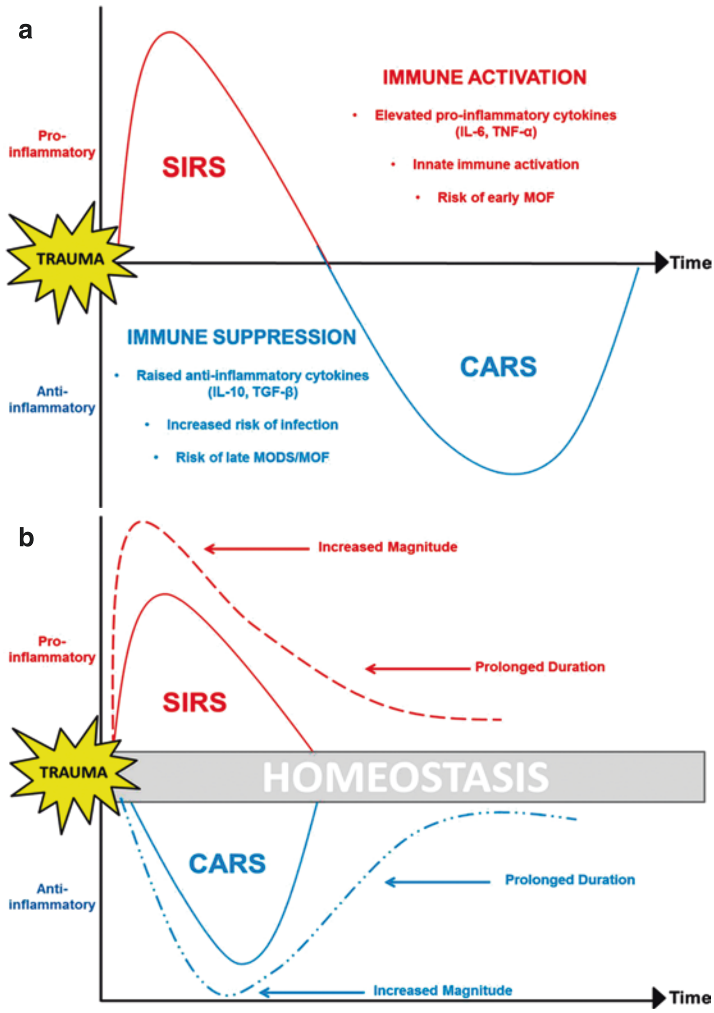

Fig. 13.1 Redefining the SIRS:CARS paradigm. (**a**) The original SIRS:CARS model of post-trauma inflammation postulated a sequential response in which the pro-inflammatory SIRS response dominated the immediate and early post-injury phase before an anti-inflammatory CARS response emerged to restore homeostasis. (**b**) The revised SIRS:CARS model favours concomitant SIRS and CARS responses that are initiated within minutes of injury. The magnitude and duration of each response influence patient outcome with those individuals experiencing "poor outcome" (dashed lines) proposed to exhibit SIRS and/or CARS responses that are of increased magnitude and/or duration when compared to patients with "good outcome" (solid lines). Ultimately, patients with "poor outcome" fail to return to homeostasis. CARS, Compensatory anti-inflammatory response syndrome; MODS, Multiple organ dysfunction syndrome; MOF, Multiple organ failure; SIRS, Systemic inflammatory response syndrome; TGF-β, Transforming growth factor beta; TNF-α, Tumour necrosis factor-alpha

One of the first studies, whose data contested the biphasic nature of the SIRS:CARS model, performed transcriptomic analyses on leukocytes isolated from 167 blunt trauma patients [48]. Within 12 h of injury, a significant up-regulation of genes related to innate immunity, inflammation and pathogen recognition was detected, accompanied by significantly reduced expression of genes involved in antigen presentation and the adaptive immune response [48]. Termed the "genomic storm", these early changes in gene expression were largely independent of injury severity, shock and blood transfusion [48]. Since this discovery, additional genomic based studies [33] as well as a comprehensive screen of the functional and phenotypic alterations that occur in the immune system post-trauma [17, 37] have lent support to the idea of concurrent SIRS and CARS responses. Interestingly, features of immune activation and suppression are detectable in the immediate aftermath of injury, with our recent analyses of pre-hospital blood samples acquired within 1 h of injury, the so-called Golden Hour revealing evidence of impaired leukocyte cytokine production and elevated numbers of immune suppressive cells occurring alongside increased neutrophil anti-microbial function and monocyte activation [17]. Thus, the growing consensus within the literature is that a concomitant SIRS:CARS model is a better reflection of the inflammatory response to injury, with both responses initiated within minutes of trauma [17, 37].

The SIRS:CARS model dictates that a CARS driven return to immune homeostasis brings an end to the trauma-induced inflammatory response. However, clinical data demonstrates that not all injured patients achieve "*status quo*", with a proportion experiencing persistent immune dysfunction [49, 50]. Termed the persistent inflammation immunosuppression and catabolism syndrome (PICS), this phenomenon is defined by ongoing inflammation (e.g. elevated C-reactive protein (CRP) levels and neutrophilia) and immune suppression (e.g. lymphopenia) [49] and is associated with a range of adverse outcomes, which include recurrent nosocomial infections, prolonged hospitalisation, increased surgical procedures and death [49–51] (Fig. 13.2). Mechanistically, the persistence in immune dysfunction has been credited to a post-injury expansion in myeloid derived suppressor cells, a heterogeneous population of immature myeloid cells that are a rich source of inflammatory cytokines (e.g. TNF-α) and potent inhibitors of T cell responses [52–54]. Replacing late onset MOF as the predominant clinical phenotype amongst ICU patients [49], PICS represents a significant clinical challenge, for which unearthing therapeutic strategies should be the focus of future research.

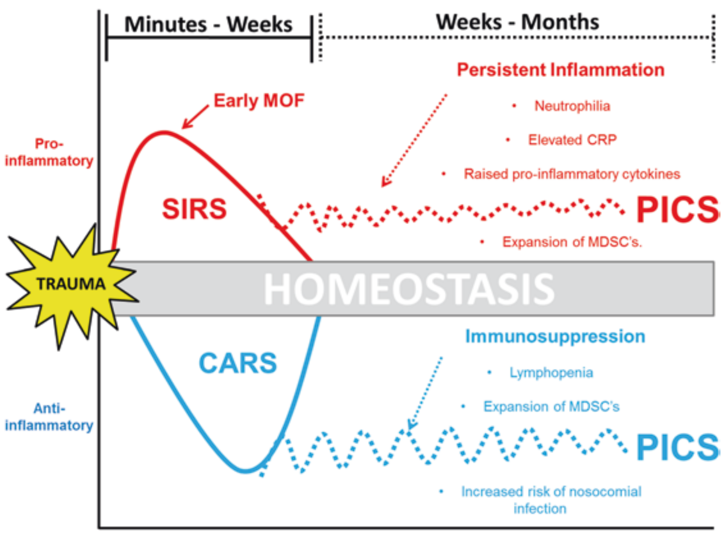

Fig. 13.2 Persistent inflammation immune suppression and catabolism syndrome (PICS). Survivors of major trauma who fail to establish immune homeostasis develop PICS, a clinical syndrome characterised by long lengths of hospital stay, recurrent nosocomial infections, poor quality of life and ultimately death. Features of persistent inflammation include neutrophilia and raised circulating levels of acute phase proteins (e.g. C-reactive protein) and pro-inflammatory cytokines (e.g. TNF-α). Immunosuppression is characterised by lymphopenia, reduced adaptive immune cell function and a susceptibility to recurrent nosocomial infections. Underlying the onset of PICS is the expansion of myeloid derived suppressor cells (MDSCs). CARS, Compensatory anti-inflammatory response syndrome; CRP, C-reactive protein; MOF, Multiple organ failure; PICS, Persistent inflammation immune suppression and catabolism syndrome; SIRS, Systemic inflammatory response syndrome. Figure adapted from Mira et al. [51]

13.3 Mitochondrial-Derived Damage Associated Molecular Patterns: Therapeutic Targets for the Treatment of Post-Injury Immune Dysfunction?

A heterogeneous collection of cytosolic, nuclear and mitochondrial-derived proteins, lipids and DNA, damage associated molecular patterns (DAMPs) are released from necrotic tissue and activated immune cells within minutes of injury [17, 37, 55]. In a seminal paper, Zhang and colleagues showed that via activation of toll-like receptor 9 (TLR-9) and formyl peptide receptor-1 (FPR-1), mitochondrial-derived DNA (mtDNA) and N-formylated peptides, respectively, promoted immune activation in vitro and in vivo, culminating in neutrophil-mediated organ injury in a rodent model of trauma [56]. Thus, this work alongside subsequent studies that have confirmed the potent innate immune stimulatory properties of mitochondrial-derived DAMPs (mtDAMPs) [57–59] provides a mechanistic explanation for how sterile traumatic injury promotes widespread systemic inflammation [56].

In cohorts of critically ill patients, prospective studies have linked elevated levels of circulating mtDAMPs to a range of adverse clinical outcomes such as MODS, SIRS and mortality [60–67]. Therefore, scavenging circulating mtDAMPs has been proposed as a potential therapeutic approach to reduce poor outcome amongst hospitalised trauma patients. This is being tested in human and rodent-based studies where nucleic acid binding polymers (NABP) such as polyethylenimine, hexadimethrine bromide and polyamidoamine dendrimers are used to test whether the capture and depletion of a variety of DAMPs, including mtDNA, can suppress trauma-induced inflammation [66, 68]. To date, results are encouraging, with data from a rat model of trauma haemorrhage showing systemic NABP treatment can reduce lung inflammation and improve lung histological appearance. In healthy animals, NABP treatment prior to mtDNA challenge significantly attenuated pulmonary inflammation by reducing cell infiltration, apoptotic cell death and oxidative injury [66].

Another therapeutic strategy that may ameliorate mtDNA-induced immune cell activation is inhibition of its receptor TLR-9. Although not tested in the context of traumatic injury, this approach, which can be achieved via systemic administration of either TLR-9 antagonists or the anti-malarial drug chloroquine, has proven successful in murine models of sepsis, protecting against infection-induced MOF and mortality [69, 70]. Importantly, these effects were observed when the drugs were administered up to 12 h after the septic insult, suggesting such approaches may be of therapeutic benefit in the post-injury phase. Recently, we and others have shown that the activity of deoxyribonuclease (DNase), a plasma residing enzyme responsible for the degradation of circulating DNA is significantly reduced for up to 28 days following blunt or thermal injury [65, 71]. Augmenting DNase activity may therefore be one method by which to combat the elevated levels of plasma mtDNA witnessed in the days and weeks following injury. A possible strategy by which to achieve this is DNase administration, a therapeutic approach that has been well tolerated in other patient cohorts when delivered intravenously [72], as an inhaled preparation [73] or via endotracheal tubes in ventilated patients [74]. Interestingly, in cases of severe trauma, specifically blast injury, significantly elevated DNase activity was found in military patients who received fresh frozen plasma prior to hospital admission when compared to those who did not, suggesting administration of blood products in the immediate aftermath of trauma may be an additional therapeutic approach by which to increase DNase activity [71].

In a longitudinal study of 166 trauma patients, analysis of plasma samples acquired at the time of hospital admission revealed significantly elevated levels of circulating mtDNA in patients who later developed nosocomial infections when compared to those who did not [37]. Although renowned for their immune stimulatory properties, this observation led to the suggestion that DAMP release post-injury may also promote peripheral immune suppression [37], a concept that has since gained support in the literature with the creation of the term “suppressing/inhibiting inducible DAMPs” (SAMPS) used to denote those DAMPs that elicit immune suppressive properties [75]. To date, two mechanisms by which mtDAMP release could promote immune tolerance have been proposed. The first, which has been demonstrated in rodents, centres around the idea that due to their chemotactic properties, mtDAMPs released at sites of tissue damage direct neutrophils towards remote injury sites and away from “authentic” bacterial foci in the lungs, thereby leaving the trauma patient at risk of respiratory-tract infections [20]. The second mechanism is based on heterologous desensitisation and altered cellular metabolism [19, 20, 56, 76]. In this scenario, mtDAMPs, pro-inflammatory agonists and pathogens activate shared intracellular signalling mechanisms within immune cells. Thus, neutrophils pre-exposed to mtDAMPs released from damaged tissue fail to respond as efficiently to “authentic” infectious stimuli in times of pathogenic challenge, resulting in impaired functional responses [19, 20, 56, 76]. Fascinatingly, in both these proposed mechanisms, it is N-formylated peptides that are the tolerising agent [19, 20, 56, 76]. Thus, targeting this specific mtDAMP could be a means by which to delay or prevent post-injury immune dysfunction. Interestingly, blocking FPR-1 signalling using the FPR-1 antagonist cyclosporin H was recently shown to protect against heterologous desensitisation and increase bacterial clearance from the lungs in a Staphylococcus aureus-induced model of pneumonia [76]. Based on these findings, a therapeutic approach of FPR-1 blockade has been suggested as a potential means of preserving immune function post-injury and protecting against bacterial infection [76].

13.4 Does Major Traumatic Injury Drive Accelerated Ageing of the Immune System?

Population based studies that have examined the long-term impact of injury on human health have shown that survivors of major trauma are at an increased risk of chronic disease and death. Compared to age and gender-matched uninjured controls, analyses have shown reduced survival rates [77–79] and increased hospital admissions for diabetes mellitus [80], cardiovascular disease [81] and musculoskeletal conditions [82] amongst patients that have suffered thermal [79–82] or traumatic brain injury [77, 78]. With these conditions often viewed as “age-associated pathologies”, an emerging hypothesis is that major trauma may accelerate the ageing process [83].

In 2013, Lopez-Otin et al. proposed the “hallmarks of ageing”, a constellation of the cellular and molecular features that are thought to contribute to the ageing process and determine the ageing phenotype [84]. Chronic low-grade inflammation (termed inflammageing) and telomere shortening (see below) are two hallmarks with relevance to the immune system [84]. Inflammageing is a sub-clinical systemic inflammatory state characterised by raised circulating concentrations of pro-inflammatory cytokines and acute phase proteins [85]. In a cohort of 977 severely burned paediatric patients, Jeschke et al. detected significantly elevated serum levels of IL-6, TNF-α, interferon gamma (IFN-γ) and CRP for up to 3 years post-injury when compared to non-burned controls [86]. At the level of immune cell cytokine production, psychological trauma has been shown to be associated with an altered inflammatory profile, with combat veterans with post-traumatic stress disorder (PTSD) presenting with significantly elevated frequencies of IFN-γ producing $CD4^+$ T cells and raised plasma IFN-γ levels up to 20 years after military service [15, 16]. Thus, traumatic injury appears to drive premature ageing of the immune system [86]. Supporting this idea are the results of several studies that have examined the impact of emotional trauma on telomere length in circulating leukocytes [87]. Hexameric repeat sequences found at the end of chromosomes, telomeres promote genomic integrity and stability. As cells divide, telomeres shorten due to incomplete replication. As such, telomere length negatively correlates with donor age [88] and cellular function, with telomere attrition associated with the age-related decline that occurs in adaptive immune cell proliferation following antigenic stimulation

[89]. On this note, significantly shorter telomere lengths have been measured in circulating leukocytes isolated from war veterans with PTSD, whose T cells also exhibit impaired proliferative responses when compared to age-matched healthy controls [90, 91]. With inflammageing and shortened telomeres associated with frailty [92] as well as the onset of cardiovascular disease, diabetes and dementia [93–97], the above-mentioned data suggests that injury, by accelerating the ageing trajectory, may predispose survivors of trauma to an early onset of age-associated diseases [83]. However, it must be noted that this field of research is in its infancy and hampered by a number of confounders, making it difficult to directly attribute trauma-induced immune changes to the accelerated ageing phenotype. Moreover, no study to our knowledge has examined the long-term effect of injury on susceptibility to infection. As such, more studies, particularly in the area of physical injury, whose effects on immune ageing are currently unknown, need to be performed to substantiate such a suggestion.

13.5 The Post-Injury Immune Response as an Indicator of Patient Outcome

As aberrant systemic inflammatory responses are thought to underlie the post-injury onset of MODS, MOF and sepsis, one might hypothesise that a patient's susceptibility to developing secondary complications is influenced by the nature of their SIRS:CARS response. If so, then one would expect the composition, magnitude and/or duration of post-injury inflammatory responses to differ between patients who experience poor outcomes from those who recover quickly. Results of prospective cohort studies that have investigated the immune response to injury at the genomic [33, 48, 98–100], protein [101–106], functional [107–112] and metabolic [113] levels suggest this may be the case.

Whilst transcriptomic-based studies may disagree as to whether it is a change in the direction of gene expression (e.g. upregulated Vs. downregulated) or a genomic signature characterised by more robust early changes in gene expression and a prolonged return to baseline that can best differentiate between patients with "good" and "poor" outcome [33, 48, 99], it is clear that transcripts associated with inflammation are a discriminating factor. Indeed, pathway analyses have revealed marked differences in the expression of genes involved in innate immunity, cell survival, immune signalling and leukotriene biosynthesis between patients with "good" and "poor" outcome [33, 48, 99]. Interestingly, with respect to MODS, the degree of differential gene expression between patients who do and do not develop this secondary complication is greatest <2 h post-injury, with very few differences evident in gene expression by 24 h [33]. This kinetic profile suggests that the gene signature of MODS is set within the hyperacute injury phase [33]. Supporting the rapidity by which immune responses can differ between patients on different clinical trajectories, we recently analysed pre-hospital blood samples acquired from trauma patients within 60 min of injury and found a potential association between elevated circulating numbers of natural killer T (NKT) cells and the subsequent development of MODS [17]. Monitoring changes in immune phenotype may also be of potential clinical benefit [28, 29, 37], with one study reporting an increased likelihood of nosocomial infection for patients in whom expression of the antigen presenting molecule HLA-DR on the surface of monocytes decreased between hospital admission and day 3 post-injury [37].

Based on the results of the above studies, the search for prognostic biomarkers of patient outcome generated during the post-injury immune response is currently underway. To date, in the settings of blunt trauma [98], thermal injury [104, 114–116], TBI [117] and critical care [67], studies have shown measures of gene transcription [98, 114], cytokine production [104, 115–117], immune cell phenotype [116] and mtDNA [67] can predict with varying degrees of specificity and sensitivity a range of adverse outcomes, which include nosocomial infection [114, 116], mortality [67, 115, 117] and MODS [104]. Of note, predictive models based on changes in gene expression outperformed those built on anatomical, clinical and physiological data [98, 114]. However, this appears to be the exception to the rule, with combinations of immune markers and clinical scoring systems appearing to provide the best discriminatory power [67, 115, 116]. One example of this is in the setting of burn injury, where a model built on inflammatory cytokine levels and the revised Baux score improved the predictive accuracy of the clinical model alone by 29% [115].

13.6 Future Directions

Although military personnel present with blast and non-blast related injury patterns that are distinct from those witnessed in civilian trauma, it is currently unclear as to whether blast injury promotes an immune and inflammatory response that differs (in respect to its duration, magnitude and/or resolution) from that which occurs following non-blast injury.

Using data collected from a prospective observational cohort study of blast and non-blast injured trauma patients (SIR Study Cohort [118]), preliminary unpublished analyses have revealed a separation of blast and non-blast injuries. Early cytokine release is higher in blast patients after correcting for age, injury severity and length of stay on intensive care. Taken with caution, neutrophil anti-microbial function is also suppressed, with the ability of neutrophils to phagocytose opsonised E.coli more suppressed in the first 6 weeks following blast injury when compared to non-blast. What underlies these early differences is currently unknown, but given that unseen tissue damage is a feature of blast injury, it is reasonable to hypothesise that blast-injured patients are exposed to a higher circulating mtDAMP load for a prolonged period of time. Given the tolerising actions that mtDAMPs have on neutrophil function, an elevated mtDAMP burden could contribute to the greater degree of immune suppression observed following blast injury. This raises the intriguing possibility that the results of studies that have investigated the post-injury immune response of civilian patients are not directly applicable to military personnel. Thus, rather than surmising that all injury types trigger comparable immune responses, military-focussed studies will be the only way to truly understand the impact of blast injury on the human immune response.

We have highlighted areas currently under investigation to ameliorate the immune response but many clinical trials in major trauma have been hampered by oversimplifying the complexity of the injury. To allow for better separation and stratification of cohorts, investigators need to understand:

1. The nature of the injury or *diagnosis*. Improved ways to quantify the amount and type of tissue damage along with any vital structures/organ systems which have a profound effect on homeostasis.
2. The pre-injury state or *substrate*; the victim's co-morbidities and their genetic robustness to deal with the injury [119].
3. The *treatment* effect, how the timing and response to any treatment affect the trajectory of the patient.

Trials in a major trauma setting have used mortality as their primary outcome. This approach misses the large trauma-related burden to any healthcare system not to mention society. Long-term outcomes have been subjective and qualitative in nature [120]. Attention should turn to defining objective outcomes, separate from mortality, to analyse efficacious treatments. Focusing on the immune response to severe injury may allow more defined outcomes to be generated and the development of treatments that will reduce the long-term burden of injury for the patient and society.

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14 Foot and Ankle Blast Injuries

Louise McMenemy and Arul Ramasamy

Abstract

Blast injuries to the foot and ankle have been a feature of military wounding patterns since World War II. These injuries are recognised to be severely disabling, and as many as one-third of casualties with this type of injury will require amputation by 3 years post injury. For military personnel, amputation appears to lead to improved patient reported outcomes. This may be due to established rehabilitation pathways and improvements in prosthetic design. Orthotic design is also proving advantageous for military limb salvage patients following complex foot and ankle injuries. In this chapter, we discuss the long term effects of these injuries as well the development of novel offloading devices that could potentially reduce the need for amputation and improve limb salvage function.

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14.1 Introduction

Blast injuries to the foot and ankle have been a feature of military wounding patterns since World War II [1]. The anti-vehicle mine and Improvised Explosive Device (IED) have become the main threats to military vehicles and their occupants in modern conflicts, including most recently in Iraq and Afghanistan [2–4].

When an explosive detonates beneath a vehicle, the blast wave from the explosion causes the release of a cone of super-heated gas and soil which impacts the under surface of the travelling vehicle (see Chap. 9). This leads to rapid deflection of the vehicle floor, transmitting a very short duration (less than 10 ms), high amplitude load into anything that is in contact with it. Most frequently it is the lower leg and particularly the foot and ankle complex that is injured [5].

14.2 The Issue

Whether caused by underfoot explosions while on foot patrol or ‘deck-slap’ injuries from under-vehicle blasts, these injuries are recognised to be severely disabling (Fig. 14.1) [6]. As many as one-third of casualties with this type of severe foot and ankle injury will require amputation within 3 years and 75% of casualties will have ongoing clinical symptoms (such as infection and persistent pain) [6]. Statistical modelling of the injuries demonstrated that open fractures, vascular injuries, and hind-foot injuries were associated with an increased risk of amputation [6]. Further subgroup analysis of casualties with hind-foot injuries confirmed that they were associated with significantly higher amputation rates and only 5% were able to return to pre-injury levels of activity at 3 years post injury [7]. This data showed that attempts to protect the hind foot from injury may reduce the risk of amputation within this cohort of casualties. The optimal treatment for these injuries, whether amputation or limb salvage, remains unclear.

Fig. 14.1 Deck-slap foot hindfoot injury

14.3 Amputation or Limb Salvage

There exist several challenges in treating patients following High Energy Lower Extremity Trauma (HELET). The Lower Extremity Assessment Project (LEAP) sought to quantify whether Limb Salvage (LS) or amputation resulted in improved outcomes following HELET. LEAP was a multicentre prospective trial and concluded that regardless of treatment, LS or amputation, outcomes are worse than population norms at 2 years [8, 9]. Furthermore, subgroup analysis, looking specifically at injuries of just the foot and ankle, found superior outcomes in amputees, when compared to patients treated with LS and a free flap and/or ankle arthrodesis [10].

LEAP finished recruiting 25 years ago in June 1997 [8], since when multiple advances had been made in the management of complex trauma patients. Improvement in lower limb prosthetic design has also enhanced outcomes for lower limb amputees [11]. LEAP excluded military patients [12] since then the conflicts in Afghanistan and Iraq have resulted in a large amount of HELET. The Military Extremity Trauma Amputation/Limb Salvage (METALS) study attempted to elucidate which treatment option was better for military patients following HELET [13]. METALS concluded that specifically for military patients, following optimum rehabilitation, amputees have superior Short Musculoskeletal Functional Assessment scores. The authors conceded that this may be due to improved rehabilitation pathways and developments in prosthetic technology not available to LS patients [13].

A plethora of research has been published concerning outcomes following LS, amputation or comparing both options, since the 7 year outcome results were published by LEAP [9]. Analysis of this literature reveals that for civilians, there is little evidence that outcomes, both physical and quality of life based, are different regardless of treatment choice. Military literature, however, seems to suggest superior outcomes for amputees, especially looking specifically at injuries of the foot and ankle. Bennett et al. [14] found superior outcomes for amputees following hind foot injuries compared to LS patients using a generic health outcomes questionnaire. The Physical Component Score was noted to be significantly lower ($p = 0.0351$) in amputees, however, almost a third of the cohort were lost to follow up resulting in a potential for selection bias. All of this combines to produce the effect that military patients with complex foot and ankle injuries, following LS surgery, are now choosing elective amputations due to persistent pain and weakness [15].

14.4 Improving Outcomes Following Limb Salvage

One of the main hypotheses for this inferior outcome is due to the biomechanical loading of the heel during gait causing load transfer and pressure on the tissues of the hind foot resulting in pain. One solution may be an advanced novel offloading lower limb orthosis known as a passive dynamic ankle foot orthosis (PDAFO) (Fig. 14.2). In the UK, the Bespoke Offloading Brace Bespoke Offloading Brace (BOB) is prescribed and in the USA the Intrepid Dynamic Exoskeletal Orthosis (IEDO). The IDEO has been shown to reduce pain and allows some patients with severe foot and ankle injuries to carry out activities that bring their quality of life close to pre-injury levels [16–18]. Of 50 patients included in the study previously considering amputation, 41 pursued LS with function augmented by the IDEO. The IDEO has also conferred improved outcomes sustained for 2 years [16].

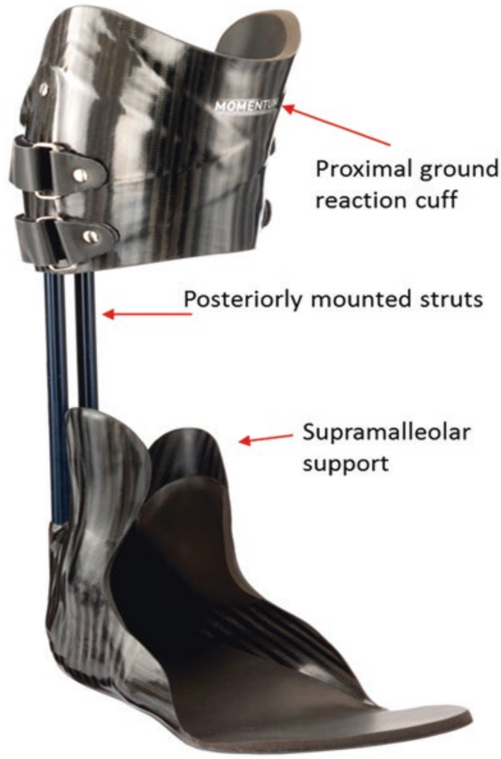

Fig. 14.2 The Momentum® or Bespoke Offloading Brace, Passive Dynamic Ankle Foot Orthosis

Despite this, the American experience indicates as many as 19% of patients who receive the orthosis still go on to amputation [16]. It is important to note that this study includes patients who went onto amputation after only a 3-month trial of the orthosis; potentially an insufficient length of time to assess the benefits offered. Due to the innovative nature of the device, long-term outcome data is not available, therefore the required trial period before progression to potential amputation has not been established.

A minimum of 12-month outcomes in the UK with the BOB highlights an improvement in ability to run independently, with a significant improvement above the minimal clinically important difference in 6-Minute Walk Test Distance. These improvements were also better than previous outcomes for limb salvage patients without augmentation with the BOB [19]. The UK experience of the BOB in 63 patients demonstrates progression to amputation in 12 patients (19%). 32 patients report a positive outcome with the BOB and continue to use the orthotic for variable periods of the day whilst 31 patients did not find the BOB useful and have abandoned the orthotic (including the 12 patients who have progressed to amputation). It was not possible to predict outcome with the BOB from patient demographics including age, BMI or smoking status, mechanism of injury or the patient's residual functional deficit.

Those patients who were candidates for amputation due to the severity of their injury (defined by the foot and ankle severity score) did not benefit from the BOB ($p < 0.001$). Also, patients diagnosed with chronic pain/neurogenic pain were less likely to benefit from use of the BOB ($p < 0.02$). Patients with a centrally derived neurological deficit did benefit from prescription of the BOB. Evidence from the UK and USA highlights the difficulties of studying a heterogeneous population and despite efforts to overcome this using functional deficit, it has not been possible to date to predict outcome with the BOB from pre-prescription characteristics and scores.

14.5 Future Research

To prevent inappropriate over prescription and protect patients from lengthy unsuccessful limb salvage with a PDAFO, it is desirable to create a clinical decision tool. To date, it has not been possible to do so with current evidence concerning both the BOB and IDEO. Therefore, future research will focus on the collection of evidence, including gait and functional performance data, with a view to the creation of a clinical decision tool to guide prescription of a PDAFO for limb salvage patients following HELET.

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15 Traumatic Amputation

Iain A. Rankin and James A. G. Singleton

Abstract

Blast-mediated traumatic amputations (TAs) have been a prevalent feature of recent conflicts in Iraq and Afghanistan, due to the increased use of the Improvised Explosive Device (see Chap. 8). Analysis of post-mortem computed tomography imaging and clinical data from these conflicts has led to the development of a new proposed mechanism of injury for TA. This mechanism accounts for transosseous TA, due to a predominant axial blast load, and through-joint amputation due to lower limb flail—resulting in peri-articular soft tissue failure and amputation. Further cadaveric animal studies have raised the possibility of the involvement of a 'sand-blast' mechanism of injury to also account for transosseous TA. A correlation between TA and fatal pelvic blast injury has been noted and suggests their mechanisms of injury may be linked.

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Abbreviations

IED	Improvised explosive device
JTTR	Joint theatre trauma registry
PBLI	Primary blast lung injury
PM-CT	Post-mortem computed tomography
TA	Traumatic amputation

15.1 The Issue

Blast-mediated extremity traumatic amputations (TAs) (as shown in Fig. 15.1) became a defining injury pattern in casualties from the combat operations in Iraq and Afghanistan in the late twentieth and early twenty-first century, with the majority due to Improvised Explosive Devices (IEDs) [1]. The driver for improving understanding of the injury biomechanics involved was both powerful and simple:

TAs are potentially lethal; IEDs caused over 50% of all UK combat fatalities in Iraq and Afghanistan—most of those killed sustained at least one TA.

A better understanding of the precise injury mechanism of blast-mediated TA would contribute to informing prevention, mitigation and clinical strategies, and research—enabling further improvements in combat casualty care. This could help facilitate improved outcomes following blast-mediated TAs by decreasing injury severity or could even decrease the incidence of such injuries.

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Fig. 15.1 An IED casualty with a left leg traumatic amputation at the level of the knee joint. The tibia is still attached but completely stripped of soft tissue

15.2 Limitations of Current Injury Mechanism Theory

Military personnel have risked TA from explosive blast since the advent of battlefield explosive munitions in the fourteenth Century. The injurious components of an explosive blast event, such as an IED strike, are categorised as primary to quaternary (see Chap. 9). These injurious components have been implicated to various degrees in the causation of TA, through several theories of its mechanism of injury.

Initial theories for the mechanism of injury of TA centred on simple limb avulsion by the blast wind, i.e. pure tertiary blast injury [2]. This was revised by Hull, a former British Army orthopaedic surgeon, through research published in 1996 [3]. Hull's mechanism of injury theory of blast-mediated TA was a summation of literature review, analysis of medical reports and photographs of injured extremities from blast casualties (the majority stemming from the troubles in Northern Ireland from the 1970s to early 1990s), live blast tests and computational finite element (FE) modelling [3–5] (see Chap. 28).

Two main points underpinned Hull's theory: (1) a paucity of through-joint TAs identified following blast injury and (2) a perceived association between fatal primary blast lung injury (PBLI) and TA in blast casualties.

A paucity of through-joint TAs was noted by Hull in blast casualty data and highlighted inconsistencies between two datasets previously thought to share a common injury mechanism: blast casualties and ejecting fast jet aircrew. Casualty data from Hull et al. showed a diaphyseal skeletal amputation level in the majority of explosive blast casualties sustaining TAs and a paucity of through-joint TAs (<2%). This contrasted with the anatomical level of extremity injury (fractures/dislocations, not amputations) in ejecting fast jet pilots [6], subject to windblast (i.e. pure tertiary blast injury) at speeds up to 1100 km/h, thought to approximate to blast wind velocities close to the seat of an explosion. These aircrew injuries, due to limb flail, tended to be through or near to joints. As such, whilst the intra/peri-articular injuries experienced by jet pilots could be accounted for by a mechanism similar to tertiary blast injury, the diaphyseal skeletal amputations experienced by blast casualties could not. A further causative mechanism of injury was hypothesised to explain this.

The potential causative role of primary blast (the shockwave) was inferred by an association, again from Northern Ireland casualty data, between fatal primary blast lung injury (PBLI) (an injury associated with primary blast—see Chap. 18) and TA. Live blast tests conducted by Hull generated long bone fractures in goat hind-limbs. These samples were shielded from secondary and tertiary blast effects, leaving primary blast injury—shockwaves—as the only likely mechanism of fracture causation. Computer modelling also supported the diaphysis as the area of greatest stress concentration following exposure to primary blast.

Thus, a new injury mechanism for TA was put forward. Hull proposed a sequence of initial primary blast injury—the blast wave coupling into the long bones of an extremity and causing diaphyseal fracture through resultant axial stress concentration—followed by tertiary blast injury—limb flail from the blast wind, completing the TA at the level of the fracture [3].

This hypothesis was formed based upon the above associations and Hull's experimental work. However, his methodology did not include any radiological imaging analysis and clinical data was extremely limited. More recently, analysis of clinical data from Iraq and Afghanistan has called into question Hulls' hypothesised mechanism of injury. Firstly, analysis of 121 combat casualties by Singleton et al. showed no statistical association between PBLI and TA, in either mounted (i.e. in-vehicle) or dismounted (i.e. on foot) blast fatalities [7]. Secondly, the decade of recent conflict in Iraq and Afghanistan generated a large cohort of survivors with TA injuries—some with multiple TAs—who, by definition, had not sustained fatal PBLI [8]. Thirdly, military surgeons with experience of managing IED casualties in Afghanistan reported seeing a number of through-joint TAs. This inferred that through-joint TAs may not be as rare as previously thought.

The introduction of routine post-mortem CT (PM-CT) imaging of UK military combat fatalities from 2007 allowed a much better analysis of injury patterns. Detailed cross-sectional imaging was performed to a standard protocol and, in the vast majority of cases, within a few hours of the fatal injuries and with no confounding surgical intervention. This provided a new tool to examine blast injury pathoanatomy at levels of detail never previously available [9]. These highlighted both a need to review understanding of blast-mediated TA mechanism of injury and an opportunity to further our knowledge in this field through new data.

15.3 The Study

The basis of the new research was clinical analysis at a level of detail never previously possible. A national prospectively gathered trauma registry (UK JTTR) and a post-mortem CT (PM-CT) database were used to identify casualties (survivors and fatalities) sustaining a blast-mediated major extremity TA (through or proximal to the wrist or ankle joint) between August 2008 and August 2010. Level of TA and associated significant limb, thoracic and other injuries were recorded.

Survivors routinely underwent emergency surgery. The majority were not comprehensively CT scanned pre-operatively, reflecting clinical priorities of lifesaving treatment. The detailed pathoanatomical analysis was possible due to PM-CT imaging, by definition only available for fatality cases. However, as shown in Fig. 15.2, survivor and fatality TA anatomical distributions were not statistically different. Crucially then, fatality TA pathoanatomy data could be considered to be representative for survivor TAs also. Tens of thousands of images were assessed to characterise the bony and soft tissue anatomy of blast-mediated TA injuries.

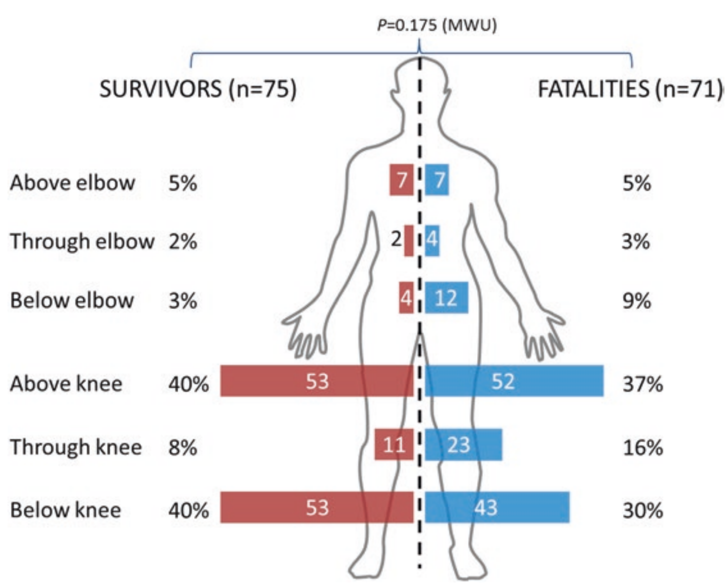

Fig. 15.2 Level of blast-mediated TAs: Survivors vs. Fatalities (Numbers in/by bars represent number of TAs, MWU—Mann–Whitney U test)

146 cases (75 survivors and 71 fatalities) sustaining 271 TAs (130 in survivors and 141 in fatalities) were identified. The lower limb was most commonly affected (117/130 in survivors, 123/141 in fatalities). The overall through-joint TA rate was 47/271 (17.3%). 34/47 through-joint injuries (72.3%) were through knee.

More detailed anatomical analysis facilitated by PM-CT imaging of the fatality group (see Fig. 15.2) revealed that only 9/34 through-joint TAs had a contiguous fracture (i.e. intra-articular involving the joint through which TA occurred) in the proximal remaining long bone/limb girdle. 18/34 had no fracture, and 7/34 had a non-contiguous (i.e. remote from the level of TA) fracture. Further analysis revealed that in many cases the amputated limb was grossly intact and had not been fragmented as may have been believed previously (see Fig. 15.3).

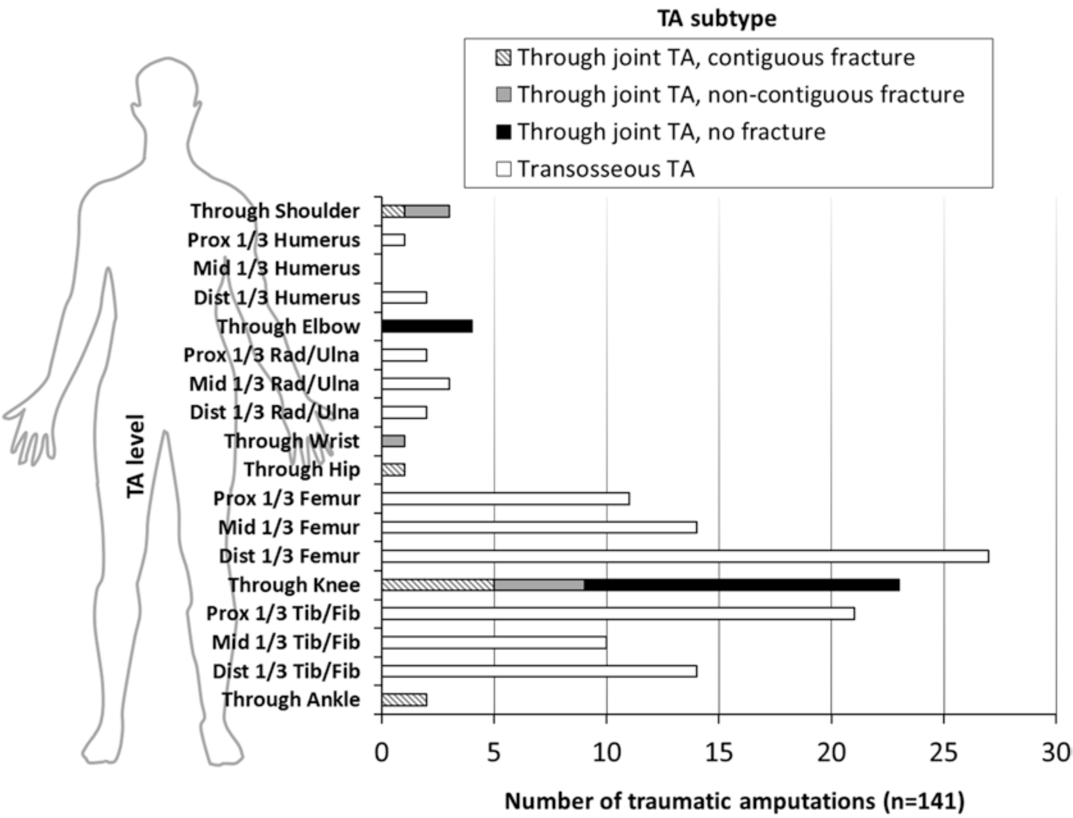

Fig. 15.3 TA incidence by level and TA subtype in fatalities

15.4 The Outcomes

This new data informed a re-evaluation of the understanding of the injury mechanism of blast-mediated TA.

Analysis of the soft tissue injury profiles of through-joint and transosseous TAs showed no significant difference. This suggests that there may not be two radically different processes occurring to generate these injuries and thus an injury mechanism theory to explain both types of TA could be valid.

The high rate of through-joint TAs (almost 1 in 4 in the fatality group), and the fact that of these, almost 3 in 4 had either no associated fracture or a fracture remote from the level of amputation, strongly inferred a flailing mechanism of injury. This contrasted with accepted theory of shockwave-mediated midshaft fractures of long bones followed by flail. A through-joint injury was inconsistent with expected fracture patterns following either primary blast (oblique diaphyseal) or close contact primary and secondary blast injury (brisance—shattering—type fractures) [10].

Furthermore, the relative significance of the shockwave/primary blast is called into question by the lack of any relationship between TA and PBLI demonstrated in this analysis of modern blast casualties. In contrast to previously held beliefs that proximity to an explosion sufficient to cause TA was lethal due to PBLI, this study has shown no such correlation, and the large cohort of survivors with TAs are a clear demonstration that such a link does not hold true with many current blast casualties. Environment of the casualty was minimised as a potential confounder by subgroup analysis of mounted and dismounted cases.

Combining all the new data analyses, the following blast-mediated TA injury mechanism is proposed:

> Dependent on the position of the extremity and the displacement produced by the explosive blast wind, in some cases the relative stability of the joint and a predominant axial load generate diaphyseal stress concentrations leading to fracture, flail at this point and transosseous TA (see Fig. 15.4). However, in other scenarios with oblique loads caused by less axial, more coronal/sagittal type extremity displacement from the blast wind, the maximal stress concentration is peri- or intra-articular, leading to primary peri-articular soft tissue failure, flail through the joint and a subsequent through-joint TA (see Fig. 15.5).

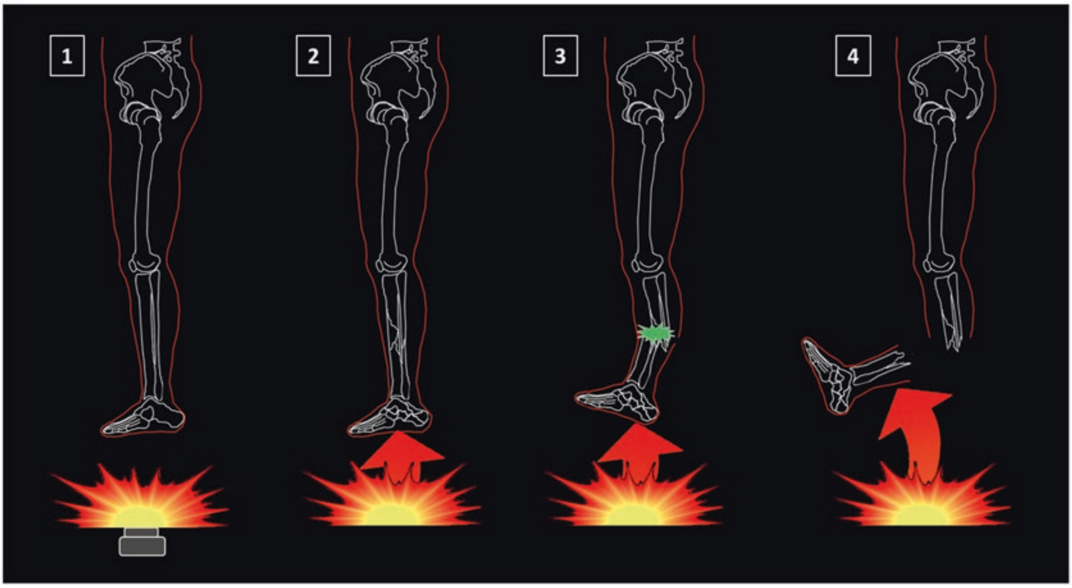

Fig. 15.4 Proposed transosseous TA mechanism from tertiary blast. (1) detonation underneath lower limb. (2) predominantly axial force transmitted, corresponding hindfoot and long bone fractures. (3) flail occurs through long bone fracture site. (4) completion of transosseous TA

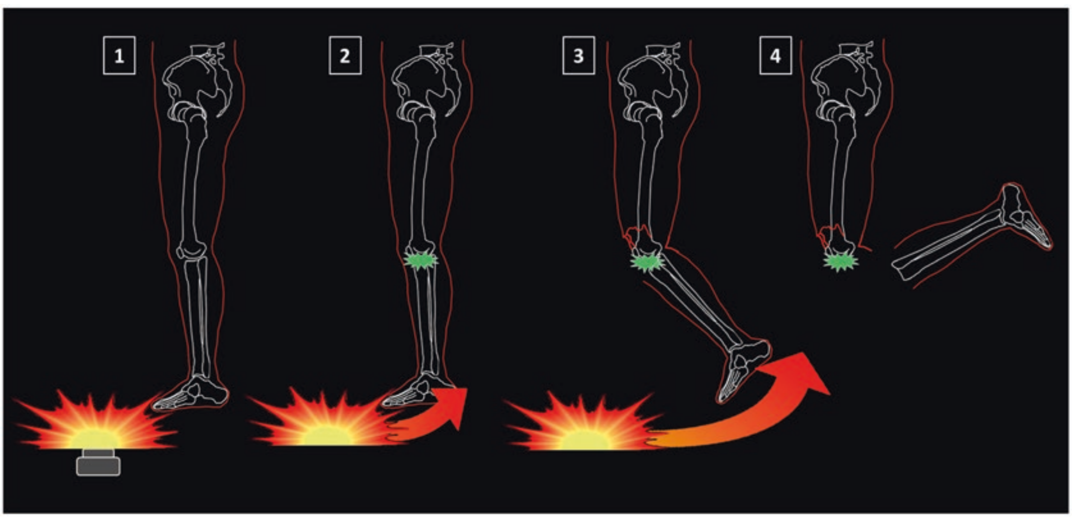

Fig. 15.5 Proposed through-joint TA mechanism from tertiary blast. (1) IED detonation, biomechanically significant offset from casualty. (2) Non-axial force from blast wind, peri-articular soft tissues loaded. (3) Peri-articular soft tissue tensile failure, gross displacement of leg. (4) Through-joint TA

Although modern blast injury data did not support a link between significant primary blast injury and TA, there is insufficient evidence to discount Hull's theory (shockwave-induced diaphyseal fracture followed by flail and TA through the fracture) as a valid injury mechanism. It is possible that there may be multiple blast-mediated TA injury mechanisms, including Hull's theory and isolated flail/tertiary blast injury. Guillotine-type TAs from explosive fragments of sufficient size and energy have also been documented—a type of secondary blast injury. All of these need to be considered in any mitigation/prevention strategy. However, this new clinical data has generated an injury mechanism theory through which tertiary blast injury can account for both transosseous and through-joint TAs. Most importantly, this highlights a target to act against to try to prevent these terrible injuries to save both limbs and lives.

15.5 Further Laboratory Studies

Subsequent cadaveric animal research has investigated the mechanism of injury in dismounted blast [11]. This study investigated a cadaveric mouse model of dismounted blast utilising a shock tube to replicate the blast wave, with a focus on the association between lower limb flail and pelvic blast injury (see Chap. 16). Whilst focusing on pelvic injury, a lower rate of TA than expected was noted. As the experimental setup utilised a blast wave biomechanically offset from the casualty, this was thought to be recreating the proposed through-joint TA mechanism from tertiary blast as described above. This research subsequently went on to investigate underbody axial blast utilising a shock tube mediated blast wave. However, with direct underbody axial blast wave loading, again an absence of TAs was noted (in particular, a lack of transosseous TA).

Subsequent tests have gone on to review the relationship between increasing blast pressure and TA. This research has failed to show a relationship between increasing blast pressure and TA. Of note, the increasing blast pressures utilised eventually caused injuries deemed non-survivable (including thoracic injury and abdominal evisceration) without increasing TA rates. As such, and in view of prior research noting a lack of link between fatal PBLI and TA, the possibility of an additional aspect of blast injury resulting in diaphyseal fracture is suggested.

The shock tube experimental platform utilised in these experiments recreated primary and tertiary blast, however, there was an absence of secondary blast injury. The authors of the study utilised a bilateral TA mouse model (having sustained a pre-crush injury to the thighs prior to the blast wind) to explore the relationship between TA and pelvic injury [11]. In all cases, bilateral above knee TA was seen following the impact of the blast wind. This enabled the proposition of the hypothesis that secondary blast injury, in the form of a ‘sand blast’, may recreate the required soft tissue and bony destruction and therefore play a role in diaphyseal fracture and amputation. This ‘sand blast’ hypothesis has similarly been suggested as a possible mechanism of injury in pelvic blast injury [12].

15.6 Association to Pelvic Blast Injury

Pelvic fractures secondary to blast injury strongly correlate with TA; they occur more frequently in those with bilateral lower limb amputation than those with unilateral lower limb amputation, and they are significantly associated with a higher incidence of concurrent TA [13]. Other authors have similarly noted that the probability of pelvic fracture increases significantly with more proximal TAs, in particular, bilateral transfemoral amputations [14]. The correlation between these two injury types suggests the mechanism of injury may be linked. This will be discussed in more detail in Chap. 16.

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16 Pelvic Blast Injury

Iain A. Rankin and Claire E. Webster

Abstract

Blast injury to the pelvis has been a significant cause of death to UK personnel in Iraq and Afghanistan. UK Military experience has shown that casualties of lower extremity blast are at a significant risk of death due to non-compressible pelvic vascular injury following pubic symphysis and sacroiliac joint disruption. Under-vehicle blast casualties are protected from fatal pelvic injury, with different injury patterns seen. The mechanism of injury in dismounted (on foot) blast has been suggested to be due to axial load via the femoral head, 'sand-blast' displacement of the pelvis or outward flail of the lower limbs. Lower limb flail in a small animal model has been shown to correlate with pelvic fracture displacement and vascular injury. Military pelvic protection introduced towards the end of recent military conflicts has reduced the risk of urogenital injury. With improved knowledge of the biomechanics of blast injury to the pelvis, future research aims to explore mitigation strategies to modify current military protective equipment to limit fatal pelvic blast injury.

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16.1 Introduction

Although the majority of wartime trauma involves the limbs, it is injuries to the head, torso and junctional haemorrhage that are the most life threatening. As haemorrhage is the most common cause of preventable death on the battlefield, it is of considerable research interest (see Chap. 9, Sect. 9.1).

In recent conflicts in Iraq and Afghanistan (2003–2014), the Improvised Explosive Device (IED) changed the character of injuries from penetrating wounds (from gunshots or fragments) to the extensive tissue loss and heavily contaminated injuries associated with close range explosions. As the majority of the devices are victim operated, lower extremity trauma is almost universal. Those with associated pelvic fractures have the worst clinical outcomes, often due to non-compressible pelvic bleeding—currently there are few opportunities for control of this in the pre-hospital environment.

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The pelvis, consisting of paired iliac, ischial and pubic bones in a ring like structure, forms the inferior margins of the abdomen and contains small and large bowel, ureters and urethra, and reproductive organs. It also has a rich network of vasculature, paired external iliac and femoral arteries and veins and multiple branches of the internal iliac leading to the pelvic organs. The pelvic bones are strongly reinforced by a network of ligaments and surrounding musculature—requiring a considerable amount of force to cause a fracture. Therefore when a fracture does occur secondary to a high energy insult, there is a high incidence of associated injuries both locally and to additional body regions such as the head and thoracic structures [1]. It is possible that if fractures could be prevented, or local bleeding controlled, then the mortality from IEDs could be reduced. One of the key initial steps is to understand the mechanism of fracture following pelvic blast injury.

16.2 UK Military Experience

The United Kingdom military maintains a prospectively recorded registry of its injured personnel, the Joint Theatre Trauma Registry (JTTR). A search of this was performed, and significant pelvic trauma caused by blast injury from the conflicts in Iraq and Afghanistan between 2003 and 2014 was identified.

In this period, there were 365 pelvic fractures associated with a blast mechanism. Following exclusion of casualties that had insufficient data available or were local nationals for whom follow-up data was not available, the overall mortality rate for the remaining 307 patients was 53%. This is significantly higher than that quoted in the civilian literature, where published mortality rates are between 8 and 14% [2–4].

As in civilian fractures, these injuries are associated with significant trauma to other body regions—which may be the cause of death rather than the pelvic fracture itself. The casualties described above consisted of two cohorts: 148 of the above casualties were noted to have more severe injuries remote to the pelvis, with an associated mortality of 71%. 159 patients without predominant injury in other body regions were identified, with a lower mortality of 36%. All deaths within this isolated pelvic trauma group were noted to have sustained a pelvic vascular injury. Military data is similar to civilian pelvic fractures, whereby a significant cause of mortality is due to associated injuries. The burden of non-compressible pelvic haemorrhage, however, is significantly higher from pelvic fractures caused by a blast mechanism than from those sustained by other causes.

An association of pelvic fractures caused by a blast mechanism with traumatic amputation and substantial perineal injury has been seen [5]. This collection of injuries was noted with increasing frequency in recent military conflicts and termed dismounted (on foot) complex blast injury [6]. Both traumatic amputation and perineal injury are found to be significantly associated with mortality when combined with pelvic fracture caused by blast. Perineal injury in particular was found from the UK JTTR data to be strongly associated with a pelvic vascular injury when in the presence of unstable pelvic fractures.

Unstable pelvic fractures are seen following dismounted blast injury, whilst a distinct difference has been noted in pelvic fractures sustained following mounted (in-vehicle) blast injury. In vehicle, casualties sustain primarily rami and sacral fractures in combination with spinal fractures—consistent with crush injury at the seat. The cause of death in these circumstances is not due to the pelvic injury. The on-foot casualty, however, is noted to have unstable pelvic fracture patterns predominantly consisting of pubic symphysis and sacroiliac joint disruption, in addition to traumatic amputation of the lower limb [7]. These injuries combined with fracture displacement are hypothesised to result in vascular disruption and death.

16.3 Mechanism of Injury

The mechanism by which dismounted blast injury results in fatal pelvic vascular injury is not clearly known. Three separate hypotheses have been suggested:

1. Axial load via the femoral head could cause load propagation to the acetabulum. Instead of acetabular fracture, the femoral head displaces the hemipelvis with a resultant superolateral separation.
2. High pressure blast wind and fragmentation may cause displacement of the pelvis in a 'sand blast' type effect.
3. Lower limb flail, as the legs are blown outwards, may transfer loads which result in lateral displacement of the hemipelvis. This has been shown in an animal model, where increasing lower limb flail angle resulted in increased probability of displaced pelvic fractures and pelvic vascular injury [8]. Flail of the lower limb has similarly been suggested as a mechanism of injury in traumatic amputation (see Chap. 15 'Traumatic Amputation'). The association between traumatic amputation and pelvic blast injury suggests that these injuries may be linked by this mechanism of injury.

16.4 Pelvic Fracture Classification

Civilian pelvic classification systems aim to direct patient treatment pathways based on severity. Common systems include the Young and Burgess classification [9], a later modification of the Pennal and Tile classification [10], which describes the fractures based on the direction of the force upon them. They are grouped into anterior posterior compression fractures (APC), lateral compression (LC) and vertical shear (VS). Pennal and Tile's classification considers stability of pelvic fractures, dividing them into grades A-C depending on the extent of rotational and vertical stability [10]. Type A fractures are considered stable, type B rotationally unstable but vertically stable, and type C unstable both rotationally and vertically [11].

Although classification systems guide treatment following civilian trauma, military pelvic fractures do not fit into these divisions clearly, meaning the classification is not fit for purpose in this patient group. To date no classification system for military pelvic fractures has been universally accepted. Analysis of CT (including post-mortem) imaging of the UK JTTR cohort did not allow for a pelvic classification system but did identify several key features. 93% of fatalities were noted to have unstable pelvic fracture patterns. Disruption of the pubic symphysis and sacroiliac joints, with lateral displacement of the sacroiliac joints, was found to be the most predictive radiological features for pelvic vascular injury. These fatal injury patterns seen most closely relate to a combination of VS and APC III, which are rotationally and vertically unstable (Tile type C).

16.5 Bleeding Following Pelvic Trauma

Bleeding may result from arterial injury or from the lower pressure venous system. Venous bleeding has been demonstrated in civilian studies to be responsible for around 90 percent of pelvic fracture haemorrhage, commonly to the iliolumbar vein or presacral plexus [12]. Arterial bleeding, when present, is more likely to lead to haemodynamic instability and death in the military environment.

There are few civilian studies detailing the exact source of arterial haemorrhage in pelvic fracture. One study of 63 patients demonstrated that superior gluteal, followed by internal pudendal, obturator and lateral sacral arteries, respectively, were responsible for bleeding [13]. A second study detailing frequency of embolisation of arteries (one of the options for treating significant bleeding) demonstrated that the internal iliac was the most commonly embolised artery, followed by the superior gluteal, internal pudendal, lateral sacral and obturator and iliolumbar arteries [14].

Military blast data has shown a higher incidence of large vessel injury to be present. A cohort of 28 military casualties who underwent damage-control laparotomy for shock in association with pelvic injury identified iliac vein injury to be present in 50%, internal iliac artery injury in 29% and external iliac artery injury in 25%. Of those who died, 86% had sustained an iliac vessel injury [15]. UK JTTR CT analysis identified pelvic vascular injuries to consist of 53% venous and 47% arterial, a significantly higher proportion of arterial injury than seen in civilian casualties.

It is important to know the type and site of bleeding, as venous bleeding can cease by stable clot formation and tamponade within the retroperitoneal space [16]. This is less likely to occur following arterial bleeding, and more direct control may be required [17]. Mechanically unstable pelvic injuries can result in an increase in pelvic volume whereby retroperitoneal tamponade may not occur, and bleeding can continue. Reducing the pelvic volume and stabilisation of the pelvis, by binder or fixation, is paramount in the management of these life-threatening injuries. Previous research at Imperial has demonstrated the importance of closure of the pelvic ring, and how improper placement of pelvic binders leads to inadequate closure of open pelvic fractures. This can lead to delays in haemostasis, and ultimately, an increase in mortality from pelvic trauma [18]. This non-invasive technique is taught on all Advanced Trauma Life Support (ATLS) and Military Operational Surgical Training (MOST) courses as a quick and reliable method for pelvic fixation in suspected pelvic fracture in the pre-hospital setting. Whilst amenable to controlling low-volume venous bleeding, it is not suitable to control large vessel injury nor arterial injury [19]. As a result, haemorrhage from pelvic vascular injury is frequently not controlled until damage-control surgery is performed, with pelvic packing and direct pelvic vessel ligation via laparotomy [20]. In these circumstances, injury mitigation is key to improving outcome.

16.6 Injury Mitigation

Death can occur soon after injury from uncontrollable bleeding as discussed above, or later from multiple organ failure—which is likely to be related to both the initial blood loss and significant contamination at the time of injury. The unparalleled survival rates seen in recent conflicts suggest that mitigation, rather than any additional improvement in treatment, is more likely to improve future outcomes. Currently, mitigation is limited to soft tissue protection from the direct effects of the explosion.

Personnel protective equipment (PPE) to the lower extremity was introduced in response to the frequency of urogenital injuries, with a historical incidence of approximately 5% [21]. This is in three tiers, a silk under layer, Kevlar™ shorts, and a further Kevlar genital protection piece [22]. Although it was only possible to compare to a historical control group, this protection has been associated with a significant decrease in urogenital injury [23]. Whilst the protection has been successful in reducing urogenital injury, future research should be targeted on modifying current PPE to provide further protection against pelvic fracture and vascular injury.

16.7 Research Direction of the Centre for Blast Injury Studies

Recent research has shown the flail mechanism to be the cause of dismounted pelvic blast injury in a small animal model [8]. Future research now aims to recreate this mechanism of injury in the laboratory using cadaveric human specimens, within a previously validated blast rig [24]. Methods to test the axial load hypothesis and sand-blast hypothesis are similarly being considered.

Subsequent finite element modelling will allow different injury mechanisms to be reproduced and identify the amount of force that would be required to prevent displacement of the pelvis, in particular, lateral displacement of the sacroiliac joints. This information can then be used in the modification and development of future pelvic PPE.

16.8 Conclusion

Blast injury to the pelvis has been a significant cause of death to UK personnel in Iraq and Afghanistan. Casualties of lower extremity blast are at a significant risk of death due to non-compressible pelvic vascular injury following pelvic fracture. Traumatic amputation following blast is significantly associated with pelvic fracture, for which their mechanisms of injury may be linked. With a knowledge of the biomechanics of blast injury to the pelvis, it is hoped that future research can explore mitigation strategies. The ultimate aim is to prevent, mitigate or improve the treatment of severe blast injury, in order to improve the morbidity and mortality of these injuries.

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Blast Injury to the Spine

17

Lucas Low, Edward Spurrier, and Nicolas Newell

Abstract

Combat-related spinal injuries have been reported since the Egyptian era. In more recent conflicts improved medical care, and possibly different wounding mechanisms, has seen an increased incidence of spinal injuries. This chapter presents a survey of the patterns, distribution, and demographics of modern spinal blast injuries, reported in the literature. The data is categorised into mounted and dismounted injuries. As different patterns of injury are observed, it is likely that different mechanisms of injury are involved for these two categories. Understanding the mechanism of spinal blast injuries allows the design of vehicles and seats to be improved in order to minimise the severity of blast to military personnel. Analysing the patterns of spinal injury also provides the possibilities of using injury patterns as markers for more severe injuries.

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17.1 Introduction

Although spinal injuries in warfare were first reported in the Egyptian era [1, 2], the devastating consequences, with long-term pain and sensorimotor disability, have been reported in relatively few papers since then. Publications prior to the Gulf conflict of 2003 made little reference to spinal injuries, but there has been greater interest more recently, possibly related to different wounding mechanisms and the increasing incidence of spinal injury (Fig. 17.1).

Fig. 17.1 The increasing incidence of spinal injury in US conflicts in key historical publications. Spinal injury received little attention before the Korean War [3–5]

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It is clear from the recent literature that blast injury causes the bulk of spinal injuries in conflict (Fig. 17.2) and that the patients affected are young (Table 17.1). Understanding the patterns of injury in blast related spinal fracture is an essential first step in understanding the mechanism of injury. Once the mechanism is understood, mitigation is possible; steps can be taken to change vehicle and equipment design to reduce the risk of injury for future generations. Identifying the most significant injuries might also support targeted treatment to improve overall clinical outcomes for these victims.

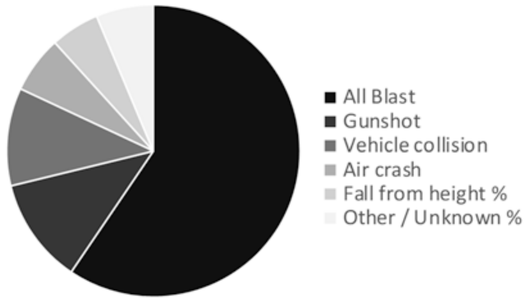


Fig. 17.2 Causes of combat spinal injury in recent literature [3–11]

Table 17.1 Demographics and cause of military spinal injury in recent literature

	Mean age (years)	Male %	Female %	All blast %	Gunshot %	Vehicle collision %	Air crash %	Fall from height %	Other/ unknown %
Bell [6]	27	98	2	56	14	7	3		7
Belmont [7]	25.8	98.5	1.5	75	20	3			3
Blair [8]	26.4	98	2	56		27		6	11
Blair [3]	26.5	98	2	56	15	25	5		
Comstock [9]	29.3	99.5	0.5	72	7	9		5	7
Possley [10]	26.5	98.4	1.6	53					
Schoenfeld [4]	27.8	92	8	83	3	3	9		
Schoenfield [11]	26.6	98	2	67	15		11		7
Schoenfield [5]	26.6	99	1	75	14.8	7.8			
Mean from all papers, unweighted	27	98	2	66	13	12	7	6	7

17.2 Injury Patterns

Unfortunately, early papers lack specific details of fracture patterns and location of injury. Barr et al. [12] described the solid blast injury mechanism in 1944, in which victims are injured when an explosion strikes a ship, accelerating the deck upwards and indirectly striking the victim. Further analysis of this form of tertiary blast (see Chap. 9) was not conducted until more recently when interest in spinal injury in military patients developed during the conflicts in Iraq and Afghanistan. Several recent studies have made use of registries of trauma patients such as those established by both the United Kingdom and United States; these registries include more detail of the fractures, allowing a more comprehensive analysis. The general distribution of spinal blast injuries reported in these studies is summarised in Table 17.2.

Table 17.2 Overview of military spinal fracture distribution from all causes

Study	Cervical		Thoracic		Lumbar	
Bevevino [13] Spinal injuries in combat amputees	5	(6%)	15	(18%)	62	(76%)
Bilgic [14] Case report of lumbar burst fracture due to anti-personnel mine					1	(100%)
Blair [3] US Casualties 2000–2009	319	(18%)	591	(33%)	857	(49%)
Comstock [9] Canadian casualties	6	(13%)	15	(33%)	25	(54%)
Davis [15] Injuries on the USS Cole	2	(17%)	8	(73%)	1	(8%)
Eardley [16] Review of British military spinal trauma	2	(5%)	14	(32%)	28	(64%)
Lehman [17] Review of the “Low lumbar burst fracture”					39	(100%)
Possley [10] Review of spinal injuries in improvised explosive device (IED) strike	279	(17%)	543	(34%)	787	(49%)
Schoenfeld [11] Review of fatal injuries in US troops	704	35%	731	36%	579	29%
Schoenfeld [4] Review of injuries in a single US unit	4	(40%)	2	(20%)	4	(40%)
Schoenfeld [5] Review of US casualties 2005–9	231	(22%)	300	(28%)	522	(50%)
Ragel [18] Thoracolumbar spine injuries in IED strike			4	(25%)	12	(75%)
Spurrier [19] Spine injury patterns in mounted UK personnel attacked by IEDs	21	(18%)	42	(35%)	55	(47%)
Newell [20] Transverse process fractures in UK personnel attacked by IEDs	7	(4%)	26	(16%)	132	(80%)
TOTALS	1580	(23%)	2287	(33%)	3092	(44%)

Using the data presented in Table 17.2 it can be seen that generally spinal blast injuries are more common in the lumbar spine than the thoracic or cervical (Fig. 17.3). This data can be interrogated further to investigate the specific levels at which injury is seen. This analysis shows that injuries are particularly common in the upper lumbar spine (Fig. 17.4).

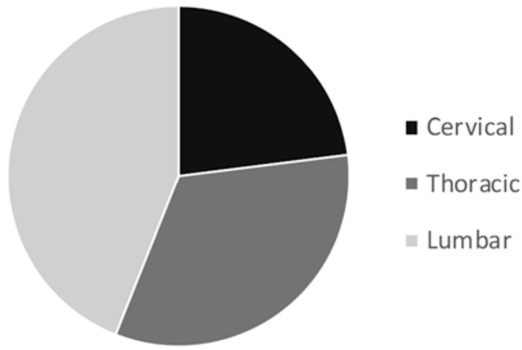
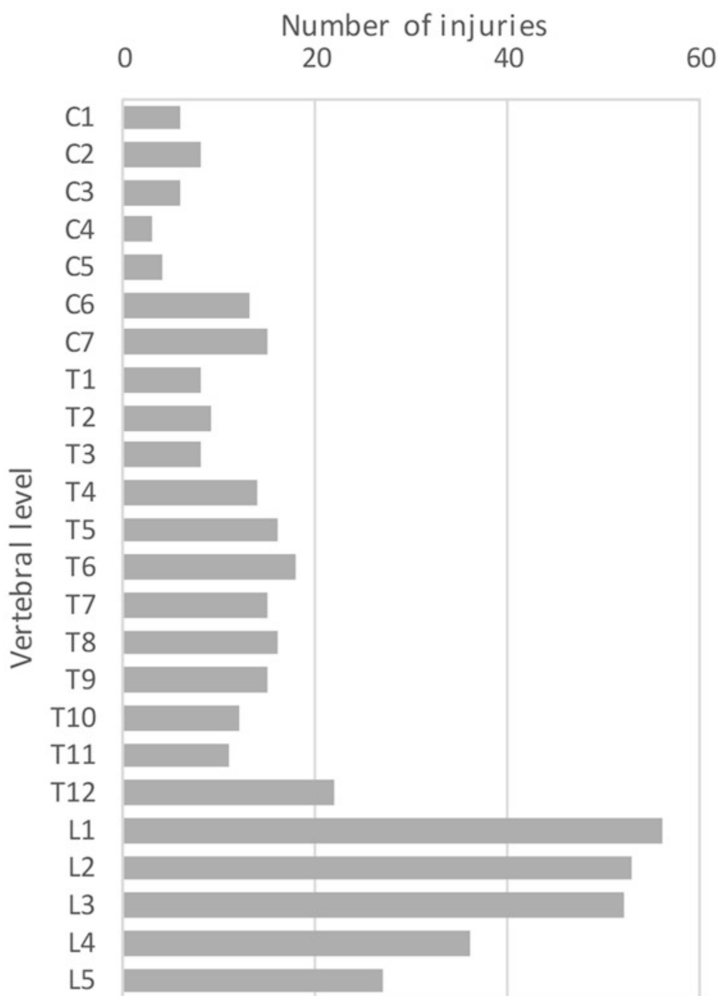


Fig. 17.3 Spinal fracture distribution in all patients [3–5, 9–11, 13–17, 19–21]

Fig. 17.4 Spinal injury distribution by vertebral level from blast injury victims using data from [4, 9, 14, 16, 18, 19]



In terms of the types of fractures presented, Table 17.3 shows that most (32%) are compression fractures, with transverse process, burst, and all sacral fractures also common.

Table 17.3 Published spinal fracture patterns in blast patients according to anatomical location and McAfee mechanistic classification of injury

	Odontoid peg	Facet fracture dislocation	Lamina	Transverse process	Compression	Burst	Flexion-distraction and chance	All sacral
Bevevino [13]		1		45	18	8		
Comstock [9]			1	14	7			
Eardley [16]					6	19	4	
Helgeson [22]								24
Ragel [18]					7	3	5	
Schoenfeld [4]	1			5	3	1		
Spurrier [23]					44	50		
Total	1	1	1	64	85	81	9	24
%	0.3%	0.3%	0.3%	24.1%	32.0%	30.5%	3.4%	9.0%

While understanding the general distribution of spinal blast injuries is important, it is difficult to develop mechanistic theories without separating the patterns of injury into groups dependent upon the likely modes of loading. For blast victims, it is common to categorise victims into mounted (in vehicle) and dismounted groups, as the effect of blast on an exposed victim (primary blast) involves a very different mechanism to the effect of a blast that imparts its force through a vehicle (tertiary blast). Studies describing the patterns of injury encountered in warfare do not always detail whether the victim was mounted or dismounted; those that do are reviewed in Sects. 17.2.1 and 17.2.2.

17.2.1 Mounted Blast Injury Patterns

Spurrier et al. [19] analysed injuries from 78 mounted casualties who had sustained spinal fractures between 2008 and 2013 from the UK joint theatre trauma registry (JTTR). The number of fractures at each vertebral level is shown in Fig. 17.5a. Most fractures in this group were lumbar, with 17% being wedge compression fractures and 15% being unstable burst fractures, both these types of injury were most common at the thoracolumbar junction. In the cervical spine, distraction-extension fractures and compression-extension fractures have been found to be common. Ragel et al. [18] analysed injuries from 12 occupants of vehicles subjected to IED attacks between January and May 2008. They found that thoracolumbar fractures were common; they also identified a high incidence of flexion-distraction fractures (seen in 38% of patients), which is considerably higher than that reported in most spine fracture series (1–2.5%) [24, 25].

In a separate study, Spurrier et al. [23] compared mounted blast injuries with ejection seat injuries (Fig. 17.5b), seeing fewer cervical, more thoracic, and fewer lumbar fractures in the ejection seat group compared to the mounted blast group, suggesting different mechanisms of injury.

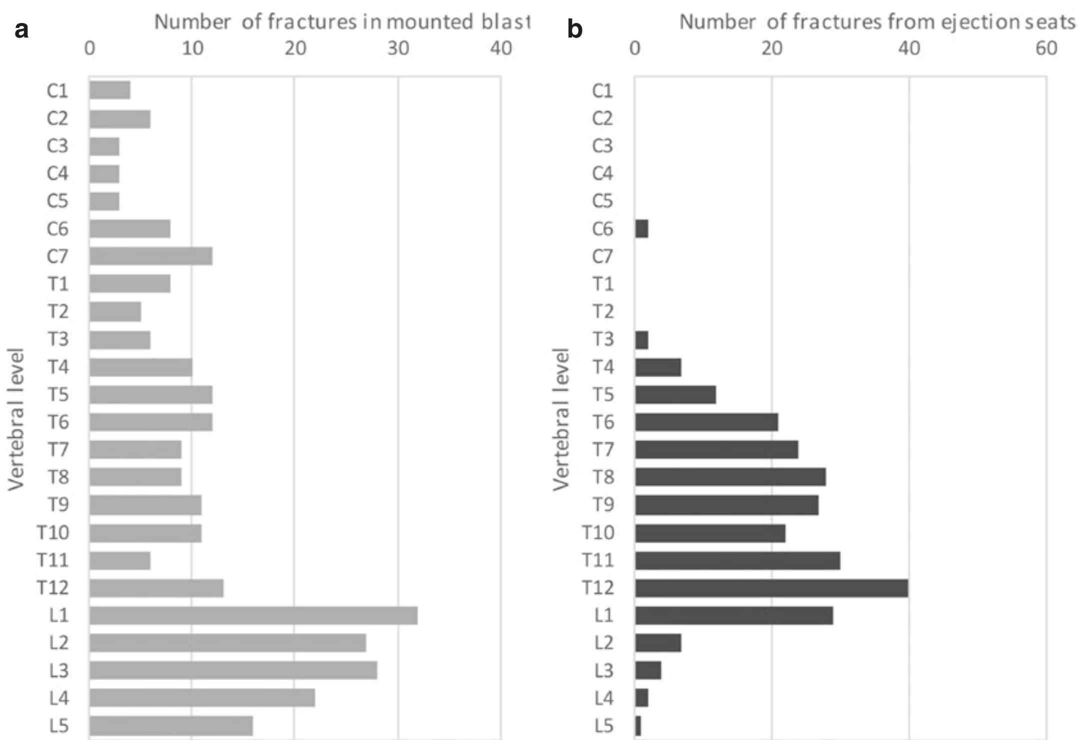


Fig. 17.5 (**a**) Number of fractures at each vertebral level in mounted casualties recorded in the UK's JTTR [19], and (**b**) levels of fractures from ejection seat incidents [23]

17.2.2 Dismounted Blast Injury Patterns

Few studies have compared mounted and dismounted injury patterns but those that have note similarities in terms of the level of fractures, with the thoracolumbar junction being the most common site of injury [10]. When the analysis delves deeper, for example, looking specifically at the transverse processes, the majority of fractures are seen in the lumbar spine in both mounted and dismounted casualties. However, more bilateral fractures (fractures of both transverse processes at the same level) have been seen in mounted casualties [20]. Furthermore, cervical spine injuries have been found to be less common in dismounted compared to mounted casualties [26].

17.3 Mechanism of Blast Spinal Injuries

Ascertaining precise mechanisms of injury from blast is difficult due to the wide range of variables that often lead to complex and relatively unpredictable loading conditions. For example, Stewart et al. [27] reported injuries sustained by vehicle occupants from a single IED attack. One fatality experienced traumatic amputations and pelvic disruption, another fatality experienced head impact and no traumatic amputation, while a third casualty, sitting close to both fatalities escaped relatively uninjured. Although categorising casualties is challenging with such complex loading, it seems sensible to discuss the likely mechanism of injury in mounted and dismounted casualties separately.

17.3.1 Mounted

The large number of lumbar fractures seen in mounted blast (Fig. 17.5a), and the fact that many of the fractures are burst [19] suggests that most casualties are seated, with loading to the spine being axial through the pelvis. The high incidence of vertebral wedge fractures reported by Spurrier et al. [19], and the high incidence of flexion-distraction injuries reported by Ragel et al. [18] suggests that loading through the thoracic and lumbar spine is anterior and that the spine is flexed at the time of loading. The reason for this could be due to flexion of the spine as a result of the legs being forced upwards by deforming the vehicle floor with the torso held rigid by a seat harness (which is supported by the high incidence of lower limb fractures in under-body blast—see Chap. 9) or by flexion of the spine around the base of the body armour and seat harness worn by occupants. The high incidence of transverse process fractures in the lumbar spine reported by Newell et al. [20] may be caused by the loading inducing high intra-abdominal pressures that cause tensile forces acting through the lumbar fascia to avulse the transverse processes. This analysis of transverse process fractures is also useful to rule out the possibility of large bending movements since bilateral fractures were common, and fractures confined to just one side of the spine, that may be expected in violent lateral bending, were not. This is likely due to occupants being restricted to some extent by restraint systems.

The more cranial injuries seen in ejection seat incidents (Fig. 17.5b) are likely due to the higher loading rate seen in blast compared to ejection seat which has been shown to shift spinal injuries inferiorly in human cadaveric experiments [28]. This has also been suggested to be due to the effects of wearing body armour that reduces mobility of the upper lumbar spine, increasing the risk of low lumbar burst fractures [17].

The most likely mechanism of injury for the cervical spine of mounted occupants is through impact with the ceiling by their head or helmet. This would explain the higher number of cervical fractures seen in mounted compared to dismounted casualties [26]. It has also been suggested that the rigidity of the rib cage, and support from body armour, may protect the thoracic spine, and increase the risk of injury at adjacent levels at the lower cervical spine [17]. The cervical distraction-extension fractures and compression-extension fractures seen by Spurrier et al. [19] suggest that the neck is subjected to buckling modes of failure. There are, however, a significant number of mounted personnel who have cervical spine injuries but do not sustain head or neck injuries (43% [5]), suggesting direct impact to the ceiling is not always the mechanism of injury.

17.3.2 Dismounted

Dismounted casualties are normally standing, and therefore, loading is more likely to be off-axial. Newell et al's [20] analysis of transverse process fractures suggests blunt trauma, violent lateral flexion-extension forces, or rapid flail of the lower extremities causing the psoas muscle to load the spine, are likely mechanisms of injury. Additionally, dismounted casualties are susceptible to serious spinal injuries caused on landing after being thrown into the air by the initial explosive loading.

17.4 Markers for Fatality

Thirty-nine percent of all US fatalities between 2003 and 2011 from conflicts in Iraq and Afghanistan experienced a spine injury of some kind [11] demonstrating that spinal injuries are common in fatalities. Most papers do not separate fatal from non-fatal injuries or do so with insufficient detail to identify any mechanisms which might be associated with fatality. Schoenfeld et al. [11] analysed the levels of spinal fractures seen in fatalities finding 52% at the cervical spine, 44% at the thoracic spine, and 30.5% at the lumbar spine. This is a higher proportion of cervical spine injuries compared to the data presented in Table 17.2 and Fig. 17.3 in this chapter (23, 33, and 44%, respectively). In particular, C1 fractures were found to be common in fatalities which was also seen by Spurrier [29] who compared fracture level in survivors and fatalities in UK spine blast casualties between 2008 and 2013 (Fig. 17.6). A greater incidence of cervical spine transverse process fractures has also been seen in fatalities compared to survivors (8.9% vs 2.5%) [20]. A recent study by Stewart et al. [27] however, showed that the link between cervical spine fractures and fatality is likely to be due to the fact that these victims also suffered a fatal head injury, rather than the spinal fractures themselves being lethal.

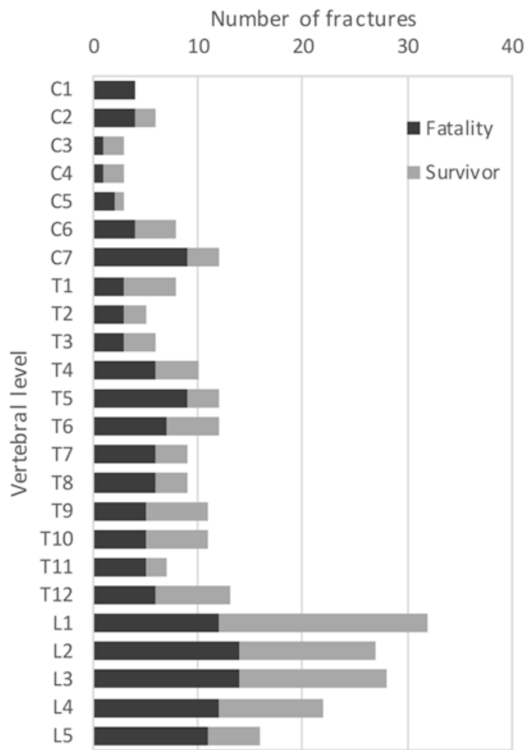


Fig. 17.6 Number of mounted victims with fractures at each level showing survivors and fatalities, taken from Spurrier et al. [29]

Newell et al. found that isolated lumbar transverse process fractures were a marker for fatality with the likelihood being three times greater when a lumbar transverse process fracture was present. This could be due to the association between L5 transverse process fractures and more severe trauma such as pelvic injuries that have been shown to be strongly associated with mortality [30, 31].

17.5 Associated Injuries

It is common for a single victim to sustain more than one spinal injury with three quarters of casualties having injury at multiple spinal locations [32]. Spinal injuries are also associated with injuries in other body regions with head/neck, and extremity injuries being particularly common [5, 11, 16, 18]. In the series of casualties analysed by Shoenfeld et al. [11], 74% of casualties with spine injuries also had extremity injuries, and 73% had head or neck injuries. Cervical spine fractures have been found to be significantly associated with skull fractures, and a trend towards an increased risk of pelvis fractures with low lumbar fractures has also been documented [19]. Isolated lumbar transverse process fractures in blast victims are significantly associated with head injury, noncompressible torso haemorrhage, and pelvic injuries in mounted casualties (odds ratios of 3.69, 4.38, and 3.28, respectively) [20]. However, a recent analysis by Pearce [33] based on data from the UK JTTR investigated associations between locations of all blast injuries and found evidence for two distinct injury complexes: the lower limb complex and the torso complex, with no injury associations between the two.

17.6 Outcomes

The Edwin Smith Papyrus [1] records the futility in early medical history of treating spinal injuries associated with paralysis, advising physicians not to attempt to treat such injuries. Little is then published with regard to the outcomes of spinal injury until the recent conflict in Afghanistan except for one series in World War 2.

Fifty-six American soldiers with penetrating spinal wounds were reported at the end of World War 2 [34]. This series followed from explosive fragment or gunshot (secondary blast injuries). Treatment was with laminectomy when there was progressive neurological abnormality or evidence of metallic or bony fragments in the spinal canal on plain radiographs. At the end of follow-up (up to 40 months), 4 of the 19 patients who were paraplegic at the time of injury had made some recovery and 22% of patients with neurological deficit made a complete recovery. However, 36% of patients with a neurological deficit immediately following injury made no recovery. Recent studies continue to show that spine blast injury often presents irreversible neurological damage [35–37].

Specific case reports of individual patient outcomes are rare. Kang et al. [38–41] reported several specific cases. One patient is described with an L5 burst fracture following exposure to blast from an IED associated with bilateral transfemoral amputations and normal neurological function in the residual limbs, but 50% occlusion of the spinal canal. He was treated with L4 to S1 fusion and achieved a pain-free outcome despite needing steroid injection for radicular pain. In this case, surgery was advocated despite the lack of neurological compromise in order to facilitate rehabilitation.

Complications following treatment of military spinal injuries have been reported in a US series [10]. The overall complication rate following spinal trauma was 9% with a high rate of multiple complications. Wound infections, venous thrombosis, and cerebrospinal fluid leak were the most common complications, and patients injured in dismounted mechanisms were at higher risk.

Although the clinical outcome of spinal fractures in blast victims is not known, it is reasonable to assume that fracture patterns which show a poor outcome in civilian injury are likely to also lead to poor outcomes in blast patients. Generally, burst fractures imply a greater risk of neurological complications and pain than compression fractures and flexion-distraction injuries [42–44]. Given that each of these fracture patterns implies a specific mechanism of injury, if the features of a vehicle, harness, or seat design that lead to each mechanism and injury can be elucidated, it might be possible to control the patterns of injury seen in subsequent blast incidents. It might, perhaps, be possible to change devastating burst fractures into minor injuries by altering the seating design and posture.

17.7 Summary

There is evidence that the incidence of spinal injury in warfare has increased in recent conflicts. While spinal injuries remain less common than limb injuries, they have the potential to cause significant disability and therefore are worthy of attention. This chapter has reviewed published clinical data to ascertain spinal injury patterns for both mounted and dismounted casualties. Using this data, theories for mechanisms for injury have been developed which may in turn be able to be used to inform biomechanical investigations that aim to investigate these mechanisms further. With this information, improvements to vehicle and equipment design may reduce the incidence of spinal injuries in future conflicts.

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Primary Blast Lung Injury

18

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Abstract

Primary blast lung injury occurs when an explosive shock wave travels through the thoracic cavity and dissipates kinetic energy, predominantly at the alveolar—air interface. This results in varying degrees of alveolar rupture, pneumothoraces and intra-parenchymal haemorrhage. Immediate fatalities are due to venous air embolism, whilst survivors present with respiratory failure and haemoptysis. Medical management frequently requires admission to an intensive care unit and can represent a significant logistical challenge. Current treatment is supportive, including drainage of pneumothoraces and mechanical ventilation. Once ventilated, experience to-date suggests that the condition stabilises and is not associated with further mortality.

As the injury is borne out of the chaos of war or terrorist attacks, clinical research is difficult. Much current research relies on modelling with either in vivo or in silico surrogates for human injury. In silico modelling is becoming increasingly sophisticated. The availability of more powerful computers and modelling software is increasing the complexity and fidelity of these tools. Such modelling is beginning to explore the pre-hospital and in-hospital medical management of primary blast lung injury with the aim of improving care in both arenas.

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18.1 Introduction

The adult human lung consists of some 500 million air sacs (alveoli), each only 1/3 mm in diameter but generating a combined surface area of approximately 100 m^2. The alveolar wall measures 0.2–0.3 μm across and is encased in a mesh of very fine and fragile blood vessels the diameter of which is just sufficient to allow the passage of red blood cells [1]. This structure facilitates effective gas exchange and acts as a blood–gas interface. Whilst elegantly adapted to facilitate rapid diffusion of gas across tissue, these delicate structures make lung particularly susceptible to injury following blast exposure. Injury from the blast wave is known as primary blast lung injury (PBLI). It is the predominant component of a multisystem syndrome of primary blast injury. As a diagnosis of exclusion, it occurs within 12 h of blast exposure in the absence of secondary or tertiary lung injury and in the presence of radiological or arterial blood gas evidence of acute lung injury [2].

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18.2 Epidemiology

Primary blast lung injury (PBLI) is encountered globally resulting from military conflict, acts of terrorism and industrial accidents. During the recent counter-insurgency campaign in Afghanistan, point prevalence surveys identified PBLI in 6–11% military casualties surviving to reach a field hospital [3, 4]. A review of all UK military casualties (survivors and non-survivors) suffered in the recent conflicts in Iraq and Afghanistan, as a result of exposure to an explosion suggests a PBLI incidence of 8% [5]. Within this group, 25 casualties attended hospital with PBLI as their only diagnosis. This figure can rise to almost 80% in non-survivors [6] in whom it is the only autopsy finding in 17% [7]. The incidence and severity of PBLI increase significantly with proximity to the explosion and when injuries are sustained within an enclosed space such as a bus or train. Following the Madrid terrorist train bombings, in which 10 explosive devices were detonated in train carriages, approximately 150 casualties suffered PBLI out of a total of 250 serious casualties [8]. A similar experience was reported following terrorist bus bombings in Israel in which 15 out of 17 casualties who survived to reach hospital suffered PBLI [9].

18.3 Pathophysiology

The blast wave dissipates its kinetic energy within the lung through the generation of shear and stress waves [10]. Transverse, low velocity shear waves result from deformation of the thoracic wall. They are responsible for the surface haemorrhage seen on lung tissue facing the explosion (Fig. 18.1) and cause random movement of tissues of differing densities around fixed points resulting in tearing of parenchymal tissue. Supersonic longitudinal stress waves generate more diffuse damage. The speed of transmission through the lung does not allow energised gas to pass along airways and thus "foaming" occurs. As the lung adopts the physical properties of foam, with its much poorer transmission of sound energy, greater kinetic energy is absorbed [11]. Rapid compression and expansion of alveoli lead to alveolar rupture and hence to the hallmark of the disease which is the formation of abnormal air-filled spaces such as pneumatoceles, pneumothoraces and venous air embolism. The stress wave reflects back upon itself as it reaches the denser mediastinum. This causes a "stress concentration" effect in lung tissue around the mediastinum leading to characteristic findings on imaging discussed below. Microscopically, severe alveolar over-distension is ubiquitous. Concomitantly, alveolar capillaries rupture and with the formation of alveolo-venous fistulae result in localised haemorrhage. This can be significant and cause immediate respiratory compromise. This extravasated blood precipitates a free radical mediated inflammatory process involving leucocyte accumulation and oxidative damage resulting in perivascular oedema. This inflammatory process continues to evolve over the subsequent 24–56 h. Pulmonary fat embolism contributes significantly to respiratory compromise and to an early risk of death, as does bone marrow embolism in casualties suffering long bone and/or pelvic injury [12]. The extent of this inflammatory process will in part depend on the overall burden of whole body tissue injury driving a systemic inflammatory response (see Chap. 13).

Fig. 18.1 Mammalian lung exposed to a sub-lethal blast overpressure. Extensive parenchymal haemorrhage can be seen on the left hand surface which was facing the explosion. Courtesy of Dr. Emrys Kirkman, Defence Science and Technology Laboratories, UK

Physiologically, the vagal nerve mediates a reflex apnoea, bradycardia and hypotensive episode [13] via stretch of peri-alveolar C-fibres. This lasts up to 15 s and is followed by a period of rapid shallow breathing [14]. This mechanism may play a significant role in non-survivors [15].

18.4 Diagnosis

Casualties will most likely be symptomatic by the time they reach a medical facility. They will present with shortness of breath, respiratory distress and haemoptysis. Tachycardia, tachypnoea and cyanosis will reflect the severity of the insult. Severe cases will develop acute respiratory distress syndrome (ARDS) described below.

The classic chest radiograph (CXR) appearance is of bilateral perihilar ("batswing") infiltrates generated as the pressure wave reflects back from the mediastinum (Fig. 18.2). The incidence of this varies considerably between series and may reflect a casualty's proximity to an explosion. Computed Tomography (CT) imaging also demonstrates increased opacification around the mediastinum. It will also reveal parenchymal haemorrhage and is more likely to reveal the pneumatoceles and pneumothoraces that are characteristic of the disease (Fig. 18.3). More detailed CT image analysis of parenchymal haemorrhage is currently being developed as a tool for the diagnosis and quantification of PBLI [16]. The process involves creating a three-dimensional image of the lungs (Fig. 18.4) and calculating the density of the imaged lung units (Voxels) to determine the proportion of lung injured by blast.

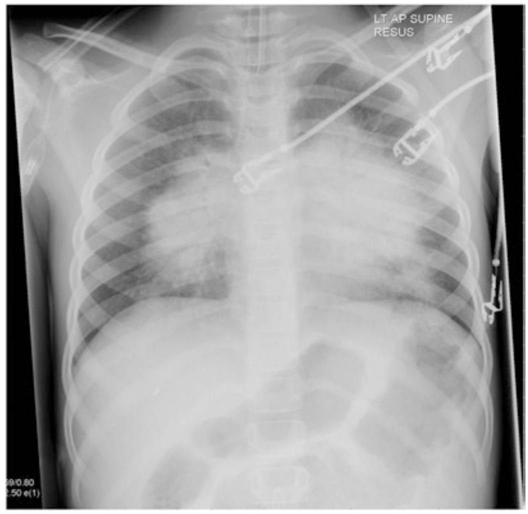

Fig. 18.2 Presenting CXR of an adolescent male exposed to an explosion in an enclosed space. The characteristic "Batswing" distribution of lung injury is seen. The patient made a complete recovery. Courtesy of Lt Col Iain Gibb RAMC, Portsmouth, UK

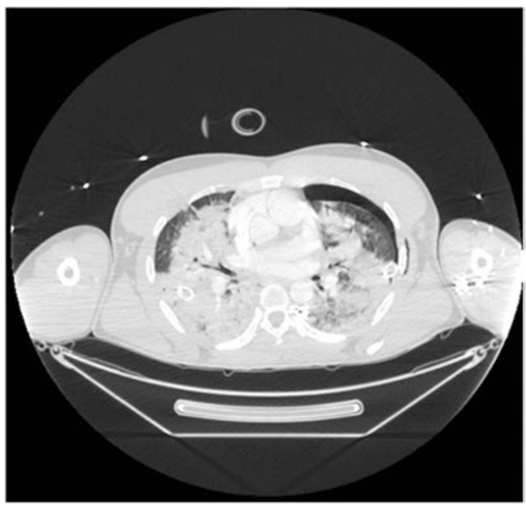

Fig. 18.3 Computed tomography image of a soldier exposed to a significant explosion. The patient required immediate intubation and ventilation requiring a fractional inspired concentration of oxygen of 90%. Shortly after intubation, the patient was referred for ECMO though this was not subsequently required. The patient made a good recovery. Significant consolidation around the mediastinum and a left sided pneumothorax is seen. Courtesy of Lt Col Andy Johnson RAMC, Birmingham, UK

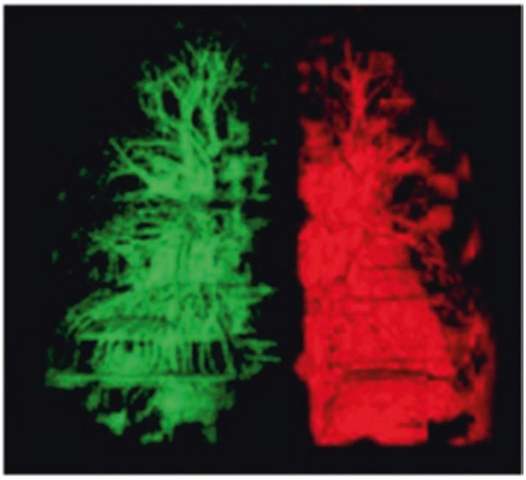

Fig. 18.4 3-Dimensional reconstruction of a casualty's lungs. Voxel density analysis allows determination of the amount of lung tissue damaged by blast

Alternative diagnosis at this stage includes pulmonary contusion (rib fractures and peripheral lung injury), tension pneumothorax, ARDS due to other causes (inhalation of toxic substances, gastric aspiration) and use of chemical weapons.

18.5 Acute Respiratory Distress Syndrome (ARDS)

ARDS is the development of non-cardiogenic pulmonary oedema within 7 days of a precipitating insult and demonstrating bilateral opacities on chest imaging. Fluid overload (i.e. renal failure) and heart failure may have to be excluded. This life-threatening syndrome characterised by hypoxia of rapid onset can result from either direct or indirect injury to the lung. It is classified as either, mild, moderate or severe depending on the inspired concentration of oxygen to arterial blood oxygen tension ratio (Table 18.1) [17]. Affected lungs demonstrate markedly reduced compliance ("stiff lungs") resulting from alveolar flooding with neutrophil rich proteinaceous material and deposition of a hyaline membrane. Histologically, the evolution of the condition is classically described as occurring in three phases, the exudative, proliferative and fibrotic phases. The first exudative and inflammatory phase consists of widespread neutrophil infiltration and alveolar flooding with haemorrhagic oedema. This cumulates in the deposition of a fibrous hyaline membrane. During the proliferative phase, fibrinous exudates become organised with the formation of collagen fibrils as a result of increased fibroblastic activity with concomitant necrosis of the lung's epithelial lining. The final fibrotic phase is characterised by further collagen deposition and intimal thickening of blood vessels.

Table 18.1 The Berlin Classification of the severity of ARDS. PaO_2, Partial pressure of arterial oxygen; FiO_2, Fraction of inspired concentration of oxygen

Oxygenation	PaO_2/FIO_2[a]
Mild	200–300 mmHg
Moderate	100–200 mmHg
Severe	< 100 mmHg

[a]In the presence of a positive end-expiratory pressure (PEEP) of 5 cm H_2O or greater

18.6 Management

In common with other primary blast injuries, the violent and unpredictable nature of explosions demands that clinical care be guided by surrogate modelling of potential therapy rather than clinical trials in humans [18]. Traditionally, such modelling has been in vivo in nature [19]. With increasing computational power, in silico models are becoming increasingly prevalent [20, 21].

Such modelling supports the use of continuous positive airway pressure (CPAP) in the pre-hospital environment. As in other forms of acute lung injury, this modelling suggests clinical benefit to the application of low-dose CPAP even when utilising ambient air (Fig. 18.5) [22, 23]. History tells us that casualties with PBLI will require supportive care in a high dependency or intensive care environment. Some 80% will require mechanical ventilation. Casualties returning to the UK from Afghanistan responded well to the low tidal volume (LTV) approach to mechanical ventilation advocated for patients suffering from acute respiratory distress syndrome [24]. This cohort of casualties required on average 4.5 days of mechanical ventilation and there was no requirement for advanced salvage measures such as extracorporeal membrane oxygenation (ECMO) [5]. This compares favourably with more conventional causes of ARDS and reflects the fact that trauma-related acute lung injury is recognised as a more responsive disease compared to non-trauma related acute lung injury [25, 26].

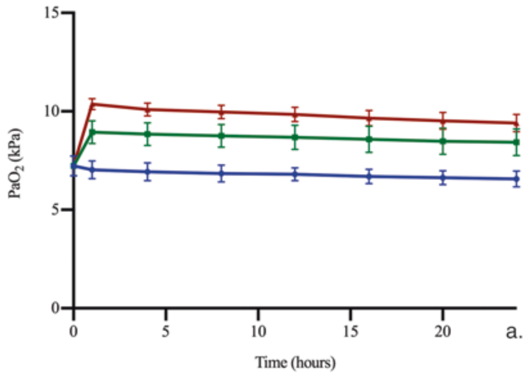

Fig. 18.5 The modelled effect of 5 cm H_2O CPAP using ambient air in in silico casualties with primary blast lung injury. The graph demonstrates the potential for clinically important increases in PaO_2 by the pre-hospital administration of low-dose CPAP without the use of supplementary oxygen. The blue, green and red lines represent 0 cm H_2O, 5 cm H_2O and 10 cm H_2O, respectively

The fundamental component of this strategy is the recognition that mechanical ventilation can significantly augment pre-existing lung injury due to a combination of barotrauma and volutrauma due to the repeated opening, stretching and collapse of alveoli during the mechanical ventilatory cycle. This injury is minimised through use of a pressure limited mode of ventilation and "splinting" the lungs open with higher than normal levels of positive end-expiratory pressure (PEEP). The lungs are ventilated at lower than normal 6–8 ml/kg lean body weight, and the inspiratory component of the respiratory cycle is extended. This is a compromise ventilator strategy with the clinician accepting relative but not life-threatening hypoxia and hypercarbia whilst awaiting resolution of the pulmonary insult. Airway pressure release ventilation (APRV) is an alternative form of ventilation available to treating clinicians and was used in British casualties from Afghanistan following return to the UK. In silico modelling suggests that it performs equally well to conventional LTV ventilation and may offer clinically important benefits by reducing extravascular lung water, driving pressure and peak pressure [27]. Alternative forms of ventilation such as high frequency oscillatory ventilation (HFOV) have been used but are dependent on local expertise and availability [28].

Prone positioning of ventilated patients with ARDS can improve oxygenation by reducing ventilation perfusion mismatch. In the prone position, the ventral and normally better-ventilated area of lung now also receives the greater proportion of pulmonary perfusion as gravity dictates that dependant tissue is better perfused. When considered, prone positioning should be instigated early and for at least 16 h a day [29]. Early neuromuscular blockade with cisatracurium should also be considered in the presence of resistant hypoxia [30]. Any pneumothoraces should ideally be drained prior to transfer to a CT scanner, though the increasing speed and improving access to cross-sectional imaging make this less vital in patients with relative respiratory stability but who may have other life-threatening injuries. It must be borne in mind that large pneumothoraces can be missed on supine CXR's. A euvolaemic to hypovolaemic approach to fluid management can attenuate pulmonary oedema in appropriate patients and reduce mortality [31].

Patients who are breathing spontaneously at 2 h post injury are unlikely to deteriorate further and develop a need for mechanical ventilation due to PBLI alone [32]. Patients who are asymptomatic at 6 h post exposure can be discharged from close medical observation. Tympanic membrane rupture is poorly correlated with PBLI with a sensitivity of 29% [33].

In the longer term, patients suffering isolated blast lung injury can expect to make an excellent recovery demonstrated by lack of symptoms, normal exercise tolerance and normal lung function tests [34].

18.7 Future Therapy

Future pharmacological therapy will target the inflammatory component of the injury. The antioxidant N-acetylcysteine (NAC) has shown early promise by modulating neutrophil mediated pulmonary inflammation in rodent models of blast injury and offers the advantage that it is currently widely used for other reasons in intensive care [35]. NAC's more potent successor, N-acetylcysteine amide (NACA), is currently being investigated. Hemin (heme oxygenase-1 inducer) and the TRPV4 family of transmembrane calcium channel blockers are also candidate treatment options with studies pending.

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19 Blast Injuries of the Eye

Robert A. H. Scott

Abstract

The eyes occupy 0.1% of the total and 0.27% of the anterior body surface. As vision is the most important sense, the significance of their injury is far more substantial. Loss of vision is likely to lead to loss of career, major lifestyle changes, and disfigurement. Eye injuries come at a high cost to society and are largely avoidable. This chapter identifies the range of ocular blast injuries in relation to the anatomical features of the eye (Fig. 19.1).

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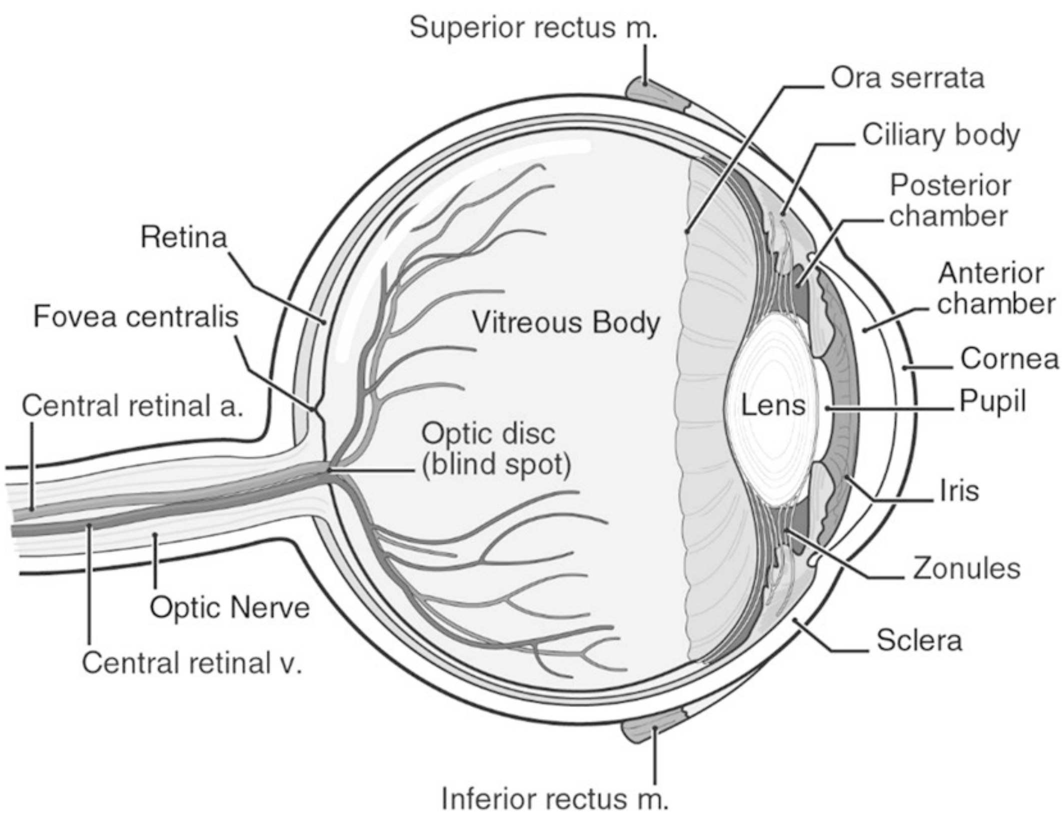

Fig. 19.1 The anatomy of the eye

19.1 Primary Ocular Blast Injuries

The eye consists of several interfacing coats that vary in their elasticity and density that can be damaged by an explosive shock wave to cause a primary blast injury (PBI). It has been postulated that reflection of the shock wave by the bony orbit amplifies this effect [1]. Improved eye protection from secondary injury and improved general survivability from explosive injuries by using/through protective clothing and combat eye protection have made pure PBI more common [2]. Factors that influence the severity of a PBI include the size of the explosion, the distance that the eye is from it, and the orientation of the eye to the blast wave.

Ocular PBI cases are found among the subgroup of casualties with closed globe injuries and intraocular lesions. The posterior segment of the eye, particularly the retina, is particularly susceptible to PBI and is affected in around 60% of cases, even though the blast wave traverses the anterior segment to reach it [3]. Ocular PBI cases have a characteristic presentation of a profoundly hypotonic eye without evidence of globe rupture, often with traumatic cataracts. This spontaneously resolves over approximately 7–10 days.

In the anterior segment, conjunctival lacerations and subconjunctival haemorrhages are common after explosions. The iris and ciliary body are damaged causing with hyphaema, iris sphincter rupture, dialysis, and spiral tears. This can lead to secondary glaucoma. Ciliary muscle atrophy causes reduced accommodation making reading difficult with associated eyestrain. Traumatic cataracts are common after

PBI, these often spontaneously resolve. Occasionally, the lens will swell and extraction is required [4].

In the posterior segment, induced posterior vitreous detachment can lead to vitreous haemorrhage, retinal tears and detachments, or traumatic macular holes [5]. Macula commotio retinae when photoreceptors are damaged leading to a whitened appearance of the retina is common after PBI. This differs from secondary blast injuries where the commotion retinae is more frequently in a peripheral retinal area. Typically, there is a profound drop in vision that slowly recovers leaving a variable pattern of residual retinal atrophy [6].

Optic neuropathy is a well recorded, if uncommon, PBI associated with large blast forces. Optic atrophy follows massive retinal atrophy, induced vasoconstriction of the optic nerve blood supply, or mechanical disruption of optic nerve axons as they leave the eye through the lamina cribrosa can. A retrobulbar haemorrhage from rupture of the vortex veins as they leave the globe compresses the optic nerve blood supply inducing acute optic nerve ischaemia [7].

The prognosis from ocular blast injuries is generally recorded as poor, with a final visual acuity of 6/12 or better achieved in 16–32% of cases. Injuries to the optic nerve, choroid, and retina carry a worse prognosis than those to the anterior segment, adnexae, or intraocular haemorrhage [8].

19.2 Secondary Blast Injuries

Secondary blast injuries (SBI) are due to the impact of fragment from the explosive device itself or from exogenous debris propelled by the explosion. They are the most common form of high explosive ocular injuries. The projectiles cause penetrating and perforating injuries to the globe and ocular adnexa. Until recently, it was rare to find a case of PBI without any evidence of contusion or any secondary injury to the eye or adnexae [9].

19.3 Closed Globe Injuries

These are common to both primary and secondary blast injuries, but secondary blast injuries are caused by the impact of an object. When a blunt object strikes the eye, the lens-iris diaphragm is displaced posteriorly centrally while the peripheral structures are expanding outwards. This causes tearing of the ocular tissues, particularly in the iridocorneal angle.

When the compressing object is larger than the orbit, it pushes the globe posteriorly to suddenly increase the orbital pressure, relieved typically by an inferior blow-out fracture of the orbit into the maxillary sinus. This phenomenon often protects the globe from injury though there is up to a 30% incidence of ruptured globe reported in conjunction with orbital fracture [10].

Corneal and conjunctival abrasions and foreign bodies (FB) are very common after blast injuries. The eyes are very painful, but recover rapidly after FB removal, patching, and topical antibiotic ointment. After blast injuries, particularly from improvised explosive device (IED) and mine explosions, there are often multiple deep stromal foreign bodies. After an initial period of inflammation, these characteristically become quiescent and the FB lies inert within the stroma, even years after the injury. Apart from problems of glare from light reflecting off the stromal FB, they are usually symptomless [11].

Traumatic iritis is common with perilimbal injection, anterior chamber cells and flare. Hyphaema is caused by bleeding in the anterior chamber and describes the layering of RBC in the inferior anterior chamber. Cyclodialysis clefts are from traumatic separation of the ciliary body from the sclera, this allows aqueous to exit directly to the subchoroidal space and causes ocular hypotony. Iridocorneal angle recession occurs when the ciliary body is torn and displaced posteriorly. It is associated with raised intraocular pressure and secondary glaucoma in 7–9% of cases [12]. Traumatic cataracts may not appear for years after the injury causing glare and loss of vision. They can be complicated due to lens zonular dehiscence, with a higher chance of subluxation and surgical complications [13].

Posterior segment contusion injuries are common. Commotio retinae is more likely to be in an extramacular position than with PBI, giving a better visual prognosis. Choroidal ruptures are common; they are typically crescent-shaped and sited at the posterior pole. Visual recovery is the rule, but can be reduced if the rupture involves the fovea, if there is choroidal subretinal neovascularisation or a significant subretinal haemorrhage.

Sclopetaria is a peripheral traumatic chorioretinal rupture from a high velocity concussion injury. There are often dramatic retinal tears with associated shallow retinal detachment. These characteristically do not require active treatment as scarring from the surrounding tissue seals the retinal break [14].

Optic nerve avulsion occurs when an object intrudes between the globe and orbital wall to disinsert the optic nerve. There is sudden, irreversible loss of vision. Traumatic optic neuropathy is a common secondary blast injury occurring directly from disruption of the nerve axons or indirectly from vasoconstriction of the pial vessels that supply the optic nerve causing ischaemic neuropathy. The ensuing neural deficit typically reduces the visual acuity, brightness sense, and colour vision with partial recovery.

19.4 Traumatic Retinal Tears and Detachments

Retinal dialyses are typical post-traumatic retinal injuries where there is peripheral retinal disinsertion between the edge of the retina and the ora serrata from sudden expansion of the ocular equator from blunt injury. The detachment evolves slowly, often years after the trauma. Eyes sustaining penetrating or open globe trauma have a high risk of retinal detachment, occurring in 10–45% of cases.

Retinal detachments associated with penetrating and perforating eye trauma are common and frequently associated with a profound retinal scarring response, proliferative vitreoretinopathy (PVR), that causes recurrent detachment and poor visual results. The mean incidence of PVR is 27% for all open globe trauma [15].

Traumatic macular holes form as a result of vitreoretinal traction over the macular area after ocular contusion, they are typically 300–500 microns in diameter, but if there is an element of retinal necrosis they can be much larger. There is a good visual prognosis with around half closing spontaneously. Most surgeons will operate to close those that do not close by 3–4 months [16].

Other retinal breaks can form as a result of an induced posterior vitreous detachment from the injury. If there is an area of increased vitreoretinal traction, typically over blood vessels a retinal tear forms often with a vitreous haemorrhage. The tears can cause a rhegmatogenous retinal detachment. Retinal tears and detachments are managed surgically with retinopexy for retinal tears and vitrectomy and internal tamponade or cryopexy and buckling procedures for retinal detachment.

Penetrating eye injuries are sharp eye injuries that have a single entrance wound from the injury. Perforating eye injuries have an entrance and exit wound. Management is by urgent primary surgical repair, this can be followed by a definitive secondary procedure at during the same procedure or at a later date depending on circumstances. Corneal lamellar lacerations where the eyewall is not breached can be directly sutured, or if there is a 'flap', a contact lens can be inserted to close the wound [17].

Globe rupture is a full-thickness wound of the eyewall from a blunt injury. The eye is filled with incompressible liquid and the impact causes sufficient pressure to rupture the eye at its weakest point, by an inside-out mechanism. Globe rupture must be excluded by ultrasound scan in all cases of hyphaema or post-traumatic media opacity that prevents indirect ophthalmoscopy of the fundus. Surgical exploration and primary repair are performed if globe rupture is suspected. Secondary procedures for intraocular haemorrhage are delayed for up to 14 days to allow the blood clot to liquefy when it can be surgically drained as part of a vitrectomy procedure [18].

Intraocular foreign bodies (IOFB) cause 14–17% of all ocular war injuries; they are a very important subset of eye injuries as they have a modifiable outcome using modern diagnostic, therapeutic, and surgical techniques. In modern warfare, a large proportion of IOFBs are from grit and stones thrown up by explosions; they are frequently multiple [19]. Appropriate eye protection significantly reduces the incidence of IOFB injuries. Surgical removal of posterior segment IOFBs is by pars plana vitrectomy. Systemic antibiotic coverage reduces the risk of endophthalmitis until the IOFBs can be removed [20].

Recovery from primary and secondary ocular blast injuries can be divided into three main stages. The first stage of active general treatment and healing lasts for 3 weeks. Poor vision at this time does not preclude a satisfactory outcome. The second stage from 4 to 12 weeks is when individual clinical patterns requiring specific treatments appear, an accurate prediction of the final outcome can be made at this stage, largely depending on the state of the macula and other chorioretinal damage. The intraocular pressure recovers at this stage. The third stage, from 3 months to 3 years can see limited further improvement, but the vision rarely deteriorates [21].

19.5 Tertiary Blast Injuries

Tertiary ocular blast injuries are from the effects of being thrown into fixed objects or structural collapse and fragmentation of buildings and vehicles by an explosion. As any body part may be affected, the injury pattern is varied and eye injuries will usually be part of a wider injury pattern, which will often be combined with other facial trauma [22].

Tertiary blast causes direct traumatic injury, as well as indirect ocular injuries. Purtscher's retinopathy is a sudden onset multifocal, vasoocclusive event, associated with head and chest trauma, causing sudden loss of vision that usually recovers over weeks and months. The appearance is of multiple patches of superficial retinal whitening with retinal haemorrhages surrounding a hyperaemic optic nerve head. Fat embolism syndrome is a potentially fatal variant that causes respiratory and central nervous system failure after long bone fractures [23].

Valsalva retinopathy is a sudden loss of vision from the preretinal haemorrhage that occurs after a sudden increase in intrathoracic pressure and is common after explosions. Spontaneous recovery is the rule [24].

Terson syndrome is a vitreous haemorrhage that occurs after an intracranial haemorrhage, thought to be related to an acute rise in intracranial pressure that is transmitted to the retina causing rupture of the papillary and retinal capillaries. It is often diagnosed when a patient recovers from the subarachnoid haemorrhage and is found to be profoundly blind [25].

Water-shed infarcts of the parieto-occipital lobes of the cerebral cortex are associated with severe blood loss and profound hypotension. The infarct occurs at the borders of cerebral circulation that are sensitive to ischaemic insults. These classically cause bilateral visual pathway damage with cortical blindness [26]. Non-arteritic ischaemic optic neuropathy where there is visual loss in one or both eyes with associated optic atrophy due to optic nerve ischaemia has a similar aetiology following explosive injuries [27].

19.6 Quaternary Blast Injury

Quaternary blast injuries of the eye are explosion related injuries or illnesses not due to primary, secondary, or tertiary injuries. There may be exacerbations of pre-existing conditions, such as glaucoma or cataracts. Chemical and thermal burns are common around the eye and adnexae in association with ballistic injuries. In thermal burns, tissue damage is usually limited to the superficial epithelium, but thermal necrosis and ocular penetration can occur. Chemical burns are blinding emergencies. Alkaline agents such as lye or cement penetrate cell membranes and cause more damage than acidic agents, which precipitate on reaction with ocular proteins [28].

19.7 Quinary Blast Injuries

These are a new entity that describes a hyperinflammatory state, unrelated to the injury complexity and severity of trauma, occurring after an explosion, particularly associated with hypercoagulability. It is postulated that they are caused by unconventional toxic materials used in the manufacture of the explosive [29]. Retinal vascular occlusion occurs in approximately 10% of ocular explosive blast injuries and has occurred where there has not been any other ocular involvement [30]. Some of these cases may represent quinary ocular blast injuries.

19.8 Summary and Incidence

According to the study of ocular injuries in British Armed Forces in Iraq and Afghanistan between July 2004 and May 2008 [18], a total of 630 British soldiers survived major traumatic injuries. Of these, 63 (10%) sustained ocular injuries with some 86% (54) were deemed to be caused by explosive blast. Of these, the most common injury report was open globe with the mean time to primary repair being 1.9 days with an average of 1.57 operations per eye.

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20 Hearing Damage Through Blast

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Abstract

Hearing damage through blast is a major problem in the armed forces. Blast can cause damage to all stages of the auditory pathway: from the outer, middle and inner ear to the different parts of the brain that successively process the complex sound information. In this chapter, we give an overview of the main stages of the auditory system, their main functions and the types of impairments that can occur to them through blast. We then describe prevention, diagnostic strategies as well as treatment and rehabilitation methods. We also discuss current research into improved prevention and treatment and conclude with a brief summary.

20.1 Introduction

Hearing damage through blast is an escalating problem in the military: it accounted for 25% of all injuries during Operation Iraqi Freedom in 2004 and was accordingly the most common single injury [1]. Auditory dysfunction in general is now the most prevalent individual service-connected disability, with compensation estimated to exceed $1.2 billion in 2016 in the USA [2, 3]. A recent report on Royal Marines returning from deployment in Afghanistan has found that two-thirds sustained severe and permanent hearing damage [4, 5].

Hearing damage can occur at several key stages of the auditory pathway (Fig. 20.1). The latter consists broadly of the outer ear which collects sound, the middle ear which acts as a lever, the inner ear in which the mechanical sound stimulation is converted into electrical nerve impulses and the central nervous system that processes and analyses the auditory signals. Blast can damage all parts of this system [6]. In the following, we discuss their impact as well as detection, prevention and treatment strategies.

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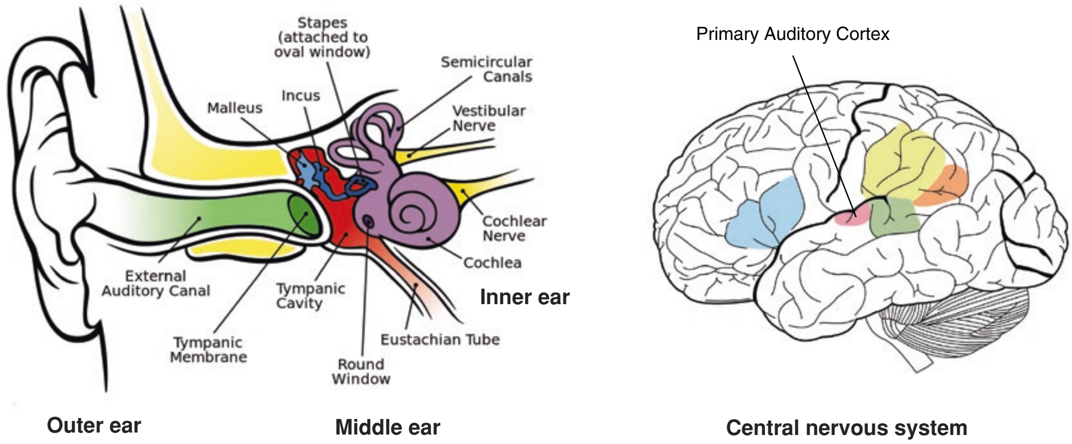

Fig. 20.1 The auditory system consists of the outer, middle and inner ear. It also encompasses parts of the central nervous system where the evoked neural signals are processed. Exposure to blast can cause damage at all stages

20.2 Blast Damage to the Outer Ear

The outer, or external, ear consists of the pinna and the ear canal. The outer ear collects sound and funnels it through the ear canal to the middle ear. It hence plays an important role in the detection of faint sound. The pinna also introduces spectral changes in a sound that are specific to the location of the sound [7, 8]. As the most prominent of these changes, the pinna introduces a spectral notch at an elevation-dependent frequency. This information can then be used by the brain to estimate the location of a sound [9].

Blast exposure can damage the outer ear, namely through burns as well as through damage by flying debris [10–12]. As long as the middle and inner ear are not damaged, such impairment of the outer ear leaves the hearing sensitivity largely unchanged. Sound localisation however will be compromised. Because the outer ear is visible and constitutes a key attribute to a human face, its damage can also have severe psychological implications [13]. Furthermore, burn patients are often treated with antibiotics that can be toxic to the ear, the mechanism of which remains poorly understood [14–16].

Detection of outer ear damage is straightforward from visual inspection. Prevention necessitates ear muffs that extend around the outer ear and hence shield it, or helmets. Treatment of outer ear damage requires surgery, namely debridement of the damaged parts of the outer ear, with repair if possible. If required, reconstruction of the amputated parts can often be successfully achieved, for instance with autologous costal cartilage [13, 17, 18].

20.3 Middle Ear Damage

The middle ear consists of the eardrum, or tympanic membrane, as well as the ossicles, the three smallest bones in the body: malleus, incus and stapes. Through these constituents, the middle ear forwards the mechanical sound vibration from the ear canal to the inner ear. It thereby acts as a lever and matches the impedance of the sound wave in the air to that of the emerging wave in the inner ear (see below). This impedance-matching by the middle ear is paramount for effective transmission of sound energy into the inner ear [19, 20].

As with the outer ear, blast trauma can cause damage to the middle ear [14]. Rupture of the tympanic membrane can occur from peak pressure levels of 130 dB or higher, which are common in a blast wave [21]. Such pressures can also disrupt the ossicles. The resulting damage can cause conductive hearing loss of up to 25 dB sound pressure level [22].

Rupture of the middle ear in a blast wave can have a protective effect on the inner ear. This is because once the ossicle chain has been disrupted, the excess pressure in a blast wave is no longer efficiently forwarded to the inner ear, and hence impairs it less.

Detection of middle ear damage involves visual inspection through the ear canal as well as tympanometry in which the reaction of the tympanic membrane to sound stimulation is measured.

Passive protection of the middle ear involves either earplugs or earmuffs. Such passive ear protection is most efficient at the high frequencies above 1 kHz where the wavelength of sound is short and where sound can be efficiently blocked [23–25]. It also affects the understanding of speech as this partly employs these higher frequencies. One problem with wearing this protection is that it not only blocks undesired noise but also desired speech signals as well as other environmental sounds that are necessary to assess a potentially hostile environment.

Recent development has focused on active hearing protection. Such equipment employs a microphone and a small speaker. The microphone records the external sound signal and plays a processed version into the ear. This can achieve noise cancellation as well as protection from overstimulation. Such active hearing protection is now employed in the UK armed forces in the form of the Personal Integrated Hearing Protection system and the Tactical Hearing Protection System [26, 27].

Regarding treatment, about 80% of ruptures of the tympanic membrane heal spontaneously [28]. In the remaining cases, as well as when the ossicles have been impaired, reconstruction of the middle ear often gives satisfactory results, at least at the lower frequencies that matter most for speech comprehension [29, 30].

20.4 Blast-Induced Impairment of the Inner Ear

The inner ear transduces the mechanical sound vibrations into an electrical signal in auditory-nerve fibres [31]. This mechanotransduction happens in hair cells, specialised cells that have a hair bundle at their apical end. Mechanosensitive ion channels inside this bundle of parallel stereocilia open and close when the bundle is displaced, and the resulting electrical signal in the hair cell can produce an action potential in the attached auditory-nerve fibres. The displacement of the hair bundles arises from the sound signal.

Due to an intricate hydrodynamics within the inner ear, every hair cell responds particularly well to a certain best frequency [32, 33]. Following a tonotopic map, hair cells from the base of the organ respond best to high frequencies (around 10–20 kHz in humans) and cells further apical respond strongest to progressively lower frequencies (down to 100 Hz, with frequencies down to 20 Hz detectable).

Blast can yield sensorineural hearing loss which is one of the most prominent causes of hearing problems [34–36]. It refers to damage of the hair cells, typically the hair bundles, which can be disrupted or otherwise damaged by overstimulation. Because each hair cell has a best response at a certain frequency, sensorineural hearing loss is often frequency-dependent. For instance, it may happen that the hair cells in a particular cochlear region are compromised, but remain intact elsewhere. This will then result in a hearing loss in the frequency interval that corresponds to the cochlear region with the damaged hair cells. Because other hair cells will still respond somewhat to frequencies from that interval, the hearing loss there will typically not be total.

The inner ear also contains the vestibular system, in form of the semicircular canals, that is responsible for our sense of balance. The semicircular canals house mechanosensitive hair cells that are similar to those of the inner ear but detect motion instead of sound. Damage to these hair cells can occur through the noise as well. Such damage is indeed often related to sensorineural hearing loss and causes problems with balance [37, 38].

Diagnosis of sensorineural hearing loss employs pure-tone audiometry in which a subject has to respond to pure tones of different frequencies and intensities. This is however not objective as it involves a subject's overt response. An objective diagnostic method employs otoacoustic emissions. These are tones produced by a healthy ear, for example in response to short clicks (transient-evoked otoacoustic emissions) or to two close frequencies (distortion-product otoacoustic emissions) [39, 40]. Lack of these otoacoustic emissions in a certain frequency region signals hearing loss there [41, 42].

Protection against sensorineural hearing loss involves passive or active earplugs as well as ear mugs as described in Sect. 20.4. Another highly promising route for the protection of the inner ear is through drugs that prevent sensorineural hearing loss [42]. Noise exposure induces the generation of reactive oxygen species (ROS) within the inner ear that, over the course of days after noise exposure, damage the mechanosensitive hair cells [43, 44]. Recent research has identified a number of drugs that potentially control the generation of ROS and that can serve as otoprotectants. These drugs include single or multiple antioxidants, such as D-methionine [45, 46], ebselen [47, 48], resveratrol [49, 50], neurotrophic factors [51] and lipoic acid [52, 53], as well as anti-inflammatory drugs such as salicylate [54], steroids [55–57] and TNF-inhibitors [58]. Although some of these drugs are being explored in animal models and others in clinical trials, none has yet been clinically approved.

There is currently no direct treatment for sensorineural hearing loss. As opposed to the hair cells of birds, for instance, mammalian hair cells do not regenerate and can currently not be replaced [59, 60]. Damage to the hair cells accumulates over life, which explains why sensorineural hearing loss is more prevalent in older than in younger people. Current treatments of this type of hearing impairment involve hearing aids that amplify sound as well as, in severe cases, cochlear implants [61, 62]. The latter are devices that bypass the outer, middle and inner ear to stimulate the auditory-nerve fibres directly. These devices have been developed to provide hearing in deaf people and are a major success of the emerging field of neurotechnology.

20.5 Damage to the Central Nervous System

The central nervous system can be divided into two main parts, the brainstem and the cortex.

The nerve signals emerging from the cochlea within the auditory-nerve fibres are first processed at different stages in the auditory brainstem. The brainstem performs sound localisation, detects onsets of sounds, and sharpens frequency selectivity [31, 63]. While the neuronal activity in the lowest levels of the brainstem, the cochlear nuclei, is relatively well characterised, the higher levels such as the superior olivary complex and the inferior colliculus remain rather poorly understood.

Importantly, the neural activity in the auditory brainstem can be used to assess the functioning of the ear as well as the brainstem [64, 65]. The neural signals of the brainstem can be measured non-invasively with only a few electrodes on the scalp. In response to short clicks, the electrodes detect a characteristic pattern of electrical, neural activity that consists of five peaks at different latencies. These peaks signal activity in different parts of the brainstem, beginning from the auditory-nerve fibres to the inferior colliculus. Their height and latencies inform the integrity of these organs. Current research investigates how these recordings can give more precise information about damage to the ear and to the brainstem, and how more complex responses, such as the frequency-following response to pure tones or the response to speech, arise [66–68].

The auditory cortex receives input from the auditory brainstem and represents the highest level of auditory processing in the brain. It is believed to be the site where recognition and processing of complex auditory objects such as speech and music occur. How exactly such processing is achieved remains poorly understood.

Recent research has shown how neural signals from the auditory cortex can give information about sound processing. For example, non-invasive electroencephalographic recordings (EEG) from scalp electrodes have demonstrated that the neural activity in the cortex traces the envelope of an attended speech signal as well as the beat of music [69–71]. Measurements from functional magnetic resonance imaging (fMRI), positron emission tomography (PET) and magnetoencephalography (MEG) give more detailed, three-dimensional pictures of brain activity in response to sound [72–74].

The central auditory system can be damaged through traumatic brain injury. This is a particular concern in the military since traumatic brain injury can occur through blast exposure or head concussions. The resulting injuries to the brain include the death of neurons as well as swelling and disconnection of axons due to shearing and stretching, which may particularly affect the central auditory system [75]. Such damage results in auditory processing disorder, the main consequence of which is a deficit with the processing of speech, and in particular with the challenging task of detecting speech in noise [76]. A recent study shows that the majority of war veterans that were exposed to blast during deployment have auditory processing disorders [6].

How to best diagnose auditory processing disorder is still a field of active research. Current diagnosis involves mostly a speech-in-noise test, such as an online test developed by the U K charity *Action on Hearing Loss* [77, 78]. Just as with pure-tone audiometry, such tests are based on a subject's behavioural response and require his or her cooperation.

Recent research has shown how non-invasive electroencephalographic recordings (EEG) of brain activity can demonstrate how a healthy brain processes speech and music [69–71, 79, 80]. This suggests that these recordings can be employed to assess hearing impairment, including central auditory processing disorder. Further research is needed to develop this technology into suitable hearing assessments.

Tinnitus means the perception of a 'phantom' tone for an extended period of time (5 min or longer) [81]. Such perception can presumably arise in different ways in the auditory brainstem and the central auditory system (somatic tinnitus) [82]. Notably, it may also arise from the inner ear itself (otic tinnitus). Tinnitus often accompanies hearing loss, and hearing loss in a certain frequency band typically produces phantom tones at those frequencies. It can severely affect a person's life, and in particular lead to insomnia, fear, withdrawal and depression.

Current assessment of tinnitus is based on a behavioural response from the tested subject, which makes testing slow and potentially inaccurate. Recent research has explored ways of automating such testing [83, 84]. More research is needed to further improve tinnitus testing. It will also be important to investigate how tinnitus can be diagnosed in an objective way. This may involve recordings of neural activity which have been shown to inform on—as well as potentially modulate—tinnitus [85, 86].

Protection and treatment of damage to the central auditory system remain intensely investigated as well. Regarding protection, safety helmets can be efficient in preventing brain injury [87, 88]. A range of pharmacological agents that can potentially protect the central nervous system and help rehabilitating it after injury has been identified in animal models, but confirmation of the effectiveness of these drugs in clinical trials has not yet been achieved [89].

20.6 Conclusion

Hearing loss from blast injury in the armed forces is a highly important emerging medical problem. Major questions remain regarding the prevention, detection and treatment of such hearing loss. Recent progress in a better understanding of the causes of blast-induced hearing loss, in the treatment of sensorineural hearing loss and in a better understanding of the role of the brain processes for hearing suggest that major improvements in all three areas can be achieved. Conquering these issues will have major implications for the well-being of military personnel.

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21 Torso injury from Under Vehicle Blast

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Abstract

The torso (comprised of the chest, abdomen, and pelvis) contains the soft vascularised organs of the body. These tissues are susceptible to injury from a variety of mechanisms, which may lead (due to bleeding, organ dysfunction, or contamination) to severe morbidity or death. Bleeding from these injuries is of concern as the internal nature of such bleeding is "non-compressible" and therefore not amenable to conventional battlefield first aid. This chapter is concerned with the occurrence of torso injury and non-compressible torso haemorrhage as a consequence of under vehicle blast. Water blast and penetrating blast injuries are not considered here. Torso injury has been shown to be a major cause of death under these circumstances, but the biomechanics of these injuries is poorly understood. This chapter discusses these injuries and describes contemporary biomechanical methods for predicting them in response to high-rate loading. Current methods are shown not to sufficiently characterise the under vehicle blast environment. Mechanisms of torso organ injury for under vehicle explosions are suggested, and more accurate injury criteria based upon cadaveric and animal experiments are proposed.

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21.1 Introduction

The chest and abdomen contain the soft, vascularised organs of the body which are susceptible to injury from most trauma mechanisms. The clinical manifestation of torso injury is dependent on the organ injured and the severity and can be considered as haemorrhage, contamination, or dysfunction.

Haemorrhage results from injury to any vascular or vascularised structure. Haemorrhage of any type may be classified as either compressible or non-compressible. Non-compressible torso haemorrhage (NCTH) describes those injuries occurring within the internal cavities of the torso which are not amenable to direct or indirect compression. These injuries are a common cause of salvageable death and a major research focus for novel methods of bleeding control [1–3]. These novel haemostatic methods may bridge the gap to definitive treatment; this bridge is essential given that survival from serious torso injury is time dependent with the death most commonly occurring within 30 min [4]. Treatment of even salvageable combat torso injuries relies not only upon effective treatment modalities but upon a robust evacuation system and point of care intervention.

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21.2 The Clinical Problem

Singleton et al. aimed to better understand the relationship between injury and death secondary to blast injury by examining likely cause of death amongst UK deployed blast casualties [5]. The academic group, which included engineers and clinicians, conducted a retrospective analysis of all UK military personnel killed by IED blasts in Afghanistan from November 2007 to August 2010. Only those casualties with available post-mortem CT (computed tomography) imaging and relevant incident data were included. A cause of death analysis was performed by the group which acknowledged that determination of a single injury as the cause of death is not often possible in the face of severe polytrauma. The group instead used the AIS system (Chap. 11) and listed all AIS4+ injuries as potentially lethal. These lethal injuries were grouped into categories based upon mechanism of death. Haemorrhage was divided into extremity haemorrhage (from upper and lower limb), junctional haemorrhage (from neck, groin, or axilla) and intra-cavity haemorrhage (intra-thoracic or intra-abdominal).

The group compared the injury patterns between mounted (in-vehicle) and dismounted (on foot) blast casualties. Injury burden between the two groups was different with mounted fatalities sustaining both significantly higher NISS and with injuries to significantly more AIS regions than the dismounted fatalities [5].

Junctional and extremity haemorrhage predominated as likely cause of death in the dismounted casualties. These injuries are described as part of the dismounted complex blast injury pattern that has been well described in the literature [6–9]. Deaths in this group are more likely to fall within the potentially survivable injury class given that bleeding could potentially be controlled with tourniquets and direct pressure.

Death in the mounted blast cohort was most commonly attributed to head injury and intra-cavity haemorrhage. Although supportive measures may reduce the effect of secondary injury [10], outcome from severe head injury in this setting is unlikely to be influenced by any point of care intervention.

Morrison et al. examined the epidemiology of NCTH in UK deployed forces. They identified 296 casualties with NCTH and explored the effect of individual organ injuries upon overall survivability using a multivariable logistic regression model. Mortality was most influenced by the presence of major vessel injury [11].

Although Morrison et al. examined battlefield injury and showed that 68.6% of casualties were injured by blast, they did not compare relative lethality by injury mechanism and did not separate mounted from dismounted blast. Secondly, their paper was intended to highlight potentially survivable injury and thus excluded heart and great vessel injury.

In contrast, Singleton et al. classified blast death by mounted or dismounted environment but did not further delineate the “intra-cavity” injury patterns and explore relative lethality of injuries [5].

Singleton et al. further examined the torso response to under body blast (UBB) by describing the patterns of lung injury [12]. They found high rates of apparent “primary blast lung injury” seen at post-mortem CT of mounted blast casualties. They also showed higher rates of primary blast lung injury from all forms of blast exposure (79% of mounted and 32% of dismounted casualties) compared with 11% of those blast exposed casualties arriving for treatment at a role 3 medical facility [13]. This difference is likely due to high rates of mortality in those exposed to overpressures high enough to cause PBLI. Significant PBLI amongst mounted casualties had not previously been described.

21.3 Pattern of Injury

Work has been undertaken to more fully describe the pattern of torso injury due to under vehicle explosions by examining the injuries sustained by UK forces from under vehicle blast during recent operations in Iraq and Afghanistan [14]. This work demonstrated that mounted blast loading may lead to injuries of the lungs, heart, great vessels, and all abdominal organs. The frequency and severity of these injuries varied (Table 21.1).

Table 21.1 Incidence of torso injuries from survivors and non-survivors of mounted blast injury

Injury type	Survivor (n = 297)	Non-survivor (n = 129)	Total	CFR	p
Splenic injury	16 (5.4%)	47 (36.4%)	63	74.6%	<0.001
Liver injury	6 (2%)	53 (41.1%)	59	89.8%	<0.001
Lung contusion	20 (6.7%)	34 (36.4%)	54	63.0%	<0.001
GI injury	8(2.7%)	33 (25.6%)	41	80.5%	<0.001
Haemothorax	8 (2.7%)	32 (24.8%)	40	80.0%	<0.001
Kidney injury	3 (1.0%)	35 (27.1%)	38	92.1%	<0.001
Cardiac injury	1 (0.3%)	31 (24.0%)	32	96.9%	<0.001
Lung laceration	2 (0.7%)	24 (18.6%)	26	92.3%	<0.001
Blast lung	10 (3.4%)	13 (10.1%)	23	56.5%	0.006
Other abdominal vascular injury	3 (1.0%)	19 (14.7%)	22	86.4%	<0.001
Thoracic aorta injury	0	17 (13.2%)	17	100.0%	<0.001
Pneumothorax	11 (3.7%)	2 (1.6%)	13	15.4%	0.192
Other thoracic vascular injury	1 (0.3%)	4 (3.1%)	5	80.0%	0.031
Pulmonary vein injury	0	4 (3.1%)	4	100%	0.008
SVC injury	0	2 (1.6%)	2	100%	0.091
Abdominal aorta injury	0	2 (1.6%)	2	100%	0.091
IVC injury	0	2 (1.6%)	2	100%	0.091
Pulmonary artery injury	0	1 (0.8%)	1	100%	0.303

CFR Case Fatality Rate, *p* value denotes significance of difference between survivors and non-survivors. Other abdominal vascular includes injuries to the renal, mesenteric, hepatic, and portal vessels. Other thoracic vascular included injuries to the azygous, subclavian, and intercostal vessels. *IVC* Inferior Vena Cava, *SVC* Superior Vena Cava

The majority of torso injuries are significantly more common in non-survivors than survivors. Injuries to the non-aortic great vessels and thoracic aortic injuries had case fatality ratio (CFR) of 100%. Given that casualties sustained a multitude of injuries, the effects of individual injuries were estimated using a multivariable logistic regression model. Mediastinal (including heart, thoracic aorta, SVC, and pulmonary vessel) injuries were the strongest predictor of death. Of this group, heart and aortic injuries predominated in number.

Thoracic aorta and heart injuries are the most statistically lethal injuries within this cohort. Cardiac and great vessel injuries have previously been described as non-survivable within the battlefield cohort and excluded from some definitions of NCTH [1, 15].

In the civilian setting, aortic injuries are relatively rare following common blunt trauma (predominantly automotive collision) and, although associated with a high mortality even when controlling for other polytrauma, are not untreatable [16].

These injuries do not occur in isolation, with a significantly larger number of injuries sustained by non-survivors of underbody blast. The "all or nothing" phenomenon of under body blast injury and severe polytrauma is difficult to quantify using conventional injury scoring systems which rely upon arbitrary injury descriptions and which are "maxed out" by a single fatal injury. Devastating injuries such as mediastinal and massive head trauma may be used as surrogate markers of absolute non-survivability, but it is important to remark that non-survivable injuries are not necessarily non-preventable.

21.4 Mechanism of Injury

The fundamentals of explosions and the sequence of events occurring during under vehicle blast are described elsewhere in this book (Chaps. 2 and 9). What has remained uncertain is how this form of loading causes torso injury. Examination of the clinical data described above and analysis of injury associations suggest that the loading environment causes two distinct injury patterns. Calcaneal and tibial fractures are related to rapid floor deflection [17]. These injuries have been demonstrated with validated physical models reflecting battlefield injury morphologies [18, 19]. In contrast, the analysis of torso injuries leads to the hypothesis that injuries to the pelvis, spine, torso, and head are due to a different and common loading mechanism through the seat. Rapid seat acceleration could be due to localised hull deformation at the point of seat mounting, internal vehicle dynamics secondary to hull deformation, or to global axial acceleration of the vehicle. The precise mechanism of load transfer through the seat is dependent upon vehicle and seat type, and requires further study, to help identify successful current, or design new mitigation strategies.

21.5 Biomechanics of Under Vehicle Blast Torso Injury

Injuries to the internal organs of the thorax and abdomen occur by a variety of mechanisms. An injury mechanism is "a description of the mechanical and physiological changes that result in anatomical and function damage" [20]. Understanding these mechanisms provides the basis for determining the response and tolerance for a particular form of loading. Injury is caused by deformation of biological tissues beyond a recoverable limit with damage to structure or alteration in function.

Non-penetrating injury generates force due to the inertial resistance of the body tissues and the elastic/plastic (deformation related) and viscous (rate related) compliance of body structures [20]. This force leads to deformation of tissue which may cause injury.

Deformation of tissues due to blunt loading is measured in strain, the change in any given dimension as a proportion of the original dimension. This is typically separated into tensile/compressive strain (stretching of the tissue along with elongation of length or crushing of tissue; this causes a change in size, not shape) or shear (opposing forces acting across a tissue in opposite direction; this causes a change in shape, not a change in size).

Each of these strain types may cause injury to organs including laceration (tearing of the structure) and contusion (internal injury to the structure without tearing of the exterior). Human tissue, like all biological tissue has viscoelastic properties, meaning that the tissue response is influenced by both degree of deformation, and the rate at which this loading is applied; it is a time-dependent effect. The consequence of this viscoelasticity is a reduced tolerance to a given stress when the load is applied at a high rate. Compressive strain of a soft organ can be absorbed if applied slowly but if applied quickly, the organ cannot deform and absorb the applied energy quickly enough with injury occurring prior to change of shape. These viscoelastic properties are of particular relevance with regard to blast injuries.

The tissue properties of the different torso organs vary, but in general they exhibit a viscoelastic response to loading (Chap. 4). The precise injury mechanism (that is, the mechanism by which excess strain is produced in the organ by the external loading) also varies between individual organs and systems.

21.6 Injury Tolerance

Given that most casualties sustain injuries to multiple structures, the overall response of the torso to underbody loading is more important than a detailed description of individual torso organ response to loading. From this overall response, the injury tolerance (the threshold of response over which injury is likely) can be ascertained. An understanding of this relationship is fundamental to designing adequate injury protection.

The understanding of this response is limited. Injury and death from falls provides an approximate estimate of the tolerance of the body from vertical loading. The variation in fall height and orientation in clinical data restricts the use of this data for the inference of precise tolerances. Equally, understanding injury tolerance to falls has clinical importance but prevention of these injuries relies more upon the prevention of the fall itself.

As the study of impact injury has been driven by the automotive advances, understanding the response to vertical acceleration and the requirement for protection from this form of loading have been driven by aeronautic advances [21].

Vertical accelerative loading has applications both to crash worthiness and ejection seat tolerances. Research into this domain began during WWII [21]. Eiband summarised the war time and immediate post war research and made it apparent that the tolerance to impact was dependent upon four primary factors [22]:

- The direction of the acceleration
- The magnitude of the acceleration
- The duration of the acceleration
- How the occupant body is supported during the acceleration

Eiband combined the available data at this time and developed tolerance curves for acceleration in different directions [22]. Research analogous to UBB was described as "headward" acceleration. The injury patterns most commonly described were bony spinal injuries. Human volunteer studies (using catapults creating g forces) were analysed and a tolerance limit of 16 g for up to 0.04 s described. Given the use of volunteers, this tolerance described the border of non-injury to minor injury. Tolerance to severe spinal injury was described using animal data with tolerance of 110 g for 0.002 s without injuries seen in pigs and 42 g for 0.048 s in chimpanzees [22].

The use of animal testing for defining human tolerance to injury has been described as a limitation of Eiband's work [23]. Despite this, animal models have been used for other impact scenarios (as described above) and although caution should be used when interpreting absolute values, these tests are fundamental for the understanding of injury mechanism and development of injury criteria. Animal models are of particular relevance to severe injury when human volunteers are not appropriate.

The pattern of human torso injury in response to under vehicle blast has not previously been described although similar injuries have been shown following air-crashes. The direction of occupant loading during such a crash may be complex with both frontal, side, and vertical forces. A review of crashes and resultant injuries described frontal impact as the principal force in most cases [24]. Wallace et al. also describe a particular case in which an aircraft was subject to purely vertical impact having following entanglement with power lines. In this instance, all six occupants died from internal injuries despite no evidence of external trauma. Wallace suggested that the protective features of an aircraft seat were ill equipped to prevent injuries when vertical impact was predominant [24]. This might represent a similar scenario to UBB.

Experiments which determine the tolerance of these internal organs to vertical loading cannot be performed with human volunteers and instead, with the caveats outlined above, are reliant upon animals or computational simulations. Historically, these experiments were performed using rocket powered sleds in a programme initiated by Stapp who used both human volunteers and animals [25, 26]. Animals were anaesthetised and restrained in a sitting position with backs against the floor of the sled. A stopping mechanism enabled reasonable control of both duration and magnitude of the deceleration. Stapp demonstrated visceral haemorrhage in pigs exposed to around 80 g for 30 ms in a + Gz (upwards) direction [26]. Cook and Mosely performed similar experiments upon eight black bears at varying accelerations and durations [27]. They described the occurrence of cardiac injury, mediastinum haemorrhage (without aortic rupture), along with lung haemorrhage and liver injury. Internal injuries were associated with fractures of the vertebral column and ribs. Severe internal injuries were seen in two bears who underwent accelerations of ~130 g for around 15 ms [27]. Cook and Mosely suggested traction of these structures in response to relative inertia as the cause of injury.

Kazarian investigated the use of primates for G_z loading [28]. In contrast to the rocket sleds used by other researchers, Kazarian used a drop tower rig in order to evaluate the spinal response of chimpanzees, rhesus monkeys, and baboons. Impact velocity was determined by the drop height with deceleration profile altered using a crushable aluminium honeycomb.

Although the primary aim of Kazarian's work was to describe spinal injury in response to G_z impact, he also noted internal injuries. A series of experiments using the above impact vehicle was performed upon Rhesus monkeys undergoing $+G_z$ accelerations between 25 and 900 g (with durations 2–22 ms) [29]. Injuries were observed in the liver, lungs, and heart. Kazarian related the internal organ injuries to the acceleration and time duration of the impact and showed that lung injury was the primary mode of injury before vertebral body fracture, liver injury, and heart injury.

Lung haemorrhage is also described in small animal models of impact loading [30, 31]. A recent rodent model of UBB (developed primarily to investigate brain injury) demonstrated lung haemorrhage to be the primary cause of death in animals accelerated beyond 2000 g [31]. Although this model is one of very few to reproduce pure whole body vertical acceleration without prior free fall stage, the animals themselves were positioned prone, and acceleration was therefore directed perpendicular to the axis of the spine. Furthermore, the animals were enclosed with no way to record their biomechanical response.

The response of the human torso to vertical loading has been evaluated using cadavers. Danelson et al. compared the response of postmortem human surrogates (PMHS) to Anthropomorphic Test Devices (ATD), using a whole body explosive driven accelerative loading fixture [32]. These tests were performed to show the biofidelity of these devices and investigate skeletal injury in response to UBB. As such, the cadavers were not ventilated or perfused and internal organ injury not described. The biomechanical response of the torso was noted, however, with up to 20% compression of the torso height in response to the loading. Importantly, the ability of the torso to compress is not a feature of contemporary ATDs [32].

What is the likely injury mechanism of the internal organs in response to UBB? Prediction of these injuries requires a mechanical hypothesis. Firstly, it is assumed from the clinical data presented that injuries occur in response to vertical loading applied through the seat. This load is transferred through the pelvis. Injury to the organs likely occurs as a result of two mechanisms: direct compression of the organs in the axial direction or by relative movement of the organs causing tears at points of attachments. Individual organs may be injured by either or both mechanisms depending upon anatomical factors. Compression of the liver and other abdominal organs due to displacement against each other, the diaphragm, or the lower ribs may cause laceration at point of compression. Tensile strain at the ligamentous and peritoneal attachments of the organs due to inertial effects may cause injuries to these tethering points. Further shear may be generated at the vascular insertions given that the IVC and hepatic veins are fixed to the liver. This mechanism as it applies to the liver is shown in Fig. 21.1.

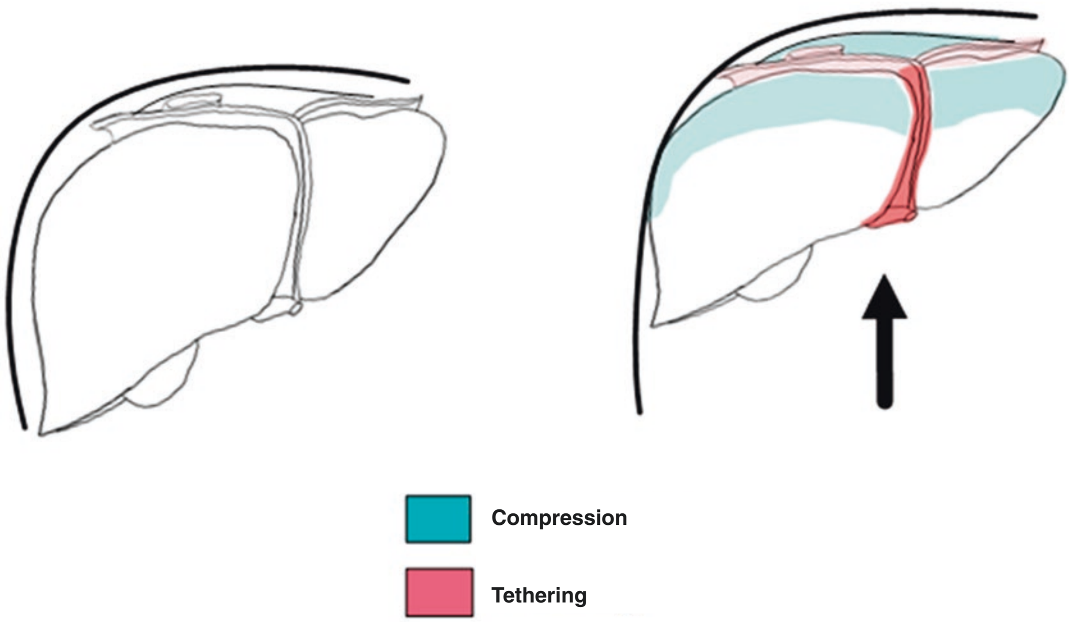

Fig. 21.1 Proposed injury mechanism for the liver following UBB loading. Cranial displacement of the viscera causes direction compression against the diaphragm and adjacent organs. Tethering of the organ by peritoneal attachments (particularly those orientated parallel to the direction of loading) would result in shear strain at these points

This craniad displacement of the abdominal contents and diaphragm would cause direct compression of the lungs (both at point of impact and at distant interfaces). This direct compression may cause parenchymal injury similar to that described as primary blast lung in mounted blast (Singleton et al., 2013). Gross disruption of the lungs, including laceration, and injuries to the great vessels and mediastinum are more likely caused by relative movement of these organs, with tethering most pronounced around the pulmonary ligament (Fig. 21.2).

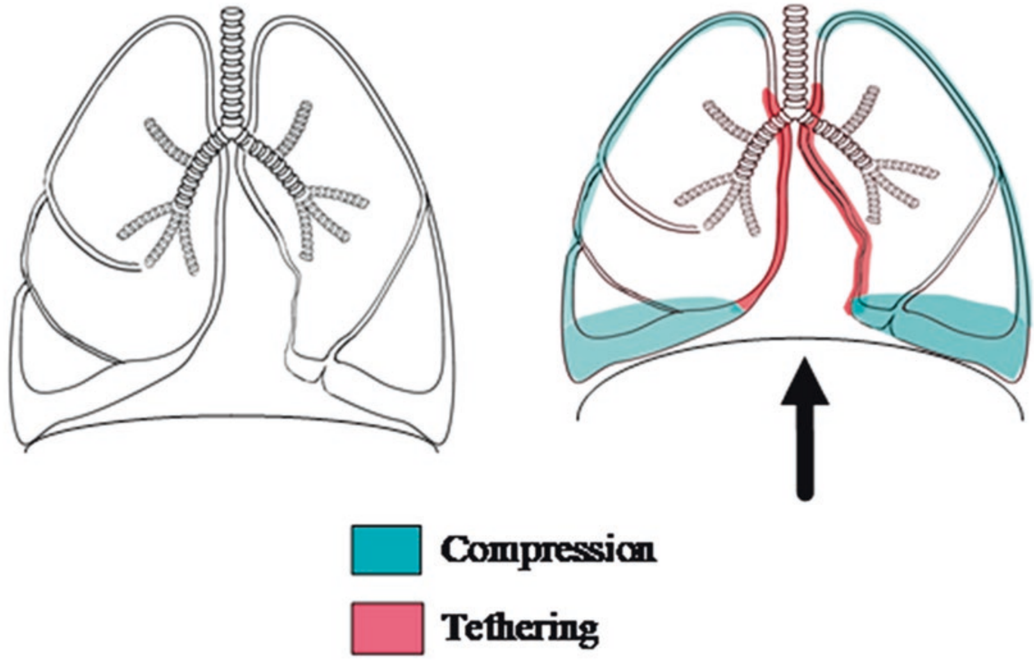

Fig. 21.2 Proposed injury mechanism for the lungs following UBB loading. Craniad displacement of the viscera and diaphragm compresses the lungs and generates shear at the points of tethering

Injuries to the heart and aorta could also arise from craniad displacement of the abdominal contents causing direct compression of the heart. As the heart is displaced vertically, the relatively mobile aortic arch is also displaced with strain generated at the aortic tetherings (including vessels) and the commencement of the retro-pleural descending aorta. The overall effect would resemble the shovelling mechanism proposed by Voigt and Wilfert [33] and is depicted in Fig. 21.3.

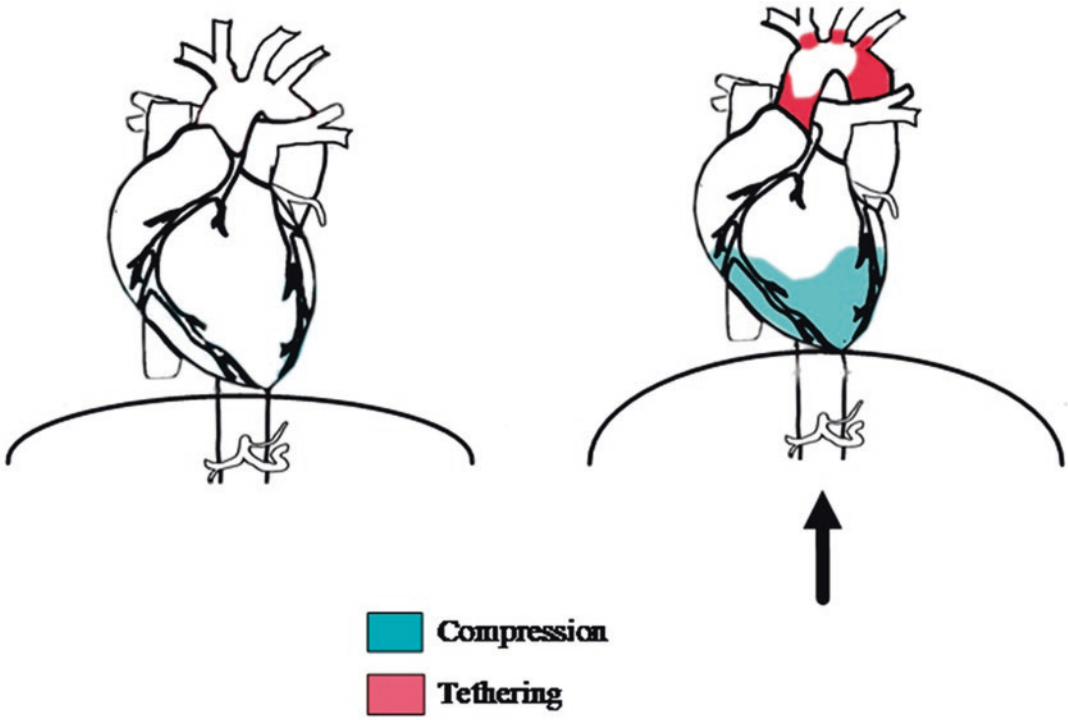

Fig. 21.3 Proposed injury mechanism of the heart and aorta following UBB loading. The pericardium is not shown

What is clear from these proposed mechanisms is that axial displacement of the organs (either together or separately) is required. As there are rate-dependent response effects due to the different viscoelastic properties of the organs, so the rate of loading as well as the magnitude of displacement of the impact will affect the outcomes.

Previous research is not able to satisfactorily predict injury in response to this loading. For example, mechanical analogues, whilst explaining the injuries, are unable to predict injury given the need to make broad assumptions about the non-linear nature of the soft tissues. One of these analogues is the dynamic response index (DRI) that has been shown to be inaccurate for prediction of spinal injury in UBB [34]. This discrepancy is likely because the DRI assumes a single spinal spring and does not account for changes at different levels. The boundary conditions of the soft tissue organs relative to one another are even more complex, and the relationship of the input loading, physical response of the torso, and resultant injury have not been established. An important aim of injury biomechanics research is not only to describe this mechanism but to accurately predict its occurrence and therefore allow the implementation of mitigation systems. Although a wide variety of animal, human volunteer, and cadaveric studies are used, the common goal by researchers is to correlate injury outcomes to some form of biomechanical parameter. These parameters, injury criteria, may then be used to predict the risk of a particular injury. Risk curves may be generated for particular injuries and injury criteria using survival analysis or non-linear regression approaches [35].

Injury criteria are used to predict probability of a particular injury in response to a particular impact scenario. The primary application of injury criteria for UBB is in the development of protective materials and vehicles. Injury criteria allow standardisation of testing methodologies and the quantification of the protection offered by a platform in response to UBB insults.

Internationally, this information has been collated by the Research and Technology Organisation of the North Atlantic Treaty Organisation (NATO). The particular NATO group tasked with the investigation of UBB has been the Human Factors and Medicine (HFM) panel 148 [36]. The goal of HFM 148 has been to "*analyse injury loading mechanisms, investigate injury assessment criteria, define test methods and measurement tools to assess vehicles (and protection systems) against the mine and IED threat. Injury criteria were defined, and the pass/fail (tolerance) levels established for the body regions are considered to represent low risk of life-threatening and disabling injuries*" [36].

The group explores available injury criteria for all bodily regions. In each case, these criteria are applied for use with one of several ATDs for both live fire and simulation testing. Tolerances are described according to each criterion such that each test results in a pass or fail for each criterion.

As discussed above, the criteria for torso injury available to the group do not include any which have been designed or validated for vertical loading. The recommendation of HFM 148 for thoracic injury criteria is the use of Compression Criterion (C) in conjunction with the Viscous Criterion (VC) for both frontal/rear/vertical loadings (with the Hybrid III ATD; described in Chap. 34) and lateral loading (with the ES-2re ATD; described in Chap. 34).

Threshold values recommended for pass/fail standards by the group are shown in Table 21.2.

Although frontal, rear, and lateral loading may play some part in injury following UBB (follow-

Table 21.2 Tolerance values for thoracic loading in response to UBB testing as recommended by NATO Research and Technology Organisation [36]

Loading	ATD	Compression criteria	VC
Frontal/Rear/Vertical	Hybrid III	30 mm (10% risk of AIS3+)	0.7 m/s) (10% risk of AIS 4+)
Lateral	ES-2re	28 mm (10% risk of AIS 3+)	0.58 m/s (10% risk of AIS 3+)

ing rolling or tipping of the vehicle), the physics of UBB suggests that the primary loading pathway is vertical. Similarly, the severe torso injuries may, in part, be caused by anterior-posterior or lateral impact but the injury associations suggest vertical loading through the seat. Previous work has demonstrated that torso injuries are observed following pure vertical loading in animals and do not require frontal chest compression. Measuring conventional compression and VC though the Hybrid III is facilitated by chest deflection sensors on the ATD [36]. These ATDs do not currently have the ability to undergo axial torso compression.

The limitations of these ATDs are further evidenced by the HFM 148 recommendations for abdominal injury risk. The group recommends the use Abdominal Peak Force (APF) measured with the ES-2re ATD for prediction of risk in cases of lateral loading. This criterion was based on localised pendulum tests causing liver injury in the anterior direction in PMHS [37, 38].

There are important limitations with the current injury criteria recommendations. Firstly, APF does not have any rate sensitivity, which has been described as an important factor in abdominal injuries [39]. Secondly, although there may (for some blast events) be a degree of lateral loading, vertical loading is the likely to predominate. The limitations to the criteria are in part due to currently used ATDs. Due to lack of measurement capabilities in the Hybrid III ATD, there is actually no recommended abdominal injury criterion for frontal/rear/vertical loadings [36].

Injury criteria should be based upon both predominant physical loading (high-rate vertical acceleration), the known material and anatomical properties of the tissues (rate dependent with tethering structures aligned in the vertical direction), and a proposed injury mechanism. The injury mechanisms postulated above take these first points into account and are an appropriate mechanical approach. They fit with previously observed measures of vertical torso deformation (as described in both anaesthetised animals and human cadavers).

Current UBB test methodology does not address these points and as a consequence may not be accurately predicting the risk of torso injury in response to UBB. Recent work has developed an animal model of UBB that seeks to address this weakness in the literature, yet this has not been fully validated to date [40].

21.7 The Future

Torso injury is a major contributor to the morbidity and mortality caused by under vehicle blast. Explosive devices, improvised or otherwise, are likely to feature in future conflicts. Preventing or mitigating torso injuries will require the development of novel materials, armour, vehicles, and strategies. These developments will only be possible if the relationship between explosive loading, the response of the torso, and likelihood of injury can be investigated. Novel torso injury criteria relevant to the rate of loading and direction of under vehicle blast must be developed and validated. Vertical compression of the torso as described during cadaveric testing may be a suitable candidate but has not been tested within cadaveric studies as an injury criterion [32]. Initial small animal work has demonstrated abdominal and chest injuries in response to high-rate axial loading through the pelvis for which rate-dependent measures of vertical compression were the most accurate predictors [41].

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Blast Traumatic Brain Injury

22

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Abstract

The aim of this chapter is to describe the background of blast brain injury from minor traumatic brain injury to moderate and severe injuries. The literature shows that there is an increasing burden of blast brain injury in military casualties due to a combination of better protection of other body regions, improved survivability, and a greater use of blast as the mechanism of injury in recent asymmetric conflicts. Management of these injuries is difficult and includes the use of surgery, medication and cognitive behavioural therapy. Current research is focused on the use of advanced pre-hospital interventions, better mitigation of these injuries, and identifying good non-blast correlates of blast brain injury. For minor blast brain injury, research into better diagnosis is focusing on the use of advanced imaging techniques, such as magnetoencephalography, diffusion tensor imaging and susceptibility weighted imaging.

22.1 Introduction and Background

Blast traumatic brain injury (bTBI) primarily occurs during military conflict or civilian unrest. It has been prevalent throughout late twentieth century and early twenty-first century conflicts, including operations in Northern Ireland, the Falklands and Vietnam where evidence suggests between 16 and 21% of personnel suffered a traumatic brain injury (TBI) [1]. Taking into account that blast was responsible for up to 81% of injuries in Iraq and Afghanistan [2–5], bTBI has been described as a common injury for combat troops [6].

bTBI is complex. The mechanism of injury can involve varying components of primary blast (blast pressure wave travelling through the brain), secondary blast (penetrating injury), tertiary blast (acceleration-deceleration effects) and quaternary blast (thermal injuries to the head and scalp) which cannot necessarily be distilled as separate injuries or patterns of injury and makes mitigation difficult to achieve.

The recent Iraq and Afghanistan conflicts resulted in large numbers of blast injuries in US and UK troops. Traumatic brain injury due to

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blast has a long-term chronic effect on quality of life and cognitive ability; it can contribute to an increased risk of neurodegeneration as well as astroglial scarring [7] and mental health disorders.

Due to the heterogeneity of injury, this chapter will divide the clinical problem, current research, management and the future focus into two sections: mild bTBI and moderate to severe bTBI.

22.2 The Clinical Problem

Blast may be causing more cases of bTBI than previously known. Head injuries from blast may be significantly under diagnosed given the limited clinical data, clinical features similar to PTSD and no definitive inclusion criteria. Research is focused on retrospective data from the Joint Theatre and Trauma Registry (JTTR) rather than prospective clinical trials. Diagnosis and identification of those patients suffering the effects of blast injury, particularly those with mild bTBI, are reliant on the self-presentation of veterans and to date, the pathology of bTBI has not been documented or categorised [8].

22.2.1 Minor Blast Traumatic Brain Injury

Over 90% of UK TBI emergency department attendances are for patients with minor blast traumatic brain injury (mTBI) [9].

Traumatic brain injury is traditionally classified into mild, moderate or severe based on the best Glasgow Coma Score (GCS) at the time of presentation. If a patient has a GCS of 13–15, then they are said to have a mTBI. In clinical practice, patients usually present with symptoms of confusion and agitation [10]. In recent years, there has been a move to use the Mayo classification system because of its focus on a broader range of symptoms as well as the fact that it enables classification in situations where there is limited initial clinical information [11]. Using this classification system, a patient who has suffered a head injury with either no loss of consciousness (LOC) or less than 30 min of LOC and less than 4 h of post-traumatic amnesia with no intracranial findings on computerised tomography (CT) scanning is said to have mTBI [12].

Historically the popular view was that the symptoms following mTBI were transitory, lasting minutes to hours. Recently, it has become clear that even once confusion resolves, a constellation of symptoms including headache, dizziness, lethargy, irritability, reduced concentration, sleep disturbance, memory impairment, anxiety, sensitivity to noise and light, blurred vision and depression can continue for several months afterwards [10]. There also appears to be an association between mTBI, increased mortality and higher rates of neurodegenerative conditions such as Alzheimer's disease and, most recently, a study has demonstrated higher rates of neurodegenerative disease in soccer players with repeated sub-clinical TBI as a result of heading the football [13]. It is clear that the current understanding of the total burden of pathology from mTBI is incomplete.

The recent conflicts in Iraq and Afghanistan saw widespread use of improvised explosive devices [13]. As the conflicts matured, personal protective equipment improved, evolving vehicle mitigation developments were introduced, and military doctors saw an increasing number of survivors of explosions [14]. Many of these soldiers had symptoms of brain injury despite not having had a recognised head injury. Further research has shown that the overpressure wave generated in a blast is itself capable of causing brain injury [15]. There is increasing recognition that smaller explosions such as those generated by building entry devices used by "breachers", and grenades are capable of causing mild bTBI [16]. Small explosions result in "sub-concussive" exposure, whereas larger ones can cause symptoms similar to concussion. Some soldiers have regular sub-concussive exposure as a result of their occupation, and in recent times, there has been more interest in the long-term clinical consequences.

22.2.2 Moderate to Severe Blast Traumatic Brain Injury

It has not been identified why bTBI is so prevalent in the military cohort of both survivors and fatalities, however, it is thought to be because the head and neck region is approximately 12% of all exposed surface areas on a warfare soldier [17]. Therefore, even where all personal protective equipment is worn, the neck and head remain highly exposed. Brain tissue and vasculature itself are susceptible to blast related injuries due to its delicate complex system surrounded by cerebrospinal fluid which can be affected by strain and rising intracranial pressure [18].

Previous UK military research around head, neck and spine injuries has predominantly involved review of the JTTR [19–21] rather than review of computed tomography, including computed tomography postmortems (CTPMs). JTTR information can lack granularity and in places accuracy. For example, many head injuries from isolated facial injury through to complex brain injury are often coded similarly. Review of such imaging modalities may provide the ability to categorise injuries more accurately in order to define the severity of head injuries.

Mounted blast was identified as a common mechanism of injury in recent UK commitments including those in Northern Ireland, Iraq and Afghanistan [22, 23]. Despite improved vehicle design mounted blast injuries continued to be seen [24]. Although the understanding of blast mechanisms has improved, there is a lack of understanding of the mechanism for brain injury, particularly for mounted blast.

Dismounted bTBI differs to those of mounted blast as there are more patients who suffer penetrating blast without the "protection" of a vehicle from energised fragments [25]. Research using the JTTR has suggested that dismounted IEDs are lower risk for head injuries due to the different mechanisms of injury [26]. Mounted blast patients are in closer proximity to hard surfaces—such as the walls and ceiling of vehicles that post an increased risk of direct head impact as well as more intensely pressure air [25].

"Future unexpected survivors" are a key area to focus on to develop novel treatment and mitigation strategies [24]. For example, research from recent conflicts has proposed that brain haemorrhage in dismounted patients could be amenable to pre-hospital intervention in two-thirds of the injuries seen [24]. Research such as this identifies areas for future focus and allows the development of prospective clinical studies in future conflicts. In the meantime, civilian surrogates should be identified that mimic or reflect the injuries seen in military personnel. One hypothesis is that acceleration-deceleration injuries such as road traffic collisions may provide good surrogates for blast brain injury [27] although current research is being undertaken to investigate this further.

22.3 Current Research/ Management

22.3.1 Minor Blast Traumatic Brain Injury

The current management of mTBI depends on the phase of the injury, whether acute, medium or long-term and the symptoms exhibited. In the acute setting, the focus of care is to rule out more severe injury and treat any concurrent injuries. National guidelines exist to help clinicians decide when patients will need radiological imaging of the brain based on their clinical presentation, past medical history and current medication use [28]. CT is the modality of choice to exclude moderate and severe TBI because it is quick, widely available and is sensitive to acute blood and fractures.

For less severe injury, acute symptoms of mTBI include confusion and disequilibrium, which can make individuals vulnerable to further injury. Initial interventions, therefore, include removing the individual from the risk of additional harm and, as in organised sport, there are guidelines to determine when an individual should be allowed to return to duty after mTBI. In most, an individual must be symptom-free at rest as well as during moderate physical exertion and have normal cognition and memory on testing before being allowed to return to the field of play or duty. Recently, it has become apparent that autonomic dysfunction, a condition in which the autonomic nervous system fails and therefore the usual functioning of organs is impaired, may be the most sensitive physical sign of injury in mTBI. Therefore, some groups are using the return to regular autonomic activity as a marker of recovery, when the heart and respiration rates return to baseline parameters appropriate to physical activity. Other symptoms of autonomic dysfunction can include impairment of the heart, bladder, intestines, sweat glands, pupils and blood vessels. Autonomic dysfunction can also cause dizziness and fainting [29]. Given that two-thirds of patients will recover spontaneously within 6 months, education about the symptoms and natural history of mTBI are essential to reassure patients and their carers [30].

In the medium term, patients may experience cognitive slowing and memory impairment, headache, vertigo and psychiatric symptoms. Cognitive slowing and memory impairment are treated with catecholaminergic and cholinergic medication such as methylphenidate or amantadine as well as cognitive rehabilitation. In the short term, headache is often caused by soft tissue trauma and is treated with a combination of paracetamol, non-steroidal anti-inflammatories (NSAIDs) and opiates. These drugs should be weaned rapidly in the first few weeks following the injury [31]. One cohort study has found that up to 50% of patients with mTBI will go on to develop persistent headaches [32]. Symptomatic migraine should be treated with NSAIDs and triptans. Patients who have frequent migraine should be offered propranolol or amitriptyline as prophylaxis. Mild TBI can exacerbate pre-existing psychiatric symptoms and can cause new symptoms. Depression is common and is treated with a combination of cognitive behavioural therapy and medication (with selective serotonin reuptake inhibitors and tricyclic anti-depressants). If further imaging is necessary at this stage, then magnetic resonance imaging scanning (using susceptibility weighted imaging and diffusion tensor imaging) is most sensitive for microstructural injury and helps to quantify the location and extent of the damage. Other modalities and tools such as biomarkers and magnetoencephalography (a neuroimaging technique for mapping brain activity using magnetic fields) are being investigated [33].

22.3.2 Moderate to Severe Blast Traumatic Brain Injury

A major problem with the classification of moderate-severe TBI in both civilian and military injuries is that classifications used within the cohort datasets are lacking utility. Both civilian and military datasets use the injury severity score (ISS). The ISS is the most severe Abbreviated Injury Scores (AIS) from the top three body regions. In this scoring system, body regions cannot be repeated if they are injured twice [34]. The New Injury Severity Score (NISS) has been developed to address this weakness [35]. A novel scoring system, the Society of British Neurosurgeons (SBNS) brain injury classification quantifies the pathophysiology of injury by grading the injuries of extradural, subdural, contusions, diffuse axonal injury, hypoxia/swelling, skull fracture from zero to three (in-press). This scoring system has been used to analyse the injuries of those military personnel from Iraq and Afghanistan. The benefits of doing so have allowed comparison with civilian cohorts, the identification of injury patterns and the hypothesis of potential mechanisms of injury in blast.

Mounted blast injuries cause significant morbidity and mortality in the military cohort. Recent work using computed tomography from casualties or CTPMs showed that in addition to direct frontal, temporal and parietal impact injuries to the skull, axial loading injur ies and flexion injuries to the spine, a significant number of personnel had subarachnoid haemorrhage in the posterior fossa, especially in the fourth ventricle [36]. This is an unusual pattern of injury and is not typically seen in civilian non-blast injury. It supports previous work [37] that found increased injury to the structures in the posterior fossa as a result of blast.

Dismounted injuries also expose the head and body to high energy penetrating injuries. However, many of the patterns of injury remain the same in the non-survivor cohort as those of the mounted injuries. Specific injuries seen in dismounted patients include blood in the fourth and lateral ventricles and cleavage through mandible and face, as well as fractures of facial bones with no intracranial bleeding and penetrating injuries from energised fragments.

Injuries seen in survivors of both mounted and dismounted cohorts are similar to those seen in civilians. For example, contusions (bruise of the brain tissue), subarachnoid haemorrhages (bleeding on the surface of the brain) and coup-contrecoup (an injury that occurs both at the site of trauma and on the opposite side of the brain) injuries. Open penetrating injuries with little surrounding haemorrhage are also seen in this cohort.

Although further work needs to be undertaken to investigate patterns of blast brain injury, these preliminary findings suggest links between mechanism and patterns of injury.

22.4 Future Focus

22.4.1 Minor Blast Traumatic Brain Injury

The breadth of research into minor bTBI is expanding. There is focus on diagnostics, including imaging and biomarkers to identifying those patients who have suffered bTBI in order to direct treatment and to protect from further exposure. There is also a drive to be able to distinguish between those patients suffering post-traumatic stress disorder and mTBI and advanced imaging is a tantalising project in this regard (for example, magnetoencephalography, MRI diffusion tensor imaging and susceptibility weighted imaging). Future work will also focus on long-term outcomes of those suffering mTBI and multiple mTBI. Animal and computational models can further assist the research into the mechanism behind blast injury.

22.4.2 Moderate to Severe Blast Traumatic Brain Injury

Future work on moderate to severe bTBI should continue to seek appropriate comparators for this injury in a time when bTBI is less common. However, where these do occur, detailed incident analysis of cases of blast injury, linked with medical outcomes and advanced imaging studies are also required to further the understanding of bTBI and so enable research on pre-hospital interventions as well as mitigation through the use of personal protective equipment, blast dosimetry or in vehicle protection.

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Heterotopic Ossification After Blast Injury

23

Neil M. Eisenstein

Abstract

Blast injury is uniquely potent at causing heterotopic ossification (where bone forms in abnormal anatomical locations), particularly in the context of military wounding and lower limb amputation. Heterotopic ossification after blast injury has been a major problem for survivors of the conflicts in Afghanistan and Iraq. Research activity directed towards understanding it, preventing it and treating it has increased significantly in the last decade. This condition can cause pain, restrict movement and erode through the skin, preventing the use of prosthetic limbs. There is no existing method to prevent its occurrence and surgery to remove it is often required, leading to morbidity and rehabilitation challenges. Current research in this area is focused on understanding the mechanism of formation, development of novel prophylaxes, early detection/risk stratification and development of a large animal model. The aim of this chapter is to summarise the current state of our understanding of blast-induced heterotopic ossification and the direction of travel that researchers are taking in this field.

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23.1 Introduction/Background

Heterotopic Ossification Definition: *Bone formation in an abnormal anatomical location. From the Greek Hetero (other, different) + Tópos (place).*

23.1.1 Background

Before the post-9/11 wars, heterotopic ossification (HO) was seen as a rare complication of major orthopaedic surgery and was of limited interest to the military medical community. However, the specific circumstances and patterns of wounding (i.e. amputations secondary to blast injury) seen in these conflicts have led to a huge increase in the prevalence and severity of this condition. Indeed, HO has become a problem of such magnitude that military surgeons have described it as an 'epidemic' and the single

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biggest barrier to the functional restoration and rehabilitation of combat amputees [1].

Interestingly, although HO poses a particularly difficult challenge currently, it has been noted as a consequence of combat wounds in previous wars including the American civil war [2]. A most eloquent description of this condition in the historical literature comes from Col E G Brackett, who described it in his combat amputee patients from the First World War:

"Excessive terminal bone production in … stumps was the rule. The most common form was an irregular mushrooming with a tendency to spurs on the inner aspect of the femur. Occasionally sharp exostoses were seen. These often were sharp enough and long enough to cause sufficient pain to warrant their removal." [3].

This historical description of HO is remarkably similar to that seen in the modern blast-injured combat casualty (Fig. 23.1).

Fig. 23.1 3D computed tomographic reconstruction of the pelvis and residual femora of a combat-injured patient. White arrows = heterotopic ossification. Reproduced from Eisenstein et al. [4] with permission from John Wiley and Sons

23.1.2 HO After Blast Injury

While any major traumatic injury can predispose patients to the formation of HO (e.g. gunshot wounds, high energy blunt trauma), there is a growing body of evidence to suggest that blast injury is uniquely potent at generating ectopic bone. X-ray-detectable HO develops in the limbs of approximately 64% of combat-injured patients (extremity injury of all causes) [5, 6]. After blast-related amputation however this proportion is significantly higher, with approximately 80% of patients demonstrating moderate or severe HO in their residual stumps. These figures are much higher than those seen after non-blast civilian trauma such as traumatic amputation (23%), brain injury (37%), distal humerus fractures (9%) and burns (0.15%) [7–9].

There are multiple cellular-, organ- and organism-level effects of blast that contribute to the potency in causing HO (see Chap. 6). Additionally, blast was a common mechanism of injury in Iraq and Afghanistan, which may explain some of the increase in the prevalence of HO observed. However, another feature of these conflicts that may also have contributed is increased survivorship [10]. It is hypothesised that, in previous conflicts, injuries of the magnitude required to generate clinically significant volumes of HO would have been almost invariably fatal and the consequence of increased survivorship amongst the most seriously injured casualties, particularly multiple blast-amputees, may be that conditions such as HO have been 'revealed'. The reasons for this increased survivorship have been explored elsewhere and are thought to include rapid casualty evacuation, improved personal protection and advances in the technical and organisational aspects of resuscitative medicine and surgery [11].

23.1.3 Mechanism of Formation

Acquired HO can form after traumatic injury (blast, gunshot wounds, fragmentation injuries, etc.), central nervous system injury and burns (inter alia). It is not at all clear whether the biological mechanism of formation of HO is the same in all of these situations. Secondary factors such as wound infection and the use of topical negative pressure dressings have been suggested as contributing to the formation of HO but their role has not been determined conclusively [6, 12, 13]. In general, our understanding of the biological mechanisms and pathways that cause blast-related HO is developing but not yet complete. It is clear from recent clinical and biochemical studies that dysregulation of the organism-level inflammatory response to trauma plays a fundamental role [14–16]. Further evidence for the role of inflammation comes from the correlation in time-course of the acute and chronic inflammatory changes post injury and the natural history of formation of HO [17, 18]. Mineral deposition has been shown to commence within days of an initiating insult with clinically relevant HO formation taking weeks or months to become apparent. There then usually follows a period of several months of progression and maturation after which the condition 'burns out' and is unlikely to progress further.

One theory that deserves mention for its lack of supporting evidence is the 'seeding hypothesis'. This is where soft tissues are seeded with small fragments of bone or other material that goes on to generate or stimulate new bone formation. While this may make some intuitive sense, there are several reasons why most researchers do not support it. First, meticulous care is given to removing fragments of bone or other contamination during the multiple debridements and irrigation that these patients undergo. It is possible that 'seeding' particles may be driven deep into tissue planes that are not debrided but this does not fit the macroscopic distribution of HO seen clinically. Second, the morphology of HO growth in the patient's limbs appears to be independent of the soft tissue wounding pattern. Finally, and perhaps most convincingly, HO has been demonstrated in cases where the bone has not been fractured (therefore, no bone fragment could have been generated or seeded elsewhere) and in others where the patient has sustained only closed injuries (therefore, no external material could have been seeded into the body).

23.1.4 Cellular and Genetic Mechanisms

Considerable work has been undertaken to identify the bone-forming cells responsible for HO. However, a highly complex picture is emerging from the data and no single cell type can be said to be the sole culprit. Instead, multiple cell types and lineages form complex networks that lead to migration and differentiation in other populations [19]. There is evidence to support the hypothesis that a muscle-resident and/or vascular endothelial-resident population of satellite/progenitor cells differentiate into osteoblast-type cells and these deposit calcium hydroxyapatite mineral into a cartilaginous scaffold formed by activated chondrocytes. Adipocytes and vascular endothelial cells also play key paracrine signalling roles and all of the cellular activity is modified by the mechanical and biochemical environment of the tissues. Osteogenic and chondrogenic gene transcripts have been shown to be up-regulated in the wounds of combat-injured patients [20]. This provides further support for the hypothesis that bone formation in HO follows the endochondral ossification model rather than the intramembranous model. A summary of the mechanism of formation of traumatic HO is demonstrated diagrammatically in Fig. 23.2, which gives the reader some appreciation of the complexity involved.

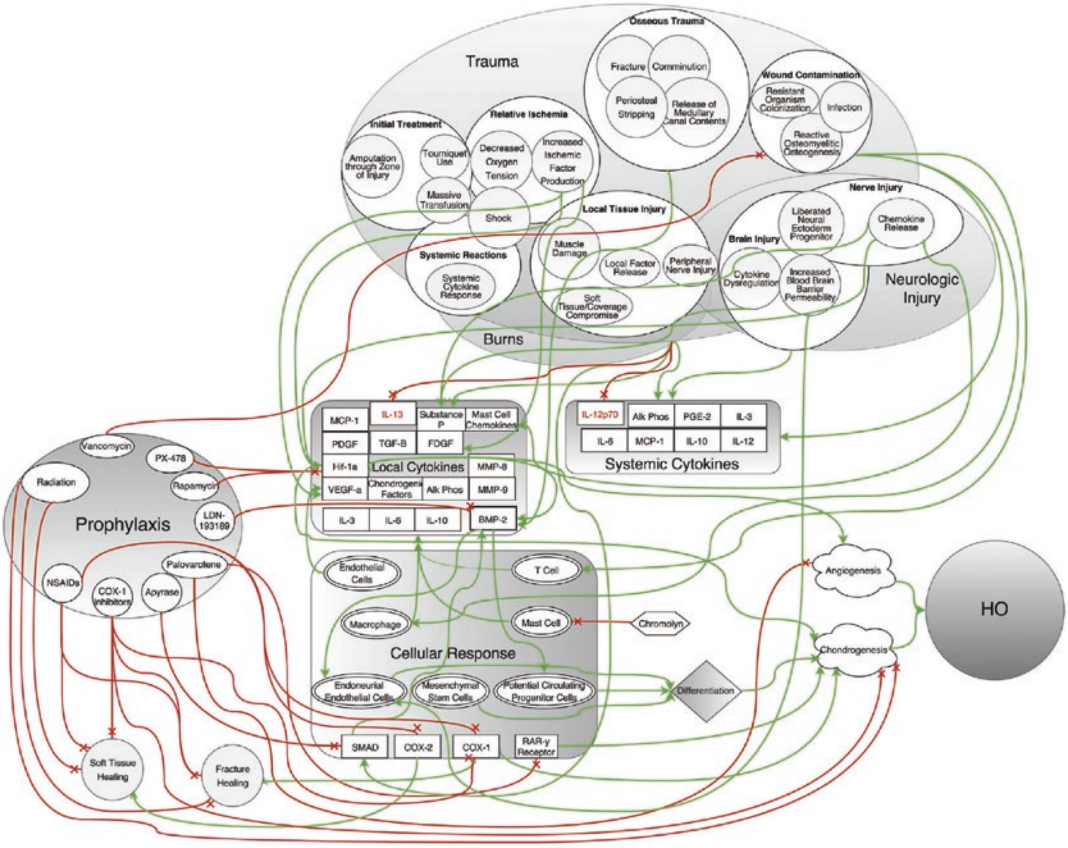

Fig. 23.2 Diagrammatic representation of the role of trauma and associated wound and systemic factors on the formation of heterotopic ossification. Prophylactic measures discussed in this article are shown and related to their mechanism of action in inhibition of ectopic bone. Reproduced from Hoyt et al. [21] with permission from Elsevier

23.2 The Clinical Problem and Current Management

23.2.1 Clinical Burden

HO causes clinical problems for patients through the presence of hard, bony, tissue in abnormal anatomical locations. When it forms close to the skin or near a load-bearing area in a limb residuum after amputation it can lead to excess shear and point loading of the skin. This causes ulceration and pain and can prevent the use of traditional socket-fit limb prostheses. If HO forms within muscle bodies or near joints it can impair movement leading to reduced function due to pain or by acting as a physical barrier. HO can also form a continuous bridge between bones leading to complete ankylosis.

Non-surgical methods of managing stump ulceration and pain include socket modification to offload affected areas, physiotherapy, analgesia and rest [1]. However, these are not permanent treatments and do nothing to prevent the HO forming or progressing.

The patient experience of HO is summarised succinctly by Evriviades et al. as follows:

> The development of HO can occur in the soft tissues very early and is sometimes found even before primary healing occurs. However, in the majority, it occurs after a number of months. The patient will typically have had an IED [improvised explosive device] injury and sustained bilateral traumatic above-knee amputations (often with other signifi-

cant injuries), and will have initially done well with limb fitting and mobilization. They may develop increasing pain in their stumps and can often feel a hard lump or spike within the stump. In severe cases, the bone has actually eroded through the soft tissues. This may prevent them from mobilizing and poses a significant challenge to the prosthetics team [13].

23.2.2 Barrier to Rehabilitation

The timing of formation of HO is particularly unfortunate for severely wounded patients (particularly those who have suffered blast amputation) because they often take several weeks to recover from their wounds sufficiently to begin mobilisation. It is at this point that HO will become apparent. If conservative measures fail to relieve symptoms sufficiently then they may require surgery. Not only is excision surgery technically demanding and comes with the risk of damage to the neurovascular structures encased in HO but, crucially, it is followed by a long period (i.e. 6 weeks) of non-weight bearing for the patient. During this time they necessarily cannot undertake the same rehabilitation activity as those who have not undergone HO excision surgery. Unfortunately, these patients may require multiple operations on their stumps and each time they will need to be non-weight-bearing for a further 6 weeks. Ultimately, their rehabilitation is slowed and their ultimate functional outcome may be worse than those without HO.

23.2.3 Lack of Prophylaxis

In civilian medical practice, there are several methods of preventing HO when performing surgery at higher risk of causing HO. These include radiotherapy to the affected area, bisphosphonates and high-dose non-steroidal anti-inflammatory drugs (NSAIDS). However, after blast amputation or other significant combat injuries likely to generate HO, none of these are appropriate. Not only is there no evidence of efficacy after combat injury, but there are also serious logistical and medical barriers to their use. NSAIDS are to be avoided in catabolic trauma patients due to potential nephrotoxic effects, gastric ulceration and the possibility of delayed fracture healing. Radiotherapy can cause delayed wound-healing and it is highly questionable as to whether it could (or should) be delivered in the theatre of operations soon enough after injury to make a meaningful difference.

23.2.4 Surgical Management

If conservative measures fail to control symptoms, operative management may be required. In one series of combat-injured patients with HO, 19% required surgical excision [6]. In this series, the mean interval between injury and surgery was 8.2 months. The timing of surgery for blast-related HO requires careful consideration of many factors. If removed too soon, there is a risk of recurrence, possibly because the underlying bone-generating process is still continuing and the additional inflammation caused by the surgery may potentiate further disease. However, if left too late, the patient has to live with debilitating symptoms of HO for longer than is necessary and may not be able to rehabilitate effectively. It is important to note that there is limited published evidence to inform the timing of surgery and it is usually a matter of multidisciplinary discussion. Preoperative planning is essential in order to minimise the risk of complications. Cross-sectional imaging (usually CT and or MRI) is obtained to define the relationship between the HO and neurovascular structures and a plan is made regarding the necessity for soft tissue coverage including skin grafting or flaps. The aim of surgery is usually limited excision of the specific volume of HO causing the patient's symptoms rather than attempting full clearance of ectopic bone.

23.3 Current Research

Research into blast-related HO is broadly divided into four areas: mechanistic understanding (discussed above), novel preventative therapies, early detection/risk stratification and animal modelling. A comprehensive review of the literature in each of these areas is beyond the scope of this chapter but the following are included to provide the reader with a general understanding of the direction of current research.

23.3.1 Novel Preventative Therapies

As our understanding of the formation mechanism of blast-related HO has developed, it has become possible to attempt to inhibit this process in a targeted and rational way. The most experimentally established prophylactic strategy is the use of retinoic acid receptor gamma (RARγ) agonism to inhibit chondrogenesis. Having established that HO develops through endochondral ossification, researchers explored whether preventing the cartilaginous scaffold might prevent ectopic bone formation [22]. A selective RARγ agonist (palovarotene) was repurposed from previous uses and, at the time of writing, was in phase 2 clinical trials (NCT02279095) as a treatment for a genetic form of HO known as fibrodysplasia ossificans progressiva (FOP). As blast-induced HO is understood to develop via endochondral ossification, it is hoped that palovarotene may demonstrate efficacy in this setting also.

Several other agents are also under investigation and many, like palovarotene, have been repurposed. One such agent is the antibiotic echinomycin; although this has so far been tested in animals only [23]. Rather than acting as an antibiotic, the mechanism of action, in this case, is through inhibition of hypoxia-induced factor 1-α (HIF-1-α), which is necessary for chondrogenic differentiation of mesenchymal stem/stromal cells in HO. Another agent is apyrase, which interferes with bone-morphogenic protein (BMP) activity through SMAD (a family of signalling proteins) signalling inhibition [24]. This agent has also been tested in animals only.

23.3.2 Risk Stratification and Early Detection

Although there are no current treatments that have been shown to be safe and effective in preventing HO after blast injury, significant work is going into developing one (see above). If such prophylaxis were to become available, it would then become advantageous to be able to identify in advance which patients were at the highest risk of developing HO so that they could be treated and monitored more intensively than the wider population at risk. Various approaches have been proposed but one that has shown early promise is the analysis of cytokine levels and osteogenic gene transcripts in serum and wound effluent at the time of first debridement. This work [21] has significant potential as demonstrated by a sensitivity of 100% for identifying those who would go on to develop HO.

Another interesting approach is the use of Raman spectroscopy to interrogate the tissues, looking for calcium phosphate compounds including calcium hydroxyapatite that indicate the early development of HO. This has been shown to be effective at detecting mineralisation in human tissues excised during debridement and can even be combined with a fibre optic system to allow percutaneous readings.

23.3.3 Animal Modelling

Multiple animal models have been used in HO research. The general pattern for these models has been a wild type or genetically modified rodent plus a physiological insult or injury such as Achilles tenotomy, burn, fracture, arthroplasty and implantation/injection of exogenous material. While some of these models are highly reliable, their fidelity in replicating the mechanisms involved in blast trauma is questionable. There have been several promising advancements in animal modelling for blast-related HO in recent years. An early rodent blast model was developed using an underwater explosive charge but this suffered from poor reliability and high mortality [25]. More recently, several advanced rodent blast models have been developed that have shown promise due to their high fidelity and reliability [26, 27]. These models allow independent variation of blast injury, muscle crush, femoral fracture and amputation through the zone of injury to investigate the relative contribution to HO development. One model, in particular, has been thoroughly characterised and has even been used to demonstrate the efficacy of Palovarotene in suppressing blast-related HO and the utility of Raman spectroscopy in detecting mineralisation before it is radiographically apparent [28]. One of the key benefits of rodent models compared with large animal models is the relatively short timescales involved and the large numbers possible in each experimental arm leading to statistical robustness. However, no matter how faithful the mechanism of injury is in rodent models, they lack certain attributes that a larger animal model could provide such as near-human bone mineral apposition rates and comparable tissue mass for realistic tissue response. Work is currently underway to develop a large animal model to address these issues [29].

23.3.4 Direct Skeletal Fixation/Intraosseous Fixation of Prostheses

Direct skeletal fixation (DSF or osseointegration) of prosthetics may provide a clinical solution for patients with existing blast-related HO. Rather than attempting to modify the HO in any direct way, DSF can be used to load the residual bone directly, rather than through the soft tissues of the stump [30]. Without soft tissue loading, HO in the stump would be less likely to cause pressure effects, skin ulceration or pain; although if the HO formed near a joint it may still lead to loss of range of motion. Cases selection is likely to prove crucial in the degree to which patients may benefit from DSF. Also, the interaction between DSF and existing HO has not been studied in detail. As such, its role is currently in the process of being defined more clearly. DSF is explored further in Part V of this volume (Chap. 43).

23.4 Future Research Focus

Researchers have to be prepared to meet the unmet clinical need faced by those who suffer blast injury and HO currently and in the future. The use of blast weapons in civilian terror attacks is likely to continue for the foreseeable future and blast injury will be an inevitable consequence of the next conflict we fight regardless of its scale. The development of a high-fidelity large animal model will be immensely useful in order to test novel therapeutics and to gain further mechanistic understanding of how HO forms after blast injury [29]. Phase 3 trials of Palovarotene are expected and early phase trials of other agents are required. Finally, the time to begin planning for the next conflict is now; designing clinical trials and research projects in advance will allow real-world testing of promising novel agents and enable progress against this challenging condition.

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24 Pathological Cascades Leading to Heterotopic Ossification Following Blast Injury

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Abstract

Heterotopic ossification (HO) is the process of de novo bone formation in non-osseous tissues and is common following blast-induced traumatic amputations. Inflammatory cytokines as well as mechanical loading play a key role in the initiation and the progression of the disease and therefore the downstream biomarkers under specific signal transduction pathways which trigger the formation of the ectopic bone in the tissues surrounding the site of injury are key. However, the current published knowledge, summarised here, shows that there is a lack of understanding of the exact signalling pathways leading to HO, thus hampering the design of specific therapeutics. Also, there are no in vivo preclinical models that replicate the full scenario of blast, fracture and amputation, experienced in conflict blast injuries. These models are required to fully dissect the triggered signalling pathways. Therefore, this gap in the literature urgently needs addressing. Replicating the full effect of the blast in preclinical models will enable a better understanding of the mechanisms of blast-induced HO development and enable the design of a specific therapeutic to suppress the formation of ectopic bone.

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24.1 Introduction

Early clinical detection of ectopic bone formation is difficult as there is currently no absolutely reliable diagnostic tool that can confirm HO in the clinic until there is calcified bone present. Therefore, the only existing effective intervention for HO resulting from complex blast injuries is the late-stage surgical excision of the ectopic bone. However, there are multiple drawbacks and risks [1] associated with this treatment, and, in some cases, the nature and location of the formed bone means that this treatment is not suitable for some patients [2]. Furthermore, HO will recur following the surgery in up to 27% of patients who have partial excision and 7% of patients who have complete excision [1]. Due to the very high prevalence of HO following blast injuries, and, because the mechanisms of HO initiation and progression in blast are not yet fully understood, the aim of this chapter is to review the key biomarkers and signalling pathways identified which trigger the development of the ectopic bone growth as well as the relevant in vivo HO models developed so far, with a specific focus on blast injuries. This will enable the elucidation of the

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molecular mechanisms that can trigger acquired HO and thus provide new opportunities for blast injury science and engineering researchers to design relevant in vivo models to investigate the mechanisms and design and test therapies for HO following blast.

24.2 Key Molecular Markers of Heterotopic Ossification

For HO to occur, stem cell recruitment, proliferation, and differentiation into osteogenic cells needs to be initiated. In vitro studies representing the mitogenic and osteogenic effects of serum from patients with trauma to the central nervous system on osteoblastic, mesenchymal and fibroblastic cells give credence to the association with humoral factors [3–8]. The good blood supply to the skeletal muscles that carries factors such as inflammatory cytokines and growth factors supports the stem cells to differentiate into the osteogenic lineage [9] thus providing evidence for the increased incidence of HO in the skeletal muscles compared to other soft tissues [10–12]. Studies on adult human skeletal muscle cells that differentiated into osteoblastic cells in vitro formed new bone when reimplanted into animals [13–15]. Studies have shown that the key trigger to initiate HO might be local inflammation following trauma, including spinal cord injury, traumatic brain injury, surgical intervention, blast and deep burns [16–18]. Inflammatory cells such as macrophages, lymphocytes and mast cells surrounding the perivascular area of the newly formed ectopic bone as a result of damage to skeletal muscle cells and tissue hypoxia can prompt the proliferation and differentiation of progenitor cells (Fig. 24.1). Yet, significant clinical and experimental evidence indicates that the bone morphogenetic protein (BMP) family plays a fundamental role in the development of acquired HO. It is well known that the BMP family, which is part of the transforming growth factor beta (TGF-β) superfamily, has a crucial role in chondrogenesis and osteogenesis through the recruitment of mesenchymal cells and stimulating their differentiation into chondrocytes and osteoblasts (Fig. 24.1) [19]. BMPs, including BMP2, BMP4, BMP7 and BMP9, have been tested previously in the context of HO by injecting them into the muscle tissue of mice [20, 21]. They have also been found to be over-expressed in human ectopic bone [22, 23] which might indicate that they have a key role in HO (Fig. 24.2). Unlike other BMPs, BMP2 is needed for mesenchymal stem cell MSC recruitment and differentiation as well as appropriate fracture healing; consequently, experiments with knock out mice showed that a lack of BMP2 leads to failure in fracture healing [24]. Other biomarkers such as Serum interleukin-6, interleukin-10 and MCP-1 and wound effluent IP-10 and MIP-1a were associated with blast-induced HO in humans [25]. Similarly, Alkaline phosphatase (ALPL), BMP-2, BMP-3, collagen 2 alpha 1 (COL2A1), collagen 10 alpha 1 (COLL10A1), collagen 11 alpha 1 (COL11A1), cartilage oligomeric matrix protein (COMP), colony-stimulating factor 2 (CSF2), colony-stimulating factor 3 (CSF3), matrix metalloproteinase-8 (MMP8), matrix metalloproteinase-9 (MMP9), SMAD1 (one of a family of structurally similar proteins that are signal transducers for receptors of TGF-β) and vascular endothelial growth factor (VEGFA) were upregulated in patients during early debridement of blast-induced HO. BMP2 has been shown to be upregulated in patients with late-stage ectopic bone debridement indicating that it might have a key role in combat-related HO [26]. In a preclinical blast model, chondrogenic markers such as collagen type I alpha 1 (COL1α1), osteogenic marker Runt-related transcription factor 2 (RUNX-2), phosphate-regulating neutral endopeptidase, X-linked (PHEX) and POU domain class 5 transcription factor (POU5F) and proteins Noggin (NOG), OCN and substance P-1 (SP-1) were upregulated in the injured soft tissue compared to the control or sham [27]. These biomarkers are summarised in Fig. 24.2.

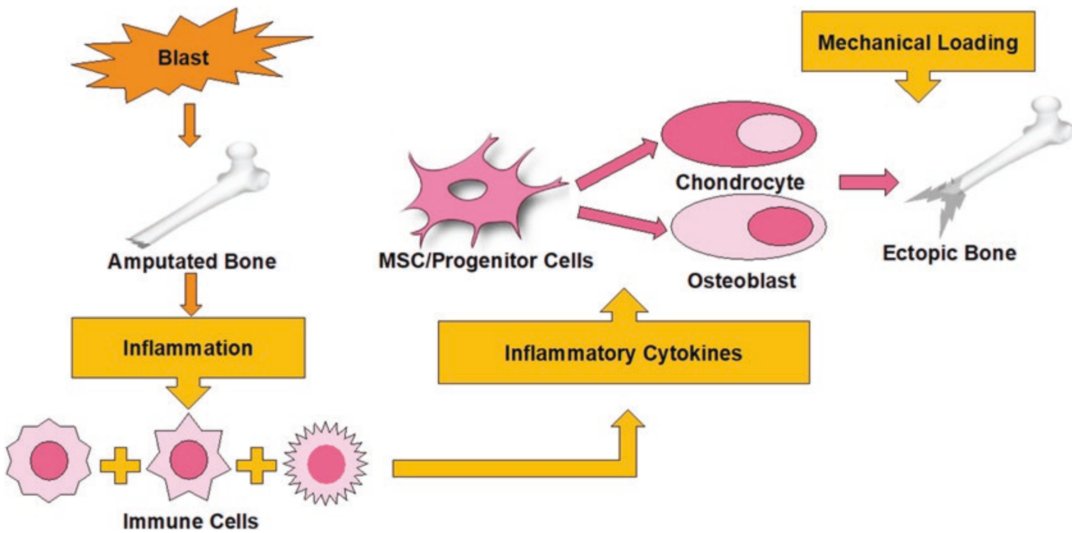

Fig. 24.1 A proposed cascade leading to HO as a result of blast injuries. Immune cells such as macrophages, lymphocytes, and mast cells can prompt the proliferation and differentiation of progenitor cells towards osteogenic lineage resulting in ectopic bone

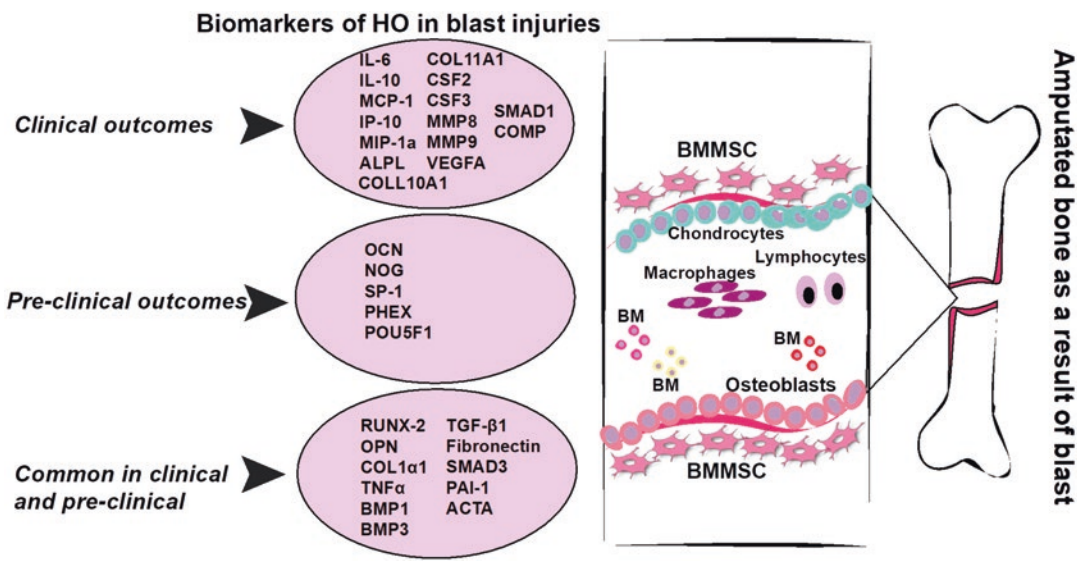

Fig. 24.2 Biomarkers identified in preclinical and clinical studies of HO induced by blast

24.3 Signalling Pathways in Acquired Heterotopic Ossification

Extensive clinical and experimental research proposes that BMP signalling plays a key role in acquired HO. Specifically, increased levels of BMP2 are associated with acquired HO [28]. In a traumatic injury model, expression of various osteogenic gene transcripts, including BMP-2, BMP-3 and SMAD1 downstream of the BMP signalling pathway was found elevated in wound tissue. Also, the expression of a type I BMP receptor, BMPRIa, was increased [26]. Type I BMP receptor BMPRIa and type II BMP receptor BMPRII also were upregulated 7 days post spinal cord injury, and phosphorylation of SMAD and translocation to nuclei was observed [29]. An in vitro study showed that low amounts of fibroblast growth factors 2 (FGF2) seemed to increase BMP2-induced ectopic bone formation via increased BMPRIb expression and SMAD1 phosphorylation while high dosages of FGF2 hindered MSC differentiation into osteogenic cells [30]. Furthermore, TGF-β2 is increased threefold in immature compared to mature ectopic bone rendering it a crucial component of HO development after THA [31].

TGF-β uses many intracellular signalling pathways as well as SMAD signalling to control a wide range of cellular functions. Quick activation of Ras (a specific family of proteins) and or Extracellular receptor kinase (Erk1/2) by BMP is also reported in myoblasts, osteoblasts and stem cells [32, 33]. ERK signalling was detected in an in vitro MC-3 T3 and in vivo mouse HO model induced by connexin 43 (Cx43). ERK signalling induced by Cx43 had an essential role in HO development because inhibition of Cx43 and ERK downregulated the expression of Runx2, bone sialoprotein (BSP) and collagen 1 (Col-1) both in vivo and in vitro. Therefore, low levels of Cx43 and ERK lowered the risk of HO recurrence [34]. Mitogen-activated protein kinase (MAPK) signalling was also found upregulated when human adipose stem cells (hASC) were treated with trauma HO^{+} patient's serum vs HO^{-} serum. Genomic analysis showed that hASCs treated with serum from individuals who developed HO had affected the expression of the activator protein 1 (AP1) signalling. Additionally, significant downregulation in FOS gene expression in hASCs treated with serum from individuals with HO was detected. Furthermore, the downstream genes associated with the MAPK were upregulated in hASCs following serum exposure from individuals with HO [35].

These signalling pathways have further crosstalk with each other to complete their role in the disease mechanism. BMP signalling is suggested to crosstalk with other key regulators of HO which are hypoxia-inducible factor 1-alpha Hif1α and mammalian target of rapamycin (mTOR) pathways in the context of HO [36]. In vitro long-term hypoxic environment (3%) exposed osteoblasts show upregulated BMP2 through the stimulation of the mTOR and HIF1-α signalling pathways (Fig. 24.3). Hypoxia increases mTOR phosphorylation and stabilises the activity of the mTOR and HIF1-α signalling pathway molecules. On the other hand, the inhibition of mTOR and HIF1-α signalling impairs the hypoxia-induced BMP2 expression [37] (Fig. 24.3). Another example is the crosstalk between the BMP signalling and Wnt B-catenin as well as the ERK signalling pathways through the BMP downstream mediator RUNX-2 [38]. There is still controversy around BMP and Wnt functions in osteogenesis, suggesting that Wnt is the major controller of osteogenesis in bone marrow-derived mesenchymal stem cells (BMMSC)s while BMP signalling is the major controller of osteogenesis in adipose-derived mesenchymal stem cells (ADMSCs) [39]. Overall, most studies suggest that the impact of Wnt signalling is less important than BMP signalling when it comes to MSCs differentiation into osteogenic cells. Figure 24.3 summarises the pathways through which BMPs and the TGF-β family can induce chondrogenic and osteogenic markers in an HO environment through cross talk with other signalling pathways.

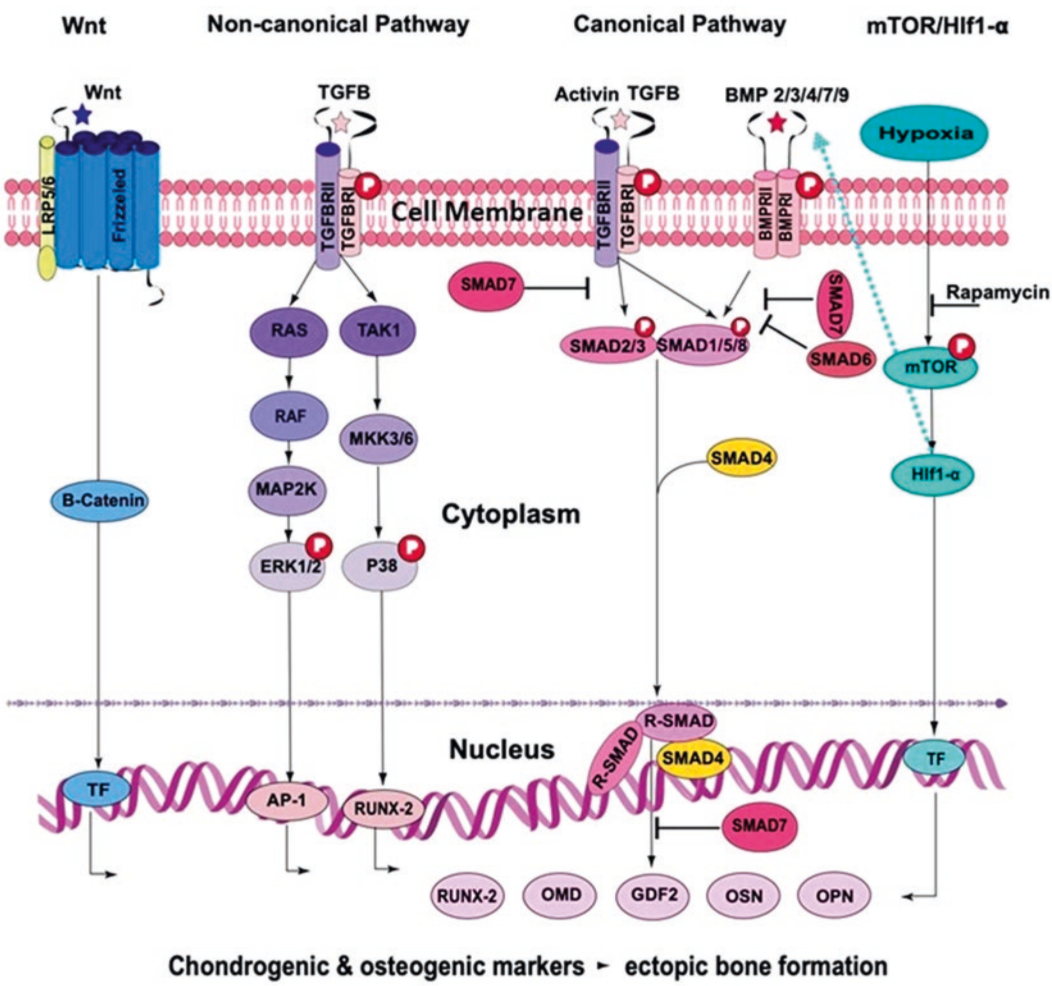

Fig. 24.3 The canonical, non-canonical pathway of BMP/TGF-β family, mTOR and Wnt signalling pathway that can lead to heterotopic ossification. The canonical pathway is mediated via different SMADs (SMAD1/5/8 and SMAD2/3) that can be phosphorylated (activated) (R-SMAD) then translocate to the nucleus and act as activators or inhibitors (SMAD6/7) to the transcription of osteogenic or chondrogenic marker genes leading to bone formation. The non-canonical pathway can be mediated via non-SMAD-dependent pathways such as mitogen-activated protein kinase (MAPK) pathway activating downstream extracellular signal-regulated kinases (ERK) and mitogen-activated protein kinase p38 (P38) which in turn will activate transcription factors (TF) leading to the transcription of chondrogenic or osteogenic markers. There might be also crosstalk between hypoxia-mediated mTOR, HIF1-α and BMP pathways which can be inhibited by Rapamycin. Moreover, BMP signalling might crosstalk with Wnt/B-catenin and the ERK signalling pathways through the downstream mediator such as RUNX-2

24.4 Discussion and Future Therapeutic Targets

Although acquired HO can arise from different types of mechanically induced trauma, it is only in recent decades that its high level of incidence post blast injury has become evident, most likely due to the very high levels of survival attained by severely wounded military personnel due to excellent medical care.

The current therapeutic interventions for HO treatment at early stages include Nonsteroidal anti-inflammatory drugs (NSAIDs), bisphosphonates, and radiation therapy [40–44], where both NSAIDs and radiation therapy are more effective when they are used in combination [45], leaving this as the only choice for HO prevention at the early stages. These existing treatments for HO are insufficient particularly for the conflict wound setting. Currently, specific drugs targeting the SMAD-BMP pathway signalling are being developed and rapamycin is promising as it may interfere with the HO at the level of two main signalling pathways through Hif1α [46] and mTOR pathways [47] (Fig. 24.3). The ideal treatment would be the complete prevention of HO, and this can only be achieved when all the pathophysiological causes leading to the ectopic bone formation due to traumatic blast injuries have been identified and the crosstalk between the signalling pathways have been studied. Understanding the molecular biomarkers and signalling pathways can help to identify therapies, yet, these studies require the early identification of HO and this is currently not possible in the clinical situation. Therefore, there is a need to have appropriate in vivo models that allow this early phase to be investigated. Current in vivo HO models are limited in that bone fracture is not always achieved as a result of the blast itself, but by using an extra tool such as a drop weight; this does not replicate the exact situation/model of combat-related HO. There is no in vivo model that best represents the blast injuries in the battlefield combining the blast and the fracture simultaneously in one insult, necessitating the development of more advanced models to better replicate the disease conditions. This would enable the examination of the early mechanisms producing HO to better understand the impact of the blast on HO development.

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Fracture Non-Union After Blast Injury

25

Major Sarah Stewart
and Lieutenant Colonel Daniel Stinner

Abstract

Fracture non-union, the failure of bone to heal, is a major cause of morbidity in the orthopaedic population. It is chiefly a consequence of disruption of the biological or mechanical environment of the fracture, resulting in either atrophic or hypertrophic non-union. Fractures sustained secondary to blast injury are particularly susceptible to non-union. The high-energy nature of blast injury results in large zones of injury, segmental defects, infection and soft tissue stripping, which together has a marked detrimental impact on the biological and mechanical environment of the fracture.

An appreciation and understanding of the likely insult to the fracture environment is critical when working up a patient with a non-union and planning appropriate treatment. Successful fracture union is reliant on the presence of four factors: mesenchymal stem cells, growth factors, osteoconductive scaffolds and an optimal mechanical environment. Current and future therapeutic strategies focus on the augmentation of one or more of these factors thereby enhancing the biological and/or mechanical environment of the fracture. Emerging treatments including stem cell and gene therapy, novel fixation devices and mechanotransductive technologies provide promising scope to better improve the rate of fracture union in this challenging subset of patients.

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25.1 Introduction

Bone non-union is the failure of a fracture to heal. The biology of bone healing relies on a series of carefully orchestrated processes within tightly defined conditions. When these biological or mechanical conditions are not achieved, the bone cannot regenerate and a non-union is established.

Two distinct types of non-union exist in the literature, determined by the amount of bone formed at the fracture site radiographically: *atrophic* and *hypertrophic*. *Atrophic* non-union is associated with inadequate biological factors and is established in the early stages of fracture heal-

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ing. *Atrophic* non-union is typified radiologically by the paucity of callus formation at the fracture site. Conversely, *hypertrophic* non-union is used to describe those non-unions where there is excessive callus formation radiographically. However, the callus formation is disorganised and outwith the fracture site, so the fracture remains non-united. It is associated with inadequate mechanical stability and is established in the later reorganisational stages of bone healing.

Management of non-union is guided by correction of those biological and mechanical factors that underpin the aetiology of atrophic and hypertrophic non-union. Atrophic non-unions where biological factors have driven the failure to heal require adjustment of the biological environment, namely through the utilisation of growth factors, osteoconductive scaffolds and mesenchymal stem cells (MSC). Hypertrophic non-unions require optimisation of their mechanical environment. Together, these factors are referred to as the 'Diamond Concept' and form the cornerstone of bone restoration and regeneration principles [1] (Fig. 25.1).

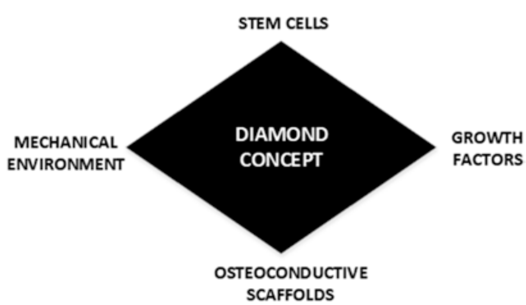

Fig. 25.1 The diamond concept of fracture healing (adapted from Giannoudis et al. [1])

Non-union causes pain and loss of function in the affected limb, resulting in significant morbidity to an individual and impairment to their quality of life (QoL). Brinker et al. demonstrated that the negative effect on QoL of a tibial non-union was comparable with the reported impact of end-stage hip osteoarthritis and worse than that of congestive heart failure [2], highlighting the degree of disability this devastating, chronic condition can cause.

The cost of treatment of non-union is also high, borne out of the difficulties associated with its treatment. Estimates place the financial cost in the tens of thousands of pounds for the medical treatment of a single fracture non-union, placing a marked fiscal burden on the health system [3, 4]. This sum is notwithstanding the indirect costs such as productivity losses, which account for 67–93% of the total costs associated with a tibial fracture [5]. Reaching a global consensus regarding the optimal treatment for non-union is therefore paramount to reduce the clinical and financial burden that this condition bestows.

25.2 The Clinical Problem

Since the introduction of gunpowder to Europe 800 years ago, combat injuries have been characterised by ballistic and explosive weaponry. Foreign substances contaminated the resulting increasingly complex extremity wounds, complicated further by shattered bone [6]. The necessity of early debridement and early amputation before infection set in defined the management of these high-energy, open fractures sustained in war throughout the fifteenth to nineteenth centuries.

The concept of limb salvage to preserve the shattered limbs sustained in combat only emerged in the early twentieth century with the advent of bone graft surgery [7]. Bone grafting enabled military orthopaedic surgeons to promote union in fractures that would otherwise have succumbed to amputation. However, with limb salvage comes the risk of non-union.

The recent global conflicts of Iraq and Afghanistan saw high rates of extremity injury. 54% of evacuated injured service personnel had sustained an extremity injury [8]. More than one-quarter (26%) of all extremity injuries involved fractures, of which 82% were open fractures [9, 10]. This alone posed a marked clinical burden and treatment challenge. The prolific use of Improvised Explosive Devices (IEDs) throughout both conflicts further complicated the management strategy. Kinetic energy is transferred to the limb from the IED blast wave itself, fast-moving ballistic fragments and surrounding debris, and from the impact of the individual with surrounding objects. This leads to extensive soft tissue damage, periosteal stripping, segmental bone loss and denudement of the vascular supply. Anything that breaches the skin barrier (be it a bullet, IED casing fragments, nails, screws or local material propagated by the blast wave) is likely to be contaminated, increasing the risk of infection which further increases the probability of the development of non-union in the event of a fracture. To summarise, the ballistic, high-energy injuries seen with blast trauma lead to high rates of bone loss, soft tissue injury and infection through a number of blast-related mechanisms, which greatly increase the chance of a fracture progressing to non-union [11–13]. Furthermore, these conflicts have been associated with increased survivability of casualties with complex wounds [14], giving rise to a cohort of survivors with devastating limb injuries that have only previously been seen in those that had perished.

Although commonly seen in a combat environment, the global terrorist attacks in recent years as well as civilian injuries sustained in warfare mean that these complex injuries are not confined to military personnel [15–17]. Moreover, the risk of non-union is further compounded by the frequently austere environment where such individuals (both military and civilian alike) acquire their injuries. Extended timelines for casualty evacuation of individuals with blast-induced orthopaedic injury often result in delayed initial management and treatment of these high-energy fractures [18, 19], further compounding the risk of non-union as tissue ischaemia and infection advance. Those fractures caused by blast and ballistic injury can therefore represent the greatest challenge to promoting union.

The decision to salvage a limb or amputate can often pose the greatest challenge to the operating surgeon. A retrospective study examined the functionality and mental health of 324 service personnel who sustained a major lower limb injury whilst serving in Afghanistan or Iraq between 2003 and 2007, which was subsequently salvaged or amputated. They demonstrated that those individuals who had undergone an amputation had significantly better functional outcomes and a lower likelihood of Post Traumatic Stress Disorder (PTSD) than those individuals who had undergone limb salvage [8]. However, other studies have shown conflicting results [20]. The decision regarding which management option to take when faced with an extremity injury therefore adds to the clinical problem. Numerous scoring systems exist to guide the surgeon's decision-making process on whether to salvage or amputate a mangled limb [21]. However, none of these has been validated in a civilian or military setting leaving the difficult decision of whether to salvage with the risk of non-union and accompanying impact on function and mental well-being, or amputate with the life-changing effects that this operation bestows on an individual [22].

A recent review has demonstrated that rates of non-union amongst combat casualties from ballistic mechanisms have largely been static over a century of warfare [23]. The rate of union amongst a cohort of soldiers with ballistic injuries from

World War II (88.8%) was virtually identical to the rates obtained in the most recent case series in the review from 2008 (89.4%) [24, 25]. Compare this to how fracture management has evolved in the civilian sector to achieve union rates of 98.1% [26], and it is evident that fractures following ballistic and combat trauma represent a uniquely challenging subset of orthopaedic injury.

25.3 Current Management

Fractures that show no evidence progression of healing for at least 3 months on sequential clinical and radiographic exams warrant investigation for non-union. There are several fracture types and treatment-specific factors that should be considered when evaluating someone for non-union. A history of a severe open fracture, which is often caused by blast or high-velocity ballistic injury, and a fracture that underwent open reduction both increase the risk of infection as a cause for non-union. Tibial fractures have a propensity of progression to non-union because of their poor soft tissue envelope. There are other patient-related factors that can impair fracture healing as well, including metabolic and endocrine disorders, in addition to the use of tobacco products. Brinker et al. found that of 37 patients evaluated for non-union, 31 (84%) had either a metabolic or an endocrine abnormality that was likely contributing to their inability to heal their fracture [27]. Therefore, if there is no clear reason for the non-union, i.e. adequate fracture stability and no signs of infection, patients should be evaluated for vitamin D deficiency and abnormal calcium regulation in addition to thyroid, pituitary, parathyroid, and reproductive hormone dysfunction. If an abnormality in one of these is identified, it should be corrected prior to considering surgical intervention for the non-union. Inflammatory markers should also be evaluated, including a white blood cell count, erythrocyte sedimentation rate and C-reactive protein to evaluate for infection. Even if an infection is unlikely, given a negative clinical workup, intraoperative cultures should still be obtained during the non-union surgery to ensure the infection is not the aetiology of the non-union.

25.3.1 Nonsurgical Management of Non-Union

While there is limited data demonstrating the success of nonoperative management of the established non-union, there are several adjunctive treatments currently available. Low-intensity pulsed ultrasound (LIPUS) can be used over the non-union site providing mechanotransduction at the fracture site in an effort to stimulate bone healing. While data describing outcomes following the use of LIPUS in an established non-union have been variable, a recent systematic review and meta-analysis demonstrated favourable results when non-infected non-unions were treated with LIPUS [28]. While used less commonly due to limited evidence [29–31], teriparatide (a recombinant form of parathyroid hormone), may soon prove to be a viable nonsurgical option for the management of non-union.

25.3.2 Surgical Management of the Non-Union

The gold standard for the treatment of a non-union involves surgical intervention. In order to plan for the surgical procedure, an appreciation for the type of non-union is important as it dictates the surgical plan. As described earlier and shown in Fig. 25.2, a hypertrophic non-union is caused by inadequate fracture stability. This type of non-union requires revision fixation to a more stable construct to promote bone healing. On the other end of the spectrum, the atrophic non-union, the fracture was appropriately stabilised, but the surrounding environmental factors (growth factors, stem cells, blood supply) are inadequate to achieve fracture healing. As a result, these fractures require bone grafting, which often is done in conjunction with either revision fixation or augmentation of the existing fracture fixation. There are several options for bone graft including autogenous bone graft, allograft bone graft, and demineralised bone matrix.

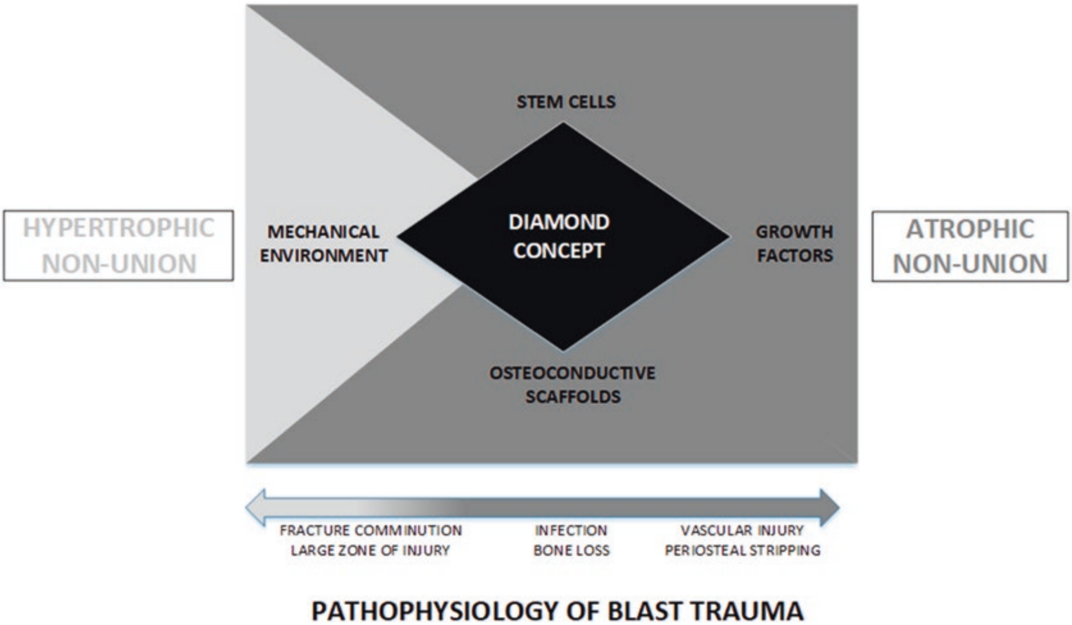

Fig. 25.2 The pathophysiological effect of blast trauma results in non-union through a number of mechanisms, leading to disruption of either the biological environment (resulting in *atrophic* non-union) or the mechanical environment (resulting in *hypertrophic* non-union)

25.4 Future Research Foci and Needs

Emerging research in non-union aims to enhance the mechanical or biological environment thus augmenting the factors required for fracture healing as described in Giannoudis's 'Diamond Concept' [1] (Fig. 25.3). This section will discuss those therapies which hold the most clinical potential, and arguably should be where future preclinical and clinical research on non-union should be focussed.

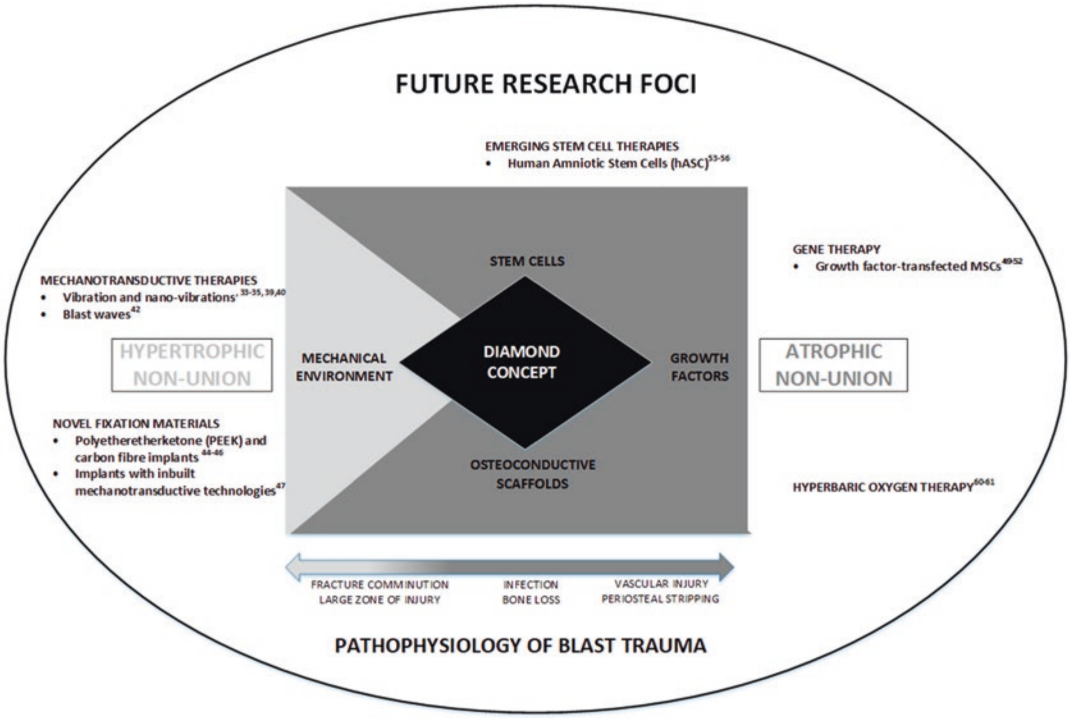

Fig. 25.3 Future research foci on non-union therapies aim to address the factors required for fracture healing as described in Giannoudis et al. 'diamond concept' model [1]

25.4.1 Mechanical Environment Enhancement

One of the most rapidly expanding areas of research in bone regeneration is mechanotransductive technologies. Mechanotransduction describes how an external physical stimulus can induce a biological response at a cellular level resulting in a change in the physiology of the surrounding tissue. The pathways involved are complex and a more detailed review of the processes can be read elsewhere [32]. Fundamentally however detection of local physical forces by specific cell receptors brings about changes in intracellular signalling pathways, ultimately altering the cell's phenotype and function and inducing a tissue response. Its role in bone physiology is well defined; the mass and architecture of bone are governed entirely by mechanical loading. Research on mechanotransductive biotechnologies focussed on osteogenesis is rapidly evolving. A number of modalities utilising mechanotransduction are already available for clinical use to treat non-union, such as extracorporeal shockwave therapy (ESWT), low-intensity pulsed ultrasound (LIPUS) and electrical stimulation (ES) [28, 33, 34].

Preclinical therapies also show promising potential. An expanding area of research is the field of vibration and nano-vibration technology. This application aims to recreate the micromovement that takes place at the fracture site during the secondary healing process. The osteogenic potential of low-magnitude, high-frequency vibration (LMHFV) has been explored in a number of animal studies [35–37]. Application of a low-magnitude (<0.1% strain), high-frequency (10–90 Hz) force produces a non-invasive vibratory stimulus which has been shown to stimulate bone, as well as enhance bone stiffness and strength [36, 38]. This systemic application of vibrations has been coined 'whole body vibration' (WBV). Although its use in improving bone density has been translated into clinical research [39, 40], application of it in a clinical setting to treat non-union remains unexplored. Nanovibrational stimulation describes the application of nanoscale vibrations in the region of 5–50 nm vertical displacements [41]. These nanomovements are small enough to induce detectable changes at the cell-surface interface. In vitro studies have demonstrated that nanovibrations induce osteoblastic differentiation of MSCs in both 2D and 3D environments [41, 42]. Application of vibrations and nanovibrations on tissue-engineered bone graft therefore represents a novel pathway in the field of bone regeneration for use in conditions such as fracture non-union.

Heterotopic ossification (HO) describes the formation of de novo, aberrant bone in extraskeletal tissues (see Chaps. 23 and 24). The blast trauma that defined the injury mechanism witnessed in Iraq and Afghanistan has been correlated with an increase in the prevalence of heterotopic ossification amongst blast-induced combat amputees at an incidence of 60–64% [43], suggestive of a link between blast waves and HO formation. This theory is further substantiated by the finding that samples from high-energy, combat wounds have high osteogenic potential, accelerating the in vitro osteogenic differentiation of MSCs [44]. This transference of energy by the blast wave into human tissue which subsequently undergoes osteogenic change represents a novel, mechanotransductive mechanism through which osteogenesis may be stimulated. This has scope to be harnessed as a clinical therapy for non-union in which delivery of a blast wave or similar high-energy analogue to progenitor cells can induce osteogenic differentiation.

The design of internal fixation devices is also evolving rapidly to better optimise the mechanical environment in which a fracture heals. The longstanding use of titanium as a plate material can preclude accurate radiographic visualisation due to its radiodensity, has potential for wear debris and corrosion, and can lead to stress shielding (a reduction in the underlying bone density) [45]. Use of inert polymers such as polyetheretherketone (PEEK) in combination with carbon fibre to provide durability is increasingly appearing in the orthopaedic implant market [46–48]. What is of particular interest is combining these novel materials with mechanotransductive properties. A recent patent application describes the use of a carbon fibre plating system augmented with an inbuilt electromagnetic stimulation capability thus allowing the generation of a mechanotransductive force directly onto the underlying fracture [49]. Although other fixation devices exist which concurrently generate electromagnetic stimulation locally at the site of the fracture, e.g. on the skin or surrounding soft tissue through an implanted cathode, this is the first device described where the mechanotransductive stimulus originates from the plate itself. Decreasing the distance between the physical stimulus and the fracture site and eliminating the interposing soft tissue between the bone stimulator device and fracture can potentially reduce the likelihood of non-union occurrence.

25.4.2 Biological Environment Enhancement

Innovations in biological therapies are also at the forefront of non-union research. A promising field of biotechnological research in non-union treatment is gene therapy. Although the importance of growth factors in bone healing has already been discussed, they are not without a side effect profile including HO, infection and immunogenic reactions likely secondary to the ‘megadoses’ required to illicit a therapeutic response [50]. Gene therapy whereby exogenous growth factor DNA is transfected into cells at the site of a fracture allows a local and continued expression of growth factors at the site of injury. MSCs recruited to a critical segmental defect and subsequently transfected with BMP-6 led to improved fracture healing in a porcine non-union model [51]. This therapeutic modality can be advanced further by combining the growth factor-transfected MSCs with scaffolds such as β-tricalcium phosphate [52], demineralised bone [53] and platelet-rich plasma [54]. This has the additional advantage of addressing a third element of the diamond concept of bone healing (osteoconductive scaffolds), as well as minimising the diffusion of the osteogenic agent away from the fracture site that can occur with local injection.

The evolution of stem cell applications in regenerative orthopaedics has also progressed significantly in recent years. Although the role of MSCs is well documented, much interest of late has been centred on the use of human amniotic stem cells (hASC) [55–58]. These confer an advantage over MSCs for a number of reasons. Firstly, their concentrations in amniotic fluid is higher by a factor of 100–1000 compared to MSCs (1% of amniotic fluid compared to 0.001–00.1% of nucleated cells in bone marrow) [59]. Further to that, they can be simply obtained during the process of amniocentesis from women undergoing prenatal diagnosis thus avoiding the morbidity and additional surgery required to obtain bone marrow. Finally, hASCs have been shown to be highly efficient at differentiating into osteogenic cells. In vitro mineralisation assays demonstrated that 85% of hASCs compared to 50% of MSCs were capable of forming osteogenic colonies [59], making these cells an attractive source for regenerative therapies for non-union.

The biochemical environment to which MSCs are exposed may play an important role in influencing their osteogenic differentiation and is therefore of particular interest in modulating it in non-union conditions. Hypoxia is a recognised limiting factor in bone healing [60, 61], and therefore optimising the oxygen exposure of fractures at a cellular level has garnered much research interest. This can be achieved through hyperbaric oxygen therapy (HBO), which causes a transitory elevation in the partial pressure of oxygen within tissues [62]. A hyper-oxygenated environment enhances both MSC survival and osteogenic differentiation in vivo [63], whilst in a critical defect model of non-union, HBO therapy improved new bone formation both radiologically and histomorphometrically [62]. HBO therefore represents a unique novel therapy for treating non-union, through its non-invasive, systemic application.

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26 Orthopaedic-Related Infections Resulting from Blast Trauma

Louise Robiati and A. Hamish R. W. Simpson

Abstract

Blast mechanisms are responsible for a large proportion of combat-related musculoskeletal injuries. These injuries include complex open fractures which are grossly contaminated and are at increased risk of developing wound infections, osteomyelitis, fracture non-union and the need for late amputation. With terrorist use of Improvised Explosive Devices on the increase globally, managing these injuries is no longer limited to the combat setting.

Eradication of infection is a key consideration when managing blast-mediated extremity injuries and is best achieved through a multidisciplinary approach. This review specifically considers the clinical factors associated with treating blast-mediated injury to extremities, focusing on strategies for minimising infection and directions for future research.

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26.1 Introduction

Orthopaedic blast injuries result both directly, from high-energy waves passing through body tissues and penetration from ordinance components, and indirectly from displacement of the casualty and surrounding objects during the blast [1]. These mechanisms result in fractures, soft tissue damage and amputations which are grossly contaminated [2]. The physics and biomechanics of blast mechanisms are often more complex than those seen in the majority of reported civilian high-energy trauma, such as motor vehicle collisions, and therefore it may not be possible to directly extrapolate civilian research findings into the management of blast injuries (see Chap. 2).

Within the current literature, the focus for acutely managing blast-mediated injuries has mostly been confined to the military, with these injuries accounting for a large proportion of the clinical workload managed in this setting. During the conflicts in Afghanistan and Iraq (2003–2011), 81% of musculoskeletal combat injuries sustained were from explosive mechanisms [3–5]. However, with terrorist use of Improvised Explosive Devices (IEDs) on the increase globally, clinicians working in the civilian environment are increasingly being called upon to manage these injuries [6].

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26.2 Clinical Problem

A 2014 systematic review of the North Atlantic Treaty Organisation (NATO) coalition forces in Afghanistan and Iraq reported that 39% of battle casualties sustained extremity injuries [7]. Fractures were reported by recent studies to comprise 15–40% of these injuries, 82% of which were open fractures [5, 8, 9]. A well-recognised complication after this type of fracture is infection [10, 11]. For example, in combat-related Gustilo-Anderson grade III open tibia fractures (open fracture with extensive soft tissue damage) rates of infection were reported to range from 23 to 40% compared to 15% in civilian cohorts with the same grade of injury [11–15]. The increased rates of infection seen in the combat setting are believed to be due to both the local and systemic insults associated with blast mechanisms of injury, with long-term consequences including osteomyelitis, fracture non-union and late amputation [16].

26.2.1 Osteomyelitis

Osteomyelitis is an inflammatory process in the bone and bone marrow caused by an infectious agent which results in bone destruction and can be challenging to treat [17, 18]. After combat-related injuries, rates of osteomyelitis are reported as ranging from 6 to 25%, compared to 8% reported in the civilian population [13, 15, 19–21]. Due to the persistence of infection and requirement for further surgical intervention osteomyelitis has been shown to complicate orthopaedic care in both the early and late phases of rehabilitation after blast-mediated injuries [22].

26.2.2 Fracture Non-Union

Fracture non-union is covered extensively in Chap. 25. A non-union can be defined as a fracture that is 9 months post-injury and has shown no radiographic progression for 3 months [23]. However, from a clinical point of view, the term is often used to describe a fracture that has no potential to heal without further intervention. Fracture non-union has a devastating impact on a patient's quality of life and can result in limb loss as well as being an increased burden on the healthcare system [12, 24, 25].

The incidence of fracture non-union in the United Kingdom (UK) and the United States (US) has been estimated at around 11,700 and 100,000 per annum respectively, with rates of tibia non-union in the civilian literature reported at 12% for closed and 24% for open fractures [21, 25–28]. However, fewer proceed to fracture union in the military population, with rates of non-union for grade III open tibia fractures at 12 months ranging from 20 to 50% [12, 13]. The aetiology of non-union is multifactorial with infection reported as a contributory factor in 38% of cases in the civilian population [26, 29].

26.2.3 Late Amputation

Late, secondary or delayed amputations are considered to be amputations performed after an attempted limb salvage through reconstruction [30, 31]. While a range of time frames from injury to amputation are reported in the literature, several studies have used the Lower Extremity Assessment Project (LEAP) study definition of 3 months from time of injury to amputation [30, 32]. Rates of late amputation from limb-threatening injuries are reported as 5% in the civilian literature and 5–22% in the military literature for grade III open fractures [13, 21, 22, 31, 33]. Infection was cited as the main reason for performing a late amputation in both the civilian and military cohorts [21, 31, 33].

26.2.4 Organisms

In combat-related open fractures, microorganisms initially cultured from the wounds have been predominantly gram-negative and include organisms such as *Acinetobacter, Enterobacter,* *Pseudomonas, Bacillus* and *Klebsiella* [13, 19]. These are consistent with findings from a UK civilian study which reported on open fractures sustained overseas, 50% of which were caused by gunshot or blast mechanisms, and repatriated to a UK level 1 trauma centre [34]. However, these findings differ from the predominantly gram-positive cultures reported in a study undertaken in German major trauma centres [35]. The time delay to culture sampling in the military and UK papers, as well as any variance between the microorganism flora prevalent in different countries may explain this finding.

Of note culture samples taken later in the military cohorts' clinical course were predominantly gram-positive and included organisms such as *Staphylococcus aureus* [13, 19, 33]. This change in flora may be nosocomial due to repeated surgeries and prolonged hospital stays [13]. Identifying these differences and changes in microbiological flora are essential for guiding changes in antibiotic regimens. They also demonstrate the importance of tissue sampling to avoid broad-spectrum therapies which contribute to multidrug resistance [36]. There is a lack of consensus amongst nations on which antibiotics to use in blast-mediated injuries however a wider range of bacterial and fungal infections should be anticipated in blast injuries compared to high-energy civilian trauma requiring additional antimicrobial cover [13, 16, 35].

26.3 Current Treatment and Management Strategies

When treating these complex injuries, clinicians are faced with the difficult decision of whether to attempt to salvage the limb or perform an early amputation. Long-term outcome studies have reported that rates of post-operative wound infections were 23% in limb salvage and 34% in amputation cohorts [21]. Burns et al. (2012) identified that 64% of culture specimens taken at initial surgery in military grade III open fractures were positive for bacterial growth, with those patients significantly more likely to go on and develop deep post-operative infections, osteomyelitis and require late amputations [13]. Given the complex nature of blast-mediated extremity injuries, increased risk of infection and the considerable complications potentially resulting from this, one of the main goals for managing these injuries in the acute setting is eradication and prevention of infection [16].

26.3.1 Antibiotics

Antibiotic administration has long been described in the literature as a critical factor in the prevention of infection in open fractures [11, 37]. There remains a lack of consensus around the optimal timing of administration after injury, duration and delivery of these antibiotics [38].

Historically it has been recommended that antibiotics be administered within 3 h from the time of injury [39]. However, a recent study demonstrated reduced rates of infection if antibiotics were delivered within 66 min from the time of injury, with these findings supporting previously reported preclinical in vivo research [39–41]. The UK national guidelines now recommend that antibiotics are administered ideally within 1 h of injury [42]. Therefore, given the potentially protracted casualty evacuation timelines in a combat setting, there is an argument for training medical personnel to provide antibiotics safely pre-hospital [43, 44]. However, the self-administration of oral antibiotics remains contentious due to concerns regarding inappropriate administration, potential adverse reactions and increased risk of contributing to antibiotic resistance [45, 46].

Antibiotics for open extremity injuries are generally administered intravenously (IV) but, in the combat casualty environment, establishing and maintaining venous access can be challenging and intramuscular (IM) or intraosseous (IO) methods may be required [16, 44, 47]. It remains unclear whether adequate therapeutic levels of antibiotics are achieved when administered IM, IO or IV in a limb with disrupted vascularity [16, 48]. Within the literature, some consideration has also been given to the efficacy of using locally delivered antibiotics through powder, liquid or antibiotic-impregnated bead formulations [16]. A meta-analysis identified that patients with grade III open fractures who received local and systemic antibiotics had infection rates of 7% compared to 27% if they received systemic antibiotics alone [48]. However, this meta-analysis identified several limitations, including the clinical heterogeneity of the studies concerning their study population, interventions, follow up and, crucially, the definition of infection [48].

Concerning the duration of antibiotic therapy, current guidelines recommend that antibiotics are continued for 72 h or until wound closure, whichever is sooner [39]. However, a meta-analysis reported that rates of infection in grade III injuries did not increase if antibiotics were only given for 24 h [49]. In the context of blast-mediated injuries, these findings should be interpreted with caution as they were based on two studies using civilian populations. They also did not take into consideration the International Committee of the Red Cross (ICRC) recommendations of continuing antibiotics for 5 days until definitive closure [49, 50].

For blast trauma, the current recommendations for antibiotic use in UK military deployed hospital facilities is Co-amoxiclav within 1 h of injury and a one-off dose of Gentamicin at the time of surgery [51]. These are mirrored by Public Health England guidelines for bomb blast victims which recommend for open fractures, 'through and through fractures' or intra-articular injuries intravenous Co-amoxiclav or Cefuroxime/Metronidazole should be administered until first surgical debridement and continued until wound closure with conversion to oral Co-amoxiclav for 6 weeks as well as a dose of Gentamicin at the time of initial surgery [52].

26.3.2 Irrigation

When managing open, infected or contaminated injuries, the adage 'the solution to pollution is dilution' is often heard. With guidelines providing recommendations on the volume of irrigation which should be used, depending on the grade of the open fracture [44]. However, in practice, wounds are irrigated with as much fluid as the operating surgeon deems necessary. Research has been undertaken to investigate whether the constituents and pressure of the irrigation alter post-operative infection outcomes [53].

Preclinical studies identified that use of irrigation fluids containing additives such as castile soap, bacitracin, benzalkonium and chlorhexidine initially resulted in reduced bacterial numbers post-operatively when compared to normal saline [54, 55]. Forty-eight hours post-operatively these studies observed a rebound effect with increasing rates of infection for those solutions containing additives [54, 55]. Authors attributed this observation to the additives having an irritant effect on the local healthy tissues resulting in them becoming necrotic, and, therefore, a favourable environment for bacterial growth [54, 55].

The Fluid Lavage of Open Wounds (FLOW) study was a clinical multicentre randomised controlled trial (RCT) with 2447 participants comparing irrigation of open fractures with castile soap or normal saline and very-low, low and high irrigation pressure rates (1–2, 5–10 or > 20 psi) [56]. They reported a significant reduction in infection rates in the normal saline group when compared to the soap group but no difference in pressure rates [56]. However, it was noted that some patients also received Negative Pressure Wound Therapy (NPWT) post-operatively which, in post-publication analysis, the authors reported increased rates of infection in these patients [56, 57]. However, they did not report any sub-group analysis for solution type or pressure or time to wound closure which may have biased their findings [56, 57].

26.3.3 Debridement

Surgical debridement excises devitalised soft tissue and bone and removes any foreign material which may become a nidus for infection [16, 39]. The 'six-hour rule' for time from injury to debridement is reportedly borne from animal experiments undertaken in 1898 which demonstrated a positive correlation between higher rates of infection and delay to surgical debridement and is often quoted in the literature and historical guidelines for the management of open fractures [39, 58]. A 2012 systematic review concluded that an association between time to surgery and rates of subsequent infection had not been demonstrated [59].

The LEAP study, a prospective observational study, identified no difference in the rate of infection when comparing time of injury to debridement of fewer than 5 h (28%), 5–10 h (29%) and more than 10 h (26%) in 315 open fractures [60]. However, they did find that a delay of greater than 2 h from the time of injury to admission to a definitive trauma centre was associated with a greater risk of infection. Brown et al. [19] also reported that time to surgery did not affect infection-related complications in military casualties with the most severely damaged extremities. Neither of these studies reported on the timing of antibiotic administration. This is an important factor as animal studies have demonstrated that a delay in antibiotic delivery (despite early surgical debridement at 2 h) resulted in higher rates of infection [40]. Although present guidance does not currently specify a time-frame, immediate debridement for highly contaminated wounds or those associated with vascular compromise is recommended, which is in keeping with military practice for blast trauma [42].

26.3.4 Compartment Syndrome

The majority of current combat extremity injuries are from explosions [3]. The resulting forces cause fractures, tissue loss and vascular injury which all contribute to the risk of developing compartment syndrome in the injured limb [61]. Compartment syndrome arises when pressure increases within a limited space and compromises the circulation and function of the tissues within that space and requires emergent decompression [62, 63]. Delays in diagnosis or inadequate decompression through fasciotomies lead to complications and poor functional outcomes [64, 65].

Ritenour et al. (2008) reported on complications after fasciotomy in the US combat casualties. This study included 336 patients who underwent 643 fasciotomies and identified 17% who required revisions and 22% who had delayed fasciotomies after medical evacuation from Iraq or Afghanistan [61]. In both the revision and delayed fasciotomy cohorts, rates of muscle excision and mortality were statistically higher than in the early, non-revised group [61]. For the revision surgery cohort, the anterior and deep posterior compartments of the lower leg were the most commonly unopened [61]. In those patients who underwent a delayed fasciotomy, the amputation rate was twice compared to those undergoing in theatre fasciotomy [61].

In the combat environment, additional factors may impede a timely diagnosis and decompression of compartment syndrome. For example, patients presenting with multiple distracting injuries, use of analgesics and sedation, oedema or delayed bleeding into compartments following adequate resuscitation, application of constrictive splints and simultaneous arrival of multiple casualties contribute to the reduced ability to identify clinical signs and perform serial examinations [61, 66–68]. Therefore, there is a need to maintain a high level of clinical suspicion for compartment syndrome in severely injured patients and early use of complete and prophylactic fasciotomies in high-risk patients should be considered [61].

26.3.5 Skeletal Fixation

When managing open fractures the main goals for treatment are prevention of infection, fracture healing and good functional outcome [69]. During the First World War deployed forward hospitals managed ballistic femoral fractures with thorough debridement and skeletal stabilisation with traction or splintage and noted a reduction in mortality rates from 80% to 20% [70]. Traction and splintage have been shown to remain a viable option today and have been used successfully in both military conflicts and in austere environments [71, 72]. Fracture stabilisation confers a variety of additional benefits including protection against further damage to soft tissues, improved wound care and soft tissue healing [69, 73].

The use of external fixators in combat fracture management continues to be an area of controversy since Bradford first reported its use on ballistic fractures in the US military hospitals during World War II [9]. It was initially indicated in patients with multiple injuries, infected fractures, or to prevent complications during evacuation [74–76]. However, in a post-war report, its use was associated with a high percentage of both infection and delayed union and was therefore forbidden and removed from hospitals [75, 77]. External fixation fell out of favour until the conflict in Somalia, where a review of the literature and resources required for managing combat-related open fractures resulted in it once again becoming the preferred method of stabilisation for US forces [78]. The purported advantages of external fixation include facilitation of transportation of wounded patients with fractured extremities, permitting access to soft tissue wounds and rapid stabilisation of the skeletal system to facilitate revascularisation procedures [78, 79]. Temporary external fixation in multiply-injured casualties may also confer systemic benefits to patients undergoing 'damage control orthopaedics' [80, 81].

Use of external fixators in ballistic trauma is not without complications. Clasper and Phillips (2005) prospectively followed up on 15 external fixators applied in the management of war injuries during the 2003 Gulf conflict. They identified that 13 (86.7%) required early revision or removal due to complications of the injury or the fixator; 10 (67%) had instability of the fixator; 3 (20%) developed pin site infections refractory to intravenous antibiotics and 5 (33%) developed pin loosening [82]. Due to the high rate of early complications, when using external fixators, this study cautioned against its universal application in war injuries [82]. Where, clinically, external fixators are favoured the authors recommended configuring a more rigid construct by using multiple pins and bars and to avoid using them for bridging fractures and if necessary acute limb shortening should be considered [83].

In a blast or combat setting use of internal fixation has been discouraged due to increased rates of infection in animal and civilian open fracture models [84, 85]. The limited availability of equipment, appropriate access to imaging and the unconfirmed sterility of theatres in a combat environment also dissuade clinicians from using this method of fixation [86].

26.3.6 Negative Pressure Wound Therapy

Surgical debridement of blast-mediated injuries can leave large wounds which may be unsuitable for primary closure. Sterile dressings are typically applied to protect the wound, but an alternative treatment is the application of Negative Pressure Wound Therapy (NPWT) [87]. NPWT are suction devices that create a partial vacuum drawing fluid which may have collected away from the wound and, in turn, encourage soft tissue healing [87].

There is contradictory and limited research reporting on the effect of NPWT on rates of infection after high-energy explosive injuries. For example, Warner et al. (2010) identified increased rates of infection in those treated with NPWT compared to those treated with NPWT and antibiotic bead pouches. However, this study was retrospective and had small study numbers [88]. Leininger et al. (2006) reported 0% of infection at 2 weeks in casualties treated with vacuum dressings. This study was also retrospective and did not undertake long-term follow up [89].

The Wound management of Open Lower Limb Fractures (WOLLF) study was a prospective multicentre RCT comparing standard dressings to NPWT for grade II and III lower limb open fractures [87]. The authors reported rates of deep infection at 30 days as 7% and 8% in the NPWT and standard dressing cohorts, respectively, and therefore did not support the use of NWPT over standard dressings [87]. Unlike the military setting, patients in this study did not require medical evacuation to treatment facilities overseas. Therefore, there may be some benefits to using NPWT if protracted aeromedical evacuation is anticipated [90, 91]. Further prospective RCTs are required in order to evaluate this as well as assessing benefits in both the military blast and civilian terrorist setting.

26.4 Future Research Directions

26.4.1 Clinical

To date, the majority of clinical research reporting on infection after blast-mediated extremity injuries has been retrospective. These studies do have inherent limitations; they are unpowered, rely on data to be charted accurately, lack control groups and are deficient in randomisation of treatment intervention with researchers not blinded to intervention [92]. Therefore, to improve knowledge in this area, prospective, randomised longitudinal studies must be undertaken. In future military campaigns, robust and comprehensive databases will be required to allow for the collection of meaningful prospective data [93]. In order to facilitate the undertaking of comparable research, the research community must validate and build on the consensus for the definition of fracture-related infection to also include definitions for late amputation, as well as criteria for diagnosis, timing and methods for microbiology sampling [30, 94]. While findings from civilian high-energy trauma research may influence clinical practice in future military campaigns, the complex nature of blast injury means it may not be possible to directly extrapolate these to combat trauma.

In addition to the areas of potential research discussed earlier in this review, an area warranting further investigation is antibiotic pharmacokinetics and pharmacodynamics. Limb injuries from blast are often associated with vascular injuries, managed with tourniquet application and resuscitated with substantial blood transfusions [3, 95]. What remains unclear is the extent to which this has an impact on the delivery of systemic antibiotics to open wounds and fracture site. Improving knowledge in this area may alter current management guidelines. For example, to ensure adequate antibiotic penetration into tissues, alternative methods of administration, higher initial antibiotic dosing or re-dosing may be required, but this has yet to be established [11, 16, 37].

26.4.2 PreClinical

On reviewing deep tissue microbiology samples from the time of revision surgery in military patients 26% had at least one organism which was the same as that cultured from samples taken at the time of injury [13]. These findings demonstrate that a proportion of deep post-operative infections are caused by the original inoculating organism [13]. Therefore, an area for further research would be to clarify if persistence of the original microorganisms could be attributed to inadequate irrigation and debridement at the time of injury or due to latent infection. With latent infection resulting from intracellular bacteria, multidrug-resistant organisms or presence of biofilms on hardware applied or inserted at the time of injury [96–98].

Translational preclinical research to date investigating interventions such as irrigation, debridement and antibiotic delivery on bacterial loads have been undertaken in animal models with critical defects [40, 55, 99]. However, a review of UK military personnel sustaining open tibia fractures on operations identified that the majority had non-critical size defects, so an alternative model is required [12]. Preclinical in vivo studies often assess an intervention in isolation and therefore do not reflect the complexity of damage control surgery. Casualties from blast mechanisms are often multiply-injured; there would be a benefit in using a poly-traumatised model such as that described by Claes et al. (2011) for investigating therapeutic interventions, although this model does not incorporate infection [100].

26.4.3 Novel Therapies

To date, research has focused on optimal strategies for local antibiotic administration, tissue decontamination and fracture stabilisation, as described above. However, other directions to consider include novel therapies such as the use of mesenchymal stromal cells (MSC). MSCs have been shown to have therapeutic potential in preclinical fracture non-union models as well as antibacterial effects in acute respiratory distress syndrome (ARDS) and biofilm models [101–103]. Therefore, their therapeutic potential in the context of orthopaedic, blast-mediated infections warrants further investigation.

26.5 Summary

Eradication of infection is a key consideration when managing blast-mediated extremity injuries and is best achieved through a multidisciplinary approach. Initial treatment strategies include early administration of antibiotics, timely and adequate irrigation and debridement of wounds, skeletal stabilisation and wound closure or dressing until definitive fixation and closure can be achieved.

Further research is required in both clinical and preclinical settings to develop best practice guidance as well as to identify potential novel therapies. These studies should endeavour to be designed and reported following the recent consensus published on fracture-related infection to facilitate the comparison of study findings.

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Part IV

Modelling and Mitigation

27 Section Overview

Spyros D. Masouros and Anthony M. J. Bull

The previous sections introduced fundamental concepts of science and engineering, described the explosive weapons and the mechanisms by which they inflict injury, and presented our current understanding, epidemiology and challenges in the management of blast injury. Over the years, engineering tools and technologies have been developed in order to aid in preventing, moderating the severity and improving upon the long-term outcomes of blast injury. The objective of this section is to present the engineering arsenal that we have at our disposal currently to achieve this and to note the challenges that are yet to be addressed.

The main tool used in engineering to analyse a phenomenon or design and make a purposeful structure is modelling: physical and computational. The first two chapters of this section present computational models and techniques used to address specific blast injury modalities, predict the severity of injury or progression of the disease and design mitigating systems. Chapter 30 provides an overview of equipment designed to replicate aspects of blast in a controlled, laboratory setting, and Chaps. 31 and 32 discuss experimental tests that replicate a primary blast injury mechanism on specific organs and systems.

The following three chapters discuss the use of physical surrogates for the study of human response to insult. Given that the ability to use human volunteers to experiment on is extremely limited when it comes to blast, use of surrogates that can simulate human response to the different aspects of blast has been necessary; the surrogates that are commonly used to understand human injury are human cadavers, animals, animal cadavers, anthropometric test devices (ATDs) and computational models. Although the use of each surrogate has strengths and weaknesses, the combined knowledge gained from using these surrogates has led to a dramatic improvement of protective systems and hence a reduction in lives lost due to insult. Chapter 33 discusses the use of cadaveric human tissue to study blast injury, Chap. 34 presents the use of engineered human-like surrogates (ATDs) and Chap. 35 deals specifically with surrogates for primary and secondary blast injury mechanisms.

The final two chapters of this section are dedicated to personal and infrastructure protection from blast.

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28 In Silico Models

Daniel J. Pope, Robert Fryer, and Spyros D. Masouros

Abstract

This chapter presents examples of computational models used to simulate injury due to blast loading. Examples from primary, secondary and tertiary blast loading that are used within our organisations are included. These are the Axelsson model for primary blast loading of the chest wall; overview of calculating the probability of injury to crowds due to a person-borne improvised explosive device; prediction of fragment penetration into the neck; prediction of injury to the lower extremity due to underbody blast (mine detonating underneath a vehicle) and evaluation of design parameters of an energy-attenuating vehicle seat for improved protection in underbody blast.

28.1 Introduction

Computational modelling is an extremely valuable tool in our arsenal for solving science and engineering problems. It allows for multiple virtual experiments to be conducted inexpensively for quantifying the response of systems to blast, or other, loading. This chapter provides some examples of computational models used in blast science and engineering, spanning injury mechanisms (primary, secondary and tertiary), applications and levels of complexity.

28.2 Axelsson Model for Blast Loading of the Chest Wall

Axelsson [1] used a single degree of freedom system to represent the response of the human chest wall exposed to primary blast loading (Fig. 28.1 primary blast loading). The thorax is represented as a dynamic system with an effective stiffness, K and damping, C that aggregate

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the response to the insult of skeletal tissue, soft tissue, liquid (blood and water) and gas in the lung. The applied force is derived from the difference between the transient blast pressure and the lung pressure.

$$m\ddot{x} + c\dot{x} + kx = A\left(p(t) - p_{\text{lung}}(t)\right)$$

where p_{lung} is the pressure within the lung and A is the effective lung area.

Solving this differential equation numerically using the numerical calculation cycle described in Chap. 5, Fig. 5.8, allows one to examine the response caused by relatively complex blast signals, such as multi-reflection loading resulting from internal explosions. This model has been used extensively to predict the probability of injury in civilian and military settings.

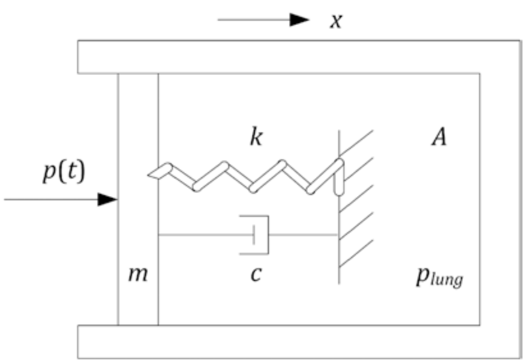

Fig. 28.1 Single degree of freedom (SDOF) representation of a thorax exposed to blast loading used to estimate lung injury

28.3 Projectile Flight and Penetration

Penetrating fragmentation resulting from the detonation of cased, military munitions or an improvised explosive device can be a significant source of human injury (secondary blast injury). Once the driving forces from the explosion have ceased to act upon it, a fragment may travel for a period within the surrounding air environment before penetrating a human at a given proximity from the explosive source. In order to assess the likelihood and severity of injury, the trajectory and velocity profiles of the fragments need to be calculated. For a fragment of mass *m,* the inertial force acting on it in the horizontal direction is balanced by retardation forces such as drag (which is a function of velocity squared), friction (which is a function of velocity) and stiffness and strength of the propagation medium. The force balance in the vertical direction is similar with the addition of gravity.

$$-m\frac{d\boldsymbol{u}}{dt} = \boldsymbol{A}\left(u^2\right) + \boldsymbol{B}\left(u\right) + \boldsymbol{C}$$

Figure 28.2 shows a practical, numerical implementation of the method within the Human Injury Prediction (HIP) software, a fast-running tool developed by the Centre for the Protection of National Infrastructure (CPNI) to predict blast and fragmentation injury risk due to explosive events in crowded public spaces [2]. Here, the trajectories of multiple fragments, produced by Person-Borne Improvised Explosive Device (PBIED), are being tracked within an environment containing a crowd of people, each represented by a cylinder with broadly human-like material properties. The approach allows the depth of penetration within the humans to be predicted or indeed determines whether the fragments are sufficiently penetrative to pass through the human with a residual velocity and then affect others standing further away within the crowd. When undertaking such analysis, the use of a temporal integration scheme with a fixed time-step, such as within the standard linear acceleration method explained above, may be inefficient as velocity changes in the fragment trajectory during transit in the air are much more gradual than those when the fragment is penetrating the humans. The HIP software makes use of a more advanced, Runge-Kutta schema—an iterative method developed for solving numerically a system of ordinary differential equations—with a variable time-step size dictated by the characteristics of the penetration medium.

a

b

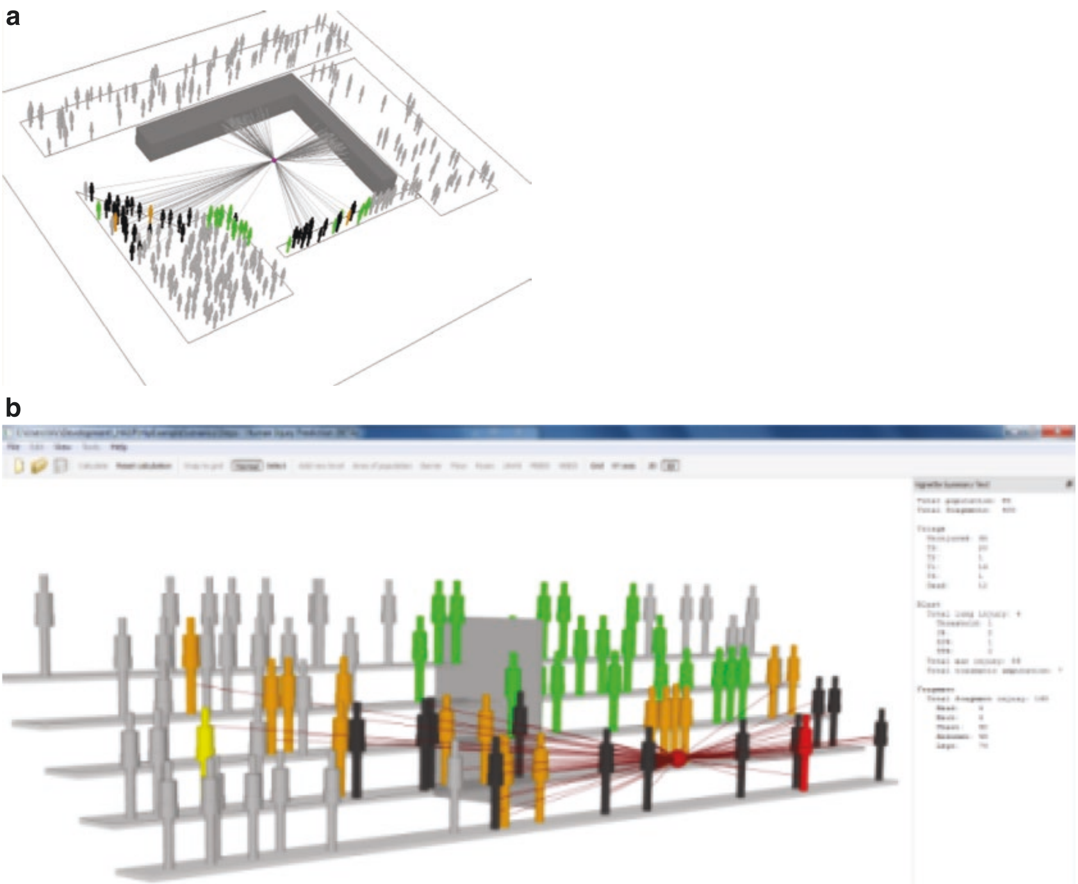

Fig. 28.2 Use of numerical fragment flight algorithms within the HIP code to represent the effect of protective barriers within (**a**) a tiered, stadium-type environment and (**b**) a crowded space during an explosive attack

28.4 Fragment Penetration to the Neck

Upon detonation blast weapons often produce high-velocity fragmentation from their casing or from other material in close proximity to the charge. Penetration of such fragments into the human body can result in serious or fatal injury (secondary blast injury); simulation of the penetration of a cylindrical fragment through a human neck is described in [2]. Using an Euler-based framework, the components of the neck (bones, veins, nerves and arteries as well as skin and muscle) are spatially generated by 'filling' the pertinent elements of the mesh with the appropriate material. In this case, either elastic, viscoelastic or inviscid fluid material models have been used in accordance with the behaviour of the component being simulated. The fragment is modelled as a rigid Lagrangian entity and fluid-structure interaction has been assigned to couple the projectile as it penetrates the neck.

28.5 The Lower Extremity in Tertiary Blast

Leg injury within military vehicles due to floor-plate ingress during a mine-loading event has been common in recent conflicts. In the example shown in Fig. 28.3a a Lagrangian, finite element (FE) model of a seated vehicle occupant has been developed using a variety of element types [3]. The bone components of the leg have been represented using a combination of shell and solid elements whilst solid elements have been used to represent the surrounding soft tissue. Contact logic is assigned between the bone and its surrounding soft tissue that could potentially interact during the deformation process. Ligaments and tendons are modelled as a collection of shell elements that join the bone components together at the appropriate anatomical locations. A piecewise linearly elastoplastic material model was assigned to bones, and a hyper-viscoelastic material model was assigned to soft tissues, in order to capture the non-linear and time-dependent effects associated with the high rates of loading seen in blast. Simulations were run of laboratory experiments on cadaveric legs using a traumatic injury simulator able to produce representative kinematics of the floor of a vehicle when attacked by a mine. The foot is located against a steel plate, which represents the floor of the military vehicle, and the dynamic phase of the simulation is initiated by assigning a velocity-time history to the plate. The model can predict the force experienced by the leg and produce the stress distributions, which can pinpoint location and time of fracture. Such models can be utilised to investigate mitigation designs inexpensively and examine human factors associated with location and severity of injury, such as posture at the time of injury (Fig. 28.3b).

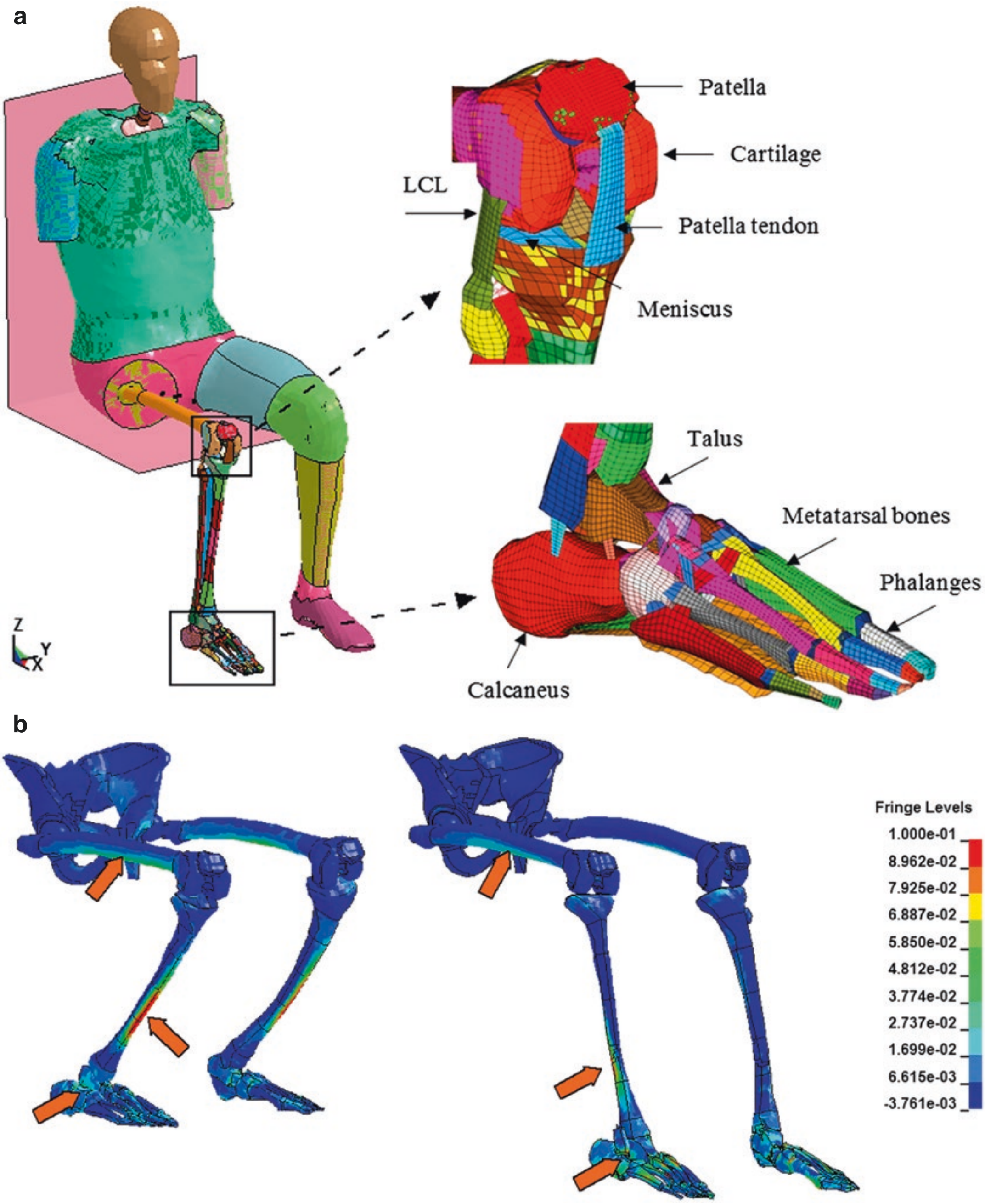

Fig. 28.3 Finite element model of a seated vehicle occupant able to predict injury to the lower limb from floor intrusion due to an under-vehicle explosion. (**a**) Overview of the model with an enlarged view of the foot and knee. (**b**) Stress distributions on the bones for two potential seated postures. The location of injury can be predicted from the value of stress at a location. Reprinted from [3] with permission

28.6 Defining the Parameters of Energy-Attenuating Vehicle Seats

Injury to the pelvis, spinal column and head is common in underbody blast. Energy-attenuating seats have been considered to mitigate injury to the upper body and deployed in current vehicles. These seats utilise stroking mechanisms to absorb energy and so reduce the load transferred to the pelvis of the occupant. In the example presented here, an FE human body model was utilised to determine seat parameters that could aid in reducing pelvic and torso injuries [4]. The MADYMO (Tass International) human body model was adapted to simulate an experimental setup of underbody blast that utilised a rigid seat [5]. The human body model was calibrated against a non-injurious and an injurious load case to result in acceptable correlations to the experimental data. The calibrated model was subsequently used in combination with an optimisation algorithm, to determine the optimal combination of the force-deflection characteristics of an energy-attenuating seat in order to reduce the injury to the pelvis and lumbar spine, as measured by the acceleration and force signals seen in the pelvis and the T12 vertebral body. Simulations were run for the human body with and without a vest in nominal and reclined postures showing a significant reduction in the resultant vertical speed of the seat after optimisation of its parameters. This study shows the utility of computational models of the human body as tools for improving the design of mitigation systems.

28.7 Conclusion

Computational models of primary, secondary and tertiary blast have been devised that vary in complexity to a great extent. Appropriate levels of complexity can be implemented using computational approaches, and these have demonstrated utility in applications such as the prediction of injury and the design of mitigation.

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29 In-Silico Modelling of Blast-Induced Heterotopic Ossification

Martin Ramette and Anthony M. J. Bull

Abstract

Computational modelling of how ectopic bone forms following blast injury can predict both the location and severity of the aberrant bone. As a loading-mediated process, this modelling uses the principle of Wolff's Law combined with mechanobiological theory to investigate the effect of loading during treatment and rehabilitation. Small changes to the loading can significantly change the volume and type of ectopic bone formed and opens the way for future mechanical therapies.

29.1 Background

Initially described during the American Civil War [1, 2], heterotopic ossification (HO) refers to the aberrant formation of ectopic bone in non-osseous tissues and is a debilitating disorder of the skeletal system. This generic condition can be linked to a wide range of genetic, neurologic and traumatic causes with equally diverse prevalence rates (see Chap. 23). However, this chapter will focus on what has grown to become a "military epidemic" [3], the frequently occurring HO developing among military casualties. More specifically, since the turn of the century, clinical studies have reported an HO occurrence rate ranging from 57% to 91% in the US and UK blast-related lower-limb amputee cohorts [4–6]. Additionally, a blast injury mechanism, amputation through the zone of injury, associated neurologic lesions and higher injury severity were all identified as risk factors in combat-related injuries [4, 7]. For the affected military personnel, HO often represents a severe disruption of their crucial rehabilitation process as it restricts range of motion, provokes pain due to neuromuscular entrapments, causes skin ulceration and prosthesis-fitting issues [1–3]. Although generic radiation and NSAIDs-based approaches can be used to treat HO, these remain controversial among clinicians and are largely contraindicated in the military setting. No satisfactory treatment currently exists for blast-related amputees (see Chap. 24 for further details about the current therapeutic strategies). Up to 41% of trans-femoral amputees require surgical excision of the ectopic bone [8, 9], a technically challenging surgical intervention that has associated induced morbidity [10]. Figure 29.1 illustrates the severity and the clinical conundrum that HO today represents for a trans-femoral amputee.

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Fig. 29.1 Clinical manifestations of HO in blast-related military amputees (**a**) coronal plane radiograph of a severe case of HO (**b**) photograph of underlying HO causing skin ulceration in the stump. Both images reproduced with permission from Potter et al. [9]

Although HO's exact pathophysiology remains elusive, there is a general consensus on the cascade of events that it follows, in a process that resembles aberrant fracture healing [11, 12]. The initial trauma leads to a dysfunctional hyper-inflammatory response [13, 14] that triggers the abnormal recruitment and differentiation of available progenitor cells, which in turn begin to deposit a calcified matrix within the traumatised soft tissue [9, 15]. This soft tissue has been shown to have osteoinductive properties [9]. The ectopic bone then progressively matures and remodels, forming an intricate structure whose exact nature remains unclear, but which seems to be a disorganised heterogeneous mixture of cortical and trabecular bone [16]. Furthermore, the newly deposited ectopic bone displays a higher level of metabolic activity than its skeletal counterpart [17].

Over the recent years, extensive research has investigated the biological aspects of this dysregulated chain of events, shedding new light on the exact signalling and differentiation pathways triggered by the initial traumatic injury, along with the origin of the progenitor cells involved in the process (see Chap. 24 for a complete review). The details of those findings are beyond the scope of this chapter. Meanwhile, other important aspects of the disease remain poorly understood, among which is the putative impact of the mechanical environment on HO. This situation is somewhat paradoxical given its established significance in healthy bone remodelling and fracture healing. The next section will underline the elements that suggest a mechanical mediation in the initiation and growth of HO.

29.2 The Effect of the Mechanical Environment on HO

While it has been reported that various sorts of stem cells intervene in the development of traumatic HO, it is commonly accepted that mechanics influences the differentiation of these cells in a dual way. Firstly, the interactions with the extracellular matrix, governed mostly by the substrate stiffness and surface topography, dictate which differentiation pathway is followed by multipotent connective tissue progenitor cells [18, 19]. Secondly, an increasing mechanical stimulus imposed upon stem cells induces the recruitment and differentiation of other available osteogenic progenitors [20, 21]. This mechanically driven response has been specifically observed in ligament cells harvested from patients suffering from an ossification of the posterior longitudinal ligament (OPLL), an ectopic bone formation in the spinal ligaments [22]. At the cellular level, HO seems to follow the same general patterns as skeletal bone in its response to mechanical loading.

Tissue-level manifestations of HO also provide arguments to support the importance of the mechanical environment in its initiation and development. An early study was able to initiate the development of HO in the quadriceps of rabbits through forced manipulations [23], causing some clinicians to advise against intense rehabilitation of affected limbs [24]. More recently, researchers have uncovered a correlation between the mechanical design of spinal disc prostheses and the type of HO affecting patients after Total Disc Replacement (TDR) with the ectopic bone formation appearing to compensate for the modified range of motion in the spine [25]. Figure 29.2 highlights these different types of HO after TDR.

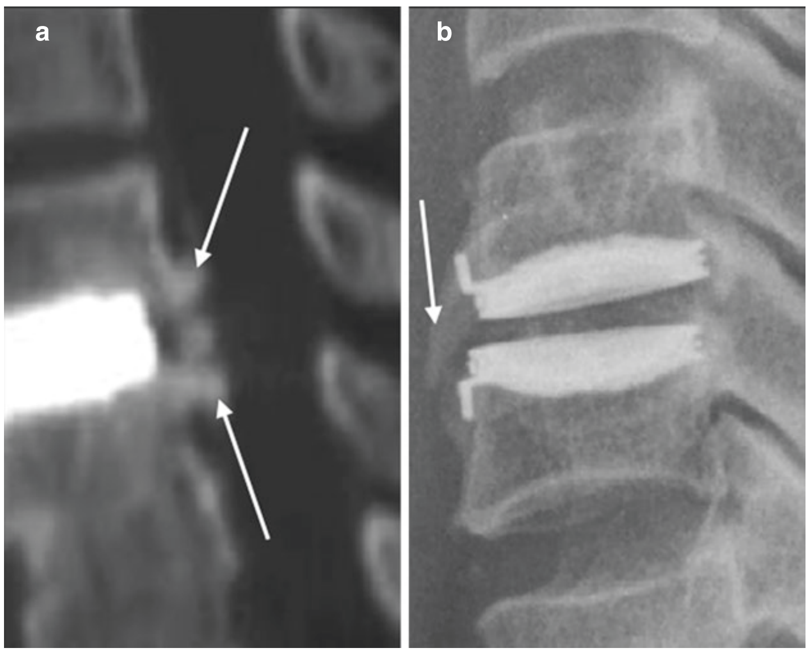

Fig. 29.2 Clinical observations of HO (white arrows) after TDR (**a**) CT-scan showing "end-plate" HO spreading in the posterior disc space (**b**) X-ray of "traction spur" HO in the anterior disc space. Both images adapted with permission from Jin et al. [25]

Altogether, this microscopic and macroscopic evidence supports the hypothesis that HO results in part from dysregulated biological processes, and in another part from an aberrant mechanical environment. Biological processes are widely investigated, yet the effect of the mechanical environment is under explored. A mechanical understanding will provide clinicians with further options to prevent, mitigate and ultimately treat this debilitating condition. For instance, such new knowledge could inform the design of prosthesis sockets to avoid putting amputees at a "mechanical risk" of developing HO or help conceive mechanical therapies to manage the disease once it is present in the stump, throughout rehabilitation.

29.3 Computational Bone Remodelling Applied to Heterotopic Ossification

An in silico approach is well-suited to investigate the impact of the mechanical environment on blast-induced HO. The advantages are that the scale effects of animal models, even large animal models (which do not, to-date, exist), are completed removed, and simulation-based approaches are, on the other hand, more cost and time effective. Also, once validated, in silico models have the ability to isolate key parameters and test out various scenarios, something that cannot be done in the same way in clinical studies given the lack of knowledge and the multi-factorial aspect of HO. Indeed, such studies are confounded by having too few subjects and too many uncontrollable variables.

Given the indications that the mechanical environment plays a crucial role in the mechanisms of HO and many benefits offered by an in silico approach, the methods of computational bone remodelling have been used in the analysis of the formation and development of HO.

29.3.1 Computational Bone Remodelling

The aim of computational bone remodelling is to transcribe the commonly accepted Wolff's law—stating that bone remodels in response to the loads that are exerted onto it—into mathematical equations describing the bone apposition and resorption mechanisms [26]. Figure 29.3 illustrates the principles linking mechanical loading and bone micro-architecture, with the example of the proximal femur [27, 28]. Those equations are integrated into adaptive finite element (FE) models that are consequently capable of simulating bone structural adaptation over time. More specifically, at any given instant and location, the FE model predicts a local mechanical stimulus which is compared to a reference value. The local bone density—and therefore its local mechanical properties—is then updated based on this comparison, mimicking either resorption or apposition, and the process can be iteratively repeated numerous times to go from a local optimisation to a macroscopic tissue adaptation simulation, as schematically represented in Fig. 29.4. This general method has been successful in mimicking the internal organisation of the proximal femur [29, 30] and vertebrae [31] callus bone formation in fracture healing [32] or stress shielding around implants [33], among other healthy bone remodelling examples.

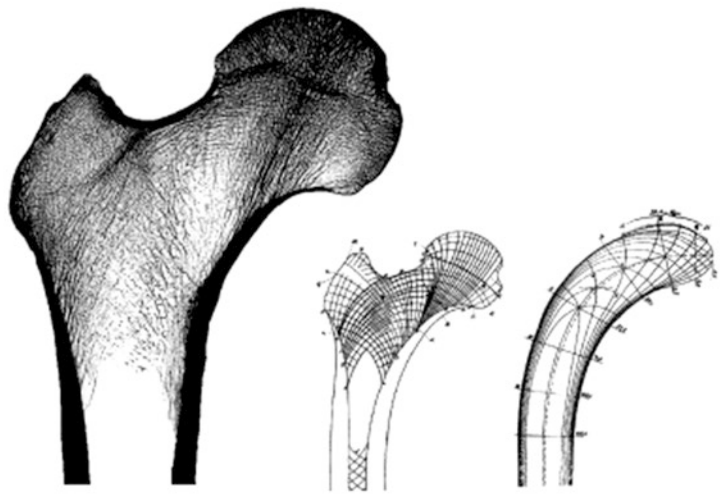

Fig. 29.3 Comparison between the normal trabecular architecture in the proximal femur, as seen in a radiograph and an anatomist drawing, and the principal stresses in a crane under load model. Reproduced with permission from Fratzl and Weinmaker [27], with parts from Wolff [28]

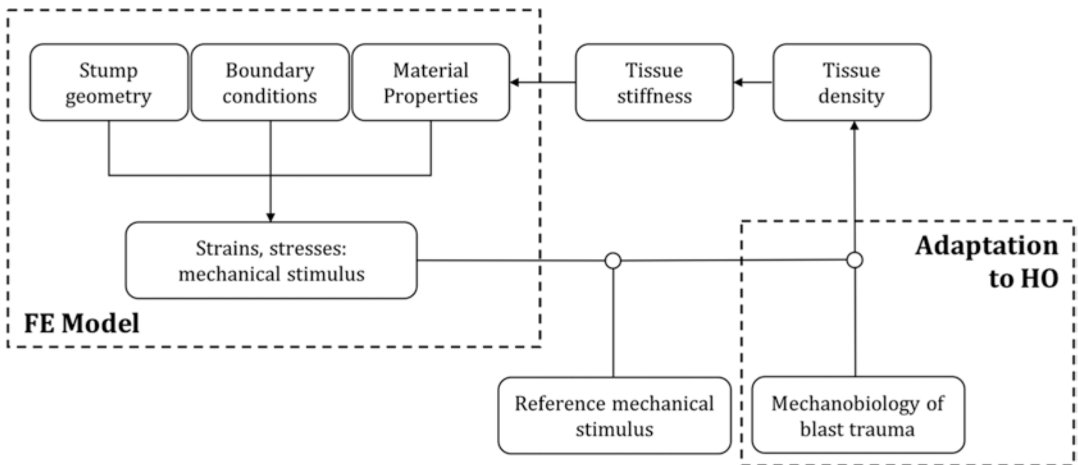

Fig. 29.4 Simplified flowchart of the general iterative mechanism driving the adaptive finite element approach to bone remodelling, and the novelty introduced to apply it to HO. Redrawn and adapted from Weinans et al. [29]

Upon establishing a computational bone remodelling model, the key parameters to consider are the driving mechanical stimulus (often linked to the strain-energy density [29, 30, 34, 35]), the dynamics of the relationship between the stimulus and the bone density rate of change [31, 35] and the properties of the associated FE model (geometry, material properties, boundary conditions, etc.).

29.3.2 Application to Heterotopic Ossification

Computational bone remodelling is now a well-established approach to most cases of "normal" load-driven adaption, yet ectopic bone formation presents a particular challenge in that it is not only bone that remodels, but it is a progressive calcification process that extends within the surrounding soft tissue. This tissue transitioning from one "state" to another calls for an adaptation of the methods presented in the previous section. On top of the usual adaptive FE and feedback mechanism, Rosenberg and Bull developed a modelling strategy aiming to include the mechanobiological specificities of blast-induced HO, introduced in Sect. 29.1. The distance to the zone of trauma, the density of mesenchymal stem cells within the soft tissue and the substrate stiffness are taken into account to represent the likelihood of a region becoming osteogenic [34, 36]. More specifically, these variables are combined in a single probability factor that acts as a regulator of a zone's level of remodelling; ectopic bone should form only in relatively stiff (i.e. osteogenic) regions with a relatively elevated density of progenitor cells, and when one of those two factors is high then the rate and extent of HO should be more sensitive to the other one. At the same time, remodelling should constantly happen in normal bone tissue. Those physiological considerations led to the relationship detailed in Eq. (29.1), defining β the remodelling regulating factor as a specific power law [34]. $P(E)$ is a probability function of the Young's Modulus, characterising the substrate stiffness, while ρ_{MSC} stands for the density of mesenchymal stem cells and is expressed as a function of the distance to the injury point. The choice of this power law ensures that in bone, where $P(E)$ is equal to 1, then β is always equal to 1 and therefore normal bone remodelling applies, while in soft tissue low values for ρ_{MSC} or $P(E)$ are enough to drive β close to 0 and prevent the formation of ectopic bone.

$$\beta = P(E)^{\frac{1}{1.5*\rho_{MSC}}} \tag{29.1}$$

The modified computational remodelling approach was tested on a plane stress square plate, as is usual in this field of research, and was validated against a previously existing model [34]. It was then applied to two-dimensional FE models of trans-femoral amputee stumps whose geometry and material properties were adapted from the literature and that included the distal femur surrounded by soft tissue, the skin, the prosthesis liner and socket [37]. The boundary conditions were derived from hip joint contact forces and adductor muscle forces, and applied onto the proximal femur. Various loading orientations were also implemented, to represent upright standing situations as well as adduction and abduction cases [37].

Further improvements were also introduced to the dynamics of the adaptation rule between the bone density rate of change and the mechanical stimulus. Indeed, contrarily to what is usually done and as shown in Fig. 29.5, this rule was modified to plateau beyond certain stimulus thresholds, thereby implementing the natural limits that are thought to restrict the bone resorption and apposition processes [35, 38]. This "saturation effect" was intended to avoid sudden, large and ultimately unrealistic evolutions of the density distributions in the stump models.

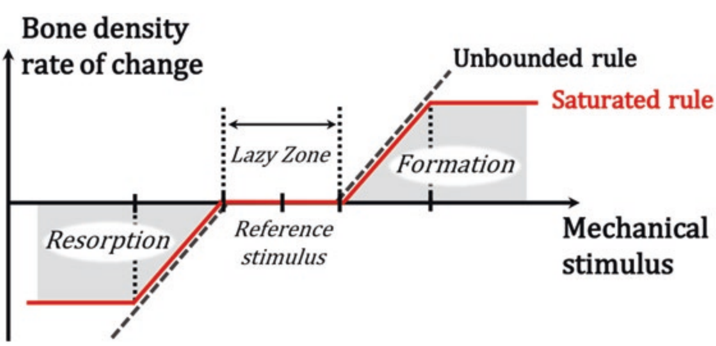

Fig. 29.5 Schematic representation of the relationship linking the bone density rate of change to the mechanical stimulus. The dashed line represents the rule commonly used in the literature. The solid line illustrates the novel "saturated" approach, where the bone formation and resorption processes are restricted

29.4 Simulating the Formation of Characteristic HO Morphologies

Adapting the general methods of computational bone remodelling to the specificities of blast-related HO yielded satisfactory results since it led to realistic ectopic bone formations within the modelled trans-femoral stumps. Indeed, by monitoring the evolution of the tissues' stiffnesses, radiograph-like images are produced and can then be compared with clinical reports of HO amputee patients reported in the literature. Those comparisons demonstrated the model's capabilities to simulate the formation of characteristic HO morphologies [37], following the classification introduced by Evririades et al. [39]. What is more, as illustrated by Fig. 29.6, the model established that adjusting the loading environment on the stump determined the type of HO that was consequently happening: forces representative of an upright standing situation led to a characteristic medial-facing hook shape, while abducted gait-like conditions caused a beetle shell HO [37].

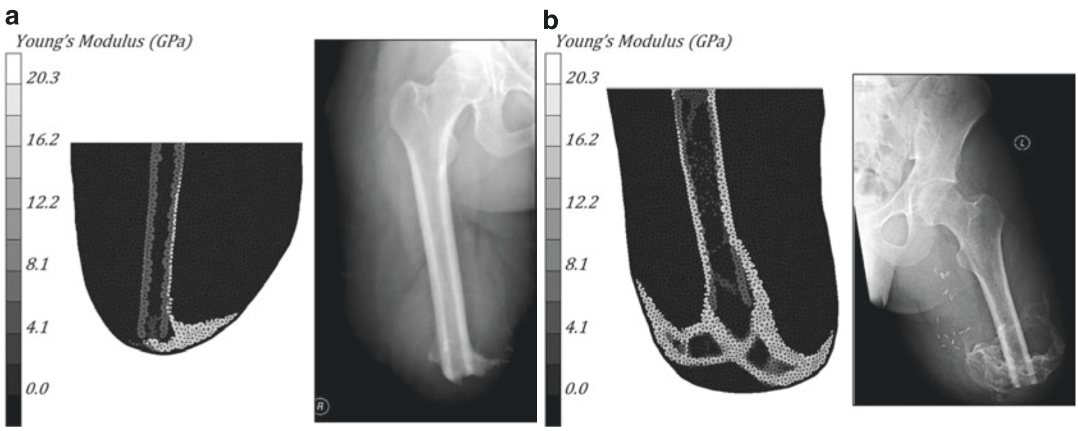

Fig. 29.6 Comparisons between the final stiffness distributions predicted by various stump models and radiographs of HO in military amputees (**a**) Upright standing loading scenario leading to a medial hook HO (**b**) 30° abducted load case showing a beetle shell type of HO. X-rays reproduced from Rosenberg [36], provided courtesy of Dafydd S. Edwards (Imperial College London) as part of his MD(Res) research

From a methodological perspective, the implementation of the "saturation effect" in the bone adaptation rule yielded a more progressive ossification pattern in the early phases of the simulation, thereby improving the physiological fidelity of the model. Indeed, the modelled soft tissue shows a gradual calcification that approximately resembles endochondral ossification, a process that is thought to underlie the development of HO in amputees [1]. As a result, the model can be used to investigate the whole pathological pathway of bone formation and not only its mature endpoint.

29.5 Discussion and Future Perspective

The model's ability to simulate the formation of characteristic clinical morphologies of HO substantiates its approach combining the mechanical predictions of an FE model with the inflammation-representing β factor. Furthermore, the observed correlation between the loading orientation on the stump and the resulting type of HO seems to confirm the disease's hypothesised mechanical mediation, a point also supported by the crucial roles played in the model by the skin stiffness and the simulation of negative pressure wound therapy [37]. In terms of ossification dynamics, the implementation of the "saturation effect" allows the model to reproduce the general process of a calcified matrix being deposited within the traumatised soft tissue and then maturing and remodelling over time. Figure 29.7 reproduces a rare series of radiographs in a trans-femoral amputee showing this progressive process, from Davis et al. [15].

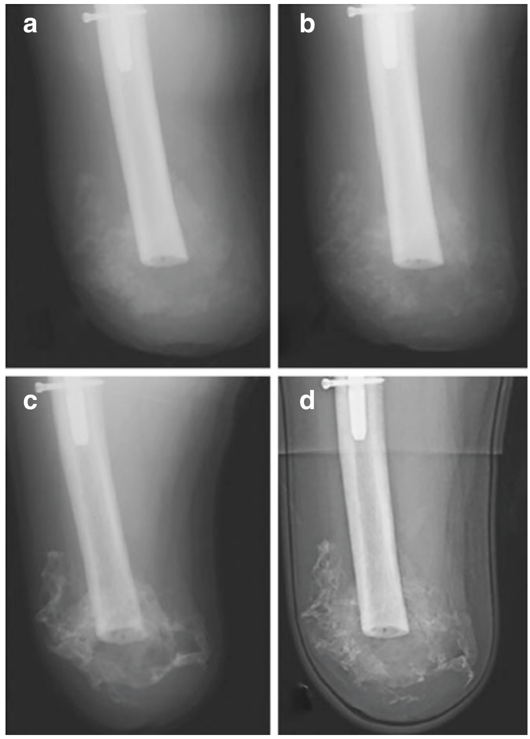

Fig. 29.7 A series of radiographs demonstrating the progression of HO in a military trans-femoral amputee, respectively, obtained after (**a**) 4 weeks, (**b**) 5 weeks, (**c**) 7 weeks, and (**d**) 20 weeks after injury. Reproduced with permission from Davis et al. [15]

Despite these encouraging similitudes between the model's simulations and clinical images, it is important to note that these comparisons do not constitute a complete validation at this point: a subject-specific anatomy, loading, and medical imaging of the HO progression would be required. Besides, the model in its current state does not offer predictions about a particular individual but is rather a scientific tool that can help better comprehend the reasons behind HO's formation. Its adaptation to an MRI-derived 3D stump model will help implement more realistic geometries and boundary conditions and will offer further verifications on the preliminary findings outlined in this chapter. Beyond these, it will also enable the investigation of clinically relevant scenarios ranging from surgical techniques (including the effect of muscle myodesis), rehabilitation strategies (how early and aggressive should it be?) and prosthesis design. Subject-specific applications could then also be developed.

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Physical Experimental Apparatus for Modelling Blast

30

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Abstract

Blast injuries inflicted by an explosive event are complex and difficult to characterise. There is a range of injury mechanisms which includes the direct insult of the blast overpressure, penetration by fragments and debris, and blunt or crush trauma induced by displacement of the body. These injury mechanisms are a result of loading at rates often beyond those that conventional testing platforms are designed to deliver, thus bespoke and well-controlled experimental devices are essential to recreate in the laboratory environment these injury mechanisms and reproduce the injury outcomes. This chapter summarises a range of physical experimental devices that have been developed for studying various blast injuries. They are classified into separate time-dependent processes of a blast loading: the primary, secondary, and tertiary blast effects.

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30.1 Introduction

The complex nature of explosive events results in injuries that may be due to multiple blast-injury mechanisms (Chap. 9). To understand the pathophysiology of blast injury, these blast-injury mechanisms need to be decoupled and re-created in well-controlled laboratory environments. This requires experimental capacity outside the loading ranges of most conventional machines as well as the ability to tune individual loading parameters to a desired blast scenario. This chapter presents the current range of platforms employed in modelling blast effects and their development and discusses their application to study blast injuries. The physical apparatus for modelling blast can be classified according to the different blast-injury mechanisms. Primary blast can be delivered by a shock tube, the Hopkinson pressure bar, a laser-induced stress wave apparatus, and by a controlled free-field explosion. Secondary blast can be simulated with a gas gun, a shock tube coupled with fragment-simulating projectiles, or by using modified rifles. Tertiary blast can be modelled with conventional drop towers or bespoke platforms such as an accelerating horizontal sled, a pendulum striker, or a vertically loading platform. The versatility of the devices is essential to ensure adaptability for specific in vivo, ex vivo, and in vitro models of blast injury, and the boundary conditions are the key to designing meaningful and biofidelic experiments

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which can be used to quantify the pathophysiology of blast injury or to test computational models for validation.

30.2 Primary Blast

Primary blast injury is caused by the shock wave from the explosion. During the explosion, the pressure of the surrounding media can reach hundreds of kilobar inside the fire ball, which dramatically drops to tens of bar just outside the fire ball and decays with distance from the source to the ambient pressure [1]. The shock wave may be especially damaging to air-filled organs such as the lungs, ears, the gastrointestinal tracts, and the brain. The threshold for lung injury is shown to be as low as 1 bar peak pressure [2]. The duration of the pulse and the number of exposures can also play a role in the injury outcome [3]. Primary blast is thought also to cause microfractures in bone [4] as well as contribute to the pathway resulting in heterotopic ossification (HO) [5, 6].

30.2.1 The Shock Tube

The shock tube is perhaps the most conventional apparatus for generating the pressure loading of the blast wave at a distance from the source of the explosion. It is capable of generating a tailored blast load in a well-controlled fashion to investigate the effects of different components (such as peak pressure, duration, and impulse) of the blast wave on the body. The Imperial College system shown here (Fig. 30.1a) is a stainless steel, 3.8 m long, air-driven shock tube with a 60 mm internal bore. It involves a driver section which is charged to a desired high pressure regulated by diaphragms in the diaphragm assembly, and a driven section remaining at ambient pressure. When the diaphragms rupture, the high- and low-pressure regions are put in contact, generating a blast wave that travels along the driven section and acts on the sample mounted in the adaptor at the end of the shock tube. The output blast pressure profile can be tailored with the diaphragm thickness, the length of the high-pressure air volume, and additional insertions such as granular beds, perforated plates, and foam [7–10]. The presented system can replicate various desired blast loading scenarios such as open-field air blast (Friedlander waveform), partially confined blast, and fully confined blast, with magnitude between 0.5 and 10 bar (Fig. 30.1b) [11].

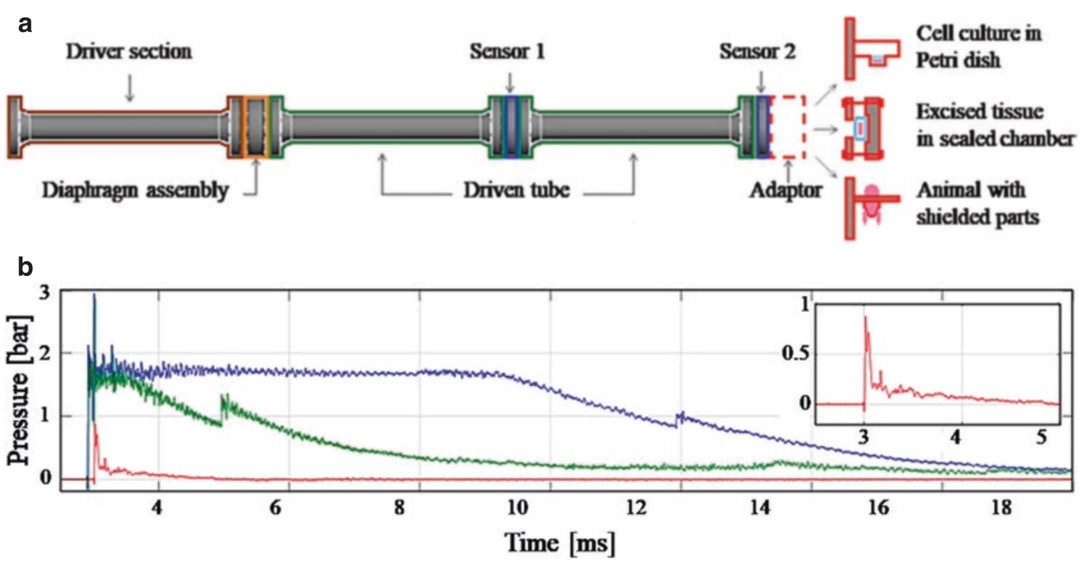

Fig. 30.1 (**a**) Schematic of shock tube with adaptors for in vitro, ex vivo, and in vivo studies. (**b**) Examples of different blast loading profiles produced by the shock tube. Adapted from Nguyen et al. [11]

Adaptors can be used to study the primary blast effects on biological samples in a multitude of scales such as whole animal models, excised tissues and organs, and cell cultures (Fig. 30.1a). Applications with a whole animal model include investigating the effect of the pulse duration on acute inflammation on a rodent limb where other body parts were shielded from unwanted complication [12]; studying the microstructural changes in significant focal injuries to rodent lungs when subjected to different pressure magnitudes [13]; investigating the effect of lower limb flail on pelvis displacement and vascular injury in a rodent model [14]; and the effect of blast exposure on the formation of post-traumatic HO in a rat model [15]. The shock tube also has applications on in vitro models of tissue and cell cultures such as the study of blast-related traumatic brain injury using rodent organotypic brain-slice culture [16, 17] and NG108–15 rat glioblastoma hybrid cells or SH-SY5Y human neuroblastoma cell cultures [18]; and investigations on the pathology of HO with in vitro cell cultures [19, 20].

30.2.2 The Split-Hopkinson Pressure Bar (SHPB)

The split-Hopkinson pressure bar (SHPB) is another conventional loading device that has been successfully modified for studying the effects of blast loading on biological tissues and engineering components. An SHPB system generally consists of three long cylindrical bars termed projectile, input bar (IB), and output bar (OB) (Fig. 30.2). Different pulses can be achieved depending on the arrangement, the geometrical and material properties of the bars, the different modes of loading (compression, tension, or torsion), and the loading conditions. In an SHPB compression test, a specimen is coaxially held in place between the IB and the OB. Upon loading, the projectile is launched on the free end of the IB generating a longitudinal stress wave that travels at the speed of sound in the material. An SHPB system typically exposes samples to loads of strain rates in the range of 10^3–10^4 s^{-1} and with tuneable time durations between hundreds of microseconds and tens of milliseconds. One of the advantages of the SHPB lies in the ability to produce closely the sample loading conditions as they are recorded in situ during blast loading.

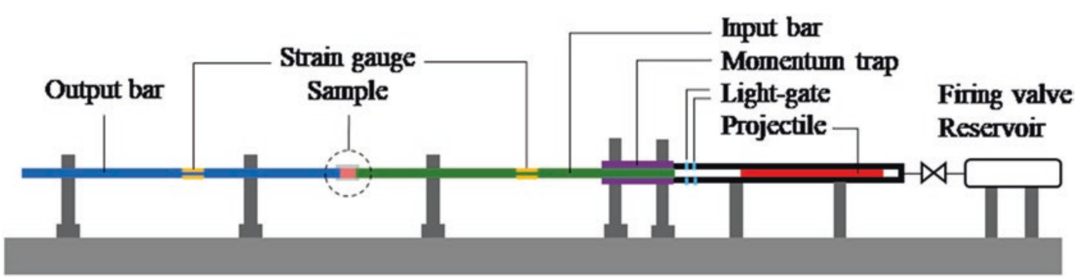

Fig. 30.2 Schematic of the split-Hopkinson pressure bar (SHPB)

A body of literature has described the use of SHPB to understand the mechanical response of biological tissues such as skin, muscle, bone, ligament, and brain tissue at strain rates relevant to blast loading [21–27]. The device can also be adapted to study the injury mechanism of an organ by the blast wave. Bustamante et al. reported the use of a modified SHPB to investigate the damage by negative blast pressure to the brain tissue due to cavitation bubble collapse [28]. In addition, the Imperial College SHPB system has been amended to study the effect of blast loading on live biological materials, especially cells. Nienaber et al. designed an aluminium cylinder in which was inserted a piston supporting neuronal cells immobilised on a coverslip [29]. Bo and colleagues developed a versatile confinement chamber enabling the mechanical loading of various cell types including mesenchymal stromal cells (MSCs) and immune cells in suspension as well as on 2D monolayer [30, 31].

30.2.3 Other Devices

There is a range of other devices which have been developed to study primary blast injuries. The approach may involve the use of explosives such as in free-field blast testing where high explosives are detonated in the open field [32–34] or in a blast tube where a small charge is used in a partially confined chamber (Fig. 30.3a) [33, 35]. While the blast effects produced by these methods are realistic, it is challenging to separate the different effects of the blast wave, especially the quaternary blast effects caused by the emissions of gas and smoke, and to limit injuries to a desired body part or organ/tissue. Other systems include a laser-induced shock/stress wave apparatus (Fig. 30.3b) to study traumatic brain injury [36–38] and damage to cells [39, 40], and the high-pressure water jet apparatus to study injury on lung tissues [41].

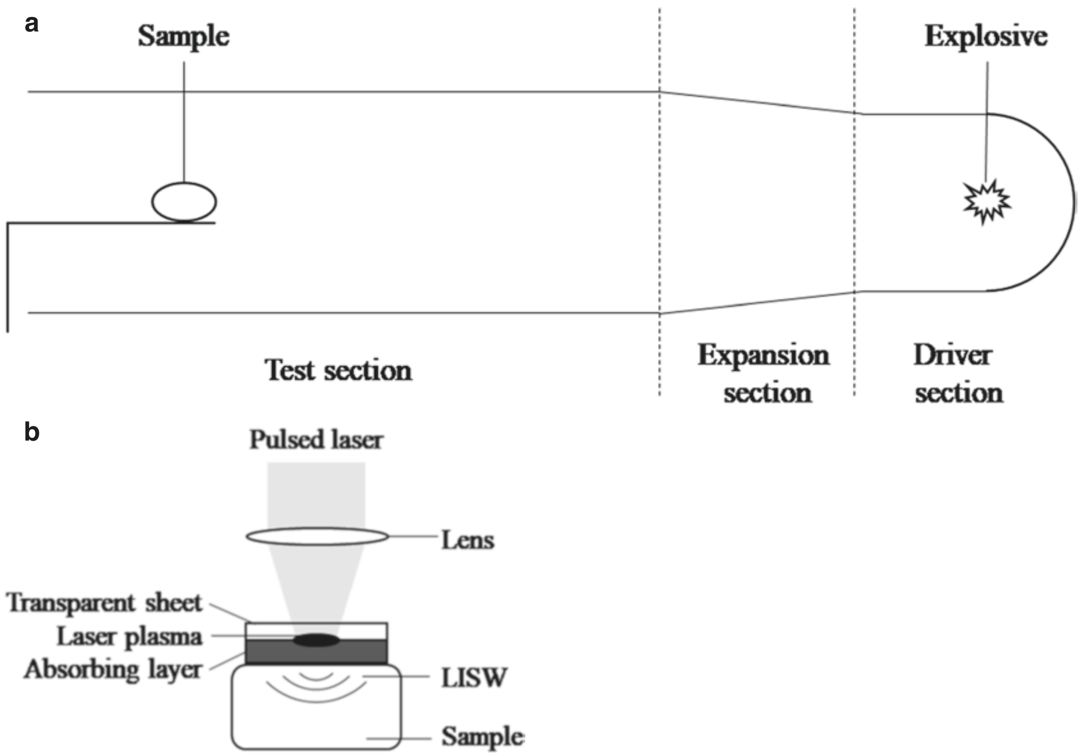

Fig. 30.3 Experimental approaches for primary blast injuries: (**a**) the blast tube [35]; (**b**) a laser-induced stress wave (LISW) apparatus [36]

30.3 Secondary Blast

Secondary blast injury occurs due to the penetration of fragments produced by an explosion (Chap. 9). The explosion generates and accelerates fragments such as glass, masonry, soil ejecta to speeds in excess of 1000 m/s, which then typically decelerate to speeds in the order of 600 m/s when striking victims [42]. This injury is the most common in modern conflict, especially to the lower extremities and face [43]. Typical velocities of fragments in surviving casualties is less than 600 m/s [44], and the most common metal blast fragments recovered are cylindrical (51%—0.78 g), spherical (27%—0.44 g), and stellate (10%—0.37 g) [45]. Multiple impacts by soil ejecta are also common, especially in casualties that present with traumatic amputation [46]. The inflicted injury depends on a range of factors such as velocity, mass and shape of the fragments, personal protective armour, and the posture of the casualty at the time of insult. A well-controlled and systematic experimental model is essential to study the mechanism of these injuries.

30.3.1 Gas Gun

The light gas gun system is commonly used for testing mechanical behaviours of materials [47, 48]. The Imperial College design presented here has been adapted to study penetrating injuries by blast fragments. Figure 30.4a is a schematic of a 32 mm-bore single-stage gas gun system whose breech can be charged up to 200 bar-litre with compressed air or helium. The system uses a double-diaphragm mechanism for firing and controlling the speed of the projectile. The prime section between the two diaphragms is charged to a low pressure, whereas the reservoir section behind the first diaphragm is charged to the desired firing pressure so that neither diaphragms are subjected to a pressure balance higher than their burst pressure. For firing, the pressure in the prime section is vented to break the pressure balance and rupture the diaphragms; the high-pressure gas then accelerates the sabot through the 3 m-long barrel to reach the desired speed as it enters the target chamber. The output velocity of the sabot is proportional to the reservoir pressure which can be controlled by the thickness of the diaphragms used. The presented system can achieve representative projectile velocities between 50 and 600 m/s.

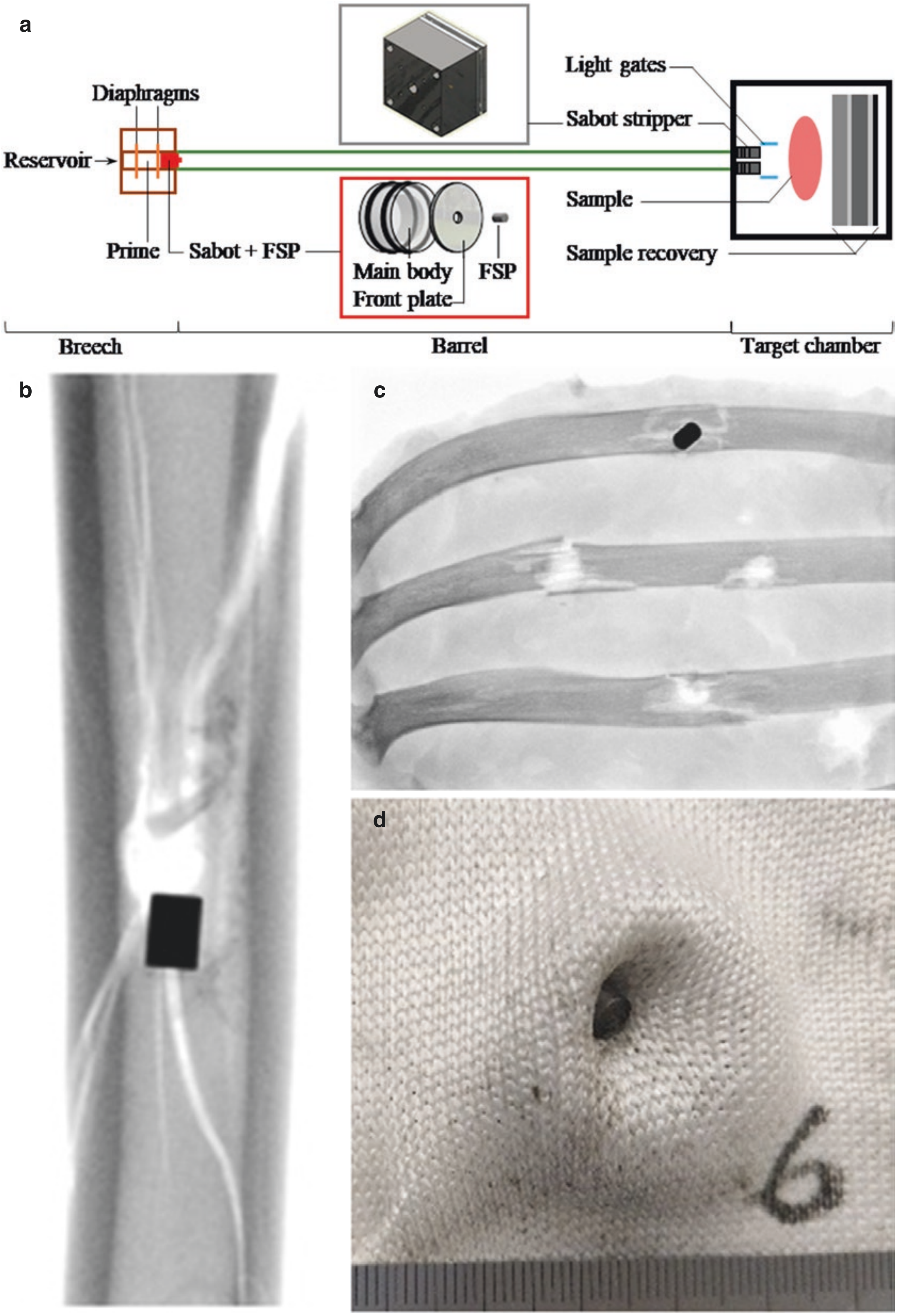

Fig. 30.4 (**a**) Schematic of the gas gun system. Examples of impacts with a cylindrical fragment-simulating projectile: (**b**) X-ray of an ovine tibia, (**c**) X-ray of a porcine rib cage, and (**d**) the FSP stopped by a ballistic-resistant material

When used to study fragment penetrating injuries, the system needs to accommodate fragment-simulating projectiles (FSPs) of various shapes and sizes, whose dimensions are often much smaller than the bore of the barrel and may be of irregular shapes, such as gravel and soil ejecta. For this purpose, the sabot may be designed with an add-on aluminium front plate to house the needed FSP in the centre. At the entrance of the target chamber, the sabot and FSP are separated from each other by a sabot stripper construction with a bore in the middle.

The gas gun system can be used to investigate the mechanism and risk of injury by different fragments to organs such as the rib cage [49] and the lower limb [50] (Fig. 30.4b, c). It can also be used to assess the performance of protective equipment against penetrating threats by fragments (Fig. 30.4d) [51]. In these studies, ensuring appropriate biofidelic boundary conditions such as pre-compressing the samples, the orientation and mounting of the sample, and the contact of samples with an appropriate surrounding medium are essential for the applicability of the results.

30.3.2 Other Devices

There are various other experimental approaches to study penetrating injuries by fragments. Hayda et al. [52] reported the use of the large-scale shock tube coupled with a glass panel to study the penetration of glass shards into ballistic gelatine covered with chamois skin simulant (Fig. 30.5a). Although the glass fragments created by the blast wave resulted in multiple penetrations, the poor energy coupling between the blast wave generated by the shock tube and the glass shards gave only 8–35 m/s of impact velocities. Other approaches involve a modified rifle with adapting barrels where the FSP is housed in a split sabot [53–55] (Fig. 30.5b) or with a secondary projectile attached to a small pin [56] (Fig. 30.5c). The first approach provides realistic movements of projectile, such as tumbling and yawing, but has large variations between shots and is not well-controlled. The latter approach, whilst shown to be useful in recreating the high percentage of animals that survive high-velocity penetrating traumatic brain injury, deviates from fragment penetration and limits penetration depth and impact velocity (<100 m/s). These experimental approaches are also challenging to adapt to different types of fragments, especially soil ejecta.

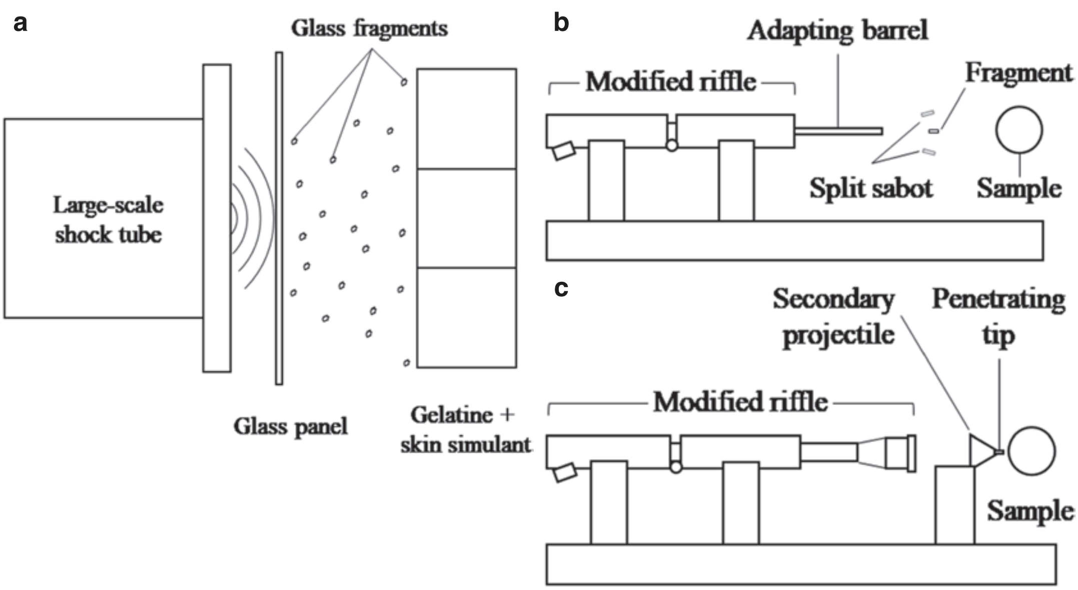

Fig. 30.5 Studies of fragment penetrating injuries: (**a**) glass fragment penetration study [52]; (**b**) modified rifle with fitted adapting barrel and split sabot [53, 54]; and (**c**) modified air rifle with secondary projectile housing penetrating tip [56]

30.4 Tertiary Blast

Tertiary blast injuries are caused when a person is physically displaced by the force of the peak overpressure and blast wind, and sustains blunt trauma (Chap. 9). Crush injuries and various injuries from collapsing buildings or surrounding structures are also included in this group [57]. Another form of tertiary blast injury termed deck-slap or solid-blast injury occurs when a structure is displaced against the human body due to an explosion. This is the most significant injury mechanism for occupants of vehicles when attacked by an explosive device and can occur at different times after the explosion [58, 59]. The vertical acceleration of the vehicle can cause lower extremity, pelvis, and spinal column injuries. The occupant may subsequently incur head and spinal column injuries as they are thrown around inside or ejected from the vehicle [58]. Afterwards, as the vehicle accelerates downwards and impacts the ground, the occupants are again subjected to further blunt or crush injuries [59]. In under-body blast (UBB) events, 96% of lower extremity casualties reported were due to solid blast [4] with the fractures observed being consistent with an axial loading mechanism. The energy involved in this process can accelerate human body to tens of m/s in milliseconds [60].

30.4.1 Drop Towers

The Gardner (or drop-weight) impact test is a traditional method for evaluating the impact strength of materials. It is characterised by the vertical dropping of a mass from a height which strikes a specimen mounted on a base plate or anvil (Fig. 30.6a). The impact energy is dependent on

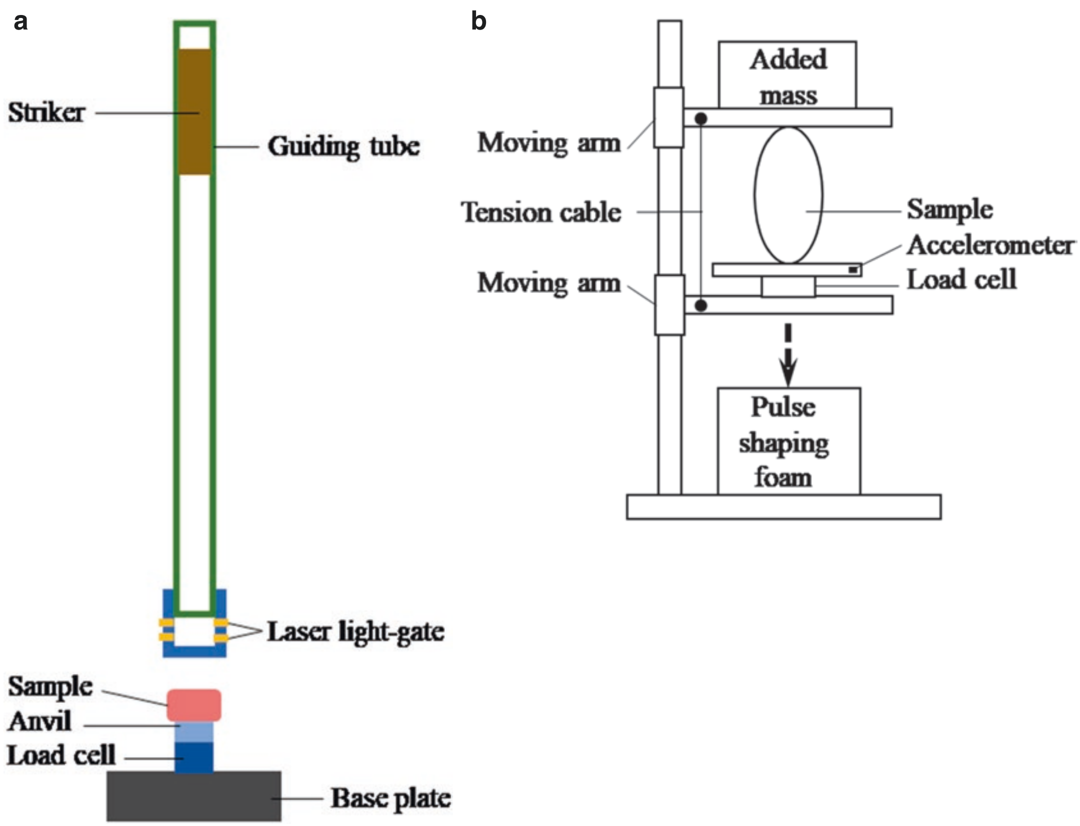

Fig. 30.6 (**a**) Simplified schematic of a typical drop-weight system. (**b**) Modified drop tower with quantified compression rates [61]

the drop height and mass. There also exist spring-assisted drop towers which utilise the energy stored in springs when compressed to achieve higher speeds at impact than what height and mass alone can achieve. This platform can be used to study injuries by blunt impact, characterise tissue at high strain rates, and investigate the impact performance of protective equipment.

The drop tower can be modified to study blunt and crush injuries to specific anatomical region of the human body such as the one reported by Stemper et al. [61] to quantify compression rates during realistic loading and compressive tolerance of the human vertebrae (Fig. 30.6b). Another example is the modified drop-weight rig for post-mortem human surrogates (PMHS) testing used by Henderson et al. [62] to simulate the loading environment for in-vehicle occupants. Bonner et al. [63] also developed a spring-assisted drop tower to investigate the strain-rate sensitivity of the lateral collateral ligament of the knee in the region of 10 and 100 s^{-1}.

30.4.2 Solid-Blast Injury Simulators

Numerous bespoke devices have been developed in recent years to study solid-blast injuries. They have been designed with the aim to deliver the acceleration pulses seen in UBB to whole PMHS or specific body parts. Some simulators mount the specimen in the vertical [64–67] and some in the horizontal direction [68–71]. These platforms are generally required to be capable of accelerating a mass to at least 12 m/s within less than 10 ms [72, 73]. The biomechanical responses of the relevant body regions can be extracted using sensors such as accelerometers, strain gauges, and load cells attached at specific locations on the specimen.

Vertical solid-blast injury simulators can be driven by a scaled explosive charge to propel a unit of deformable floor with seats which simulate the environment inside the vehicle (Fig. 30.7a). Examples of such platforms are the Accelerative Loading Fixture (ALF) [64] which has been used to compare Anthropomorphic Test Devices (ATD) to PMHS response in UBB as well as the response of female to male PMHS [74], and the Test Rig for Occupant Safety Systems (TROSS) [65] which has been used to test restraint systems as well as the mechanisms of lower limb and head injuries using ATDs [75]. Such devices can provide accurate loading parameters but not always reproducible. Repeatable loadings can be generated with well-controlled rigs powered by mechanisms other than explosives. The Anti-vehicle Underbelly Blast Injury Simulator (AnUBIS) was developed at Imperial College to apply UBB loads to cadaveric lower limbs [66] (Fig. 30.7b), where the specimen rests on a plate which is pneumatically accelerated upwards to a target velocity. AnUBIS has been used to assess the effect of posture on severity of injury [76, 77]. The Lower Limb Impactor (LLI) uses a spring powered plate that impacts the surrogate leg which is held in position using a small wire. The LLI can achieve different peak velocities of the plate by changing the compression of the spring, and has been used to study the response of ATD legs fitted with combat boots [78]. Other devices include the Vertically Accelerated Load Transfer System (VALTS) which enables three types of loading scenarios independently or in conjunction to fully characterise the effects of the UBB environment on whole-body PMHSs or surrogates [79, 80], and the vertical accelerator (Vertac), with the load-application section incorporating a falling weight to induce a vertical impact load at the load-receiving end, which has been used to study the response of legs with and without combat boots [81] and that of the cervical spine [82].

Horizontal solid-blast simulators employ a range of operating mechanisms such as sliding impactors, pendulum impactors, or linear impactors. The Odyssey simulator is a sliding impactor that consists of a supine, rigid seat with independently articulating seat and foot platens that allow the feet and pelvis to be accelerated axially by means of a pneumatically driven carriage loaded with adjustable mass transfer rams [68]. Yoganandan et al. [83] also developed a custom buck, fixed to the platform of an acceleration sled (Fig. 30.7c) to study the dynamic response of PMHSs from simulated seat loading acting at the

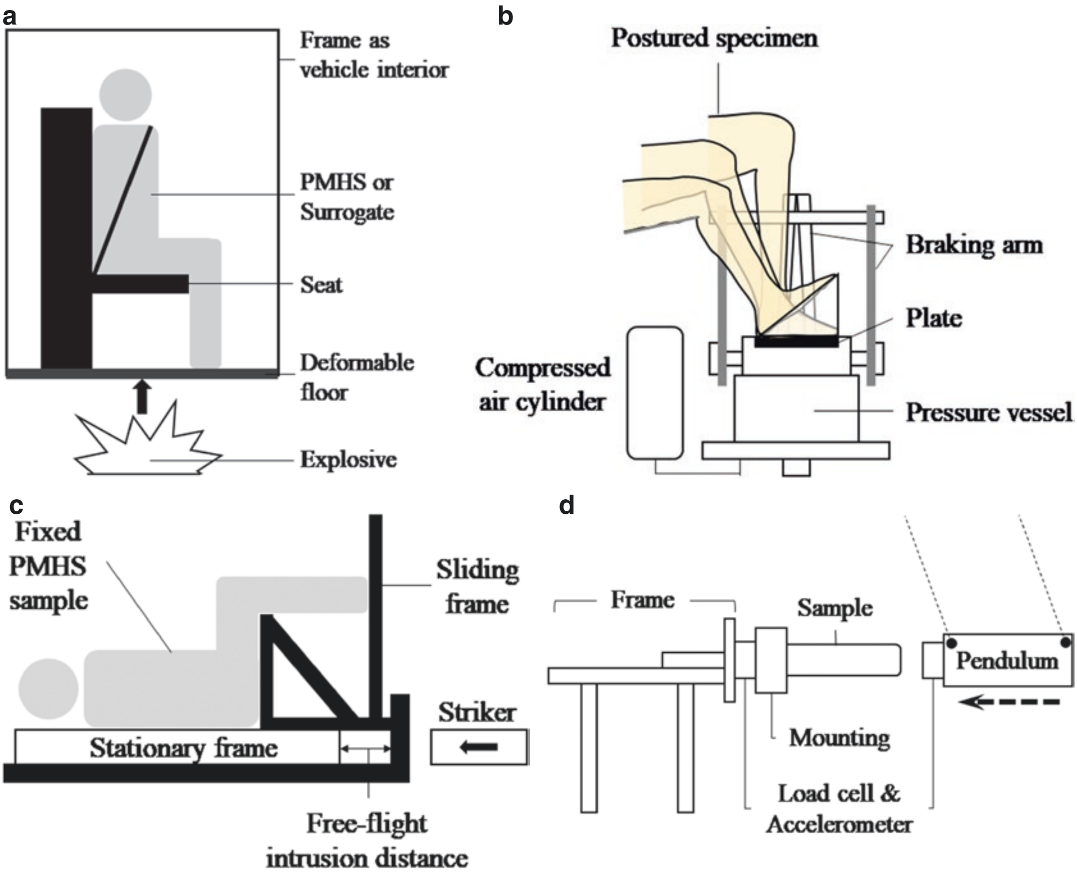

Fig. 30.7 Examples of solid blast-injury simulators: (**a**) Explosively driven vertical platform [64, 65]; (**b**) AnuBIS system for testing different postures of the leg; (**c**) Acceleration sled impacting the PHMS pelvis [83]; (**d**) Axial loading to PMHS legs using a pendulum striker [84]

pelvis. Pendulum impactors have been used to load lower limbs, fixed to a mini-sled, to describe their injury tolerance [69, 84, 85] (Fig. 30.7d). Examples of linear impactors include adaptations of horizontal sleds to simulate floorplate intrusion by accelerating the floorplate into the specimen at a target impact velocity and rapidly decelerating it [70], which was also used to identify lower limb injury criteria due to axial loading. Similarly, Quenneville [71] used a pneumatic linear actuator to evaluate the performance of floor mats in terms of the protection they may offer to reducing the probability of injury to the lower extremity in high-rate axial loading.

30.5 Conclusions

Experimental devices for modelling the injury mechanisms to the human body in an explosion have been developed and used recently due to the increase in use of explosive weapons in both military and civilian theatres. Field testing using actual explosive charges is probably the most representative of a real explosive event. It can deliver different elements of the explosion that contribute to the injury, including the shock wave, fragments and soil ejecta, blunt collisions, and thermal exposure [34]. This type of testing has been employed successfully such as with the

ALF rig to investigate the vertical loading in UBB [64] and the system to study brain motion due to live-fire blast loading from a charge 2.3 m away and 1 m from the ground [86]. The main challenge from field testing is its repeatability which is very sensitive to a range of factors such as the preparation of the explosive charge, the height or depth of burial of the charge, and its distance from the target, to name but a few, which result in differential responses in the near-field, midfield, or far-field, and complex blast wave interactions with surrounding objects (obstacles, surface, sensor mountings, experimental set-up) [87]. The ability to isolate different components of a blast and understand their contribution to the overall resultant injury has brought about the development of laboratory-based experimental devices, which are well-controlled and versatile with fast turnaround time and high reproducibility.

Investigations into the pathophysiology resulting from blast injuries have involved the adaptation of existing- and creation of new-experimental platforms by various injury biomechanics laboratories around the world. The versatility of these designs enables them to be easily adapted to study different parts of the body and various protective equipment. In addition, platforms that reproduce different aspects of the explosion can be used in conjunction to model injury mechanisms such as traumatic amputation or pelvic disruption where both primary and secondary blast effects contribute to the injury sustained.

The experimental results obtained from these platforms can be combined with the outcomes from computational simulations (Chap. 28) to provide more complete insights of the injury mechanisms. With appropriate biofidelic boundary conditions and meaningful injuries generated, the experimental results can be used for validation of relevant computational models. Validated computational models of blast injury can aid in optimising experimental design, produce measurements that are not possible to obtain in physical tests, and perform an array of virtual experiments.

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In Vitro Models of Primary Blast: Organ Models

31

Hari Arora and Anthony M. J. Bull

Abstract

Primary blast injury research at the organ level aims to identify local injury mechanisms caused by shock wave loading. In vitro and ex vivo organ level models of blast bridge the gap between length scales (e.g. from cellular to whole body) and provide suitable platforms for studying the effects of blast on isolated systems prior to any further study, e.g. potentially in vivo or for evaluating efficacy of armour systems.

Specific damage mechanisms have been studied using a combination of exposure to real explosives, lab-based blast simulators, shock tube devices and other bespoke test arrangements to deliver shocks to isolated tissues (Chap. 30). The effects of magnitude and duration of primary loading have demonstrated their relative degradative effects on isolated organs such as the brain, lung and whole bones. As methods of organ preservation are advancing and injury diagnostics are improving in other disciplines, so other organ models of primary blast such as for the kidney and eye are now being developed.

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31.1 Introduction

With primary blast, when a shock wave interacts with the body, some of the energy is reflected and some absorbed by the body. As tissue within the body possesses both elastic and viscous properties (as well as some organs being multi-phasic in nature), their response to blast loading is complicated and difficult to predict. Different parts of the body, specifically organs, react differently to impulsive loading. This is due to a combination of factors: their unique structure, which responds in a certain way to a mechanical stimulus; and the unique stress-strain state experienced in that part of the body, due to the loading from a given blast wave and the support conditions of that organ. This can lead to local injury development within a given organ which can also result in consequences to the whole system (e.g. inflammation). Multiple injury sites generate increased burden on the system. This can lead to added complications in their treatment when the effects of multiple injuries superimpose. Although in vivo blast models continue to dominate the existing literature, these models tend to analyse whole body responses and may not focus on identifying physical injury at the organ/tissue level. Isolated organ

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experiments, termed either ex vivo or in vitro models, maintain the architecture and functionality of the tissue for a short period of time and constitute a close representation of the in vivo state [1]. This section focusses on research assessing primary blast evaluation of the body, identifying injury mechanisms at an organ level.

31.2 Brain

Traumatic brain injury (TBI) is a leading cause of mortality and disability in injured service personnel. Blast induced TBI (also referred to as blast induced neurotrauma, BINT) is a specific area of research focusing on how the impulsive nature of a primary blast changes both the physical status of the brain and any neural functionality. Duckworth et al. [2] give an overview of the differences between the three common types of battlefield TBI from closed-head injuries (brain impacting against the skull wall—tertiary blast effect), penetration injuries (secondary blast effect) and finally blast TBI (primary blast effect).

An extensive summary of TBI models in use has been collated by Sundaramurthy et al. [3], who then extended the studies to explore the effect of positioning of test samples within and in front of a shock tube. The optimal position of the head for ensuring primary blast induced neurotrauma was found to be within the shock tube with the dominant mode of stress wave transmission, determined numerically, to be through the cranium. Chavko et al. [4] reinforced this observation via direct measurement of intracranial pressures during blast loading. Intensity of pressure wave seemed relatively undisturbed by the skull, with the decay of cranial pressurisation varying with loading orientation. Further numerical work by Panzer [5], on a brain and head model, showed the dependence of pressure arising within the brain on the peak pressure of the blast wave. The largest brain strains were shown, however, to be a controlled by the shock wave duration or impulse. In addition to transcranial loading, blast induced TBI has also been hypothesised to occur via a thoracic pressurisation [6].

Advanced methods such as diffusion tensor imaging (DTI) have revealed the extent and severity of mild blast induced TBI. Scans were performed on extracted brains in rats after they were subject to transverse cranial loading in a shock tube [7] and pigs subject to blast from free-field explosives [8]. This method has previously been exploited in vivo in other areas of TBI and brain degradation assessment [9, 10] as well as ex vivo in the human brain. Therefore, there is the capacity to use the current state-of-the-art imaging methods such as DTI to map in detail the injury profile (damage pathways) in blast TBI whether in vivo or ex vivo. This is in addition to traditional imaging modalities that would highlight only structural disruption such as conventional MRI.

In addition, high-speed photography methods can be used to capture shock wave profiles and their interaction with the body. Sarntinoranont et al. [11] used living tissue slices from a rat brain, attached them to a ballistic gelatin substrate and subjected them to high strain rate loads of 1584 ± 63.3 psi, using a polymer split Hopkinson pressure bar (PSHPB) (Chap. 30). Simultaneously, they used real-time high-speed imaging and noted cavitation due to a trailing under-pressure wave. Neuronal injury was quantified at 4 and 6 h. post blast. Ouyang [12] and Connell et al. [1] exposed isolated sections of guinea pig spinal cord white matter to a shock wave produced from a small-scale explosive event (Fig. 31.1). The latter study explored dose response with regard to input shock pressures and the resultant functional and anatomical deficits. Direct exposure to the blast wave compressed nervous tissue at a rate of 60 m/s and led to significant functional deficits. Results also showed that a relationship exists between the magnitude of the shock wave overpressure and the degree of functional deficits [1]. Damage to the spinal cord was marked by increased axonal permeability suggesting that compression from the shock wave results in acute membrane disruption [1]. Methods using brain tissue segments can enable high throughput and reproducible injury dose to enable screening of protective or preventative measures [13].

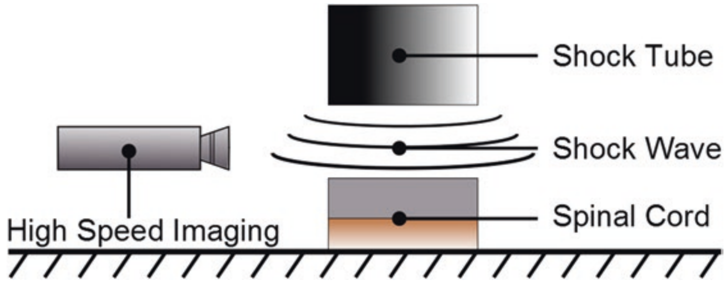

Fig. 31.1 Schematic of ex vivo spinal cord experimental set-up

Work such as these complement the studies ongoing in vivo (Chap. 32), which forms the majority of active research in the area, to isolate and identify injury mechanisms worth pursuing.

31.3 Lung

Shocks interact more intensely with larger surface areas of the body. Therefore, regions such as the thorax and abdomen can be more susceptible to a larger reaction from primary blast. Moreover, these delicate internal membranes are not equipped to sustain significant forces. There are large differences in impedances within these organs and with their surroundings, which further contributes towards their likelihood of internal rupture and bleeding during a blast event.

Study of respiratory mechanics has often involved the use of ex vivo perfused lung ranging in scale from rodents to humans. Extracted lungs are often degassed and filled/washed with saline before experimentation in order to effectively eliminate surface tension effects and mitigate sources of degradation. Such processes are expected to help maintain the lung in a condition similar to that occurring in vivo—although the mechanics are naturally disrupted by extraction and loss of surfactant [14]. In terms of models for blast, specifically, there are few models in the literature at the organ level, much like the other organs of the body. Fung et al. [14] explored the effect of transpulmonary pressure on the wave speed through different animal lungs. A water jet was used to generate the characteristic incident pressure-time profiles observed in blast. Measurements were taken on the front and rear surface of the lungs as shown in Fig. 31.2. Since the square of the wave speed is proportional to modulus and inversely proportional to density, transpulmonary pressure had a significant effect on wave speed (as it controls the stiffness of the lung). This is due to the composition of the lung tissue and its mechanical behaviour under differing degrees of pre-tension. The influence of having a perfused lung was also highlighted as this significantly affects the density of tissue. Similar methods to [11], previous outlined for TBI study, were employed to stabilise and measure shock wave attenuation of rodent lungs, in the context of surrogate organ development [15].

Volumetric imaging methods (such as microCT) have shown increasing relevance to damage quantification in lungs. The use of synchrotron sources to image fast enables highly viscoelastic tissues such as the lung to be imaged (in 3D)—without appreciable motion blur and without the need for fixation. Various researchers [16–21] have developed micro-CT and other tomography-based methods to study microstructural characteristics and mechanics in healthy and diseased lungs. Such methods have been adopted to quantify residual tissue mechanics in primary blast lung injury research [22]. Methods for blast lung injury evaluation (beyond in vivo models) currently concentrate on isolated tissues being exposed to shock waves such as in [23, 24] to complement clinical observations and in vivo models currently in general use.

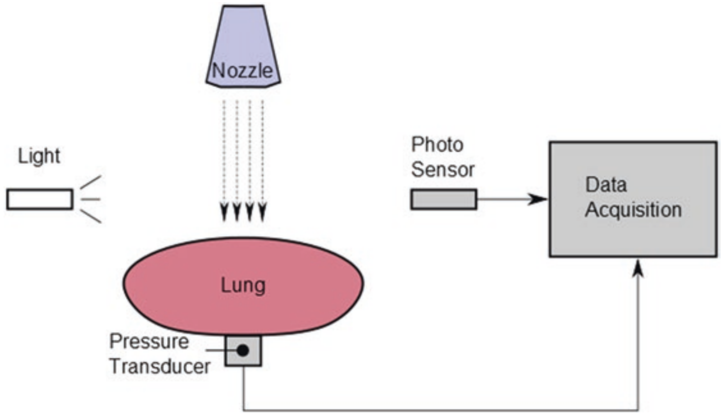

Fig. 31.2 Water jet system for simulating shock waves on lung tissue (adapted from [14])

31.4 Abdominal Organs

Primary blast effects on soft organs focus primarily on the lung and the brain. However, gas-containing sections of the gastrointestinal tract are also vulnerable to primary blast loading. Observed injury mechanisms include immediate bowel perforation, haemorrhage and mesenteric shear injuries, solid organ lacerations, and testicular rupture (Chap. 9). Most studies in this area are clinical observations, and abdominal injuries are commonly reported in combination with other injuries. Some studies using isolated perfused kidneys, researching treatment for kidney stones, have been extended to look at the effects shock waves on kidney integrity in various animal kidneys such as pig [25] as well as human [26]. Köhrmann et al. [27] were able to evaluate vessel lesions by microangiography to determine the size and number of damage sites formed in the different areas of the organ subjected to focused ultrasound waves. The variation in the different patterns of lesions observed helped to characterise the pathway of the shock wave. Light microscopy revealed dose-dependent necrosis of tubular cells up to macro-scale parenchymal level defects [27]. This platform can be scaled-up to explore higher intensity shocks in addition to the multiple doses of low-level shocks, relevant to ultrasonic treatments.

31.5 Eye

Tissue damage from primary blast injury can be an important cause of trauma to the ocular system (see Chap. 19) which can result in severe vision loss and injuries to the peri-orbital area [28]. Ocular injury occurs in up to 28% of blast survivors. The most common injuries include corneal abrasions and foreign bodies, eyelid lacerations, open globe injuries and intraocular foreign bodies [28].

Glickman et al. [29] subjected ex vivo porcine eyes to blasts produced by a shock tube and demonstrated that this approach could be used to detect trauma-induced biomarkers. Similarly, Sherwood et al., [30] also subjected porcine eyes to a range of primary blast energy levels and showed that the incidence and severity of damage in the exposed eyes increased with impulse and peak pressure. Moreover, these data also suggested that primary blast alone can produce clinically relevant ocular damage in a postmortem model. These models are relatively new and require further validation; however, they could become useful in determining direct effects of primary blast on ocular trauma.

31.6 Summary

The bulk of research on primary blast effects conducted currently lies at the two extremes of in vitro cellular models and in vivo models. Organ level research in vitro or ex vivo is predominantly being covered as preliminary experiments to in vivo work (or might simply be unreported). Neurotrauma research is fairly advanced, including research on whole explanted brains, brain sections, and spinal cord. Soft organ research, such as for lungs and the abdominal organs is less commonplace mainly due to the complexities involved in extraction and perfusion, as well as the ability to replicate in situ loading characteristics in the excised models. Although other niche areas, such as ocular models, are now beginning to be developed with demonstrable utility in blast research.

Other major areas including musculoskeletal injuries are predominantly studied in vivo or clinically, rather than in isolated ex vivo or in vitro. The effects of primary blast on bone lead to micro-fractures, produce bending forces on long bones and can result in fracture and amputation [31]. Whole organ models for bone are required, since geometrical effects of the bone significantly influence the fracture type, which, in turn, will affect the design of mitigation technologies. Such models can additionally be used to understand forensically the type of loading that produced certain injuries in casualties [32–34]. The requirement for organ level models to isolate, identify and enable the replication of injury mechanisms is consistent and significant across hard and soft organs/tissues in the body. Consideration of the injury environment, threat level and organ biomechanics is important and, therefore, makes this a challenging area of research. Furthermore, efforts to standardise these experimental methods for improved repeatability and translation from one laboratory to another remain a key activity in the area of primary blast injury model development [35–37].

This area is progressing due to both the advancing techniques for preserving tissues and organs in a viable manner as well as the increased understanding from organ transplant research. This will certainly lead to the possibility of more sophisticated models being developed in the future for blast injury research. Several models from other areas of physiological research have been and can be extended towards blast injury research. These need to address key concerns, including maintaining the extracted tissues over appropriate long time periods [11] and ensuring appropriate perfusion and physiologically relevant support conditions during a given experiment, as raised by Fung et al. [14] amongst others. Ex vivo models however, can assist researchers in gaining a better understanding of the key mechanisms of blast injury by mitigating any confounding factors associated with in vivo models [1], i.e. systemic factors, providing significant control over such additional variables. Current work in conjunction with the rise of computational methods (see Chap. 28) help govern the nature of appropriate loading of excised organ models. Such methods will improve understanding of stress wave transmission and, therefore, understanding of blast injury development.

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32 Modelling Blast Brain Injury

Rita Campos Pires and Robert Dickinson

Abstract

The consequences of blast traumatic brain injury (blast-TBI) in humans are largely determined by the characteristics of the trauma insult and, within certain limits, the individual responses to the lesions inflicted (Maas et al., Lancet Neurol. 2008;7:728–41). In blast-TBI, the mechanisms of brain vulnerability to the detonation of an explosive device are not completely understood. They most likely result from a combination of the different physical aspects of the blast phenomenon, specifically extreme pressure oscillations (blast overpressure wave), projectile penetrating fragments and acceleration–deceleration forces, creating a spectrum of brain injury that ranges from mild to severe blast-TBI (Hicks et al., J Trauma. 2010; 68:1257-63). The pathophysiology of penetrating and inertially driven blast-TBI has been extensively investigated for many years. However, the brain damage caused by blast overpressure is much less understood and is unique to this type of TBI (Chen et al., J Neurotrauma. 2009; 26:861–76). Indeed, there continues to be debate about how the pressure wave is transmitted and reflected through the brain and how it causes cellular damage (Nakagawa et al., J Neurotrauma. 2011; 28:1101–19). No single model can mimic the clinical and mechanical complexity resulting from a real-life blast-TBI (Chen et al., J Neurotrauma. 2009; 26:861–76). The different models, non-biological (in silico or surrogate physical) and biological (ex vivo, in vitro or in vivo), tend to complement each other.

32.1 In Silico Models

Computer simulation represents a valuable link between laboratory experiments and the study of human cases. These models may provide a better understanding of the damaging mechanisms resulting from the blast wave and may reduce the number of trial-and-error tests involving laboratory animals [1, 2]. A comprehensive computational model of blast-TBI should be multidisciplinary and should be validated by data from animal tests [2, 3].

32.2 Ex Vivo Models

Human cadaveric models have been used to determine the anatomical response to blast-TBI [4]. However, the biomechanical properties may

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be altered *post-mortem,* and the models lack the post-injury physiological response [4]. Through the understanding of the biomechanics of the forces that act upon the human head, these models provide valuable information for the development of protective equipment and helmets [5].

32.3 In Vitro Models

In vitro models offer several advantages in the study of blast-TBI, allowing for a carefully controlled experimental environment, both in terms of cellular characteristics and regarding physiologic conditions, such as temperature, pH and nutrient concentrations, which reduce experimental variation and result in greater reproducibility. The time scale for the completion of in vitro studies is usually shorter than for in vivo experimental protocols, making them less expensive than more complex models. The in vitro models also permit a precise control and characterisation of the injury biomechanics. Arun and colleagues have described an in vitro model using human neuroblastoma cells exposed to a pressure wave from a compressed air-driven shock tube [6]. They showed a transient increase in the permeability of neuronal nuclear and plasma membranes. The authors suggest that the blast exposure disturbs the integrity of the cell membrane and hypothesised that this may underlie the development of acute tissue damage seen in blast-TBI victims. Simple cellular models lack the heterogeneity of cell types and synaptic connectivity found in the intact brain. The in vitro organotypic brain-slice culture technique represents an intermediate model between the single cell and the whole organ. A thin slice of brain tissue can be kept alive in vitro for many days or even months preserves the different cell types (e.g. neurons and glia) and maintains synaptic connectivity mirroring that in the intact brain [7]. Organotypic hippocampal slice culture models have been used to investigate mechanisms of injury in other types of brain injury including blunt-traumatic brain injury and ischemic brain injury [8–10]. Effgen and colleagues demonstrated that a shock wave from a compressed air shock tube induces cell dysfunction and death in hippocampal organotypic slice cultures [11, 12]. Campos-Pires and colleagues have recently described a high-throughput in vitro blast-TBI model using organotypic cultures to investigate injury mechanisms and evaluate novel neuroprotective treatments [13, 14]. Nevertheless in spite of the aforementioned advantages, in vitro models are relatively simple biologic systems of cells or tissue that do not fully mirror the in vivo situation. Cultures are typically obtained from young animals whose cells may not be fully differentiated and may exhibit a different phenotype from the mature tissue. Moreover, the artificial controlled environment is not the same as the cells experience in vivo and this can affect their morphology and function. Therefore, novel in vitro findings must be validated and confirmed by in vivo tests in whole animals. Thus, while in vitro models are a powerful tool for mechanistic analysis and a convenient, cost-effective method of screening, they cannot completely replace animal studies.

32.4 In Vivo Animal Models

Animal models are the gold standard translational research method for blast-TBI, allowing hypothesis testing in the controlled laboratory environment [15, 16]. Despite the usefulness of human clinicopathological analysis in providing evidence of correlative association, the use of animal models is the most important tool in investigating causal mechanisms of disease [16]. An animal model's value and relevance are directly proportional to its adequacy in recapitulating the histopathological features and/or neurological deficits of the corresponding human disorder [16]. Irrespective of the research questions to be addressed, a clinically relevant blast-TBI model should fulfil the following criteria [17]:

1. the injurious mechanical component of the blast should be clearly identified and replicated in a controlled, reproducible, and quantifiable manner;

2. the inflicted injury should be reproducible, quantifiable, and mimic components of human blast-TBI;
3. the injury outcome should be chosen based on morphological or histological and/or behavioural parameters; and
4. the mechanical properties of the injurious pressure wave should predict the outcome severity.

A considerable number of animal models have been used in blast-TBI research. In the following paragraphs, critical aspects of blast-TBI models will be described based on the available literature.

32.4.1 Animal Species

Blast-TBI models have used many different species of animals, from mice, rats and rabbits to ferrets and pigs [4]. Rodents, mainly rats, remain the most commonly used animals for modelling human blast-TBI [17]. The relatively small size and lower cost of rodents permit repetitive measurements of relevant experimental parameters that require relatively large numbers of animals. Rat models allow better monitoring and control of physiological parameters (e.g. blood pressure, blood gases). Mice offer the additional possibility of conducting tests where genetic manipulation is possible. Due to ethical, technical and/or financial limitations, blast-TBI studies are less feasible in phylogenetically higher species [17]. However, rodents have a lissencephalic cortex, i.e. they lack the gyri and sulci found in the human brain; it has been suggested that this characteristic makes rodents less than ideal for modelling complex injury-induced changes in functional outcomes and is a factor that most likely affects the brain's mechanical response to a pressure transient and an acceleration impulse [18]. Pigs and non-human primates have brains more similar to humans. However, the cost of larger animal models and, in the case of non-human primates, ethical issues and availability limit their use [19]. The choice of a species also affects scaling considerations, namely the selection of the pressure wave parameters. Skull size and geometry (including the anatomy of the orbits and sinuses), skull biomechanical properties and histologic characteristics of bone affect how external forces act upon the brain [1].

32.4.2 Generation of Overpressure Waves

32.4.2.1 Free-Field Explosives

In free-field blast testing, blast waves are generated using high explosives in an open field (Fig. 32.1a). Experiments with explosives in the open field have been used to determine thresholds of mortality and injuries in air-filled organs, such as the lungs. These experiments provided fundamental data on blast magnitude-response curves (the Bowen curves) [3, 20, 21]. Free-field explosive tests allow realistic experiments in large animals that are more similar in size to humans, closely replicating real-world blast conditions [3]. There are, nonetheless, some significant drawbacks. Large adjacent structures such as buildings or vehicles can reflect the pressure wave, potentially exposing the specimen to a complex blast waveform, often more damaging than the initial waveform. The explosives produce by-products (e.g. heat, noxious gases, fragments), which summed to the overpressure damage may cause complex blast injuries due to the possibility of penetrating and burn injuries for example [4]. To summarise, this set-up offers less experimental control over the physical characteristics of the blast (when compared to a shock tube, see below), making it more difficult to quantify the overpressure transient and represents an expensive and time consuming protocol [4, 19].

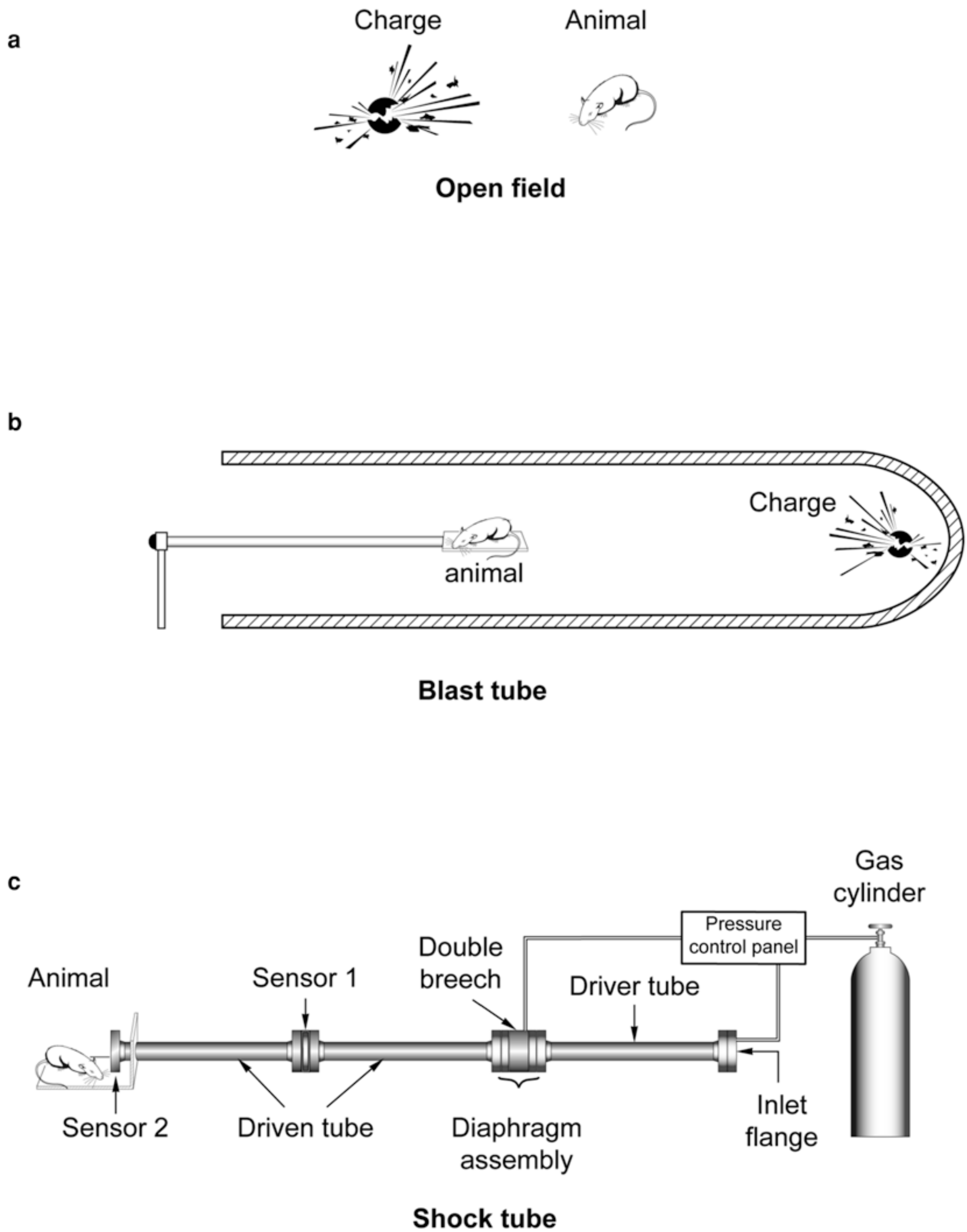

Fig. 32.1 Different methods of generating blast overpressure waves in in vivo models of blast injury. (**a**) Open field experiments using an explosive charge outside of a laboratory. (**b**) In a laboratory setting, a blast tube can be used with an explosive charge. (**c**) In the laboratory, a gas-driven shock-tube results in a more reproducible and controlled shock wave

32.4.2.2 Blast Tubes

During the 1950s, blast tubes were often used to study materials' responses to high pressures. The blast tube was modified for studies with rodents in the 1990s [3]. A small explosive charge is detonated in a conical or parabolic driver section, and the blast wave is allowed to propagate down the driven section (Fig. 32.1b). By-products of the explosion, such as smoke and gases, may contribute to quaternary blast effects [3]. Also, the handling and storage of explosives in the laboratory environment demand strict safety considerations. For these reasons, shock tubes are typically preferred over blast tubes for blast-TBI testing in the laboratory setting [4].

32.4.2.3 Shock Tubes

Most published blast studies have used shock tubes to investigate the effect of blast-TBI using animal models [1]. A shock tube is a long tube with constant cross-section (Fig. 32.1c). The device is divided into two chambers separated by a membrane or diaphragm: the driver section of high-pressure gas and the driven or main section of low-pressure gas (atmospheric pressure). Compressed gas (typically air, helium or nitrogen) is loaded into the driver section. The diaphragm either ruptures spontaneously, or rupture may be triggered electronically (in the so-called double breach configuration), at a given pressure dependent on diaphragm thickness and material. Upon rupture of the diaphragm, the discontinuous pressure differential between the driver section and the driven section creates a shock wave that propagates along the driven section towards the test specimen [3, 4]. With an open-end configuration, the animal or specimen can be placed either inside or outside the tube. Testing inside the tube ensures that the shock wave is planar, but the diameter of the shock tube must be large enough for the specimen. Outside the tube, the specimen should be placed as close to the tube as possible, to ensure the shock wave is planar [4]. Shock tubes can be designed to reproduce any characteristic of an ideal blast wave in terms of magnitude of peak overpressure, duration of positive-phase, impulse and shape of the pulse [3, 4].

The shock tube has several advantages over the use of high explosives. Shock tubes allow blast overpressure effects to be studied in isolation. Shock tubes can produce a variety of repeatable pressure transients that closely resemble free-field blast waves, in a controlled laboratory environment (Fig. 32.2a). Furthermore, shock tube testing is more economical and safe compared to either free-field blast or blast tube testing. However, shock tubes can only reproduce certain aspects of real-life explosions; while they replicate the ideal pressure wave, they can't model the non-ideal complex blast wave and they are unable to reproduce other real blast effects such as thermal injury [4, 19].

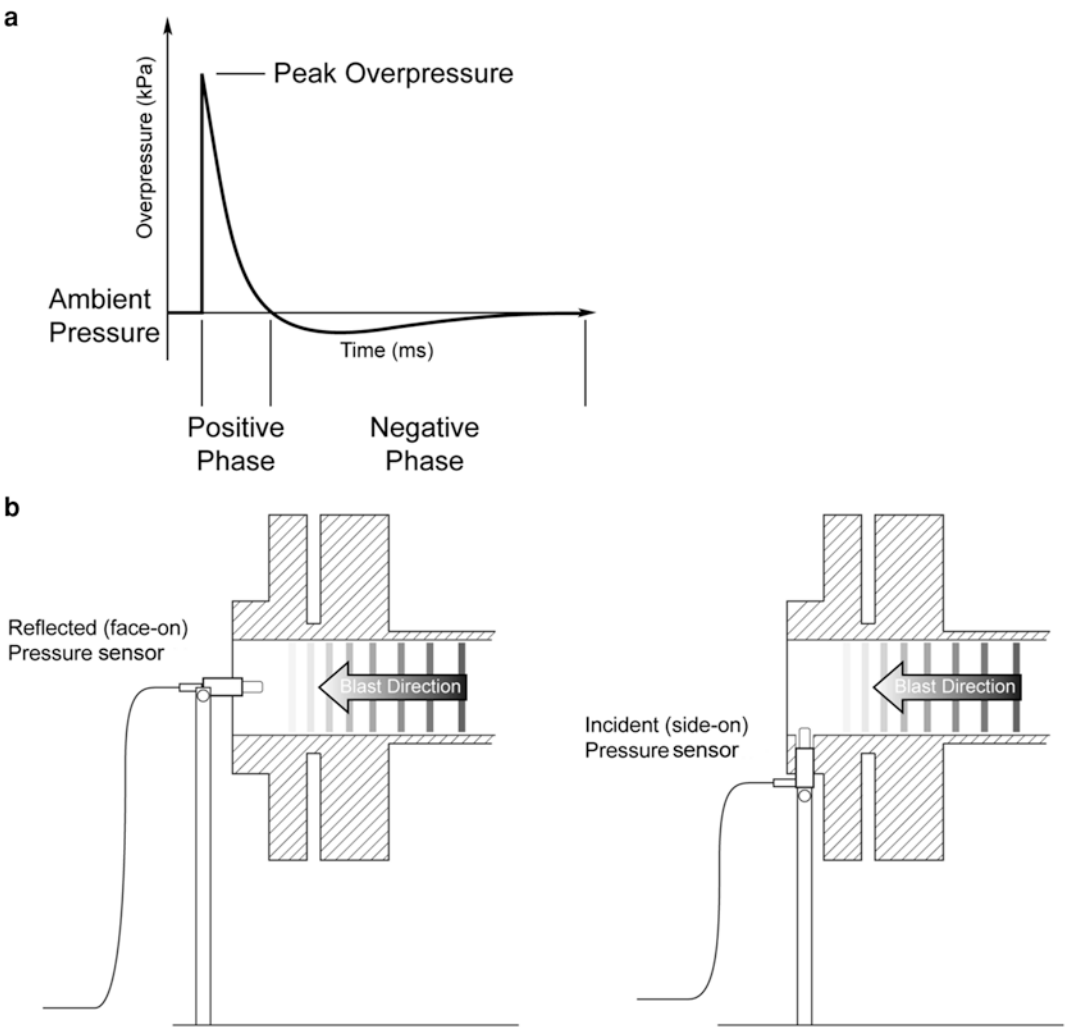

Fig. 32.2 (**a**) Schematic of a typical Friedlander waveform produced in a shock-tube experiment. (**b**) The magnitude of the peak overpressure for a given shock wave depends on the orientation of the pressure sensor relative to the shock wave, with face-on configuration giving a higher value than a side-on configuration

32.4.3 Critical Aspects

32.4.3.1 Anaesthesia and Analgesia

Most authors anaesthetise their animals before and during the pressure wave exposure (an exception is Ahlers and colleagues [22]). Different methods of anaesthesia have been used, the majority allowing the animals to be under spontaneous ventilation during the procedure. Inhalation general anaesthetics, such as isoflurane, are often used due to their advantages of providing effective anaesthesia with rapid onset and rapid recovery at the end of the protocol [23, 24]. These drugs may be supplemented with an opioid-based analgesia (for example, buprenorphine). Another common choice is the administration of intraperitoneal drugs, as the combination of ketamine and xylazine [24, 25]. This method requires less equipment than an inhalational anaesthesia technique, has the advantage of providing concurrent analgesia (ketamine) but it is less versatile in terms of anaesthetic induction and recovery times. Anaesthetics are usually potent respiratory and cardiovascular depressants, so the researchers using them need to be familiar with their side-effects. Some particular anaesthetic side-effects should also be considered carefully according to the goal of the experimental protocol. For example, some drugs, such as medetomidine, induce profound hypothermia [26], while others, such as ketamine, may modify neurological impairment [27], which are aspects that should be taken into account in blast-TBI studies of neuroprotection.

32.4.3.2 Pressure Wave Characteristics

The characteristics of the pressure wave are determined by the device used and in the case of a shock tube, by the thickness, material and number of the diaphragm(s) used. Animals have been exposed to blast waves as low as 36 kPa [15] or as high as 500 kPa [28], but typically are exposed to blast waves with peak overpressures between 150 and 340 kPa [17]. The overpressure wave is sustained for varying durations of time, typically from 2 to 10 ms [28, 29]. The magnitude of the peak overpressure and wave duration used should be based on the severity of the injury being modelled (mild, moderate or severe blast-TBI) and should take into account the hypothesis being tested [15]. An important consideration is that the reported characteristics of a given overpressure wave depend on the orientation of the pressure sensor relative to the wave (Fig. 32.2b). For example, the face-on peak overpressure reading of a shock wave will be higher than a side-on reading for the same wave [4].

32.4.3.3 Animal Head Orientation Relative to the Direction of the Pressure Wave

The amplitude and duration of the pressure wave to which the animal brain is exposed depend on the orientation of the animal relative to the pressure device [15]. Different orientations of experimental animals relative to the direction of the pressure wave have been reported. The most common are the frontal or head-on orientation in which the animal faces the source of the pressure wave and the transverse or side-on orientation in which the body of the animal is perpendicular to the source of the pressure wave (Fig. 32.3). The frontal orientation results in greater overpressure exposure than does the side-on orientation [25]. However, Ahlers and colleagues reported different functional outcomes based on orientation in the shock tube, including greater impairment in gross motor function for rats in the side-on position [22].

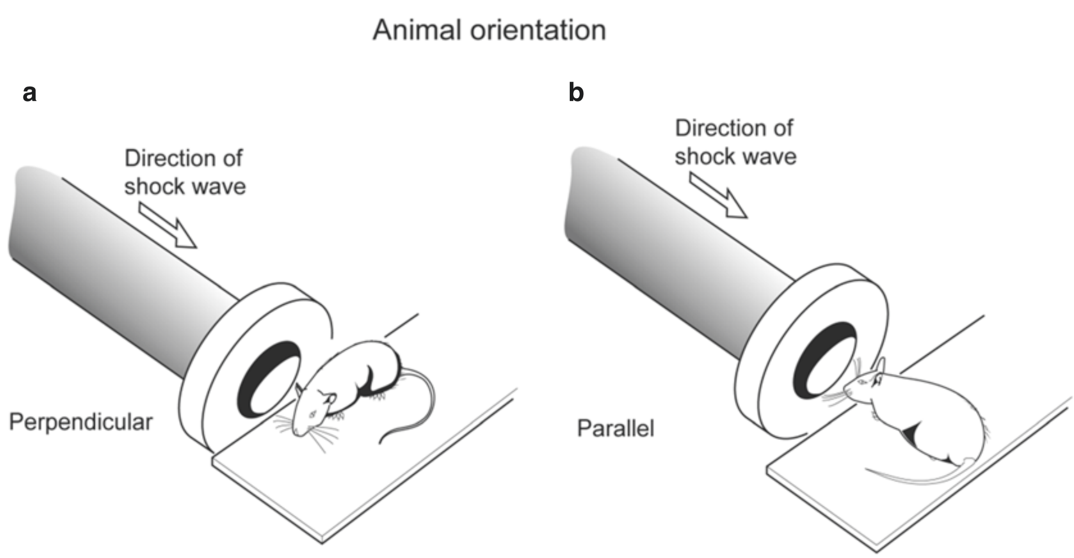

Fig. 32.3 The orientation of the animal relative to the shock tube is a key variable in the experimental set-up. Shown are perpendicular (**a**) and parallel (**b**) configurations

32.4.3.4 Head Mobile Versus Head Restrained

Several recent studies have highlighted the value of investigating how the presence or absence of head motion affects histopathologic injury patterns and neurobehavioural deficits. Goldstein and colleagues [30] reported learning and memory deficits in mice with unrestrained heads exposed to a blast wave. Notably, restrained head movement resulted in the disappearance of functional deficits, implicating blast-induced acceleration–deceleration of the head ("bobblehead effect") as the main pathogenic mechanism by which the blast exposure induces brain injury [30]. Gullotti and colleagues [29] have shown that minimising head acceleration led to an increase in survival rate and a decrease in the duration of loss of righting reflex following blast. However, increase in duration of loss of righting reflex was achieved by significantly increasing the peak blast overpressure [29]. When planning the experimental design of a blast-TBI study and interpreting its results, the head mobility is a very important feature of the protocol that must be considered according to the intended goals of the study.

32.4.3.5 Head Only Blast Exposure (Thorax Protection) Versus Whole Body Blast Exposure (no Thorax Protection)

Some investigators use chest protection on their animals arguing that the use of appropriate shielding isolates the effects of blast injury to the brain from injury to the body, particularly the lung. Different materials have been used, such as Kevlar fabric [23], plastic tubes [28] or metal cylinders [29, 31]. A considerable number of blast-TBI investigators do not use chest protection [24, 32, 33]. Long and colleagues reported that rats wearing protective vests when exposed to 126 or 147 kPa overpressure were more likely to survive 24 h after the procedure compared with rats not wearing vests [23]. Some investigators have examined the use of protective shields on animals placed inside the shock tube and found that their use does not significantly reduce the effects of the pressure wave [15, 25].

32.4.3.6 Single Blast Versus Repeated Blasts

Most of the experimental blast-TBI studies published so far have used protocols consistent with single blast exposure. However, multiple blast exposures are common in the war zones even in subjects not known to have suffered a TBI [19]. Repetitive blast exposure in military service personnel has been associated with long-term neuropathology and psychiatric disturbances, including cognitive impairment similar to what is seen in athletes with repetitive concussive injury [34, 35]. These findings suggest that repeated blast overpressure waves have a synergistic effect causing cumulative brain damage, as shown by Calabrese and colleagues [36].

32.4.3.7 Outcomes

Outcomes relevant in animal models include physiological, pathological and behavioural parameters [19]. Animal models of blast-TBI will be more useful from a translational perspective if the experimental emphasis focuses on reproduction of clinically relevant endpoints, accelerating the development of new preventive, diagnostics, treatment and rehabilitative strategies for blast-TBI victims [16]. Blood–brain barrier disruption, brain oedema and vasospasm, neuronal degeneration, axonal injury, glial cell activation, chronic neuroinflammation and subsequent cognitive deficits, including memory impairment and anxiety-related behaviours have been shown both after human and experimental blast-TBI [16, 17, 37, 38]. An additional aspect that is usually overlooked by many investigators studying TBI models is the measurement of physiological variables before and after blast-TBI, including arterial blood oxygen and carbon dioxide partial pressures, pH, heart rate, blood pressure and core body temperature. These variables are extremely important in determining pathophysiological responses to injury and therapy, both for acute outcomes and long-term outcomes [39].

Although overlapping clinical features of blast-TBI models and related findings in humans suggest common pathobiology, the underlying pathophysiological mechanisms and interactions are poorly understood. Furthermore, the temporal course of the acute and chronic stages following blast-TBI injury is largely unknown [16]. One aspect of particular note is that blast-TBI patients often present with cognitive and anxiety-related symptoms similar to those resulting from post-traumatic stress disorder (PTSD). In these blast-TBI patients, it can be difficult to disentangle symptoms resulting from the physical blast and those related to being in the stressful environment of combat or civilian blast situation. In a controlled laboratory environment where the animal is anaesthetised it should be possible to study isolated blast-TBI with minimal confounding PTSD pathophysiology. Together these aspects emphasise the importance and clinical relevance of developing reproducible validated animal models that can be used to understand the mechanisms of brain injury following exposure to blast.

32.5 Conclusion

A broad range of experimental animals and models are being used in blast-TBI research. Early models focused mainly on biomechanical aspects of brain injury, while more recent ones are targeted towards improving the understanding of the damaging injury processes initiated by blast-TBI [17]. However, translational research in this area has been fraught by a number of methodological issues. The shock tube devices (or alternative methods) and blast injury conditions, as well as specimen or animal mounting, degree of head restraint and location relative to the driven section can vary significantly between different laboratories. Blast waves are characterised by several parameters including the peak overpressure, duration and impulse. Different shock tubes may produce pressure waves with differing characteristics leading to different biological effects. Many studies do not report the full pressure wave data, or report only peak overpressures, but not duration or impulses. Some laboratories that do report pressures omit key set-up information, such as the pressure sensor orientation. This makes interpretation and comparison between studies extremely difficult. Blast-TBI is particularly challenging due to scarce exposure data from actual operational/clinical situations [18]. Reliable in vivo blast-TBI experimental studies, complemented by in silico and in vitro models, led by multidisciplinary teams, are of great importance not only in the identification of the complex mechanisms leading to short- and long-term functional deficits, but also in guiding novel approaches to diagnosis and treatment modalities.

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Post Mortem Human Tissue for Primary, Secondary and Tertiary Blast Injury

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Abstract

Blast injury studies require the ability to replicate the blast injury insult in a controlled manner. This enables the biological, physiological and physical (including anatomical) effects of blast to be understood and mitigation to be evaluated. Post mortem human tissue can be used to simulate the physical response to blast of living human tissues. Animal models are an appropriate surrogate for some physiological, biological and physical effects (Chap. 32 'Modelling Blast Brain Injury' and elsewhere), although these have significant anatomical differences to human tissues. The benefits of using post mortem human tissues include anatomic fidelity and material (or tissue) response. Reservations with the use of such tissues include variability in size, shape and mechanical response, post mortem tissue changes, biohazard and storage, inability to model physiology and lack of availability. Ethical issues should also be considered and addressed. Once the experimental cadaveric model has been deemed appropriate, then it can be applied to many areas of blast injury research, with recent examples in primary, secondary and tertiary blast.

This chapter outlines the benefits of using post mortem human subjects, issues with their use, and then presents a number of examples of how they can be used in blast injury research.

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33.1 Benefits of Using PMHS

Post mortem human subjects (PMHS) are used widely in blast injuries studies. The main benefits of these are their anatomic fidelity and material responses.

Anatomic fidelity enables the geometrical and inertial effects to be analysed and taken into account. For example, the analysis of traumatic lower limb amputation mechanisms such a flail requires a hemi- or whole-cadaver set-up. There is no animal model that provides the same physical response to flail due to the mass and moments of inertia features of a human cadaver, the local geometrical features of the articulations of the hip and knee, as well as the tissue responses of the connective tissues such as ligaments, muscles, bone and skin. However, animal cadaver models have been successfully used to investigate some aspects of blast injuries, and, where physical and anatomical effects are important, animal cadaveric studies can be used the refine the number of PMHS experiments required. Also,

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scaling laws can be derived to enable the transfer of learning from animal studies to human studies [1]. Examples of animal whole-cadaver studies include those assessing blast flail in mice [2] and secondary blast injuries to the thorax in pigs [3].

Tissue response effects are of interest in many blast injury scenarios. All biological tissues are viscoelastic, meaning that they change their loading and deformation response dependent on the loading rate (Chap. 4 'Behaviour of Materials') [4]. Although the viscoelastic response of biological tissues at the loading rates seen in blast is not known for most tissues (a rare example is human ligament material [5]), what is known is that the tissue response at low loading rates of animal tissues is not the same as that for humans. For example, porcine skin is frequently used as surrogate for human skin, yet its dimensions and loading response are known to be significantly different from the human response [6].

Therefore, PMHS provide the opportunity to replicate the human in vivo physics-based responses (*mass, moments of inertia*), *geometry* (articular shapes and tissue insertions) and *material property effects* (load deformation, stress strain, viscoelasticity, failure properties).

33.2 Issues with Using PMHS

PMHS have some issues associated with *preservation*. It is not known how tissues change their properties after death. However, changes are known to develop with time post mortem; therefore, storage means have been developed that minimise these changes. These include freezing or preserving in alcohol or formaldehyde. Although preservation in a medium can significantly reduce infection risk, fresh freezing is the only method of preservation that has been shown not to significantly change mechanical properties of certain tissues such as bone, ligament and whole joints [7–11]. Multiple freeze-thaw cycles do not cause significant tissue degradation [12], yet the time that specimens are kept at room temperature should be minimised to less than 96 h as longer than that can cause tissue degradation [13]. There are no data available in the literature on the effect of long-term storage on material properties at the high loading rates seen in blast.

The *availability* of PMHS is not consistent. In particular, although specific tissues are readily available through reputable tissue banks, access to whole-cadaver tissues is limited. In addition, where PMHS are available, they are not always available within the *age range* or *size range* of interest. Geometry, including body segment parameters (mass, centre of mass, moment of inertia), changes with age [14] as do the properties of tissues [15]. Also, there is great variability in PMHS that reflect the variability in living human subjects. This variability is a strength when planning and conducting experiments as it reflects reality, yet it reduces the replicability of experiments and thus frequently necessitates large number of experiments to account for this variability. These experiments are expensive and complex.

There are a number of further practical difficulties in using human tissues, such as the *biohazard implications*, with the transmission of the viruses that cause hepatitis and AIDS a real possibility [16, 17]. Suitable donor harvesting, storage, and handling protocols can easily address these points [18], however, there are documented cases of disease transmission through handling PMH biological samples [19].

As with all use of human tissues, there are *ethical issues* to consider. Historically, there were abuses including body snatching for anatomical work, but practice and the legal framework for tissue use have changed over centuries. For example, dissection was part of the death sentence for murderers in the sixteenth century; this changed to the authorisation of the dissection of deceased persons whose family members could not afford to pay for a funeral in the nineteenth century. The principle of donation of bodies for research was developed in the United Kingdom in the twentieth century leading to Acts of Parliament requiring consent [20]. Full ethical discussion is not appropriate in this short chapter, but there is consensus that consent must be given freely and without compulsion and that it must be given by an appropriate person; either the deceased themselves ante-mortem, or an alternate person who is aware of the wishes of the deceased [21, 22]. Secondly, there is consensus that the human tissues must be used only for ethical activities, although what constitutes such ethical activities is often debated [23]. Finally, there is consensus that the storage, use and disposal of the human tissues must be done with reverence and care [24].

33.3 Examples of Using PMHS for Blast Injury Research

PMHS can be used to *validate other blast injury models*. For example, in silico models as presented in Chap. 28 are able to model greater variability, differences in posture and effects of various protection and mitigation measures than PMHS. Therefore, they are an efficient experimental tool. However, computational models need validation, and the typical approach is to instantiate the computational model with the specific geometry and material properties from a single (or multiple) PMHS. This PMHS is then tested, and the response is measured. The computational model has the same tested conditions applied (boundary and loading conditions), thus enabling the PMHS to validate the outputs of the computational model. An example of this approach is in the work of Rebelo et al. [25] in which a lower limb computational model was validated for simulation of underbody blast using PMHS blast injury experiments. This computational model was then used to investigate the efficacy of blast mitigation measures. The response of other surrogates such as anthropometric test devices (Chap. 34) can also be validated using PMHS. For example, the WIAMan project seeks to develop an anthropometric test device for use in under body blast testing of ground vehicles [26]. This has been extensively validated using PMHS [27]. A less explored use of PMHS is to employ these to validate or verify the utility of animal models. This could serve to reduce the number of animals used in research and is advocated by reputable organisations such as the National Centre for the Replacement Refinement and Reduction of Animals in Research (http://nc3rs.org.uk).

Surrogates using human tissues can be used to develop *injury-risk curves*. This is described in Chap. 34. These have been used in creating injury-risk curves and injury criteria for under body blast (e.g. [28, 29]), side impact blast loading [30] and secondary blast [31].

Other studies using PMHS answer specific questions on blast injury, many of which are focused on under body blast. These include foot and ankle fractures and the effect of posture [32], the mechanism and timing of failure of different regions of the spine [33] and the differential effect of gender of blast injury response [34]. An example of primary blast applications is the investigation of the mechanism of blast induced traumatic brain injury [35].

33.4 Conclusion

Post mortem human subjects are used widely to answer specific questions in blast injury mechanisms, and protection, yet they are costly, show great variability necessitating multiple experiments, and have some practical and ethical considerations that limit their use. Greater use of these surrogates is in the areas of validation of other models and the creation of injury-risk curves. No suitable alternatives have been found for these applications and thus the use of post mortem human specimens at the tissue, organ and whole body level is likely to continue.

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34 Surrogates: Anthropometric Test Devices

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Abstract

This chapter presents and discusses the use of physical surrogates for the assessment of human injury. The focus is on anthropomorphic test devices (ATDs), otherwise known as dummies. These have been developed in order to evaluate occupant protection in response to impact loading. They are used extensively in the automotive industry to quantify occupant safety and the defence industry for military vehicle assessment. Their main design objective is to be robust and repeatable. Most ATDs do not assess failure directly; instead, they are heavily instrumented with transducers which record data during testing; these data are then analysed to determine the injury risk in the human.

34.1 Introduction

ATDs are designed to be biofidelic meaning that they aim to represent the geometry, mass, mass distribution, kinematics, and kinetics of the human body for a given application [1]. ATDs must exhibit both internal biofidelity (comparable deformations, accelerations and articulations of the body regions) and external biofidelity (meaning similar response when interacting with the surrounding environment). This does not, typically, mean that they are fidelic in terms of failure or biological response. This is because their main design objective is to be robust and repeatable. As such, probability of human injury is determined by comparing the data recorded on transducers mounted on the ATD with relevant injury criteria; injury criteria are parameters (for example, a force) that correlate with injury severity and are associated with an injury mechanism. The goal of a vehicle or a mitigation system design is for the ATD's response (quantified by the injury criteria) for all test conditions at regions of interest to be below a certain value (sometimes termed the injury threshold) that corresponds to a certain risk of injury [2].

Standardised ATDs and equivalent injury thresholds for blast-related loading only exist for assessing the protective efficacy of light armoured vehicles. There is a limited amount of surrogates for investigating the effects of primary blast; these include the Facial and Ocular Countermeasure Safety (FOCUS) head. The external geometry of the FOCUS head-form is designed to replicate a 50th percentile male soldier across the three branches of the US military (Army, Navy and Air Force). The FOCUS head-

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form is capable of measuring forces imposed onto facial structures using internal load cells. Other surrogates for assessing primary blast effects exist within the boundaries of national authorities, but none of these has been standardised. For the remaining chapter, only ATDs that feature in international standards are presented.

34.2 A Brief History of ATDs

There are various types of ATD, each designed to be biofidelic under a specific type of impact loading, mainly for automotive crash testing. For example, the Hybrid III 50th percentile male ATD is commonly used to evaluate automotive restraint systems in frontal crash testing. It was designed to mimic human responses for forehead impacts, neck bending, distributed sternal impacts and knee impacts. The Hybrid III midsize male dummy has been improved since 1976 to enhance biofidelity at the hip and ankle joints. THOR (Test device for Human Occupancy Restraint) is another ATD used for frontal crash testing. Compared to the Hybrid III, the distinguished features of THOR are its two-segment thoracic spine, a human-like rib cage and more human-like neck and ankle structures. The SID (Side Impact Dummy), developed in 1979, is based on the predecessor of the Hybrid III (the Hybrid II) without arms and shoulder structures. It was designed mainly to measure injury risk to the head, chest and pelvis when the body is impacted from the side. The SID-HIII is the SID dummy with its Hybrid II head and neck having been replaced with Hybrid III head and neck. This improved the biofidelity of its head and neck response. The EUROSID1 is the European side impact dummy developed in 1986 and finalised in 1989. The BioSID (Biofidelic Side Impact Test Dummy) was designed to the International Organization for Standardization (ISO) impact response biofidelity guidelines for the head, neck, shoulder, thorax, abdomen and pelvis in 1989.

34.3 ATDs Used to Assess Occupant Safety in Blast

In contrast to the automotive industry applications, currently there is no accepted ATD designed specifically to assess injury associated with mine/IED detonation. The Warrior Injury Assessment Manikin (WIAMan) dummy is currently under development by the US Army; this will be the first ATD specifically designed for an explosive threat. North Atlantic Treaty Organization (NATO) task groups (HFM-090/TG25, HFM148/RTG, HFM-198/RTG (Human Factors and Medicine (HFM) Research Task Group) (RTG)) have suggested the use of the most relevant existing surrogates with appropriate injury criteria when assessing blast injury. These were consulted to inform the NATO standard that encompasses vehicle assessment under blast loading; STANAG (Standardization Agreement) 4569 'Protection Levels for Occupants of Logistic and Light Armored Vehicles' [3].

As current ATDs have been designed to be biofidelic in automotive impact conditions, their validity in predicting injury risk under blast conditions is uncertain. In addition, these ATDs are direction specific. Selection of the appropriate ATDs by the NATO groups was based on the location of the IED with respect to the vehicle occupant, independently from the seating orientation in the vehicle. The standard Hybrid III dummy was chosen for scenarios where the IED is located underneath, in front or at the rear of the ATD. The EuroSID-2RE dummy (ES-2re) was chosen for the scenario where the IED is located laterally with respect to the ATD. This point highlights how an ATD has to be 'tuned' for a specific threat or insult and cannot be easily extrapolated to a broader set of threats.

Specifically for injury assessment of the lower leg, the Military Lower Extremity (MIL-Lx) dummy was designed in 2009 with the objective of being biofidelic under a set loading condition that might occur in an explosion under a military vehicle. It now features in the STANAG 4569 standard as an additional option to the original Hybrid III lower leg, albeit with a different injury criterion.

The next sections elaborate on the ATDs used in the vehicle IED/mine qualification tests.

34.3.1 The Hybrid III ATD

The Hybrid III (H3) 50th percentile male ATD was developed for automotive frontal crash tests. It represents the average male of the USA population between 1970 and 1980 and has the following characteristics:

- stature (standing position): 1.72 m;
- body mass: 78 kg; and
- erect sitting height: 884 mm.

These figures may not any longer accurately represent the current USA population or, indeed, the global population. For example, the northern European population is significantly taller than the H3 50th percentile male ATD [4, 5]; the size difference between the used 50th percentile H3 and the real user population is especially important for the head clearance and thus the risk of head/neck injury due to head contact. The use of the existing 95th percentile H3 does not solve this problem as the difference of this version of the dummy with the user population is large. Therefore, the HFM-148 task group recommended that attention be paid to the head clearance and to quantify the space required for free vertical head motion without contact. This information can be used to advise the desired free head clearance for the required tallest population for the tested vehicle.

The H3 ATD can be instrumented with transducers to measure accelerations, forces, moments and displacements in several body parts. It can withstand a range of loading conditions and is re-useable. The total body mass of the 50th percentile H3 ATD is still close to the mass of the current population when looking at a mixture of male/female 50th percentile population. The use of personal protective equipment (PPE) can increase the human body mass significantly and thus might have an influence on the response observed. There is, however, no standardised method yet to take body mass variability into account.

The standard H3 50th percentile ATD is required to have instrumented legs (tibia and ankle and foot) for IED/mine vehicle qualification. It has a curved spine in order to represent the seating posture of a driver in a car. The spine may be replaced by a Hybrid II straight spine to simulate standing and lying postures. The HFM148 task group has suggested the use of the standard version of the Hybrid III (curved spine) for the most realistic seating posture in the majority of the current seats in military vehicles. It is known that the spine configuration has an influence on the load transfer into the upper body. However, this is deemed to have little influence on the pelvis acceleration, which is used as input for the lumbar-spine injury risk assessment. More investigations need to be carried out to analyse the response of the two available spines (straight/curved) for future development of the spine injury criterion. For standing or lying positions, the same measurement method could be used to assess injury risk assessment, but it is not known whether the injury criteria and risk curves are still valid for these conditions.

34.3.2 The Eurosid-2RE ATD (ES-2re)

The Eurosid-2RE (ES-2re) ATD is a side impact dummy developed in the European automotive community to expand the capabilities for the crash safety protection measures. The ES-2re 50th percentile male ATD represents the average adult male, but without lower arms. It has the following anthropomorphic characteristics:

- mass: 72 kg;
- erect sitting height: 909 mm;
- shoulder width: 470 mm; and
- pelvis lap width: 366 mm.

As with the 50th percentile H3 ATD, there are differences in the anthropomorphic data compared to the current population. The same comments as the H3 with regard to the size of the ATD hold for the ES-2re. The most salient point, however, is the difference in shoulder width, because this determines the free space to the side wall. The narrower the free space is, the harder the impact to the shoulder will be in some loading situations. This means that when the shoulder force in the 50th percentile ES-2re shows low risk of shoulder injuries (<10%), the risk for a real occupant with wider shoulders might be higher. Accurate measurements of side wall intrusion during side blast could help to formulate ATD requirements.

The head of the ES-2re is the same as the H3 head, albeit the neck has a different design in order to mimic the kinematic behaviour of the human under lateral loads. There is an upper neck load cell with which loads through the neck can be measured. Compressive force can be measured on both shoulders.

The thorax of the ES-2re incorporates a rear rib extension bracket on the impact side of each rib that, together with a rear rib extension guide, provides more realistic interaction with vehicle seatbacks. The deflection caused by lateral impact is measured in each rib. In the abdomen, the lateral force at three positions is measured and summed to get the total force for the injury risk assessment. In the pelvis, the lateral force is measured at the pubic symphysis.

The standard ES-2re ATD comes with non-instrumented tibias. For IED/mine vehicle qualification tests, both of its lower legs (including tibia, ankle and foot) have to be replaced by instrumented lower legs.

34.3.3 The MIL-lx

The Military Lower Extremity (MIL-Lx) was designed with the objective to be biofidelic under a set loading condition that might occur in an explosion under a military vehicle (Fig. 34.1). The design of the leg is based on both the original H3 leg and the THOR-Lx leg and optimised for measurements of the vertical force through the tibia.

The said biofidelity of the MIL-Lx was achieved using a lower leg injury assessment study at Wayne State University. The loading regime for under-vehicle blast detonations was used as input for PMHS testing. The same set-up was used to develop and tune the MIL-Lx by aiming for a good correlation in the load-time (impulse) response at approximately 50% injury risk.

The biofidelic tuning of the MIL-Lx was achieved primarily by the use of a rubber compliant element in the tibial shaft. The axial load measured at the upper tibia load cell (located above the compliant element) is the one that correlates to the load measured on the PMHS study at Wayne State University and therefore is the one to be used for the prediction of risk of lower leg injuries; this is in contrast to the H3 lower leg where the measurement used for injury assessment is the axial load measured at the lower tibia load cell.

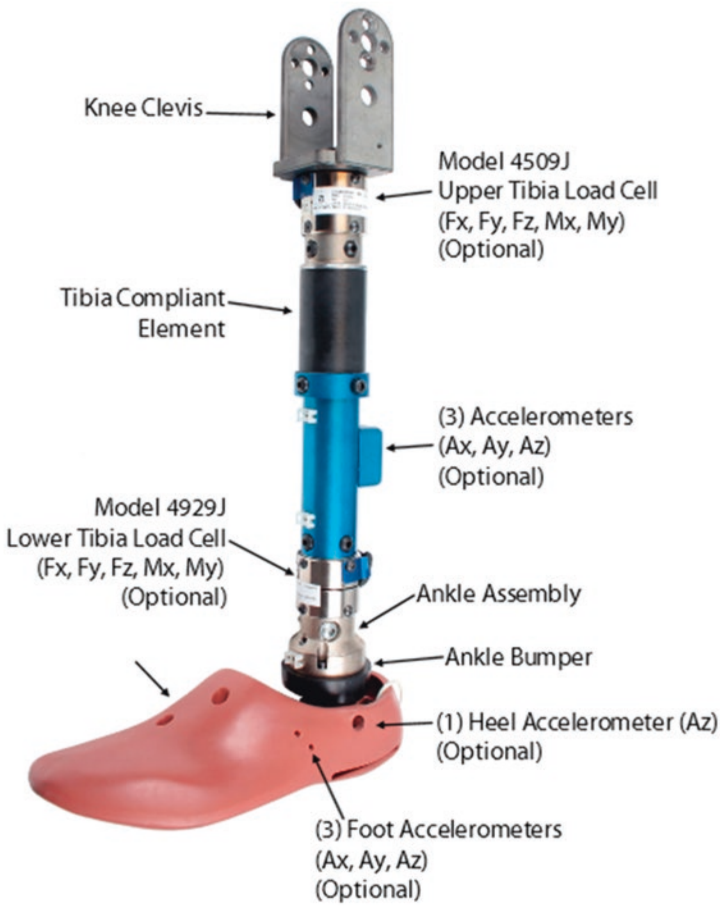

Fig. 34.1 The Military Lower Extremity (MIL-Lx) (adapted from NATO HFM-148/RTG [5])

The Hybrid III, Eurosid-2RE and MIL-Lx are commonly referred to as multi-use surrogates. They are designed to be used for multiple impacts at loading conditions beyond those normally intended for testing evaluations. As discussed above in Sect. 34.2, ATDs lack frangibility by intention of design and therefore do not assess failure directly.

34.3.4 Warrior Injury Assessment Manikin

The Warrior Injury Assessment Manikin (WIAMan) is a novel ATD designed by a consortium of the US Army, academia and industry to assess vehicle occupant protection in underbody (under the vehicle) blast scenarios. The ATD is specifically designed for upright military seats, with the option to adjust for reclined and slouched postures. While WIAMan is a ground-up new design, it is loosely based on the Hybrid III 50th-percentile male frontal automotive crash test dummy that has been modernised to represent today's soldier. Specifically, anthropometry and posture are based on a 50th-percentile male soldier [6] and the Seated Soldier Study from the University of Michigan Transportation Research Institute (UMTRI) [7]. The new ATD is intended to provide more realistic posture, seat interaction, and dynamic response for these loading modes and for military seating scenarios than the Hybrid III family of ATDs that are currently used. Dynamic response of the anatomic regions of the WIAMan is based on target biofidelity corridors, generated from biomechanical PMHS tests. Every component of the dummy, from the fasteners to the flesh, was designed to improve its biofidelity and performance. The three-dimensional shape requirements for the dummy were determined by full-body surface scans of individuals to allow for the appropriate installation of personal protective equipment. The WIAMan is instrumented with over 150 sensors such as load cells, accelerometers and angular rate sensors at key locations with a high-tech data acquisition system that is entirely contained within the dummy. This system will assess the relevant human kinematic response and injury-related measurements along critical load paths.

34.3.5 Frangible, Single-Use Surrogates

Frangible, single-use surrogate lower limbs have been specifically designed to assess the effects of underbody blast. Frangible surrogates may consist of anatomically correct synthetic bones and soft tissues to mimic closely the human anatomy. Examples of these include the Complex Lower Leg (CLL) and the Frangible Surrogate Leg (FSL) [8–10]. Developed in Canada, the CLL was claimed to show realistic injury patterns and biofidelic response under footplate velocities of 3.4–8.5 m/s [9]. More recently, experiments conducted at Wayne State University have shown that the CLL failed at lower loads in comparison to cadavers [5]. The FSL has been compared with human cadaveric data in a landmine experimental set-up; the findings showed good correlation with respect to gross bony damage, but low biofidelity in soft tissue and cancellous bone response [8]. The FSL has not been validated for under-vehicle blast research. More research needs to be carried out before single-use surrogates can be implemented in test standards.

34.4 Injury Risk Assessment

Physical injury occurs when loading to the human body causes damage to anatomical structures and/or alteration in normal function. The mechanism involved to cause such damage/alteration is then called an injury mechanism.

An injury criterion is defined as a physical parameter or a function of several physical parameters which correlates well with the injury severity of the body area under consideration for a specific loading condition. Parameters that can be measured when testing may include the linear acceleration experienced by a body part, the global forces or moments acting on the body or the deflection of a structure.

Injury risk curves are used to define the probability of exceeding the severity of a set injury. Examples of such curves are shown in Fig. 34.2 [5]. The vertical axis shows the injury risk as a function of an injury criterion (horizontal axis). These are usually produced from series of PMHS, animal, or human volunteer experiments under appropriate loading and boundary conditions. Sometimes, anthropomorphic factors such as age and gender are included in the injury risk curves. The data set on which the injury risk curves have been developed, however, is usually unable to account for such factors. Depending on the shape of the injury risk curve, a small difference for tolerance level could result in large differences in injury risk. Tolerance level or injury criterion level is defined as the threshold of the injury criterion corresponding to a specific risk to sustain a specific injury severity (range).

The injury severity can be defined using an injury scale which is a numerical classification of severity of an injury and can be related to injury type and body part affected. The Abbreviated Injury Scale (AIS) was developed in the 1970s to score the severity of injuries in victims of road traffic accidents on a scale of 1–6, based on the likelihood of the event causing a fatal injury and is discussed in full in Chap. 11 [5, 11]; higher AIS levels indicate an increased threat to life. The NATO HFM-090 task group suggested the use of the 10% probability of an AIS 2 or greater (AIS 2+) injury to establish the acceptable injury threshold for the different body regions of interest; this is included in the standard STANAG 4569 for light armoured vehicle assessment. Accepting a 10% risk of AIS 2+ implies a lower risk (less than 10%) of higher severity (AIS 3+) injuries and a higher risk (more than 10%) of AIS 1+ injuries. Higher injury severity levels or other injury severity scales with refined discriminating capacities are used if the available AIS2+ information is lacking or if AIS 2 injuries are considered not to be acceptable. Table 34.1 [5] gives an overview of the injury criteria and corresponding ATD that should be used for the different body parts recommended by the NATO HFM-148 task group

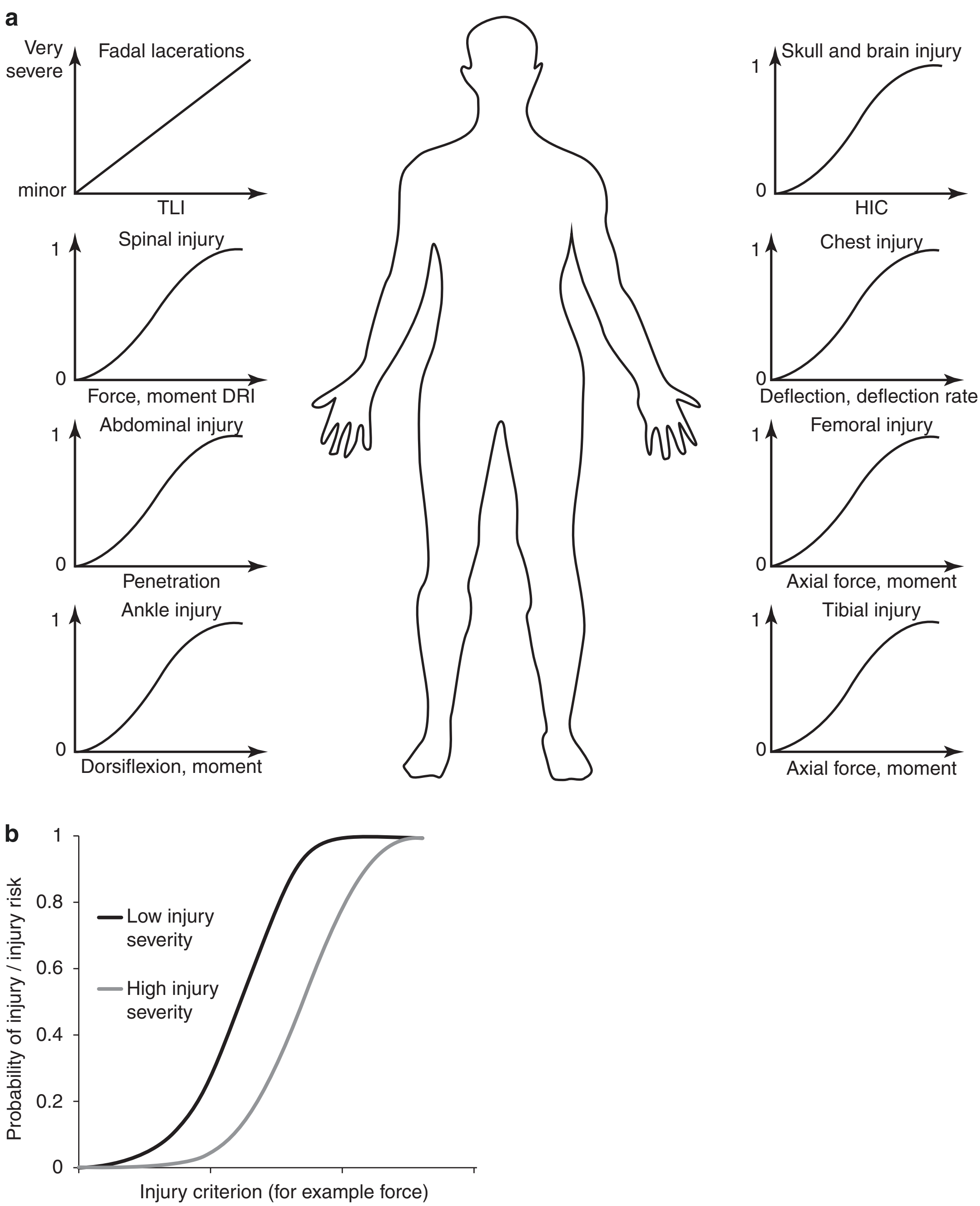

Fig. 34.2 Examples of injury risk curves suggested for (**a**) different body parts (adapted from NATO HFM-148/RTG [5]) and (**b**) different injury severities

Table 34.1 Injury assessment reference values and corresponding ATD to be used in order to assess injury risk for the different body parts according to NATO. Adapted from NATO HFM-148/RTG [5]

Body region	Injury criterion	Metric	Pass/fail level	ATD
Head	Head injury criterion	HIC15	250	H3 or ES-2re + MIL-LX
Neck	Axial compression force	Fz-	4.0 kN @ 0 ms/1.1 kN >30 ms	ES-2re + MIL-LX
	Axial tension force	Fz+	3.3 kN @ 0 ms/2.8 kN @35 ms/1.1 kN > 60 ms	
	Shear force	Fx+−/Fy+−	3.1 kN @ 0 ms/1.5 kN @25–35 ms/1.1 kN > 45 ms	
	Bending moment (flexion)	Moc_y+	190 Nm	
	Bending moment (extension)	Moc_y-	96 Nm	
Neck	Axial tension force	Fz+	1.8 kN	ES-2re + MIL-LX
Shoulder	Compression force	Fy	1.4 kN	
Thorax (ribs) (upper/middle/lower)	Rib deflection criterion	$RDC_{lateral}$	28 mm	ES-2re + MIL-LX
Thorax	Thoracic compression criterion	$TCC_{frontal}$	30 mm	H3 + MIL-LX
Thorax	Viscous criterion	$VC_{frontal}$	0.70 m/s	H3 + MIL-LX
Thorax	Viscous criterion	$VC_{lateral}$	0.58 m/s	ES-2re + MIL-LX
Abdomen (front/middle/rear)	Abdominal peak force	F_{total}	1.8 kN	ES-2re + MIL-LX
Spine	Dynamic response index	DRI_z	17.7	H3 or ES-2re + MIL-LX
Pelvis	Maximum pubic force	Fy	2.6 kN	ES-2re + MIL-LX
Upper legs	Axial compression force	Fz-	6.9 kN	H3 or ES-2re + MIL-LX
Lower legs	Axial compression force	Fz-	2.6 kN	H3 or ES-2re + MIL-LX
Internal organs/lungs	Chest Wall velocity predictor	CWVP	3.6 m/s	H3 or ES-2re + MIL-LX

34.5 Summary

Human surrogates have been utilised throughout the years to aid in the understanding and prediction of injury under impact in order to improve protective systems and inform mitigation strategies. Commercial ATDs used in standardised test procedures against explosive threats have been designed for automotive applications, and therefore their response in blast does not represent optimally the human body. Cadaveric tests are used to derive injury risk curves in blast-related loadings (Chap. 33); these are ever more utilised currently to improve our understanding of the response of current ATDs and to develop new ATDs or parts thereof that are biofidelic in blast-related loadings. These developments will improve occupant safety through the ever more accurate evaluation of military vehicle design and mitigation strategies.

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35 Physical Models for Assessing Primary and Secondary Blast Effects

John Breeze, Debra J. Carr, and Peter F. Mahoney

Abstract

Physical models are used in experiments to assess the effects of primary and secondary blast with respect to the performance of munitions, understanding wounding and the effectiveness of personal protective equipment (PPE). Such models include post mortem human subjects (PMHS), animal surrogates and specific tissue simulants. Each model has its own advantages and disadvantages, such that accurate representation of the mechanical properties of live human tissue currently requires a combination of such models. This chapter summarises physical models used for assessing primary and secondary blast injury. A short introduction is provided, followed by a consideration of blast testing, fragmentation testing and combined blast/fragmentation testing.

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35.1 Introduction

Physical models are used in experiments to assess the effects of primary and secondary blast with respect to the performance of munitions, understanding wounding and the effectiveness of personal protective equipment (PPE). Such models represent a single region of the human body (e.g. thorax, head or a leg) or individual anatomical components (e.g. lungs, skin and muscle). Post mortem human subjects (PMHS; human cadavers) have been utilised as physical models as their anatomy is the closest to that of a live human. However, their use has declined due to regularity issues and in some countries is considered unethical [1]. Some organisations are able to use PMHS and have reported results from blast, or simulated blast testing, e.g. [2–4]. Some care must be taken in the interpretation of results from PMHS due to physical and mechanical changes that occur post mortem (see Chap. 33). Alternative physical models used include live anaesthetised animals, cadaveric animals and specific human tissue simulants. Ethical considerations need to be addressed when using living animals in many countries; cadaveric animals prepared to human food consumption are usually less problematic, but also less representative of the mechanical properties of live tissue.

A typical munition might be a metallic container filled with explosive or a container filled with metallic objects (e.g. nails, nuts, ball bearings) and explosive. Munitions may be traditional military munitions (e.g. artillery shells, mortars, grenades) or improvised explosive devices (IEDs). On initiation, the explosive fractures the container and the expanding gases accelerate the fragmentation towards a target which typically suffers multiple fragment impacts of varying depths of penetration (DoP) [5–7]. The fracture pattern of many munitions is random; therefore, the fragmentation will be produced in varying shapes and mass [6]. Initial fragment acceleration is dependent upon mass; heavier, faster fragments will have more capacity to inflict damage (higher initial kinetic energy). Smaller, lighter fragments will tend to have faster initial velocities, but they will be affected more by aerodynamic resistance. This is linked closely to kinetic energy density, i.e. kinetic energy divided by presented cross-section area. If the person is close enough to the device when it initiates, injuries may combine the effects of blast and impacts from fragmentation [8] (Fig. 35.1).

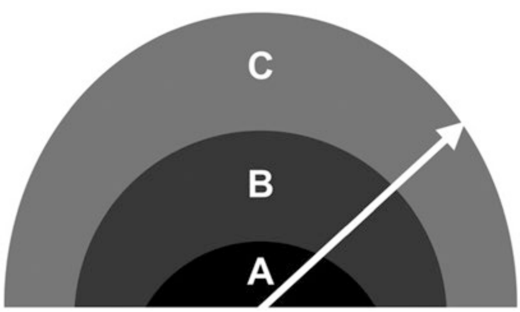

Fig. 35.1 Representation of primary and secondary blast effects. (**a**) Fatal blast and fatal fragmentation, (**b**) Blast plus fatal fragmentation, (**c**) Fragmentation only, Arrow = Distance from initiation

35.2 Physical Models for Primary Blast Experiments

Primary blast testing is usually conducted using explosives in an open area range or using a shock tube. In the open, data collected is typically pressure-time values from pressure gauges mounted at varying distance from the device, e.g. [9]. The lethal radius of primary blast is defined using a 50% lethality rate for pressure, which is accepted to be 896–1241 kPa [10]. Injury tolerance values cited in the literature include 103 kPa for a 50% rate of eardrum rupture and 552 kPa for a 50% rate of lung injury [10].

The primary model used by many research groups for assessing the likelihood of lung injury from blast is an animal, usually anesthetised during testing and sacrificed by exsanguination under the influence of the anaesthetic post-testing for post mortem inspection. Swine, sheep and small animals have all been used, e.g. [11–16]. In the 1990s, an apparatus to assess the risk of blast injury to the lungs without the use of animals was developed in the United Kingdom (UK) [17]. The peak acceleration of swine thoracic wall was related to the severity of injury to the lung and this data used to develop the thoracic rig. A good summary of the Ministry of Defence's research in this area is available [18]. Other research groups have used instrumented anthropometric mannequins [19–21] or representations of the human body [22] for assessing lung injury.

Work has been conducted to develop models for the assessment of blast injuries to the leg using buried charges; these models include gelatine with or without animal bones [23, 24] and various metallic and wooden rigs [25, 26]. Many nations have developed mechanical test equipment to simulate vehicle underbelly blast. Comparisons between human cadaveric material and mechanical surrogates are rare [27–29].

Shock tubes have also been used to investigate injury, e.g. [13, 30–32]. The targets, which are often instrumented, range from animals (swine, sheep and small animals) to human torso shaped manufactured using synthetic materials [32] and representations of the head, e.g. [33–35].

35.3 Physical Models for Assessing the Effectiveness of Secondary Blast

Secondary blast injuries occur when fragments from a device or the surrounding environment (e.g. in the instance of a buried charge) impact the human body. How a projectile interacts with, penetrates into and perforates through (if it does) the human body depends on many factors including (i) the type and shape of projectile, (ii) the kinetic energy at impact, (iii) where on the body the projectile impacts and (iv) any clothing or PPE that might be worn. Wounding patterns are impossible to predict without knowing the full details of the incident, and even then, variability will exist. It is not the intention of this chapter to provide an in-depth discussion of the physics of wound ballistics; that topic is covered in Kneubuhl's classic text [36]; however, a short summary is required. Wounding occurs because kinetic energy is dissipated in the body due to the projectile interacting with it. When a projectile penetrates the human body, a temporary cavity is formed which collapses to leave permanent damage; the shape and size of the temporary cavity are influenced by the elastic properties of the soft tissue type in which it forms and whether the projectile deforms, fragments, or tumbles. If the projectile remains in the body, then all of the kinetic energy of the impact event is transferred to the tissues. That fragments (e.g. steel ball bearings which do not typically deform nor yaw) form a temporary cavity as well as a permanent tract comprises some researchers not familiar with the field (Figs. 35.2 and 35.3).

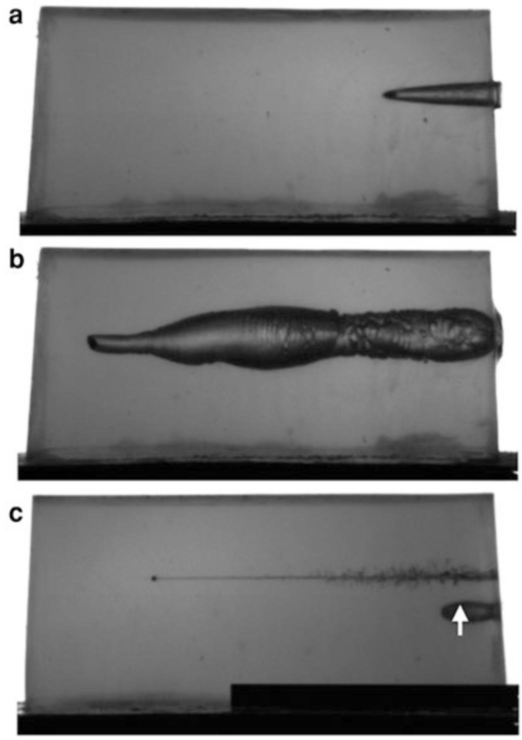

Fig. 35.2 High-speed video stills of a 20% gelatin block being penetrated by a 5 mm spherical FSP (**a**), demonstrating temporary cavity (**b**) and permanent cavity (**c**). Arrow marks the position of a temperature probe

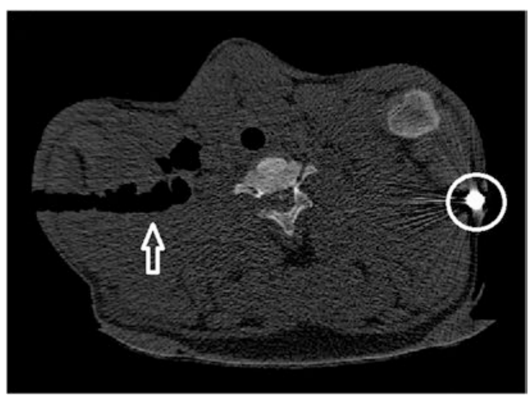

Fig. 35.3 Penetration of a 2.84 g chisel nosed FSP into swine tissue creating a permanent cavity (arrowed; maximum diameter of 20.4 mm). After hitting the vertebra, the FSP had insufficient residual velocity to perforate the specimen and was lodged under the skin (circled). Provided by JB

Experiments that investigate the effectiveness of fragmentation fall into two categories (i) those that consider the penetrative performance of a single fragment and (ii) those that investigate the performance of a device which when detonated produces multiple fragmentation (i.e. combined blast/fragmentation effects).

35.3.1 Single Fragment Testing

Single fragment testing is typically used for (i) investigating the performance of fragment protective equipment (e.g. body armour or helmets) or (ii) wound ballistics studies [37, 38]. Experiments rarely use single actual fragments [39], but more commonly fragment simulating projectiles (FSPs; chisel nosed, right circular cylinder or spherical) (Fig. 35.4) [38]. The size and hardness levels of FSPs reportedly represent a range of fragments recovered after detonating 20 mm, 37 mm and 105 mm High Explosive (HE) artillery shells [40].

Fig. 35.4 Examples of actual fragments from a Mortar Bomb 81 mm HE L41 (left) and 1.1 g chisel nosed FSP (right; circled). Photograph: DJC

Assessing the performance of fragment protective equipment is conducted according to NATO AEP 2920 using steel FSPs and provides a measure of the protection offered by such items [38]. These experiments do not usually involve simulants to investigate penetrative wounding. However, there has been some research published that has considered wounding that might occur when clothing or PPE is overmatched by bullets as an example [41–45].

In wound ballistic studies, blocks of gelatine are often used to investigate the effects of penetrating impacts using a material that it is assumed represents (some part of) the human body [37]. Gelatine blocks of 10% concentration by mass (conditioned to 4 °C) resulted in a depth of penetration (DoP) to within 3% for selected bullets compared to those in living swine thigh tissue [46, 47] and similar DoP to swine torso [48]. Breeze et al. (2013) reported similar DoPs for three sizes of FSPs in 20% (by mass) gelatine and swine cadaveric leg and neck tissue at velocities ranging between 112 and 1652 m/s [49]. A comparison of wounds caused by 4.8 mm diameter ball bearings (1150 ± 5 m/s) in 10% (by mass) gelatine and the legs of anesthetised swine reported similar trajectories in both targets, DoP within 1% and the pattern of temporary cavity formation and collapse being similar, but the maximum size and duration larger (12%) and longer (24%) in gelatine [50]. Post-testing, dissection of the gelatine block and measurement of permanent damage usually occur in addition to analysis of (usually) high-speed video. Other simulants that have been used in a similar way to gelatine include synthetic materials such as PermaGel™ and Clear Ballistics Gel®. However, ballistic impacts into these materials produce different damage when compared to gelatine blocks [51, 52].

Many researchers have recognised that projectiles might interact with structures other than muscle during a penetrating impact. Therefore, human and/or animal bones are often combined with gelatine to produce a target with improved biofidelity. Examples include the use of human, swine and deer femur [53–55], ribs [56, 57], human whole skulls [58], or use of flat bones to represent the skull [43]. A number of polymeric bone simulants exist and have been used to represent skulls [59–61] and other bones [62, 63]; these may be a simple geometric representation or anatomically accurate (Fig. 35.5). Reproducing actual events can involve complex experiments that include accurate representations of a physical model and a scenario as well as appropriate expertise to design the experiment and analyse the data obtained, e.g. [64, 65].

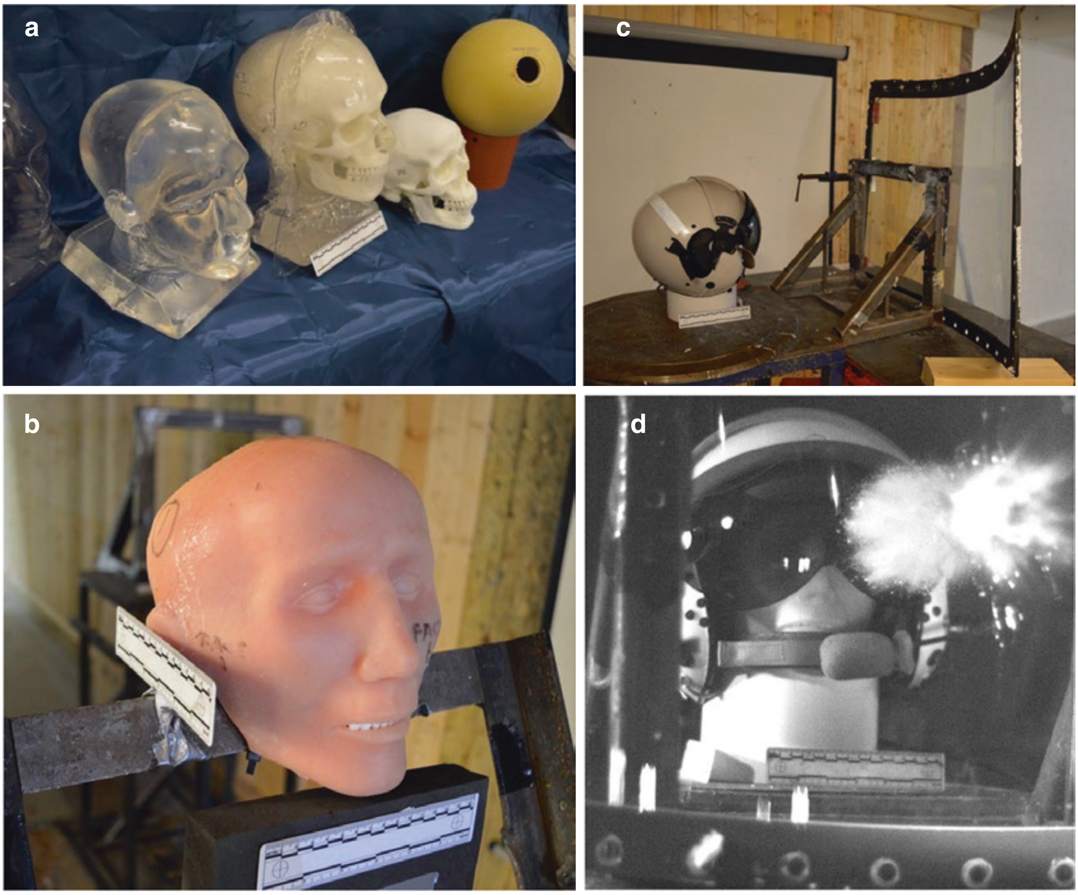

Fig. 35.5 Examples of head models of different complexity and the development of a scenario. Photographs: PFM. (**a**) Clear Ballistic Gel head, anatomically correct skull with PermaGel™ tissue made using a mould from the Clear Ballistic Gel head, anatomically correct skull, SYNBONE®sphere (left to right), (**b**) Soft tissue layers over the anatomically correct skull in Fig. 35.5a, (**c**) Model in Fig. 35.5b with an aviation helmet, mounted behind a military helicopter windscreen, (**d**) Image from high-speed video of a ballistic impact through the military helicopter windscreen on to the model in Fig. 35.5c

Minimal work has been conducted that considers the penetrative effect of single fragments from a wounding perspective; exceptions include the use of gelatine [49, 66], PermaGel™ [66, 67], animal tissue [49, 50, 68, 69] and strawboard [39].

35.3.2 Combined Blast/Fragmentation Investigations

Combined blast/fragmentation testing typically utilises outdoor arena trials or tests in a containment facility; both types of testing are expensive requiring specialised regulated facilities. Tests conducted outdoors are typically in the form of an arena trial—the device is placed centrally, and the targets arranged around at varying distances. These targets are usually packs of strawboard which are used to collect fragments and assess their spread and penetrative capability, e.g. [70–73], but a representation of the human body (usually anaesthetised or cadaveric swine) and / or PPE or other equipment may also be used. Three millimetre thick strawboard is typically used in packs of 20-layers [74]. Some open area ranges do not allow contamination by fragmentation and therefore testing is conducted in a containment building. These buildings are rated to an amount of explosive that can be detonated inside them. The space inside the building is usually limited, but targets can be arranged to assess munition performance with respect to wounding and/or performance. Examples include performance of PPE and a consideration of wounding when challenged with mortars, grenades and buried charges [9, 39, 75, 76].

An alternative to using arena trials or a containment facility was proposed by Pinto [24]. Buried charge blast testing of antimine footwear was simulated by firing sand at a 20% (by mass) gelatine leg (cast in a Wellington boot) in a Rarden shell using a pressure housing fitted with a 1135 mm long 30 mm diameter barrel. The slug of sand produced reportedly resulted in a similar sand velocity, impulse and damage patterns compared to those observed when blast testing using a buried charge. Other work of interest includes the use of buried charges to test blast resistant vehicles, PPE and scaled testing of pelvic protection using charges buried in different media, e.g. [26, 76, 77].

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36 Tertiary Blast Injury and its Protection

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Abstract

Tertiary blast injury results from the interaction of the body with solid structures due to bodily or structural displacement caused by blast. This chapter presents briefly the types of tertiary blast, the epidemiology of tertiary blast injury and key advancements in protecting against it.

36.1 Introduction

Tertiary blast injury is the result of the human body displacing due to the force of blast, the blast wave and the following wind potentially carrying debris (Chap. 9). A form of tertiary blast injury is also what is termed 'solid blast' injury, which is the displacement, due to an explosion, of an object against the human body. An example of solid blast is the vehicle hull encroaching into parts of the human body which may be in contact with, or in close proximity to, it. Another example of solid blast is behind-armour blunt trauma (BABT); BABT is the injury caused by the encroaching of body armour into the human body, such as a thoracic plate deflecting and deforming the chest wall. Limb flail causing traumatic amputation may be considered also as a form of tertiary blast; this mechanism of injury will not be discussed in this chapter and the reader is directed to the recent work by Rankin et al. [1] and to Chap. 15. This chapter focusses on tertiary blast injury due to the interaction of vehicles and infrastructure with the human body.

36.2 Vehicles

Explosive devices, in the form of anti-vehicle (AV) mines, buried artillery rounds, above ground explosives, shaped charges, and car bombs, to name but a few, have been common threats to military vehicles. They are all designed to debilitate personnel and incapacitate vehicles. A detailed overview of the threat, the mechanism of injury, the epidemiology, and the injury assessment methods and criteria for military vehicles is given by the unclassified NATO Human Factors in Medicine (HFM) reports published over the last 13 years (HFM-090 in 2007, HFM-148 in 2012, and HFM-198 in 2016), with the latest one (HFM-271) due to become available in 2021. The relevant NATO standardisation agreement (STANAG) for assessing the protection levels of military vehicles is the STANAG 4569 AEP 55,

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Protection Levels for Occupants of Logistic and Light Armoured Vehicles.

The explosive threat to lightly armoured civilian vehicles is commonly a grenade; it is usually aimed at disrupting the mobility of the vehicle rather than injure its occupants. The protection levels required for lightly armoured civilian vehicles can be found in the BSI PAS 300 (2018 is the latest edition) and is not going to be discussed any further in this chapter.

Explosive devices attacking a military vehicle are usually buried in the ground or placed on the roadside. The result of the detonation is the loading of the vehicle hull relatively locally causing it to deform at the first instance (local effects), and, depending on the size of the blast, its location relative to the vehicle, and the mass of the vehicle itself, to be displaced as a whole a few milliseconds later (global effects). It has been established that injury to the occupants in almost all incidents occurs due to local effects, at the first few milliseconds of the event.

The injury profile of the occupant depends on the type of vehicle and the protection it offers. The most commonly injured body regions reported from recent conflicts are the lower extremity and the spinal column [2–4]. Injuries to the head and torso are the most severe; approximately half of the fatalities can be attributed to head injury and one-fifth to torso trauma [5]. Of the 426 UK mounted casualties in recent conflicts secondary to an explosive mechanism with an AIS ≥2, 297 (70%) were survivors and 129 (30%) were fatalities. Table 36.1 shows the distribution of injury across body regions [4]. Looking into associations between injured body regions (Table 36.2), there is a statistical correlation between injuries to the calcaneus and the tibia, but not between injuries to the leg and any other body region. Conversely, there is statistical correlation between femoral, pelvic, spinal, head and torso injuries [4]. These data suggest that there are two distinct mechanisms of solid blast injury in the mounted casualty; one stemming from the floor of the vehicle affecting mostly the legs and one from the seat affecting the upper body.

Table 36.1 Incidence of injuries (by region) of severity AIS 2 or higher amongst UK casualties ($n = 426$) in vehicles attacked by a mine in the recent war in Iraq and Afghanistan (2003–2014) [4]

Body region injured	Number of cases (% of total in-vehicle casualties)
Calcaneus	97 (23%)
Tibia	145 (34%)
Femur	49 (12%)
Pelvis	85 (20%)
Spine	171 (40%)
NCTH	111 (26%)
Head	135 (32%)

NCTH non-compressible torso haemorrhage

Table 36.2 Two sided Pearson's χ^2 correlations between injuries in the mounted casualty. Statistically significant correlations ($p < 0.05$) are denoted in bold [4]

	Calcaneus					
Tibia	**<0.001**	Tibia				
Femur	0.08	**0.04**	Femur			
Pelvis	0.85	0.71	**<0.001**	Pelvis		
Spine	0.49	0.38	**0.001**	**<0.001**	Spine	
NCTH	0.10	0.61	**<0.001**	**<0.001**	**<0.001**	NCTH
Head	0.06	0.83	**0.006**	**<0.001**	**<0.001**	**<0.001**

Arguably the best summary of what may protect a vehicle from AV mines is detailed in Ramasamy et al. [6] who analysed data from 2212 mine incidents involving 16,456 casualties that occurred during the Rhodesian bush war (1972–1980). The analysis showed that improvised vehicle modifications in the form of V-shaped hulls, increased ground clearance, increased wheel-axle length, and increased vehicle mass all reduced significantly injury occurrence and fatality rates. Combinations of these features can be seen in contemporary armoured vehicles.

36.3 Infrastructure

Explosive attacks (bombings) inside or in proximity to buildings have been common in terrorist attacks on civilian targets. The main blast-injury mechanisms in bombings, whether they are in confined or open spaces, are primary and secondary rather than tertiary.

Edwards et al. [7] carried out a review of all terrorist bombings since 1970, using the Global Terrorism Database (GTD) and a rigorous literature search. Building collapse, which is a tertiary blast-injury mechanism, albeit a rare result of these attacks causing only 3.9% of all injuries documented, is the most fatal. The number of injuries that were attributed to a tertiary blast-injury mechanism were 12.7% of all injuries documented per incident; these injuries were skeletal fractures or traumatic amputations.

The incidence of terrorist bombings in the last decade has increased dramatically compared to previous decades [7]. Therefore, it is not surprising that ensuring resistance of infrastructure to blast loadings has been in focus in recent years. Key aspects of blast resistance are to generate standoff (that is, ensure that the explosive will not detonate close to the building) and prevent collapse. Reinforced concrete and steel-framed structures are known to be blast resistant; the incidences of building collapse causing tertiary blast injury have been very rare and due to large amounts of explosives positioned in key structural locations [8]. Therefore, recent developments have been mostly focused on building façades and particularly the glazed areas, which have been more susceptible to blast loading and their rupture could be a significant cause of injury primarily due to a secondary blast-injury mechanism. There exist standards and guidelines for quantifying the blast resistance of buildings and for calculating the potential risk an explosive threat might pose to a building. The interested reader is directed to the excellent book entitled 'Blast Effects on Buildings', edited by David Cornie, Geoff Mays and Peter Smith [9].

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37 Optimising the Medical Coverage of Personal Armour Systems for UK Armed Forces Personnel

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Abstract

Historically, personal armour was designed to prevent death, but there is an increasing recognition that prevention of those injuries causing long-term morbidity is also required. To optimise the coverage of future personal armour systems for UK Armed Forces personnel, the required medical area of coverage has been defined, which can subsequently be modified by factors such as tactical considerations on the ground, ergonomics, weight restrictions and equipment integration. Protection levels of personal armour should be selected to correspond to the threat that will be encountered, within the constraints of acceptable human factors considerations, however, this is outside the scope of this chapter. This chapter summarises medical coverage for each body region.

37.1 Introduction

Historically, personal armour was designed to prevent death, but there is an increasing recognition that prevention of those injuries causing long-term morbidity is also required [1]. For example, ballistic eyewear has been worn for many years and has been shown to be very effective at mitigating ocular injuries [2], but more recently pelvic protection was introduced due to the long-term morbidity from genital injuries [3]. However, any protective system will be a compromise between the degree of protection and the encumbrance or 'burden' on the wearer [4]. To optimise the coverage of future personal armour systems for UK Armed Forces personnel, the required medical area of coverage has been defined, which can subsequently be modified by factors such as tactical considerations on the ground, ergonomics, weight restrictions and equipment integration. This chapter summarises medical coverage for each body region. Protection levels of personal armour should be selected to correspond to the threat that will be encountered, within the constraints of acceptable human factors considerations, however, this is outside the scope of this chapter.

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37.1.1 Essential and Desirable Medical Coverage

The exact medical coverage requirements for UK body armour have not been openly published until recently, previously making objective comparisons between designs more difficult. In 2016, Breeze et al. introduced the terms *essential* and *desirable,* in relation to coverage of anatomical structures to enable such comparisons [1]. The essential and desirable medical coverage provided by elements of personal armour are medical judgments and should be independent of the level of ballistic protection. (Table 37.1) [5].

Table 37.1 Essential and desirable coverage definitions; adapted from [1]

Essential medical coverage	Those anatomical structures that, if damaged would likely lead to death prior to definitive surgical intervention being available, for example, bleeding from the thorax that cannot be compressed and requires surgical access (thoracotomy) to arrest it. This was based upon a premise that damage control surgery would be performed within 60 min of injury
Desirable medical coverage	Those anatomical structures potentially responsible for mortality which, if damaged, would cause morbidity necessitating lifelong medical treatment or that result in significant disability. This includes physiological disability as well as psychological disability, for example, damage to the lower parts of the spinal cord (lumbar or sacral parts) that may result in significant loss of function of limbs, or damage to the genitalia that may result in psychological trauma

37.1.2 Vulnerable Anatomical Structures

Anatomical structures were identified that, if damaged, were highly likely to result in death within 60 min (essential medical coverage) or would cause death after that period or result in significant long-term morbidity (desirable medical coverage). This was based upon a premise that damage control surgery to arrest haemorrhage would be performed within 60 min of injury [1]. This was representative of the mature medical evacuation chain present in the latter part of Operation HERRICK in Afghanistan [6]. Strictly defining the time to surgery and not just 'time to medical care' as used in the past was important as surgery is the only means of arresting non-compressible haemorrhage. Fluid resuscitation, compression and novel haemostatic agents are important adjuncts but are not a substitute for surgery. The essential and desirable coverage structures are defined in the Tables 37.2 and 37.3, and the medical rationale for these definitions are available in the relevant papers [1, 7–13]. It is recognised that to date much of the research, as demonstrated by these tables, has focused upon male service personnel. More recently, data has been collected by the Royal Centre for Defence Medicine using the CT scans of injured female UK service personnel; these will be incorporated into the design of future personal armour.

Table 37.2 Anatomical structures comprising essential medical coverage in males [1, 7–13]. The thoracic trachea has been added since the initial medical consensus

Head and face	Neck	Torso (thorax and abdomen)	Arms	Pelvis and thigh
Cerebral hemispheres	Spinal cord (C1-C5)	Heart	Axillary arteries	Iliac arteries
Brain stem	Carotid arteries	Aorta	Brachial arteries	Femoral arteries
Cerebellum	Vertebral arteries	Vena cava		Aorta
	Larynx	Liver		Inferior vena cava
	Cervical trachea	Bronchial arteries		Iliac veins
		Pulmonary arteries		
		Pulmonary veins		
		Spleen		
		Subclavian artery		
		Subclavian vein		
		Thoracic trachea *		

Table 37.3 Anatomical structures comprising desirable medical coverage in males [1, 7–13]

Head and face	Neck	Torso (thorax and abdomen)	Arms	Pelvis and thigh
Eyes	Oesophagus	Oesophagus	Median nerve	Testes
Optic nerve	Pharynx	Pharynx	Ulnar nerve	Anus
Nose	Vagus nerve	Lungs	Radial nerve	Rectum
Lips	Brachial plexus	Main bronchus		Sacral nerve
Ears	Vocal cords	Kidneys		Femoral nerve
	Spinal cord (below C5)	Intestines		Urethra
		Thoracic spinal cord		Ureters
		Spinal nerves		Penis
		Pancreas		Perineum
				Spinal nerves L4 to S5

37.2 Medical Coverage Definitions by Body Region

This book chapter focuses on the medical coverage of personal armour by body region and not the level of protection required. Ideally, personnel would be protected from every threat from every angle, however, this is not practically feasible. External anthropometric landmarks on the skin surface have been identified that define the medical coverage boundaries of each of the body regions that correspond to the internal anatomical structures. The boundaries of personal armour will be eventually determined by the 'trade-offs' in the specific requirements that include medical coverage, level of protection (by severity of threat), mobility, integration, interoperability, comfort and acceptable thermal burden. To assist with determining such trade-offs, coverage has been further described in terms of threshold and objective measures (Table 37.4). For the torso, medical consensus has recommended that coverage should additionally be differentiated into the feasibility of haemorrhage control in the pre-hospital environment, in combination with external anthropometric landmarks [14]. These definitions can be used subsequently to derive and provide the measurements for sizing personal armour within a population [15].

Table 37.4 Definitions of threshold and objective areas of coverage

Threshold coverage	The minimum area that must be afforded coverage by personal armour. This corresponds to those structures comprising essential medical coverage as they are highly associated with mortality. Currently, this is mortality in the first hour post injury prior to damage control surgery
Objective coverage	The maximum area of coverage beyond which the advantages of additional medical coverage rapidly diminish. These anatomical structures covered are less likely to result in death but are additionally associated with long-term morbidity

37.2.1 Head and Face

Medical coverage of the head and face (including the eyes) was originally defined in two separate papers [7, 9]. However, more recently, and with the development of the concepts of threshold and objective coverage, the medical coverage of the head and face is considered as a single anatomical area (Table 37.5). This is a pragmatic approach, as for the foreseeable future combat helmets will provide the threshold coverage, with additional components provided to increase the medical coverage provided (see Sect. 37.4.2). The only exception to this is eye protection, which is worn independently of the remaining components. The head and face can be defined as the area above the lower border of the mandible, and in front of Zone 3 of the neck. A visor, for example, is generally considered to protect facial structures, but additionally covers the brain from lower angle trajectories [16] (Fig. 37.1). Currently, head and face coverage is related to three landmarks on the head surface, which broadly relate to margins of the brain. Research is underway to quantify the relationship between these landmarks and the brain, as was done for the torso [15].

Table 37.5 Area of coverage definition for head and face

Threshold coverage	Coverage should comprise the cerebral hemispheres, cerebellum and brainstem. The margins of the brain can be broadly related to the external landmarks on the head, including the nasion, external auditory meatus and superior nuchal line, as shown in Fig. 37.1
Objective coverage	All areas of the head and face

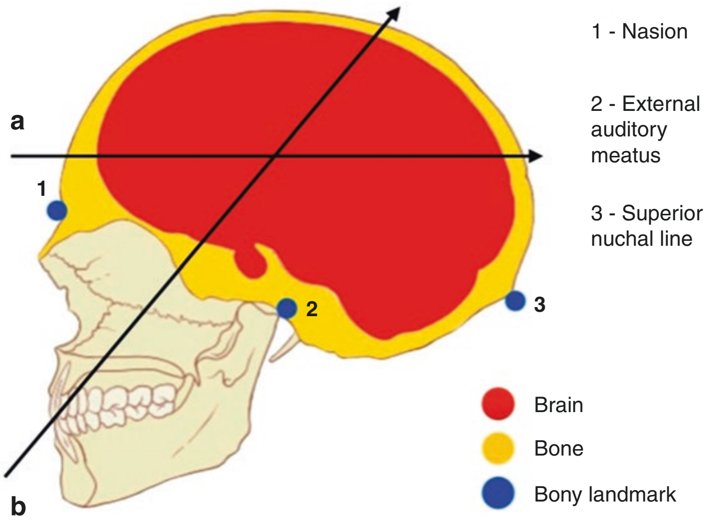


Fig. 37.1 Margins of the cerebral hemispheres and cerebellum related to external landmarks on the head. The brainstem is not shown for clarity. This coverage applies to horizontal trajectories (**a**). It is unrealistic to protect from lower angle trajectories (**b**). Reproduced from [7] with permission of BMJ

37.2.2 Neck

The neck is defined as the area below the base of the skull and above the clavicles and suprasternal notch. Essential and desirable coverage of the neck is not determined according to anatomical structures alone [17]. Instead coverage utilises surgical zones (Table 37.6) reflecting the difficulty of performing damage control surgery to the neck in an austere environment and acknowledging the potential human factors considerations of providing coverage in these zones (Table 37.7 and Fig. 37.2).

Table 37.6 Essential and desirable medical coverage of the neck utilise surgical zones

Zone	Description
Zone 1	Suprasternal notch to the cricoid cartilage
Zone 2	Cricoid cartilage to the lower border of the mandible
Zone 3	Behind the face at the level of the lower border of the mandible to base of the skull

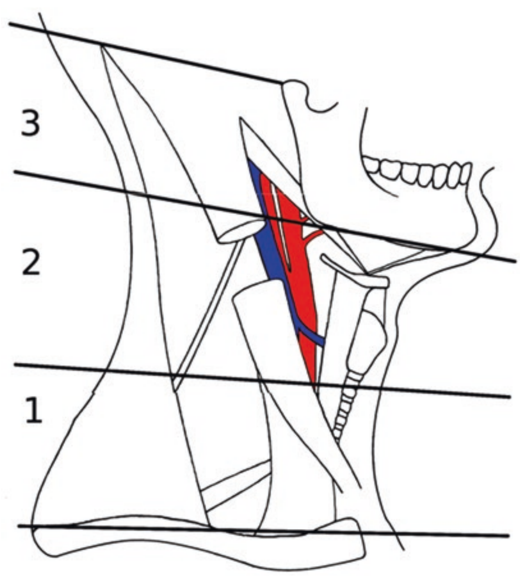


Fig. 37.2 Surgical zones broadly relate to the difficulty in performing damage control surgery to underlying structures in the neck in an austere environment

Table 37.7 Area of coverage definition for the neck related to surgical zones

Threshold coverage	Coverage must be afforded to neck zones 1 and 3
Objective coverage	Coverage of all three neck zones

37.2.3 Torso

The torso is considered a single anatomical area and comprises the thorax and abdomen [1]. It is defined as the area below the suprasternal notch and above the superior border of the iliac crest; at the junction with the arms, it is bordered laterally by the axillary fold. Coverage of the torso is measured using three external landmarks shown in Fig. 37.3. These landmarks are the suprasternal notch (1), lower border of tenth rib (2), and the iliac crest (3). Analysis of Computed Tomography (CT) scans of UK male service personnel have demonstrated that these external landmarks correspond within tolerable anatomical variations to internal landmarks of the boundaries of those structures comprising essential coverage; namely the aorta, heart, liver and spleen [15] (Tables 37.8 and 37.9). The dimensions of hard armour can subsequently be related to these external landmarks (Fig. 37.4)

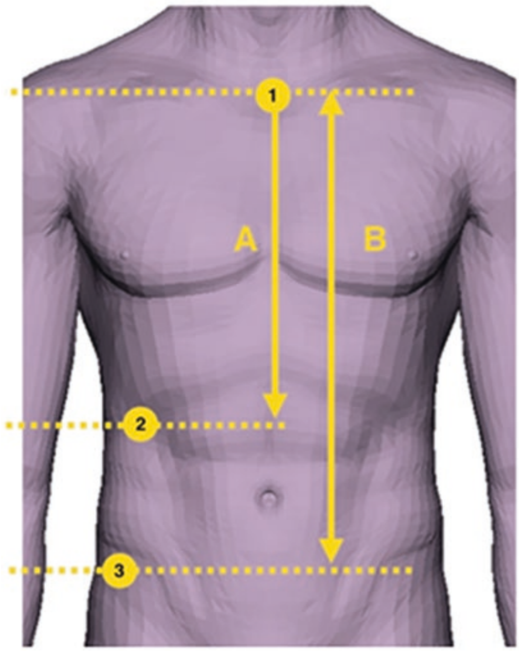


Fig. 37.3 Suprasternal notch (1), lower border of tenth rib (2) and iliac crest (3). Threshold height is (**a**), objective height is (**b**)

Table 37.8 Boundaries of those structures comprising essential and desirable medical coverage for the torso

Serial	External landmark	Internal corresponding anatomy
1	Suprasternal notch	Top of aorta (aortic arch)
2	Lower border of ribcage (tenth rib)	Lower border of liver
3	Iliac crest	Bottom of aorta (bifurcation of aorta)

Table 37.9 Area of coverage definition for Torso (Thorax and abdomen). DCS=Damage Control Surgery [14]

Term	Definition	Application to torso	Surface landmarks
Threshold coverage	Absolute minimum area that must be afforded coverage by personal Armour. Less than the threshold would not be considered to provide an improvement over current capabilities, and is not medically recommended	Anatomical structures if injured would result in life threatening haemorrhage and lead to death within 60 min without surgical intervention	Suprasternal notch to lower border of ribcage (tenth rib)
Objective coverage	The additional area to be covered as far as practicably possible to increase operational utility. Examples include increasing the time to DCS, but whilst also not impeding mobility, weight and excessive thermal burden	Anatomical structures if injured that would result in haemorrhage that could potentially be slowed down by treatment in the pre-hospital environment to enable subsequent surgery	Suprasternal notch to superior border of iliac crest

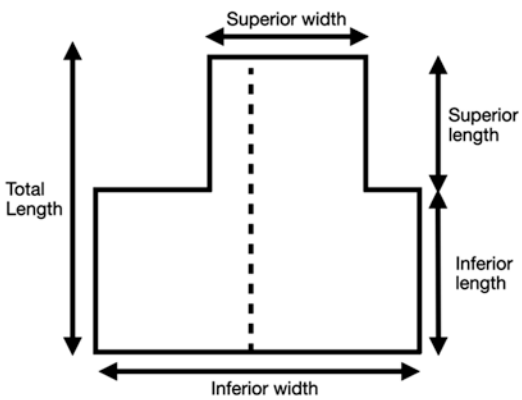


Fig. 37.4 Descriptions of how hard armour dimensions can be defined to meet threshold coverage; the dotted line reflects the midline of the individual. Extending the top width to produce symmetry and increase coverage, but yet enable mobility, is an example of objective coverage [14]

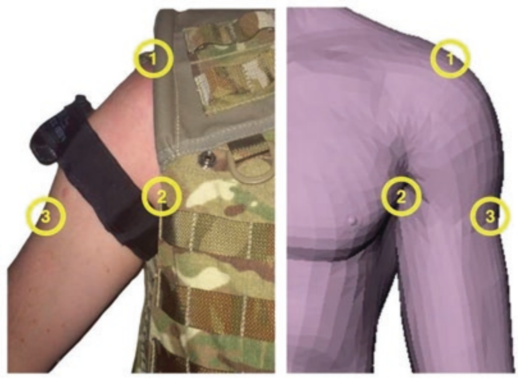


Fig. 37.5 Landmarks of the arm coverage, with example of application of tourniquet (left)

Table 37.10 Area of coverage definition for the upper arm. Derived from [13]

Threshold coverage	Coverage must be afforded from the acromion to 40 mm below the deltoid insertion ensuring a tourniquet remains in place and is effective
Objective coverage	Same as threshold area of coverage

37.2.4 Upper Arm

Coverage of the upper arm includes the axilla [8, 13]. The junction between arm and torso coverage is the anterior axillary fold, as a tourniquet cannot be advanced further up the arm beyond it (Table 37.10). The boundaries of arm coverage are determined using the three landmarks depicted in Fig. 37.5. These are the acromion (1), the anterior axillary fold (2) and the deltoid insertion (3).

37.2.5 Pelvis and Thigh

The pelvis and thigh are currently considered in terms of protection as a single area. This area is defined as the area below the iliac crest, extending towards the knee (Table 37.11). Coverage of this region is determined from the two landmarks [12], as shown in Fig. 37.6. These are the iliac crest (1) and the ischial tuberosity (2).

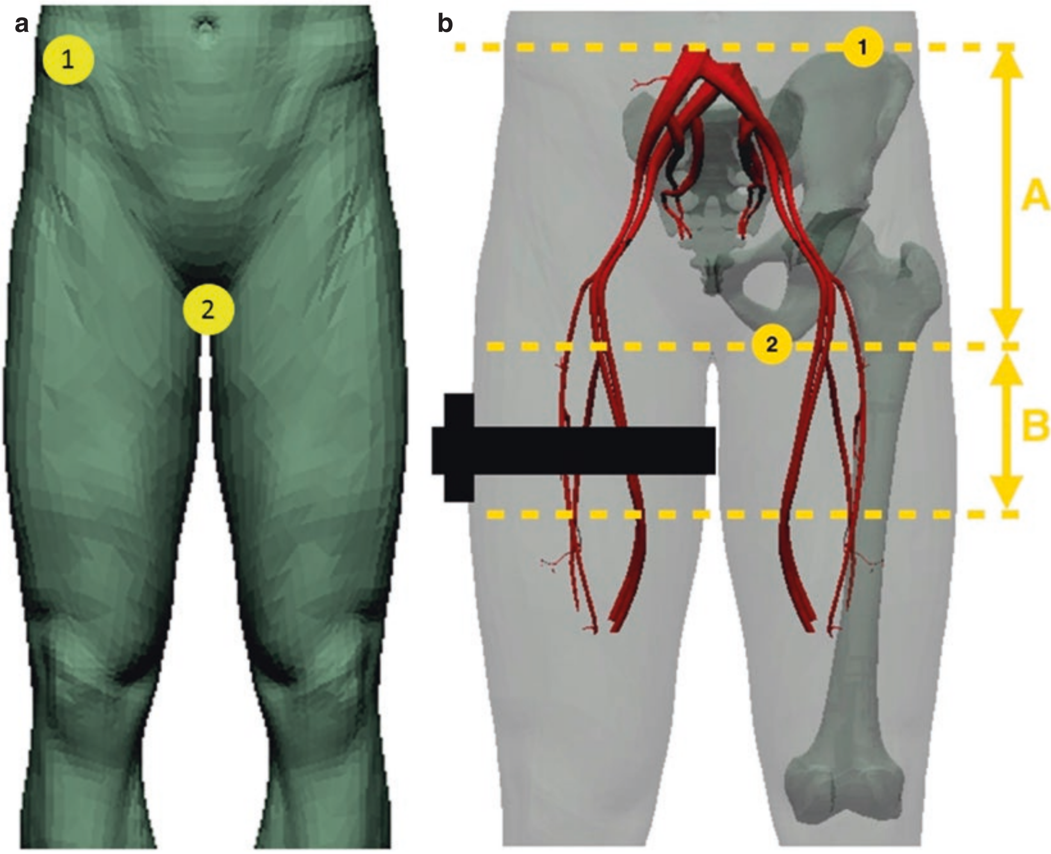


Fig. 37.6 Landmarks of the pelvis and thigh (**a**). Example of application of tourniquet (**b**)

Table 37.11 Area of coverage definition for the pelvis and thigh. Derived from [12]

Threshold coverage	Coverage must be afforded from the iliac crest to 100 mm below the ischial tuberosity, ensuring a tourniquet (or two adjacent tourniquets) remains in place and is effective.
Objective coverage	Same as threshold area of coverage.

37.3 Computerised Comparisons in the Anatomical Coverage Provided by Different Armour Designs

To inform decisions on the 'best' armour solution to procure or use, it is imperative to be able to objectively compare the coverage of multiple armour solutions [18]. However, it must be noted that the optimum solution is not just the one that offers the most medical coverage. It is vital to assess the human factors implications alongside coverage. There are two tools used by UK (Ministry of Defence) MOD to compare objectively medical coverage of personal armour [19] (Table 37.12).

Table 37.12 Summary of the two tools currently used by UK MOD to compare objectively armour coverage

Tool	Description
Coverage of Armour Tool (COAT)	COAT is a simple shot line tool. Geometrical elements that model the Armour and body are represented. Vulnerable structures from Tables 37.2 and 37.3 are selected; the tool then calculates the percentage of coverage from azimuths and elevations that are selected by the analyst. An example grid is shown in a screen shot of the tool in Fig. 37.7. The percentage coverage of different Armour solutions for areas of the body can then be objectively compared. This method assumes that all vulnerable structures are equally important and that uncovered shot lines pass through the entire body.
Weapon Target Interaction (WTI)	WTI is a terminal effects vulnerability model. It simulates the penetration through a human geometry, determines the volume of damaged tissue and outputs injury scores that are commensurate with the Abbreviated Injury Scale. The injury scores for grids of shot lines, from user defined azimuths and elevations can be calculated. The scores can then be weighted for different levels of injury to determine an objective coverage score. Examples of a rigid Armour plate providing threshold level of medical coverage from 0° and 20° azimuths both at 0° elevation are shown in Fig. 37.8.

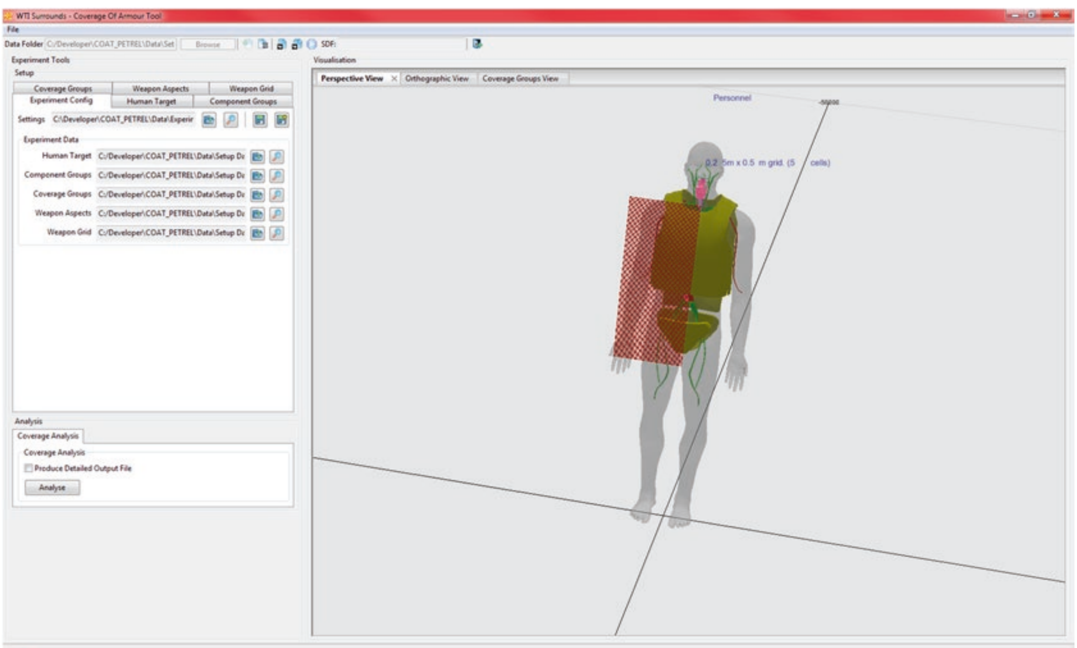


Fig. 37.7 Screen shot of Coverage of Armour Tool being used to assess the medical coverage provided by an armour system

Fig. 37.8 Example of output from Weapon Target Interaction model. The blue area represents shot lines being stopped by the hard plates; the other colours correspond to Abbreviated Injury Scale 1–6 as shown on the right of the diagram

37.4 Coverage of Personal Armour Issued to UK Armed Forces Personnel

The development of personal armour for UK Armed Forces personnel has been already reported in multiple formats [3, 4, 19–25]. This section provides a few key examples of how the tools and techniques described in earlier sections have been used by the UK MOD.

37.4.1 Coverage Provided by Hard Plates as Shown in COAT

Figure 37.9 shows the area of coverage provided by a front Enhanced Combat Body Armour (ECBA) plate and a front OSPREY plate superimposed over those structures defined as requiring essential coverage (Table 37.2). Both these hard plates are currently in-service and used by UK Armed Forces personnel. Figure 37.9 shows that the ECBA provides complete coverage of the heart and that the OSPREY plate provides complete coverage of the heart, liver and spleen and partial coverage of the great vessels. Complete coverage of the inferior part of the aorta within the abdomen is not possible due to human factors constraints, and the COAT allows this to be visualised.

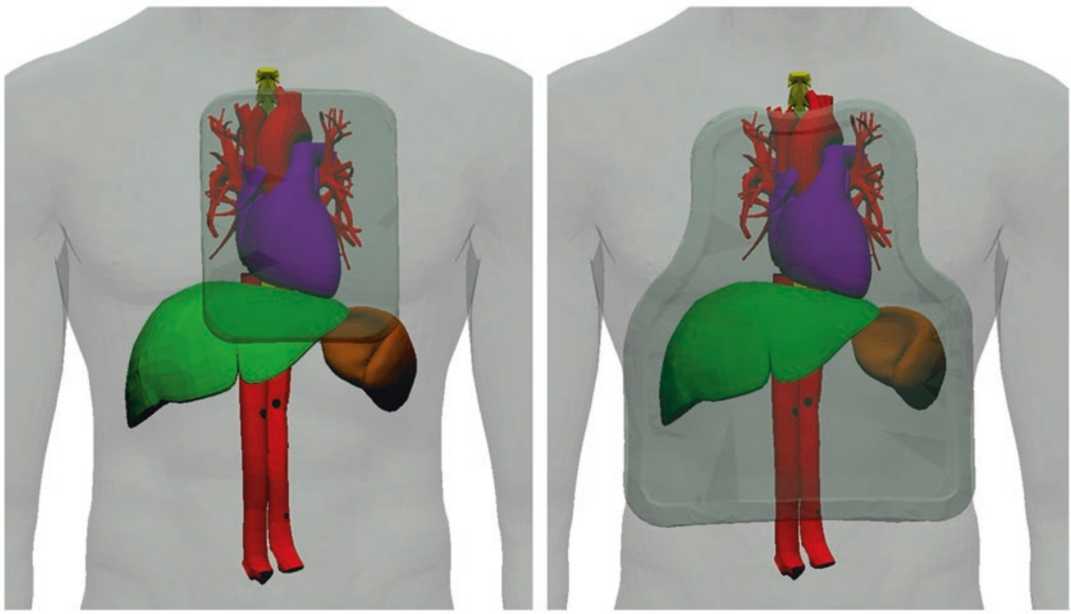


Fig. 37.9 The ECBA plate (left) and the front OSPREY plate (right) superimposed over those structures requiring essential coverage: heart (purple), aorta and vena cava (red), liver (green) and spleen (brown)

37.4.2 Medical Coverage Provided by the VIRTUS Helmet

Another example of using COAT is in visualising the area of medical coverage provided by the current VIRTUS helmet (as worn by UK Armed Forces personnel) (Fig. 37.10). The brain comprises the cerebral hemispheres, cerebellum and brainstem. The VIRTUS helmet provides complete coverage of the cerebral hemispheres (cerebrum) and cerebellum, and partial coverage of the brainstem [26]. Although some of the brainstem is not covered by the helmet itself, the additional coverage can be provided by nape protection. Both a visor and nape protection are provided as part of the VIRTUS helmet sub system [9, 16, 20] to meet these coverage requirements.

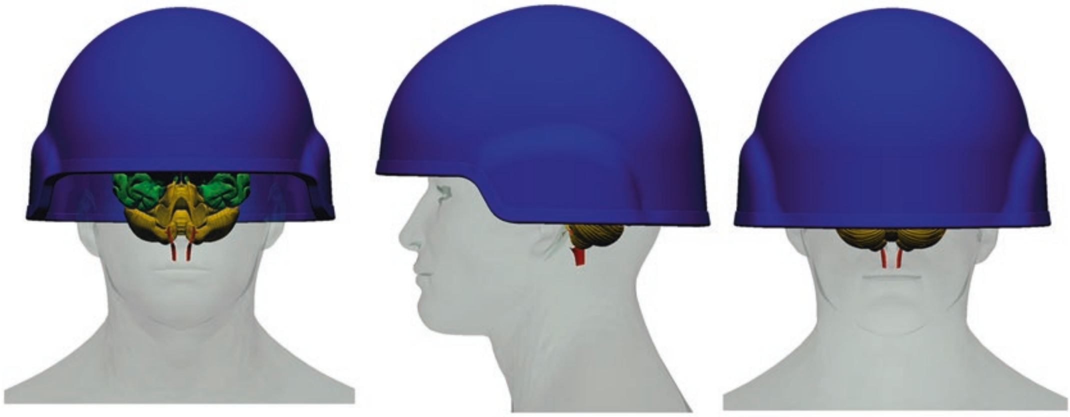


Fig. 37.10 The VIRTUS helmet superimposed over those structures requiring essential coverage: cerebellum (green) and brainstem (brown). The cerebral hemispheres are not visible. Adapted from reference [26]

37.4.3 Quantifying the Coverage Provided by Side Plates

ECBA plates can be used in the VIRTUS body armour and load carriage system as front and rear plates or inserted into pockets on the sides of the VIRTUS soft armour to be used as side plates (when the OSPREY plates are used as front and rear plates [25]). Using COAT, it was possible to quantify that ECBA plates provide an additional 35% coverage of the anatomical structures defined as essential coverage (Table 37.1) when compared to using front and rear OSPREY plates alone (Fig. 37.11). It was also possible to visualise the coverage provided as shown in Fig. 37.12.

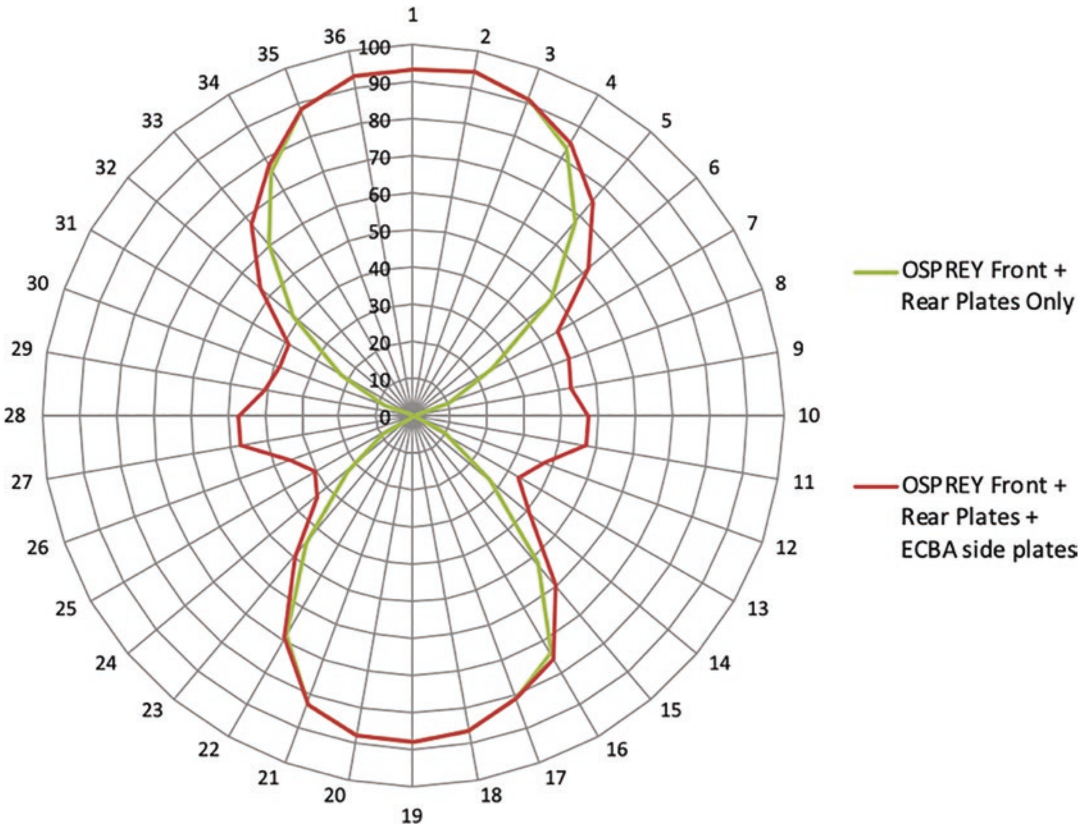
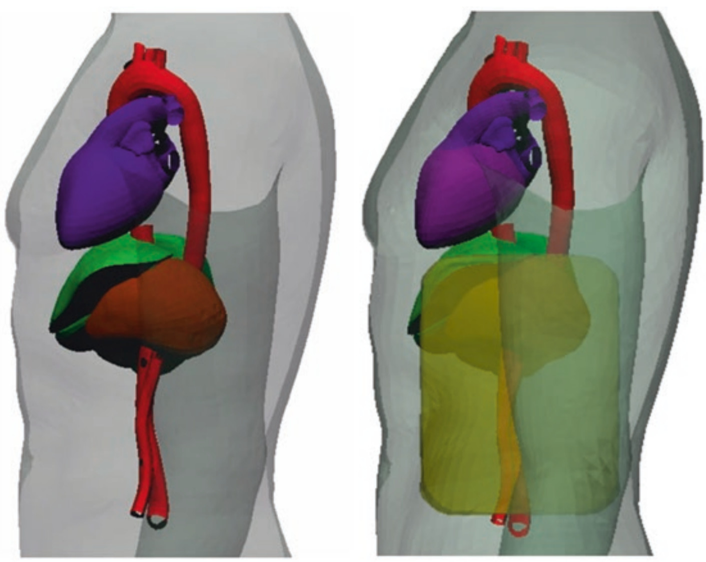


Fig. 37.11 COAT Azimuth plot demonstrating the coverage of anatomical structures comprising essential medical coverage using front and rear plates alone (green line) with addition of current ECBA side plates (red line)

Fig. 37.12 Anatomical structures comprising essential medical coverage from the side (left) and visualisation of the coverage provided by ECBA plates when worn as side plates



37.4.4 Optimising the Coverage of Arm Protection

The VIRTUS body armour and load carriage system used by UK Armed Forces personnel is continually evolving in response to new medical evidence and user feedback [20]. For example, it was possible to reduce the length of the brassards (used as arm protection) by over 30% using the coverage definitions shown in Table 37.11 and data from medical analysis of CT scans [13] to meet the objective coverage requirements (Fig. 37.13). The length of the new brassard was based upon the 99th percentile male and was introduced in 2017. Optimising the brassard resulted in the reduction in overall weight of the VIRTUS system, improved integration and interoperability, and reduced thermal burden.

Fig. 37.13 A pictorial representation of how upper arm coverage was determined to optimise the VIRTUS brassard from the sum of distances (**a–d**). (*1*) Acromion process, (*2*) anterior axillary fold, (*3*) deltoid insertion, (*4*) 40 mm below deltoid insertion

37.5 Conclusions

The area of medical coverage and subsequent sizing of UK personal armour systems for UK Armed Forces personnel is currently based on the interrogation of anonymised Computed CT scans of injured male military personnel undertaken at University Hospitals Birmingham [13, 15] in conjunction with measurements derived from a variety of anthropometric data sources [27, 28]. Tools and techniques used to calculate and visualise the area of coverage provided by personal armour have been described as well as examples of how some of these tools are used.

The medical area of coverage and anthropometric landmarks for the torso have been nationally and internationally accepted and have been adopted in the UK Home Office Body Armour Standard Guidance (for UK Police Services) [29] and in the North Atlantic Treaty Organization (NATO) Standard on Design Criteria for Body Armour Carriage Systems [30]. Helmet coverage definitions have been adopted in the NATO Standard on Non-Ballistic Test Methods and Evaluation Criteria for Combat Helmets [31]. Furthermore, a medical area of coverage UK Defence Standard is in preparation that will form the basis of all future body armour procurement for UK Armed Forces personnel.

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Part V

Rehabilitation

Check for updates

38 Section Overview

Alison H. McGregor

This section builds on the journey of blast injury and the success of work to prevent and mitigate injury and the success of our clinical management of blast injuries described in Parts III and IV. These approaches generated unexpected survivors who presented with complex injuries. For these people to live life to the full their long-term health needs consideration.

The section explores the personal experiences of surviving and managing severe injury from the perspectives of military amputees and the clinicians working with them to achieve their high function. The focus is on those with significant limb trauma and amputation exploring their orthotic and prosthetic considerations, rehabilitation pathways, aspects of musculoskeletal and bone health and the management of pain. It outlines ongoing issues that need to be reconciled and considers how science can help us determine the answers and to predict future problems.

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39 Survive to Thrive

Dave Henson

Abstract

Recovering from the realities of traumatic blast injury is a long, complicated, and often painful process. However, with appropriate planning, this process can be optimised. Upon injury, the creation of an outline recovery plan including targets and goals should be established as soon as possible, drawing on the experience of peers for guidance. It should be understood that rehabilitation takes time, but with the appropriate input from the team and peers around you, this can be minimised as long as there is the required level of adherence to the exercises. Prioritise medium- to long-term ambition, as this provides scope to adjust to the new physical state whilst still achieving targets. This space creates the opportunity for the identification of longer term development targets as the appreciation of self-identity develops.

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39.1 An Anecdote of Adjustment

I was 4 months into my tour of Afghanistan when I lost both of my legs, 13 February 2011. Part of my personal preparations for deployment to combat was to come to terms with the psychological concept of death or injury. Strangely enough, and probably selfishly, the idea of death was easier to deal with because in the event of that situation, it really had very little to do with me. The complexities surrounding the realities of blast injury were more difficult to fully comprehend, but the preparations put me in a much better position to deal with the concept of the short-term aftermath of injury.

Adjusting to immense and immediate life changes is not, and never will be, easy. Consultants make substantial sums of money advising corporations on how to understand and manage organisational change, and the difficulties in doing so are well recognised. Adjusting to any life-changing injury is the same, but on a much more painful and personal level, and often comes without ‘consultant’ input. I was fortunate in my experiences, and the ‘consultants’ I had were my utterly inspirational peers and colleagues, and the therapists who had had to deal with similar injuries and situations so many times before and, unfortunately, after me. In an effort to try and translate my experiences to a wider audience, this short anecdote forms my recipe for success and details the lessons learned and the

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details that have made adjusting and coping to this new situation a little easier.

Prior Planning and Preparation Prevent Poor Performance

Identify the realities of after-injury; understand the basics of rehabilitation and recovery. A properly prepared psychological foundation maximises recovery potential.

In 2011 in Afghanistan, the IED (see Fig. 39.1 for an example of a pressure-plate trigger) was the main threat and, seemingly, the Taliban's weapon of choice. Looking at the Defence Statistics for the British Military's combat operations in Afghanistan, over half of all our operational deaths, and over a quarter of all in theatre injuries, were caused by explosions. It seemed like every day on the news someone was getting killed by an explosion, or news was leaking back to us about some friend or colleague losing limbs.

Fig. 39.1 Pressure-plate switch of an IED wrapped in plastic

As part of our pre-deployment training (PDT), we conducted extensive first-aid training in case of injury, so the logical follow-on thought process leads to considerations of life without limbs. Certainly, you could at least gain an appreciation for what the rehabilitation technology and pathway were like and what the Defence Medical Rehabilitation Centre (DMRC) treatment consisted of. For me, prior processing of the potential realities meant that dealing with the reality was slightly easier.

Our job on 13 February 2011 was the clearance of two small compounds that had been used as firing points by both enemy and friendly forces. As a result, it was highly likely that these compounds were seeded with IEDs in case of future use by International Security Assistance Force (ISAF) troops. In Afghanistan, I was in charge of a Royal Engineer Search Team (REST). It was our job to plan and conduct operations to find IEDs. For this task, our job was to ensure that these compounds (example in Fig. 39.2) were safe in order that local farmers could begin working the land to kick-start the local micro-economy.

Fig. 39.2 One of the compounds that required clearance, showing the internal courtyard

The first compound clearance passed without incident. In the second compound, the external courtyard was cleared, and I crossed over it to gain sight of the infantry unit that we were working with. As I returned, I stood on a pressure-plate IED that we had failed to find during our entry phase. I was blown into the air and landed back down on my head. I forced myself into a seated position and looked down at the wreckage of my legs. There was no immediate pain, just shock. My boots were still intact, but the soft tissue from the top of my boots up to my thighs was ruined. It was an appalling sight. Thanks to the training they had received, and their presence of mind, my team were incredibly quick to act. Whilst they applied immediate first aid, the CASEVAC (casualty evacuation) process was initiated. The adrenaline began to subside, and the pain began to set in. It was a strange pain, incredibly intense and like being crushed by a huge weight. I had shrapnel injuries in my hands, arms, buttocks, and chin. The skin had been stripped from the backs of both thighs. My tibia and fibula had been shattered. However, even with the extent of my injuries, I was stabilised quickly. After 20 min, the helicopter arrived, and after another 15 min, I was on the operating table in the hospital at Camp Bastion. The injuries I sustained are shown in Fig. 39.3.

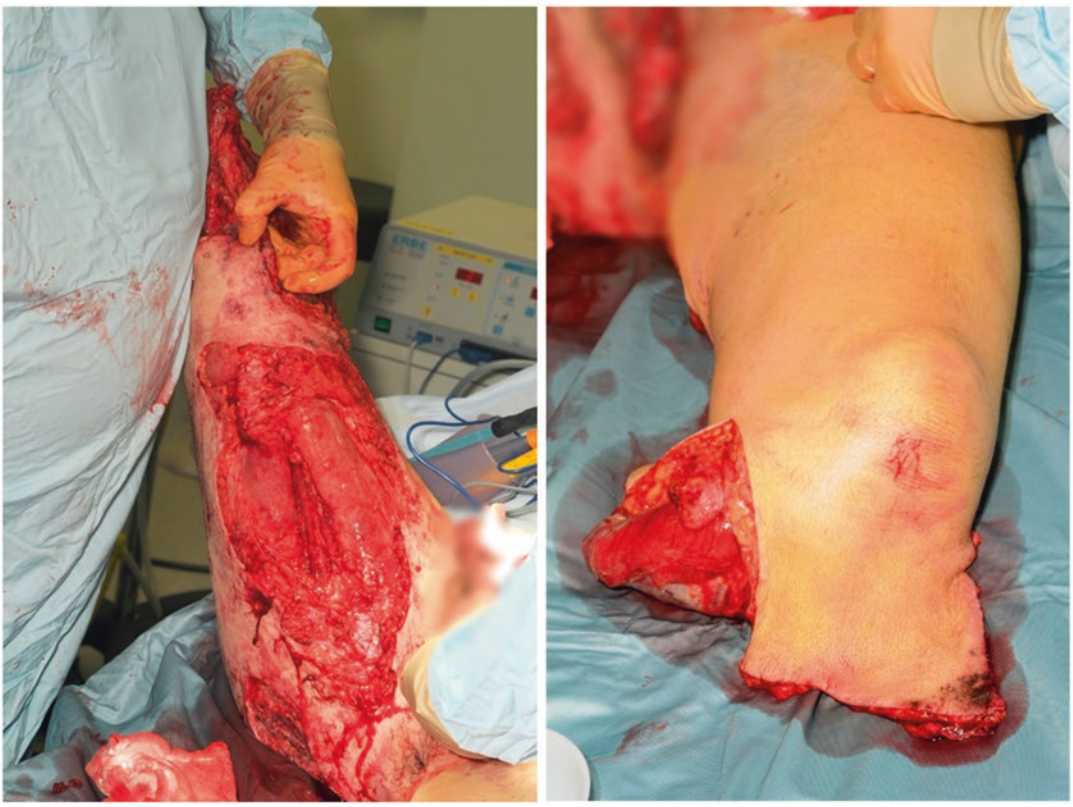

Fig. 39.3 Wounds on return to the UK

In this immediate aftermath of injury, I was lucky enough to experience the results of some hard-earned lessons. The first was the 'platinum ten minutes'—a constituent part of the 'golden hour'. In my case, this involved stemming catastrophic bleeding, but also included considerations for airway blockage, breathing difficulty, and other circulatory concerns. The second was the composition of the crew on the medical emergency response team (MERT). Included on this team was an anaesthetist who prepared me for surgery on route back to the hospital, ensuring that as little time was wasted before I was ready for the surgical team to operate on. The third was the layout of the hospital, an emergency department with eight resuscitation bays situated immediately adjacent to four operating theatres, with adjoining medical imaging facilities to improve patient flow. Fourthly, I had benefitted hugely from the lessons in infection control and improvements in debridement techniques and my residual limbs were left open, with vacuum dressings to reduce infection risk.

The swift stabilisation of my condition on the ground and subsequent use of techniques to minimise injury impact and infection risk, coupled with pre-deployment psychological preparation, placed me in a very favourable position for a successful recovery, which began shortly after exiting the Intensive Care Unit at the Queen Elizabeth Hospital in Birmingham, UK. I had been flown back to the UK and kept in the ICU for a week. Once sufficiently stable, I was moved onto the ward and into a four-man room. As soon as possible, we were encouraged to begin mobilisation exercises. These started with simple movements, such as transitioning from lying to a seated position (adjusting for altered body centre of mass), transferral into a wheelchair, self-care including washing and ablutions, and progressing to self-mobilisation in a wheelchair. The achievement of independence in these activities of daily living led to morale improvements and increasing positivity. It becomes possible to go to the shop or meet visitors in a location away from your hospital bed.

For me, and many others in my position, I was fortunate enough to receive a peer mentor at this stage. This was a colleague who had already been through injury and the subsequent recovery journey. These visits allowed me to identify and empathise with the reality of the situation in a much more tangible manner. Having someone with experience explain everything, from the variations in prostheses that were available and rough timelines for how soon they could be achieved, through to explanations about likely career routes and sporting opportunities, was incredibly helpful. From this, it becomes possible to create the outline recovery plan. As the patient, this is the opportunity to identify and discuss with your support network what your anticipated and ambition-based timeline for recovery looks like and to identify early-stage challenges and goals.

Rehabilitation Takes Time

- It is a team game.
- Do the rehabilitation exercises.
- It can still be painful.
- It will take time.
- Practice makes perfect.

Upon discharge from the hospital, the real rehabilitation work begins. For me, I was fortunate enough to become a patient of the DMRC, but others will have a different route. Figure 39.4 shows two key milestones in rehabilitation. As the journey from this point is dependent on an individual's health service provider, below are some key learning and development points:

A. It is a team game. You, your consultant, your prosthetist, your therapist, your social worker, your support network—you are all in this together. As the patient, you cannot afford to sit back and be dictated to about how your rehabilitation will progress. This would provide you with no ownership of your rehabilitation targets and objectives, and therefore makes the achievement of them much more difficult. You, as the patient, need to recognise your majority stake in what your rehabilitation looks like—this is your life and your body, and no one is going to rehabilitate it for you.

B. Move, stretch, and strengthen as much as possible. Become as fit as you possibly can be. Do the exercises you have been prescribed. Walking with prosthetics is incredibly hard, and to be as successful as possible, you need to be as strong and as fit as you possibly can be. A positive rehabilitation outcome is earned; it cannot be gifted, and it is hard work.

C. It still hurts. In these early stages of rehabilitation, your limbs are still healing. That invariably means there is pain somewhere. You have to accept a certain level of pain; otherwise, you cannot progress as rapidly as you will want to.

D. It will take time. Your body is still adjusting to its new postural scheme. It has no idea how to provide balance and stability in static tasks, let alone dynamic, so be fair to yourself and understand that your body needs to adapt to how it is now configured.

E. Practice makes perfect. If at first you don't succeed, try, try, again! I am sure that there are other clichés that can be applied here, but the message is simple—walking with prosthetics is hard, but it is worth it. That being said, there are movement patterns to develop that you have never had to perform before, and there are muscles being recruited for tasks they have never had to solve before. You will fall often, and you will fail often. But the rewards are plentiful and worth it. Stick to the plan, do the exercises, wear the legs, and it will happen.

Fig. 39.4 Progressing from training limbs to full prostheses

Get a Life

Prioritise progression and short- to medium-term ambition. Allow yourself space to identify new motivation and how your new physical state matches your own perception of identity.

If available resource allows (i.e. from a financial perspective), focus should transfer to short- to medium-term goals, such as those involving sporting ambition, arduous endeavours, fund-raising challenges, training, or interest-based courses. Personally, the option was sport, and this was an easy route to take. Sport is something that is inherently understood by service personnel as it forms a core component of life in the military. This is no different when dealing with recovery from injury. The DMRC uses sport in its everyday rehabilitation programmes as standard. For me, the routine of rehabilitation was broken up by sport. It started with swimming and quickly progressed to running, which was one of the activities I craved over the weeks in hospital and the months of the slow, boring, but necessary exercises designed to get you back to walking. I was fitted with running prostheses (running blades) 10 months after being injured. My first run was across the length of the prosthetics fitting room—less than 10 m—and my next few were not much longer. The feeling of going fast again was almost addictive and soon became my biggest motivator. I began to relish the prospect of the next run and maintained a training log to make sure I could track how I was progressing. One thing which I noticed early on was that people automatically assumed that the running prosthetics provide you with a huge performance advantage and that you are now likely to be heading to the Paralympic Games. This is obviously a huge marker for the positive progression of disability sports but does become a little frustrating for the beginner amputee runner and makes you feel as if you should be able to run instantaneously! As a bilateral amputee with no knee joints, the mechanics of running were so different to how I had been running for the first 26 years of my life that learning how to run with any kind of consistency took a long time, with multiple prosthetic issues taking me back a few steps before being able to move forward again. However, progress can always be measured, and in these early days of running, I measured mine in metres, not miles. I would run to the end of my road and back, a distance of 1500 m, but I started by doing it in 50 m sections. I would run 50 m and walk 50 m and so on until I made it back home. I was also fortunate enough to test out the running track at the Olympic Stadium in London in April 2012 (Fig. 39.5). It took a long time, it was painful, and I fell over a lot, but it still felt worth it. As I got better, 50 m became 100 m, 200 m, and so on. Over the course of a year, I managed to progress to a confidence level such that I was able to leave my house and go for a run, unsupported, and not be concerned about my own abilities to make it back home. This was a hugely positive step for my independence in physical exercise, but it took almost a year!

Fig. 39.5 Exhibition race in the London Olympic Stadium as part of the London 2012 readiness testing, just 4 months after being fitted with running prostheses

The time and effort spent on relearning this basic skill was utterly worth it, and I progressed into sports competitions at an amateur level. The competition element provided a huge buzz, but at this time in my recovery, the focus was tilting towards understanding the position of my career and professional development. I had achieved what I had set out to with running, in that I was independently able to exercise myself in this way, whenever and wherever I wanted. But my medical discharge from the military was imminent, uncomfortable, and nerve-racking. One thing that was clear for me, however, was that I felt it important for my own sense of well-being and identity that I left the military in as good a physical state of health as possible. Obviously, a good state of physical health is a good thing at whatever stage or walk of life you are at, but for me, this would be measured by passing the running component of the British Military's basic fitness test. This comprised being able to run 1.5 miles in under 10 min 30 s. So, whilst I was concentrating on my professional transition from military to civilian life, I was also training hard to pass this test. I succeeded (just) a month before my discharge and ran the distance in 10 min 28 s. Unsurprisingly, this was a huge morale boost for me and ensured a confident exit from the military. The process of preparing for this test meant that my general running speed had also improved. At the time, the first Invictus Games was on the horizon, and I decided to enter the 200 m sprint competition. In terms of tracking progress, I won this competition, and the time I achieved in the process was such that, had I competed in the London 2012 Paralympic Games, I would have made the final of the 200 m. This acted as a huge catalyst for me and underlined the power of SMART (specific, measurable, achievable, realistic, time bound) targeting techniques when underpinned by a genuine self-accountability. As the Rio 2016 Paralympic Games were just under 2 years away at this point, I decided to place a priority on my sport and see whether

I could achieve success at these Paralympics. 2015 saw selection to the GB team for the Doha 2015 IPC Athletics World Championships. 2016 saw my second Invictus Games, selection to the GB team for the Grossetto 2016 IPC Athletics European Championships, and selection to the GB team for the Rio 2016 Paralympic Games (Fig. 39.6). Success, by any measure, but made sweeter through the achievement of a gold medal at the Invictus Games, a silver at the European Championships, and a bronze at the Paralympics. I would go on to complete the suite by gaining an additional bronze medal at the London 2017 World Para Athletics Championships.

Fig. 39.6 Final UK training session ahead of the Rio 2016 Paralympic Games. Picture copyright Roger Keller

Focussing on the Short Term Allows Space for the Development in the Long Term
The understanding of how your new physical state aligns to your self-identity is incredibly different and can only properly be achieved in a state of confidence in personal abilities.

Whilst it is pleasant to recount these sporting successes, the reasons why these achievements are so important are for what they provide to the psyche. Prior to attending the Rio Paralympics, I was often referred to as 'Dave Henson, who was injured in Afghanistan', or similar descriptions that focussed on my injuries or my past career, as opposed to the present or the future. This inevitably resulted in an outlook that was unfairly focussed on the past. On returning from Brazil, however, there was a distinct change in my descriptor, and I became 'Dave Henson, GB Paralympian', or similar. Almost overnight, I transitioned from someone who was restrained by an incident of the past into someone with both a present and a future. This provided me with the space, confidence, and opportunity to begin properly focussing on the positive aspects of where I saw my life going, what opportunities were still available, and where else I could achieve.

To summarise, I afforded myself the best chance of early acceptance in the event of a life-changing injury in a conflict zone by genuinely dealing with the prospect and treatment of it prior to operational deployment. In hospital, early movement is key. As soon as it is safe to do so, start doing it. At this stage, sitting up and moving about—anything that gains independence—should be celebrated as positives. As soon as you are able, do more and move more. The hard work of returning to walking seems endless in the early days. It is an incredibly physically and mentally demanding task that requires complete focus and dedication. It is highly likely that there is still physical pain in these early stages, but the benefits of pushing through this pain barrier and making rehabilitation gains whilst the central nervous system is still in this highly plastic state cannot be overstated! Once you have gained independence in living, independence in life is key. Allow yourself time to focus on short- to medium-term goals and targets that you find highly rewarding and that fit within the SMART framework for goal setting. Success within this framework enables the development of confidence and the ability to begin identifying how your new physical state will interact in the long term with your identity as a person. Once this new self-understanding is achieved, true transition away from injury becomes possible.

39.2 Short Biography

I joined the British Army in 2008, commissioning into the Corps of Royal Engineers and deployed to Afghanistan in 2010 as a Royal Engineer Search Advisor. In February 2011, I lost both of my legs to an improvised explosive device (IED) blast. I used sport as a catalyst for my recovery and have competed at the Warrior Games, Invictus Games, IPC World Championships, and the Paralympic Games, winning medals in all competitions. I completed my PhD in Amputee Biomechanics at Imperial College, London, in 2020. I am a trustee of the Invictus Games Foundation, the Explora Scholarship Fund, and the Armed Services Trauma Rehabilitation Outcome (ADVANCE) Study Charity. I am the Veterans' Advisor to the Centre for Blast Injury Studies and the co-chair and founding member of the CASEVAC Club. I am married to Hayley and have three daughters. I was awarded the MBE in 2014 for services to the military, and I am an Honorary Fellow of the University of Hertfordshire.

40 Rehabilitation Lessons from a Decade of Conflict

Peter A. Le Feuvre, Kate E. Sherman, and Andrew P. Wareham

Abstract

Blast causes severe and complex injury patterns and significant rehabilitation challenges. By 2011/12, the peak of the Afghanistan conflict, complex trauma admissions into the Defence Medical Rehabilitation Centre Headley Court were equivalent to the total admissions into specialist inpatient NHS rehabilitation for the whole of England. These high casualty numbers enabled the rehabilitation specialists to evolve practice and challenge expectations. The service was built upon existing principles, namely early assessment, exercise-based rehabilitation, cross-disciplinary working, active case management, and rapid access to specialist opinions and investigations. Rehabilitation commenced at the earliest possible point in the intensive care unit in the deployed setting. This then progressed through to the inpatient trauma ward to the delivery of outpatient rehabilitation even while the patients were still in hospital. Finally, the integration of medical rehabilitation and transitional support agencies is critical in the support of the casualty in the final stages of their recovery.

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40.1 The Rehabilitation Challenge

Blast injury typically encompasses more than one body region. Proximity to the explosion, and differing environmental conditions will vastly alter the presentation and severity between individuals, even for those involved in the same incident [1]. This variable injury presentation prevents a formulaic response. In this chapter, we will explore the principles of rehabilitation that the UK military employed as we sought to promote and support the recovery of the many combat casualties who fell within this complex case scenario.

The complexity of injury sustained by amputee survivors is summarised in Table 40.1, highlighting the very high burden of injury and the long rehabilitation period [2]. Patient volume also increased rapidly in the early stages of the conflict in Afghanistan. In 2006 the Defence Medical Rehabilitation Centre (DMRC) Headley Court, had four ward-based beds for Complex Trauma, 6 months later, one consultant-led team

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managed 20 beds. By July 2009 there were 30 complex trauma beds, which increased to 60 beds, 6 months later [1].

By 2011/12, the peak of the Afghanistan conflict, complex trauma admissions into DMRC were equivalent to the total admissions into specialist inpatient NHS rehabilitation for the whole of England. However, unlike most NHS admissions, multiple limb loss and orthoplastic, spinal and neurological injury were combined within one patient, demanding a breadth of expertise and facilities within the rehabilitation service. Currently, such a facility does not exist within the NHS. Now that improvement in survival in the NHS following major trauma has been seen, as military lessons have influenced the rationalisation and coordination of NHS trauma services [3, 4], investment in complex trauma rehabilitation services is needed if we are to optimise recovery. The recent need for charitable support to enable victims of the Manchester Arena Bombing to receive the rehabilitation they needed further underlines that capacity building of rehabilitation capability should be a critical NHS priority [3, 5, 6]. Likewise, it is also vital for the Defence Medical Services to maintain a foundation of military rehabilitation during peace time so it can grow to support military operations as and when it is needed [1].

Table 40.2 contrasts the injuries suffered by two soldiers, one in a mounted patrol and one dismounted. Both presentations illustrate the complex predicament faced by the rehabilitation specialist when seeking to support the individual and move their recovery forwards. Traumatic amputation has been referred to by some as the hallmark injury from recent operational commitments [7]. The focus upon amputation can be demoralising for those who did not suffer in this way, but nevertheless experienced equally disabling and life changing injuries. However, this chapter will focus on how defence rehabilitation evolved their rehabilitation practice to meet the challenge presented by the amputee, primarily because they represent a cohort whose experiences can inform future military and civilian care. Furthermore, their injuries have implications that will continue through life. As such, the lessons learned during this operational surge must also inform ongoing and onward veteran provision.

Physically fit military personnel had chosen to pursue an active and adventurous profession. This drive was rarely lost, whatever the injury and subsequent disability. Their expectation, fueled by the example of others, was to continue such a lifestyle post-injury (see example in Chap. 39). Their injury presentation, coupled with their own drive and expectation, biomechanical and prosthetic skill acquisition, as well as co-morbidity and psychosocial factors, presented a unique rehabilitation challenge. It made the rehabilitation team rethink what could be achieved.

The human challenge faced by Defence Rehabilitation has many similarities to complex challenges faced in very different scenarios and industries. To quote Dael Wolfe [8]:

> *It may turn out that [the space program's] most valuable spin-off of all will be human rather than technological: better knowledge of how to plan, coordinate, and monitor the multitudinous and varied activities of the organizations required to accomplish great social undertakings*—[8] (p. 753).

This post-hoc analysis of the Apollo mission chimes perfectly with the lessons presented in this chapter. The complexity of the human challenge faced by rehabilitation providers across NATO forced a rethink of organisational processes and structures, clinical approach, and collaboration. International knowledge sharing and collaboration, cross-organisational working, and multi-sector support were matched by individual ingenuity and sheer dedication from patient and clinician alike. The lessons outlined here detail how UK Defence Medical rehabilitation evolved its practice. This practice arose from a deeply embedded philosophy of care that provided the foundation of what was to come.

Table 40.1 Amputee new injury severity score (NISS)/Number of admission/Length of stay in specialist rehabilitation (data extracted from Ladlow et al. 2015. (p2050–2051) [2]

		Amputee groups			
Injury characteristics	Unilateral	Unilateral non-op	Bilateral	Triple	Total Amputees
Injury severity					
NISS (± SD)	28 ± 11.3	N/A	44 ± 11.7	57 ± 5.6	40 ± 15
NISS (95% Confidence Interval)	23–33	N/A	39–50	52–61	35–44
Number of admissions (± SD)	1144 ± 5.2	8 ± 4	13 ± 4.3	12 ± 3.7	11 ± 4.7
Length of rehab (months ± SD)	39 ± 15.2	20 ± 10.6	33 ± 10.2	44 ± 8.5	34 ± 14

Table 40.2 Injury presentation comparing exposure to blast in vehicle vs foot patrol: case studies of two soldiers (#=Fracture)

Case Study 1. Blast injury—vehicle (mounted) patrol	Case Study 2. Blast injury—foot (dismounted) patrol
# right maxilla	Right hyphaemia
# right mid humerus	Left globe disruption with intraocular
Right elbow dislocation	Haemorrhage
# left radius and ulna	Complex facial lacerations
# sternum	Degloving of left maxilla
Compression # L2	Devascularisation of right forearm and hand with brachial artery tear below elbow and ulnar artery division at the wrist
Unstable burst # L3	
Small right pneumothorax	
Cardiac contusion	Complex right-hand trauma with loss of P2 thumb, P2 index, open PIPJ ring, and distal pulp loss little finger
# right tibial plateau	
# right lateral malleolus	
# right 3,4,5 MTPs	Left orchiectomy
# right calcaneus	Traumatic left trans-tibial amputation
# left fifth MTP	Traumatic right trans-femoral amputation

40.2 Principles of Military Rehabilitation

Defence rehabilitation is an occupational health service, seeking to rehabilitate injured military personnel back to duty or to support their transition from service to civilian life [9]. This vocational focus, set within a broad military context, directs the resourcing and the philosophy of the approach taken.

Military cultural ethos promotes courage, determination, intensive training, and group socialisation over individualism [10]. The nature of the military task and training requires personnel to have a high level of fitness. When faced with injury, this patient demographic, socialised and conditioned for intensive effort, focused upon their vocational outcome and demanded a functional goal orientated approach which restored them back to fitness and function as quickly as possible [11, 12]. Recovery is as much about preventing further physical and psychological decline during recovery, as it is about enabling tissue healing [1]. In order to expedite return to function and activity, defence rehabilitation models its delivery around five core principles:

- early assessment;
- exercise-based rehabilitation;
- Cross-disciplinary working: the interdisciplinary team;
- active case management; and
- rapid access to specialist opinion/investigations.

These principles sought to bring momentum and a functional and vocational focus to the rehabilitation provided. This robust model of care, which together with the supporting infrastructure and equipment, provided the foundation of what subsequently developed with the advent of war [1]. Significant realignment of services and team structures were needed to support the development of the complex trauma team at DMRC. However, the rehabilitation of the combat casualty started in the Intensive Care Unit (ICU) within the deployed setting, and it continued as the casualty was evacuated back to Queen Elizabeth Hospital, Birmingham (QEHB). Critical care and trauma rehabilitation specialisation were needed. However, the core question was; could military rehabilitation apply these principles to a very different injury presentation being treated, in the acute phase, within a very different context?

40.2.1 Early Assessment

In a specialist musculoskeletal (MSK) military rehabilitation centre, it is a relatively simple process to organise a cross-disciplinary assessment. It is a much more complex process to do the same in a large hospital, where specialist rehabilitation practitioners and ward-based clinicians are separated by geography, culture, and work schedules. Once the casualty arrived back in the UK, rehabilitation started in the ICU in QEHB and continued as they moved to the trauma ward. Current trauma guidelines recommend a rehabilitation assessment within the first 48–72 h following admission [3, 6]. However, this is often not feasible in the initial stages of admission where the focus is upon saving life and limb followed by the laborious process of cleaning and debriding wounds in order to commence reconstruction [7]. Gaining access to the patient was one challenge; the spread of injuries required a diverse specialist knowledge of assessment and treatment, and this input was required over several weeks as the casualties recovered on the ward. The specialist outpatient setting is more familiar with patterns of recovery, whereas the inpatient clinician will usually only see the patient for a few days before they are discharged home. Given the length of stay, we sought to start outpatient care whilst the casualties were still inpatients. This matter required sensitive handling, as it challenged traditional boundaries of practice, yet it was critical in enabling the identification of rehabilitation need and to address potential long-term factors that would limit rehabilitation before it was too late.

The complexity of injury in blast can have profound ongoing functional ramifications, especially with tendon healing in the upper limbs, neurological issues, and scarring. To facilitate recovery, early mobilisation and engagement with simple daily functional tasks are important. However, without key specialist knowledge, progress can be unwittingly impaired. Simple interventions such as providing pressure garments or resting splints, ensuring the position of an external fixator allows for transfer to and sitting in a wheelchair, or identifying a patient who needs to be fast-tracked for further input can make a difference. In the hospital setting, military funding was provided to augment inpatient provision with outpatient specialist input. Hand therapists, neurological/spinal specialists, and burns therapists all worked with the inpatient team in critical care or on the ward. These relationships not only enhanced patient care, but they also enhanced the skill base of the inpatient team. This in turn informed the wider medical team's provision at an early stage.

Specialist support also came from DMRC. A fast-track prosthetic service was developed for upper limb amputees. Prosthetic casting commenced as soon as possible. The aim was to fit an upper limb prosthesis within 2 weeks of amputation [13]. The rehabilitation consultant also attended a weekly ward round at QEHB, thereby providing early awareness and assessment of needs in preparation for their subsequent admission to DMRC. For a previously fit population of critically injured military personnel, enabling access to early specialist assessment and treatment informed surgical and medical planning during the formative reconstructive period. Furthermore, it developed the skills of the inpatient medical team. Together, these outcomes enabled the casualty to progress towards the functional independence they longed for.

40.2.2 Exercise-Based Rehabilitation

Exercise-based rehabilitation uses physical training to aid recovery, retaining a clear focus on the vocational outcome it seeks to achieve. Within UK Defence Rehabilitation, this is the primary role of the Exercise Rehabilitation Instructor

(ERI). Knowledge of exercise physiology, military styled group therapy, and tissue healing enables the ERI to modify their exercise prescription according to the injury presentation. This role also promotes a military ethos ensuring that the rehabilitation context maintains a vocational focus.

Traditionally, the application of exercise-based rehabilitation within the military occurs within an outpatient or residential rehabilitation setting predominately focused upon MSK injury. The complexity of blast injury invariably leads to the patient requiring extensive input and time within critical and high dependency care. The critically ill, polytraumatic presentation requires prolonged periods of immobility, mechanical ventilation, and sedation. The adverse effects upon the physical capacity of a patient following mechanical ventilation, prolonged bed rest, and the administration of pharmacological agents have been well documented [14–16]. Physiological changes such as reduced oxygen uptake, skeletal, respiratory, and cardiac muscle disuse atrophy, insulin resistance, postural hypotension resulting from fluid loss, decreased stroke volume, decreased bone density, joint pain, and loss of functional range in joints are similarly known [14, 17, 18]. In addition, the systemic inflammatory response syndrome, which occurs with major trauma, creates a dramatic catabolic effect in patients already systemically compromised by injury [19, 20]. Reduced oxygen uptake following a period of bed rest was also proportionally greater in those with a high VO_2 max [21]. As such, one might argue that soldiers conditioned to the extreme physical demands of operational service suffered proportionally greater reductions in their cardiovascular fitness compared to an age matched civilian patient [21, 22].

The complexity of blast injury is further challenged by the systemic support such injuries demand and the impact of this support upon body systems. The challenge, in the acute stages of care for the rehabilitation specialist, is how one might abate the systemic consequences of critical illness and bed rest. The importance of early exercise-based rehabilitation commencing in critical care is now widely accepted thanks to the work performed during this era [15, 16]. Effective early intervention depends upon achieving sufficient intensity, volume, and frequency of the targeted intervention, and this requires the correct skills mix and ratio of staff. Staffing models in the hospital setting, therefore, reflected recommendations for critical care (patient: physio 6:1) in an effort to ensure each patient could receive an hour of input per day [23]. This was enhanced further by posting an ERI onto the ward, which had never been trialled before. Specialist training was provided, and support from fellow therapy staff was given. The creation of this role was significant in providing a functional exercise-based focus, as well as supporting casualties with a military peer who understood their needs and spoke their language. A dedicated occupational therapist (OT) was also employed to ensure that wheelchair provision and home adaptations could be fast-tracked. In this way, independence around the hospital was achieved early, and the length of stay was reduced.

40.2.3 Cross-Disciplinary Working: The Interdisciplinary Team

The Apollo program [8], the war on terror [24], and the success of many digital companies are all prominent examples where a collaborative, decentralised team approach overcame complex threats or circumstance [24]. Such an approach was mirrored at DMRC. Complex trauma clinicians, managers at DMRC, and their patients were asked what components of the rehabilitation service were, in their opinion, critical to ensure the successful recovery of the combat casualties. One consistent response was given across all focus groups and interviews—the Interdisciplinary Team (IDT). These examples of collaboration, and many more like it, align with the findings in our research and evidence from other quarters of healthcare [24–27] in which the solution to the challenge is found within the human context in which team interaction occurs [28, 29].

At DMRC, the surge of complex operational casualties challenged traditional healthcare delivery, where the multi-disciplinary team (MDT) consists of professionally autonomous departments providing parallel outputs for the same patient. These complex operational injuries needed a different approach or human context in which cross-disciplinary collaboration and support would flourish. The complex trauma team arose out of this crisis and adopted an interdisciplinary approach. The previous traditional multidisciplinary departmental structure was described by participants within our research as tribal, repetition of clinical effort was frequently reported, and cross-disciplinary solutions were rarely achieved. In contrast, the IDT was dynamic, cross-disciplinary working defined roles, and there was a strong ethos of mutual support. What started as an experiment quickly overturned traditional practice. Despite escalating patient numbers, the success of this approach was obvious. Clinical departments (OT/physiotherapy/social work, etc.) were abolished, and clinicians were co-located as interdisciplinary clinical teams focused upon provision for a specific patient group. Complex Trauma, Neurology, Upper Limbs, Spines, Lower Limbs, and Medicals (later named Specialist Rehabilitation) were created. The professional profile and clinical composition of each team were determined by the needs of the patient.

Explanation of this team dynamic requires us to be clear on what each term means [29]. The multi-disciplinary team (MDT) refers to a team in which different professionals, working in parallel, pursue their own professional goals, but coordinate their independent activity through regular meetings [29]. The IDT is marked by a more integrated approach, in which all professionals share the same goal, and their contribution is collectively agreed. Territorial boundaries are also broken down by co-locating clinicians and through joint assessments and treatment sessions [29].

Collaboration is the central goal when managing a complex scenario, and an interdisciplinary context appears conducive to achieving this outcome [28]. However, an interdisciplinary approach is not without its risks [28]. A growing understanding of social cognition and group processes teaches us that context is only part of the answer. Nurturing the concepts which make up collaboration is critical to enable it to flourish [28, 29].

Within our research and in the wider published literature ‘sharing’, partnership, interdependence, and empowerment are components critical to collaboration. ‘Sharing’ is a concept that includes working towards a unified set of goals sharing decision making, responsibility, and skills [29]. The concept of ‘partnership’ emphasises a genuine cohesive network, which operates with mutual trust and respect, open communication, and honesty. Partnership was reported in our own research between clinicians and patients. Interdependence between professionals occurs when cross-disciplinary sharing and partnership arise from a recognition of the unique contribution of each professional. The willingness to blur professional boundaries and collectively approach a problem creates interdependence [29]. Empowerment or transitional leadership refers to the allocation of responsibility according to expertise and the treatment goal under focus [24]. In our own research, empowerment was linked with a mutual trust and interdependence within the IDT. Genuine creativity and innovation arose from this collective collaboration.

These concepts need a context in which to flourish. Geographical proximity of members was key, as was accessible non-hierarchical leadership. However, mutual trust seems to be the common thread linking each concept. Many authors feel that successful collaboration and cross-disciplinary working start with the leader. Case studies show that the leader who adopts a non-hierarchical position embracing an ‘*Eyes-On, Hands-Off*’ integrative style is more likely to nurture an ecosystem of collaboration within the IDT and across the rehabilitation continuum [24].

A comparison of accounts from clinicians working at DMRC within an MDT versus IDT arrangement suggests that the experience of teamwork and leadership was more positive for members of the IDT. Statements reflecting a sense of resilience were also common for IDT members. These findings are supported in the wider literature [30]. Thematic analysis of accounts from members of the IDT also shows a positive approach to exploration, innovation, and failure. Failure was embraced as a part of learning, and seen as a necessary part of evolving practice, especially where there is little evidence and the outcome is uncertain [31].

This section has centred upon the team configuration at DMRC. However, the intensive medical and surgical input required for this cohort meant that upon arrival in the UK their care journey commenced within the NHS. Even following admission to DMRC, ongoing surgical episodes were often needed. In short, there was a dense interdependence between the acute NHS setting and specialist military rehabilitation. Cultural, organisational, and geographical difference may well have undermined the collaboration and interaction needed between the military and NHS. This was averted by a concerted effort to build connection and communication. From this partnership, the practical steps taken to enable this strategy are presented in Sect. 40.4.

40.2.4 Active Case Management

Active case management seeks to maintain momentum in the recovery process. It aims to identify issues before they arise or offer solutions when they do. As discussed, the emphasis of military rehabilitation upon functional outcome focuses the provider and the recipient upon physical forms of recovery. For this cohort, physical recovery benefited the patient psychologically and socially, whilst also minimising loss of physical capacity. The risk, however, was that the psychological and social impact of injury would be neglected.

In the hospital setting, the term ‘Military Bubble’ was created. It came to represent a protective process which sought to envelope the patient and family, overwhelmed by events. Families’ welfare, social and financial needs were actively managed. A military unit within the hospital centrally coordinated the social and psychological well-being of both the patient and their family.

Upon transfer to DMRC, the family’s needs became the responsibility of the soldier’s regiment. Rehabilitation now became the soldier’s main occupation. The IDT became the cross-disciplinary forum in which physical, psychological, and social recovery were actively monitored and managed. Veterans rated this clear delineation of responsibility and the geographical colocation of the team as a significant factor that enabled them to focus their own cognitive resources upon recovery. In short, they knew someone was looking out for their needs, and they knew who to approach if they had a problem.

At the outset of military operations, the eventual discharge destination for combat casualties was uncertain. As it became clear that many were unlikely to return to military duty, transitional arrangements to assist their medical discharge and reintegration into civilian life were developed. This process was not part of the rehabilitation service, yet transitional needs and ongoing medical rehabilitation issues are often entwined. The measured consequences of this separation remain unclear; in our own research, veterans prioritise issues arising from transition above all others they currently face.

40.2.5 Rapid Access to Specialist Opinion/Investigations

The principle of active case management is a philosophical stance in which momentum is sought through the active networking of specialist opinion, investigations, and treatment provision. It relies upon access to timely funding, a network of specialists able to provide opinions, ongoing investigations, and adaptive equipment, but above all, it requires a nominated principal coordinator or coordinating unit. Active case management must be able to interact with the military chain of command, as well as civilian medical and social support in order to access specialist services and opinions, so they can enact solutions and guide the journey of recovery. The blast-injured amputee can face several orthoplastic issues, which will impact their ability to use prostheses. Soft tissue breakdown, the development of heterotrophic ossification, painful neuromas or recurrent pockets of infection can complicate socket fitting [1]. In addition, damage to peripheral nerves often needs specialist wisdom to know when to wait for recovery and when to intervene [32].

Active case management and rapid access to specialist opinion or investigations go hand in hand. The accountability and coordination implied within active case management make best use of the resources needed to access specialist opinion and investigations. There is no point having one without the other [1]. Together, they will minimise stasis and maintain the momentum of recovery. This chapter will now outline how these principles were applied as the rehabilitation provision evolved at QEHB and DMRC.

40.3 Hospital-Based Rehabilitation of the Blast-Injured Amputee

In 2001, the Queen Elizabeth and Selly Oak Hospital (later QEHB) became the primary receiving unit for all military patients injured overseas. Providing medical support to military operational deployments, QEHB works hand in hand with the integrated Royal Centre for Defence Medicine (RCDM). During the conflict in Iraq and Afghanistan, casualty numbers tested this partnership. With the fight for survival won, rehabilitation brought a shift in focus towards achieving function, recovery, and a quality of life. The subsequent success achieved in the rehabilitation of military casualties revealed the potential benefit on offer for civilian patients and society as a whole if rehabilitation services could be aligned and adequately resourced [6].

Reflecting upon lessons learned during the Vietnam conflict, Brown (1994) [33] identified the importance of commencing rehabilitation as early as possible, creating centres of excellence, limiting convalescent leave, and introducing recreational and motivational activities. A retrospective analysis of the rehabilitation lessons from Iraq and Afghanistan mirrors this analysis.

Hospital-based rehabilitation following blast injury aims to promote early independence, increase function, reduce pain, and eliminate complications from the injuries sustained [34] and was facilitated by patient-orientated goal setting, clear communication, and the IDT. Military and civilian therapists along with a Rehabilitation Coordination Officer (RCO) connected and coordinated the team and rehabilitation pathway. Military funding enhanced resources provision. Key performance indicators ensured timelines of provision. With resource and equipment provision in place, quality improvement became a matter of building trust across disciplinary collaborations, between and within clinical disciplines. The result for the clinical team was a dynamic system of rehabilitation that could navigate the unexpected. The result for the patient was that combat casualties achieved functional goals upon discharge which far exceeded their civilian counterparts [2, 35].

40.3.1 Early Rehabilitation in Critical Care

The adverse effects of bed rest and the systemic challenge of major trauma have been previously documented. Combat casualties were assessed at the earliest opportunity, and early exercise-based rehabilitation commenced as soon as patients were physiologically stable [14, 18]. Commencing early mobilisation and an exercise-based physiological challenge within the ICU have been shown to reduce hospital length of stay, mortality, the deleterious effects of prolonged bed rest, improve longer term functional outcomes [15, 16, 18, 36–38], and provide financial benefits in terms of both the immediate acute care burden and the ongoing rehabilitation requirement [16, 18, 37, 39, 40].

Some key advances were achieved. For example, for those with blast lung, knowing whether and how to apply exercise-based activity, potentially creating further oxygen debt, was a dilemma. The IDT introduced heated humidification [41–43] for patients considered ready to mobilise, but for whom oxygen delivery during exertion was an issue. This resulted in improved oxygenation highlighting the success of the IDT in moving clinical practice forwards.

Another advance was the expansion of the ERI role into a civilian NHS ward setting. The success of this strategy and the acceptance of the ERI as part of the medical rehabilitation team by NHS colleagues enabled the incorporation of this role into the roving military rehabilitation team. The military rehabilitation team was now fully equipped to provide one-to-one exercise prescription and group therapy in the burns and trauma wards, or ICU. This was further enhanced with equipment such as the MOTOmed® (an over bed bike), provision of mobile weights and exercise bands, or even taking patients to the rehabilitation gym from the ICU.

It has been asserted that the benefit realised following the introduction of a cross-disciplinary rehabilitation strategy lies more in the cultural shift which takes place on the ICU [16]. The introduction of the early rehabilitation team at QEHB provided rehabilitation specialists with a place and a voice in the daily planning meetings. Nursing staff shared mobility and rehabilitation goals. Collaboration therefore grew between anaesthetics, nursing, and the mobility team [16]. Augmenting the team in critical care with a military physiotherapist supported the outreach of rehabilitation from ICU into the burns, trauma, and neurology wards. The roving and networking function of the military physiotherapist also brought about 'in-reach', as discussed. Their role on the pain management team, interaction with hand therapy, spinal cord and neurological specialists as well as links to DMRC enabled these specialisms to input into critical care.

40.3.2 Early Function and Progress to Ward-Based Rehabilitation

The IDT provided a forum that reminded clinicians of the broader context of recovery. The impact of mild traumatic brain injury associated with exposure to blast, communication issues, vision loss, loss of sexual organs, testosterone replacement regimes, nutritional status, sleep disturbance, fatigue, and pain control issues must all be addressed in concert when commencing intensive rehabilitation [34]. Ongoing surgical intervention was clearly a necessary part of this process, but it interrupted momentum. Prioritisation and scheduling of clinical intervention were therefore a daily routine involving all clinicians.

The emotional ups and downs involved in recovery needed to be managed with care. Arrival on the ward was viewed by patient and family as a positive step forward. However, whilst some casualties could mobilise independently or take trips outside in their wheelchair, others were confined to the ward or an isolation area. Such restrictions could be demoralising, as were medical setbacks such as infection, wound break down, pain, and further complications [33]. Equipping the patient and their family to understand the process, allowing them to challenge decisions whilst explaining alternative rehabilitation approaches, was critical at this time [34]. Mental health support to the staff also equipped them to monitor atmospherics and to manage situations as they arose.

Successful rehabilitation requires trust between clinician and patient, but trust requires the investment of time, and clinical time on a ward full of competing priorities can be a challenge. Many casualties were junior soldiers, and few would frankly communicate their issues to a senior military officer who was their surgeon. Even fewer would do this when in front of many clinicians conducting a ward round. Their need for a trusted clinical advocate was clear. The military rehabilitation specialist had been present at each stage of the patient's journey, spending significant time and thereby building a rapport, as well as having some insight as to what lay ahead for them. For this reason, they frequently found themselves in a role where the patient sincerely opened up about their thoughts and fears. At such a time of overwhelming uncertainty, support for the patient from peers, fellow patients and trusted clinicians was even more critical.

For the soldier who has trained and fought with others, camaraderie is a powerful source of support and motivation at a time of uncertainty. Many knew each other or were injured in the same incident, and for those who did not, their proximity and circumstance quickly helped them to build bonds of trust and kinship that continue long after. Patients who needed to be isolated due to infection suffered notable psychological strain and needed greater clinical support. At these times, clinicians could call on welfare support within the 'military bubble'. Interestingly, within our own research, veterans reflect on the isolation they felt when the trauma ward moved to a location with predominately single rooms as opposed to the eight bed bays in the previous location. The introduction of the ERI to the ward brought significant therapeutic gain in such circumstances. Their military approach, using familiar language and humour, drawing on group therapy, brought connection, providing a social and supportive element. Bringing patients together in this way encouraged what has become common in the field supporting physical and mental health, the buddy-buddy system, where both looked out for the other [44].

40.3.3 The Rehabilitation Coordinating Officer

The Rehabilitation Coordination Officer (RCO), the military Registrar on the ward, the civilian OT employed by the MoD, but working in the NHS, as well as the military physiotherapist working across disciplinary settings, are four examples of strategic posts that supported the IDT approach in the hospital setting. These individuals knew the team at DMRC, they were key points of contact when questions arose, and their presence thereby ensured continuity and collaboration between organisations.

The RCO was the most significant of all these roles. They were the focal point for information regarding a patient's ongoing treatments, multiple surgeries, their rehabilitation, and future care requirements. They acted as the patient's advocate to ensure that their needs and wishes were considered, and medical plans contributed to the overall process of recovery. They were also a key information source and support for the patient and their family. The RCO role was cited as an example of best practice by the Regional Networks for Trauma NHS Clinical Advisory Group Report in 2010. This recommendation has been taken on by a number of major trauma centres [3, 6]. Further collaborative initiatives can be found in Table 40.3.

Table 40.3 Strategic collaborative initiatives (From Pope et al. (2017) [34], p. 125)

Practices	Summary of key initiatives promoting cross-disciplinary working practice
MDT meetings and communication	Weekly MDT meeting with agreed treatment plan, proceedings managed by the military registrar. Attendees: Consultants (including radiology, pain, and microbiologists), nursing, rehabilitation (including RCO), mental health, and welfare
	Weekly rehab IDT meeting specifically for goal setting, including occupational therapist (OT), physiotherapist, ERI, and nursing staff
Therapy sessions	IDT therapy treatment—Joint therapy sessions conducted in the therapy gym with OT, physio, and ERI present. An emphasis placed on function
Teaching across professional groups	Teaching sessions carried out to ensure nursing staff and healthcare assistants were skilled handling complex patients, including safe transfers during silent hours
Orthotists	Close working with technical specialists to allow custom made adaptations to standard equipment thereby increasing function. For example, dorsi-wedges adapted to maintain Achilles tendon length without compromising k-wire placement

Table 40.3 (continued)

Practices	Summary of key initiatives promoting cross-disciplinary working practice
Strategic military clinical appointments	Rehabilitation coordination officer (RCO)—Ensured communication between all stakeholders for planning current and ongoing rehabilitation requirements
	Designated military registrar—Link between consultants and MDT. Ward-based to allow regular access and continuity across multiple clinical teams
	Military physiotherapist ICU/burns outreach: Principal clinical liaison between burns, trauma wards as well as specialist outpatient rehabilitation services (e.g. hands, neuro)
	OT: A dedicated OT to work with military complex trauma providing an enhanced service ensuring access to wheelchairs within 24 h
	Addition of military personnel and professional development officers in critical care and trauma posts, enhancing service while maintaining competencies
	ERI: A unique role within the MOD, delivering exercise, and recreational therapy in preparation for inpatient rehabilitation at DMRC

40.3.4 Innovating Practice

Innovation born from this shared awareness helped orientate the service towards the unique needs of its patients. The development of a functional outcome measure that was sensitive to the patient population was one example. The need to measure progress and to communicate ongoing patient problems is especially important in the complex scenario [45]. Established outcome measures were designed for a very different population and presentation to that of the soldier [12]. Understanding the information needed at DMRC led to the creation of a military specific 'Pre-Prosthetic Function Outcome Measure' (PPFOM). This outcome measure was internally validated against the Amputee Mobility Predictor Questionnaire (AMPQ), and good inter-rater reliability was also shown (unpublished) [34]. Practices which evolved at QEHB during the recent conflicts in Iraq and Afghanistan are captured in Table 40.4 [34]. Not all interventions are novel, but the scope of provision was.

Table 40.4 QEHB/RCDM Therapy practice development (From Pope et al. 2017, p. 126, Table 2) [34]

Area	Issue	Solution/impact
Transfers	Type of mattress	Requested early adoption to optimal mattress firmness.[a] This is mutually beneficial in facilitating early mobility and to reduce the need of a pressure-relieving mattress
	Bed mobility	Appropriate provision of 'high-low' hospital bed/cot sides/grab handles to facilitate independent bed mobility
	Dependent transfer required (see Fig. 40.1a)	Bed to plinth day one post-operation[b] on wide Bobath plinths with appropriate slide sheet use. Improved motivation by leaving ward and attending gym
		Developed forwards-backwards transfer use, to allow early transfer out of bed. Use of one-way glide sheets. Early issue of banana transfer boards
	Hoisting amputee patients	Ensured procurement, training, and use of appropriate amputee slings. Adequate stock kept ensuring no delay
	Complex soft tissue injuries (Fig. 40.2)	Use of Vicair® AllRounder buttock cushion for significant blast injuries to the buttock/pelvis. Allows early mobility while protecting weight-bearing skin, reducing friction during transfers. It also facilitates sitting balance for hemipelvectomy patients (Fig. 40.3b)
		Close liaison with burns and plastics team, including substantial use of topical negative pressure (TNP) dressings to reduce shearing forces and increase early mobilisation

(continued)

Table 40.4 (continued)

Area	Issue	Solution/impact
Gym	Attendance	Early daily gym attendance starting day one[b] with ERI. Plinth gym exercises instead of bed exercises on the ward. Change of environment, routine, and independence increased morale
	Clothing	Normalisation avoided hospital gowns and encouraged use of shorts and T-shirt during physical training. A rucksack was provided to carry attachments, i.e. TNP (provided by TroopAid charity)
	Tolerance	Split rehabilitation, i.e. exercise therapy session/cardiovascular fitness a.m. and therapy intervention or functional rehabilitation p.m., focussing on strength, range of motion and graded exposure
	Motivation/group therapy	Recreational and group therapy led by ERI included competitive games, encouraging group cohesion, mutual support, and distraction therapy.
	Breakfast club	To mark the end of the working week, nursing staff brought all military patients to the ward gym for a group exercise session. Afterwards staff and patients were given breakfast in the hospital canteen. This initiative brought about group cohesion (staff and patient), normalisation, and exercise adaptation, as well as enjoying breakfast
Seating	Limited by external fixators (ex-fix) (see Fig. 40.1b)	Surgeons taught seating requirement so they could position pelvic ex-fix bars to enable seating and transfers. Adaptations to chairs, i.e. tilt and space recline/removal of arm supports
		Two plinth gapped bridge to allow prone lying for amputees with pelvic ex-fix and avoid hip flexor contractures, or side lying modified Thomas test stretch allowing ex-fix to hang over edge of plinth
	Standard seating unsuitable	Occupational therapist facilitated bespoke specialist seating and pressure mapping as early as possible, i.e. for hemipelvectomy patients
Function	Prone nursing (see Fig. 40.3a)	Prone wheeled trolley made by military medical engineers to allow independent mobility for patients who required prone nursing due to significant buttock wounds
	Wheelchair independence	Early assessment and provision of wheelchair (designated stock), including short-term provision of electric wheelchair to allow early independence
		Unconventional transfer methods taught, i.e. wheelchair to shower chair, in/out bath, on/off floor into a wheelchair[b]
	Amputee mobility (see Fig. 40.3b)	Bridging work using wedge/BOSU® trainer to balance residual limb/s. Plinth under parallel bars to practise sitting for bilateral amputees. Progressed to sitting on BOSU®/gym ball, exercising or using Wii™
		Early use of floor to plinth/chair/bed transfer and step/blocks transfers
		Negotiated weight-bearing status with surgeons, balancing multiple injuries, and function, i.e. early kneeling on through knee amputations or upper limb stump lever use to facilitate independence with transfers
		Tilt table with Femurett/pneumatic post-amputation mobility aid weight-bearing, progressed to parallel bars to allow early standing
		Innovative ways to enable stairs use (patients often declined stair lift) such as seated stairs manoeuvres[b]
	Outdoor mobility	Exposed to outdoor/multi-terrain/obstacles in preparation for discharge, incorporating trips out of hospital to allow integration into society

[a]As Waterlow scores dictate
[b]If no contra-indications, i.e. wounds, skin grafts, or pressure concerns and balancing independence/risk

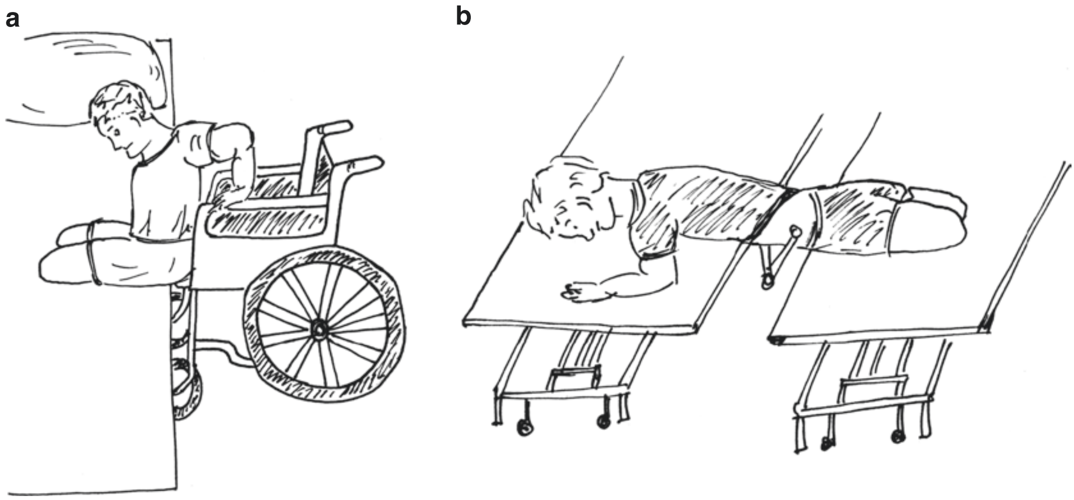

Fig. 40.1 Adaptations made in the hospital or acute stage (**a**) Forwards–backwards transfer. (**b**) Prone lying with pelvic external fixators (from Pope et al. 2017 [34] p. 127)

Fig. 40.2 Adaptations made in the hospital or acute stage. Vicair® AllRounder buttock cushion

Fig. 40.3 Adaptations made in the hospital or acute stage. (**a**) Prone trolley. (**b**) Exercising on BOSU. (Images from Pope et al. 2017 [34] p. 127)

40.3.5 Pain Management

Management of acute pain has evolved considerably from the point of wounding through to UK hospital care. From fentanyl lollypops delivered on the battlefield through to early use of neuropathic medications, epidurals, and nerve blocks [46–48], the evolution has largely centred upon the way in which these agents and modalities have been used [49]. The focus should never be upon only removing the pain. What really matters is whether the intervention enables the patient to ‘do more’ [49]. If the agent or modality does not allow the patient to ‘work, rest and play’; the team must think again [49]. Pain must not be allowed to impede the momentum of rehabilitation. Coordination of clinical activity and early identification of critical cases were essential to avoid this. As a result, a rehabilitation representative became a core member of the military pain team. Cross-disciplinary collaboration sought to coordinate rehabilitation activity and to titrate and time analgesia according to symptoms. Identification of cases where symptoms were not effectively managed led to early investigation and intervention by the medical team [48].

In the amputee, phantom limb pain (PLP) has been notoriously difficult to treat, with potentially catastrophic consequences for function in long-term sufferers [49]. Amongst military amputees, up to 70% reported symptoms in the acute phase of recovery [50]. Conventional treatment has also shown a low incidence of success [51]. Nevertheless, a growing pathophysiological understanding of PLP together with a more holistic appreciation of intervention strategies appears to produce some encouraging longer term outcomes. Recent evidence has shown on discharge from DMRC, combat casualties reported either no pain, or pain that is controlled.

PLP is a neuropathic condition with a complex aetiology. Factors within the central and peripheral nervous system have been shown to contribute to its presentation, whilst the patient’s context and interpretation may further magnify or dampen their pain [52]. Treatment must therefore acknowledge this complex interplay between pathophysiological, social, and psychological factors [52]. As a result, there is no ‘one size fits all’ treatment approach [49]. In an effort to support clinicians caught in this fog of uncertainty, and to move assessment and treatment planning beyond the biomedical sphere, an evidence-based military pain hierarchy was developed (Fig. 40.4) [49].

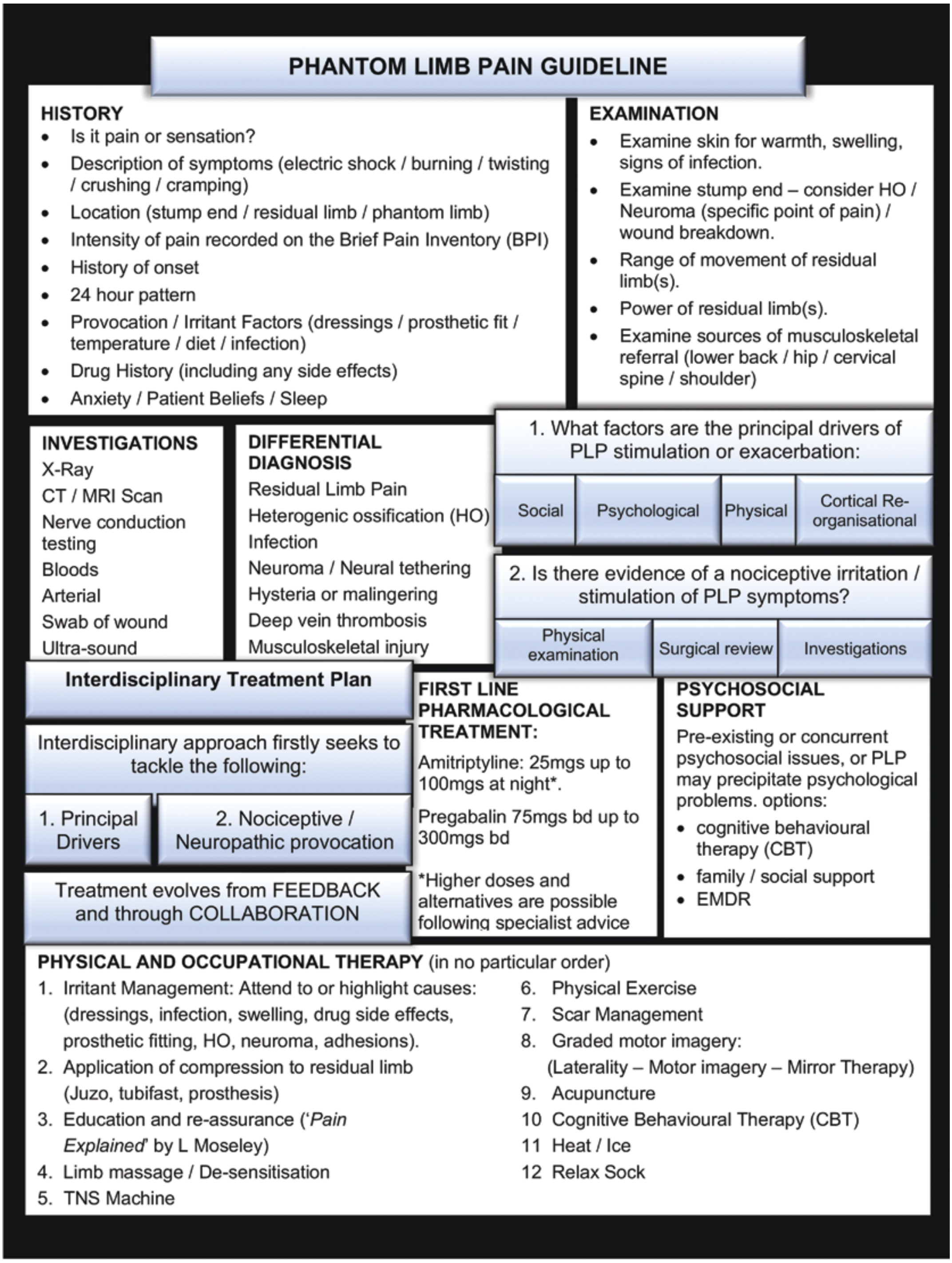

Fig. 40.4 A summary of treatment for phantom limb pain (PLP) (from Le Feuvre & Aldington 2014 p19) [49].

The approach seeks to equip clinicians with an appreciation of how rehabilitation strategies may influence PLP and to help identify if other initiators exist. Transcutaneous-electrical nerve stimulation (TNS), mirror therapy, exercise therapy, compression socks, massage, desensitisation, distraction acupuncture and Entonox administered under a Patient Group Direction (PGD) were common strategies used at QEHB [34]. Observational work and case study reviews combined with this guideline to help the team problem solve on a case-by-case basis.

Consistent with the principles of defence rehabilitation, the emphasis of pain management was upon the restoration of function and strength, in order to allay cortical and cognitive factors believed to influence the individual's perception of pain [53–55]. Adjunct treatments targeting PLP were then presented as a tool to enable the casualty to self-manage. The importance of practices such as pacing activity and sleep hygiene were given equal importance.

Where there was extensive scarring and skin grafts, initial restorative management looked to desensitise and mobilise scar tissue. The aim was to normalise stimulus and sensation and therefore reduce the pain response. Compression garments were a central element to this approach. The irregularity of some residual limbs meant that bespoke compression shorts were sourced. 'Relax socks' for night pain also had some anecdotal success [56]. Although there is little available published evidence to support the use of compression to relieve PLP [57], anecdotally, benefits were reported [58].

Graded Motor Imagery (GMI) has a growing body of evidence justifying its use as a treatment for neuropathic pain, including PLP [59, 60]. The regime promoted by the Neuro Orthopaedic Institute (NOI) was followed. 'Recognise' online software and the use of the 'flash cards' were started early. Where the patient seemed adept at mental imagery, this was taught as a self-management tool. Mirror therapy (MT) alone or as part of the GMI approach seems to have some support when treating PLP, especially in upper limb amputees [55, 60, 61]. The practical challenge during their hospitalisation was to find an environment where it could be performed in relative peace. In addition, few casualties were without injury to the contralateral limb. However, MT was especially helpful in cases where the phantom limb remained in a fixed position. Its use appeared to free the painful 'stuck' limb [34, 62]. Mental imagery or other local forms of stimulation may then offer ad-hoc relief but did not work for all patients.

Western acupuncture in the management of PLP [63] had mixed results. It tended to be an early option when dressings were 'bulky', mirror therapy was not appropriate, or the patient was unable to participate in active rehabilitation. Early use of 'battlefield' acupuncture has been reported in the US [64] and was trialled at QEHB with some anecdotal success reported.

Recent work is now suggesting further surgical options for pain management with the development of targeted muscle reinnervation (TMR) [65] and regenerative peripheral nerve interface (rPNI) surgery [66]. Developments around pain management are key to follow as this is an area of great impact to patients.

40.4 Military Rehabilitation of the Blast-Injured Amputee

Military rehabilitation is built around a model of sport and exercise medicine, placed within a biopsychosocial model, driven by a goal-centred approach. As discussed, this approach sought to encapsulate the vocational requirement with individual characteristics and motivational targets, packaged in an active scaled military training format. Whilst most amputee rehabilitation in the UK originates from a background of vascular and diabetic care, the traumatic nature of the amputations to a young and previously extremely fit population, meant that a sport and exercise therapy approach was innately beneficial to maximise the rehabilitation of this very different cohort of amputees [33, 67]. The rehabilitation team consisted of both military clinicians, familiar with the vocational demands of soldiering and deployment, and clinically specialised civilian staff who provided continuity and depth of clinical experience.

Blast injury amputee rehabilitation is not just about prosthetic provision. The complex nature of these injuries meant that therapy had to address many other concomitant conditions, such as scar management, fracture healing/weight-bearing status, Peripheral Nerve Injury (PNI), brain injury, Post-traumatic Stress Disorder (PTSD), and other mental health conditions [7]. Pain management strategies, as previously mentioned, included medication, activity pacing, cognitive strategies, cognitive behavioural therapy (CBT), acceptance and commitment therapy (ACT), GMI, and manual therapies.

Patients were admitted for residential blocks, after which they returned home for a designated period. Activity during these residential blocks was frenetic. Communication, negotiation, and collaboration between many potentially overlapping therapies were required. However, during periods at home, activity was at the discretion of the patient. As an example, in early rehabilitation at DMRC nutrition was an especially important element as many patients arrived from QEHB still in a catabolic physiological state [1, 7, 20]. As previously fit individuals, they had become used to eating whatever they liked without gaining weight; as their metabolic rate settled, and they were unable to exercise as much, they started to rapidly gain weight. This would alter stump shape and size, affecting the fit of their prostheses, as well as their overall health. Weight gain would further complicate their capacity to take part in skills training, altering their centre of gravity. Specific nutritional guidance from the dietician for each patient ran concurrently with education sessions run by the Dietician, ERI, and OT to complement the advice and training programmes supplied by the ERI and cooking/food preparation skills taught by the OT.

The service described throughout this chapter evolved over time. 'Eyes on hands off' leadership devolved authority to the clinicians as evidence of trust in their skill [24]. The IDT had the peer support to be able to face failure and the leadership backing to seek solutions. Keeping problem solving near to the clinical environment led to joint problem solving resulting in further innovation of practice. The solutions were owned by the team, who had created them.

The Phoenix service is an example of this clinician led innovation. As patients achieved higher level functioning, and their expectation of the service grew, so did the challenge facing clinicians. Patients voiced concern that the prosthetic service was not sufficiently responsive in dealing with socket related problems, especially with challenging residual limbs and bilateral amputees. With the growth in the team, and the turnover of staff, passing on the accumulation of knowledge was a challenge. In response, the clinical lead prosthetist and physio formed a problem-solving team to support other staff: the 'Phoenix service'. The service increased teaching and supervision for newer staff. It also supported existing staff and patients, in particular demonstrating what is mechanically due to the prosthesis and what is mechanically required from the human body (e.g. range of movement, muscle control, fitness/endurance, and the impact of losing muscles and joints). Where traditional prosthetic casting and fitting processes were not working, due to the complex nature of the injuries, this service supported prosthetists to consider non-traditional or even 'dated' options. Newly developed casting techniques from around the world were introduced. A refined fitting process commenced in which both physio and prosthetist worked together to ensure that alignment, posture, limb position, and control were optimal. The training effect, supervision, and collaborative problem solving quickly became routine normal practice. The need for the Phoenix service was no more, it had evolved into everyday practice.

40.4.1 Measuring Progress

Goal setting and outcome measurement provided direction to the rehabilitation process. Long-term goals were broken down into short-term functional (admission) goals. However, specific admission goals also existed which enabled clinicians to tailor intervention around a particular purpose or activity that the patient wished to accomplish. Often these shorter-term goals focused on adventurous training, sports, or activities, which promoted personal development.

The Canadian Occupational Performance Measure (COPM) [68] was used with all patients, and this helped communicate the focus for each individual, across the team. Outcomes were measured to assess individual change both objectively and by patient report in almost every aspect of their care, e.g. physical, quality of life, and independence. No single outcome measure was able to cover the array of needs and conditions, therefore multiple measures were used. The IDT amputee toolbox (see Table 40.5) was an assortment of outcome measures which were developed from various sources, including work completed by the British Association of Chartered Physiotherapy in Amputee Care (BACPAR) [71]. Table 40.5 outlines the purpose of each outcome measure which was used, ideally, on a once-per-admission basis.

Whilst this formal clinical process suited most, there were some who struggled to see a way forward following injury. Where their grief, loss, social, or other psychological stressors prevented engagement in the formal rehabilitation process, an alternative approach was needed. This sometimes led to holding off on specific physical goals and admission planning, whilst services such as mental health and social work took the lead. Support from organisations specialising in adaptive sport, adventure training or activity/hobby-based training was helpful in providing a community away from the clinical space where they could process what had happened to them. This process often helped them to re-engage, at least in part with the formal rehabilitation process with a new focus in mind [74, 75].

Table 40.5 IDT Toolbox of outcome measures used with amputees

Outcome measure	Abbreviated name	Purpose	Resource
Canadian Occupational Performance Measure	COPM	Therapist administered Self-perception of performance and satisfaction in everyday living	www.thecopm.ca
Generalised Anxiety Disorder assessment—7	GAD7	Self-administered Screening and severity measure for generalised anxiety	Spitzer et al. [69]
Patient Health Questionnaire—9	PHQ9	Self-administered Monitors depression severity	Kroenke et al. [70]
Reintegration to Normal Living Index	RNLI	Self-administered VAS on daily function and self-perception	www.sralab.org/rehabilitation-measures/reintegration-normal-living-index
Special Interest Group Amputee Medicine	SIGAM	Self-administered Mobility grading	BACPAR [71]
Amputee Mobility Predictor Questionnaire	AMPQ	Therapist administered Functional status +/− prosthesis	Raya et al. [72] BACPAR [71]
6-Minute Walk Test	6MWT	Therapist administered Performance-based measure of functional exercise capacity	ATS 2002 [73]
Timed Up & Go	TUG	Therapist administered Measure of balance, sit↔stand and walking	BACPAR [71]
EuroQol 5 Dimension	EQ5D	Self-administered Perception of mobility, self-care, usual activities, pain/discomfort, anxiety/depression	www.euroqol.org
Disabilities of Arm, Shoulder and Hand	DASH	Self-administered Questionnaire on physical function and symptoms	www.dash.iwh.on.ca
Prosthetic Limb Users Survey—Mobility	PLUS-M	Self-administered Self-report mobility measure	www.plus-m.org

40.4.2 Periodic Intensive Residential Rehabilitation (PIRR)

The admission pattern developed into a programme of 3–6 weekly residential admissions, with the patient returning home in between these. This system enabled intensive blocks of residential rehabilitation with rest and recovery periods at home. Momentum was maintained by keeping the period of home leave relatively short during the first few residential blocks. This time of rest with family and friends was also an opportunity to practice skills in a real-life setting [1]. On re-admission, they could then highlight what had worked and what they had struggled with which directed their subsequent rehabilitation. As skills and physical capacity developed, longer periods were spent on leave away from rehabilitation. Individuals were able to use extended leave from rehabilitation to carry out return to work programmes or transitional courses. Admission planning was administratively complex, but it also improved the throughput of patients. On average, a bilateral amputee completed rehabilitation in 33 months and required 13 admissions (Table 40.1) [1]. Triple amputees had 12 admissions over an average period of 44 months [1, 2].

The residential nature of rehabilitation was known as the 'Headley Bubble'. New and challenging skills could be worked on in a rehabilitation environment amongst peers with similar injuries and clinicians who understood their situation. However, the 'bubble' contrasted dramatically with the home/local environment, which was not designed around the patient. Families and the public were not used to seeing these injuries, and they did not know how to respond or to support. Many patients struggled with this contrast. Community work carried out by OT, physiotherapy, and prosthetics, with input from social work and mental health professionals was frequently required to support this transition.

The rehabilitation programme ran weekly from Monday to Friday, starting at 08.15 and finishing at 17.00. Rest periods were scheduled according to need. Their weekly timetable (Fig. 40.5) was populated with appointments from all treating therapists and could be modified at any point to ensure it worked for all involved. Short notice appointments such as scans, revision surgeries, or external referrals were also scheduled and communicated to the team [35, 67]. The weekends were deemed rest periods, or home leave. Social activities were organised by the Military Liaison Officers or charities if the patients were not able to return home. These activities, whilst not part of the therapeutic effort, helped to encourage reintegration into society, built social confidence in different settings and required the patients to mobilise outside of the 'safe' DMRC environment.

<table>
<tr><th>Time</th><th>Monday</th><th>Tuesday</th><th>Wednesday</th><th>Thursday</th><th>Friday</th></tr>
<tr><td>0815-0830</td><td>Parade</td><td>Parade</td><td>Parade</td><td rowspan="2">Breakfast Club 08.00-09.00</td><td>Parade</td></tr>
<tr><td>0830-0900</td><td rowspan="2">Physio / Prosthetics</td><td rowspan="2">IP Team B</td><td rowspan="2">Physio</td><td rowspan="2">Long Leg Class</td></tr>
<tr><td>0900-0930</td><td>IP Team B</td></tr>
<tr><td>0930-1000</td><td></td><td rowspan="2">Prosthetics</td><td></td><td></td><td rowspan="2">OT</td></tr>
<tr><td>1000-1030</td><td rowspan="2">Glutes & Core</td><td rowspan="2">Horticulture</td><td rowspan="2">Swimming</td></tr>
<tr><td>1030-1100</td><td rowspan="2">Yoga</td><td></td></tr>
<tr><td>1100-1130</td><td rowspan="2">Driving Assessment</td><td></td><td></td><td rowspan="2">Circuits</td></tr>
<tr><td>1130-1200</td><td></td><td>CPN</td><td>SW</td></tr>
<tr><td>1200-1230</td><td>Lunch</td><td>Lunch</td><td>Lunch</td><td>Lunch</td><td>Lunch</td></tr>
<tr><td>1230-1300</td><td>Lunch</td><td>Lunch</td><td>Lunch</td><td>Lunch</td><td>Lunch</td></tr>
<tr><td>1300-1330</td><td>Lunch</td><td>Lunch</td><td>Lunch</td><td>Lunch</td><td>Lunch</td></tr>
<tr><td>1330-1400</td><td rowspan="2">Long Leg Class</td><td rowspan="2">Hydro</td><td rowspan="2">OT</td><td rowspan="2">Physio / Prosthetics</td><td></td></tr>
<tr><td>1400-1430</td><td></td></tr>
<tr><td>1430-1500</td><td rowspan="2">Wheelchair skills class</td><td></td><td>Ward Round</td><td></td><td></td></tr>
<tr><td>1500-1530</td><td rowspan="2">Physio</td><td></td><td rowspan="2">ERI 1:1</td><td></td></tr>
<tr><td>1530-1600</td><td rowspan="2">ERI 1:1</td><td rowspan="2">IP Team B</td><td></td></tr>
<tr><td>1600-1630</td><td rowspan="2">Rec Therapy</td><td rowspan="2">Rec Therapy</td><td></td></tr>
<tr><td>1630-1700</td><td></td><td></td><td></td></tr>
</table>

Fig. 40.5 Timetable of rehabilitation provision

Intervention was provided either one to one or in groups, depending on the activity or the patient's needs. So not only did they have individual physio, OT, ERI, social work, and mental health sessions but they could also participate in, for example, aquatic therapy, yoga, mindfulness, relaxation, breakfast club, wheelchair skills, horticultural therapy, prosthetic skills class (also separate classes for different types of prostheses), boot camp, and high-activity group. Incidentally, the breakfast club was an idea developed from RCDM. It was an applied practical session on food preparation, diet, and nutrition, learned whilst making their own breakfast. For each of these group activities, the clinical intent and education goal was multiplied by the military humour and group connection which marked each occasion. With this combination of sessions, they had the opportunity to learn not only from clinicians, but also from each other. The patient partnership and informal mentoring that occurred between patients were an essential element in their acquisition of skills, preparing them for the real world. The ethos was to 'try and see'. It allowed them to work out what skill or strength they needed to develop. They would then build this into their rehabilitation programme. Empowerment was the goal.

In addition to the above functional ethos, there remained core basic rehabilitation needs, which required a one-to-one approach. This included work on ensuring joint ranges were full, especially immediately proximal to the amputation, so that gait and function were not impeded. Control of posture, the trunk and limbs were equally vital and the physio and ERI would create exercise programmes specific to an individual's remaining musculature so that any required adaptations could be considered to minimise any undue stresses on joints and soft tissues. These exercise adaptations could then be built into the group and functional sessions to enable inclusivity and to minimise risk.

Alongside these rehabilitation strategies were the inclusion of work elements, both technical and sport related. The relationship between positive health and mental well-being and participation in work, sport, and fitness is widely recognised [76, 77]. The military ethos promotes these values. Many of the patients lived by this belief but they were unsure how this could now be achieved given their injury state. The OT was critical in bridging this gap. Light and heavy workshops for skills such as crafts, carpentry, and car mechanics all the way through to accessing the unit's Rehabilitation Squadron Workshop for access to technical engineering skills/re-training were possible. Battle Back was also developed as a joint military and charity initiative to reintroduce patients to the adventurous training activities they would previously have performed in their military role. It also introduced them to new sports, they may not have tried before. Skiing, snowboarding, shooting, fishing, climbing, and horse-riding were just some of the activities on offer. This was initially set-up alongside the rehabilitation process to ensure that it tied in with specific rehabilitation needs and involved a clinician attending the activity to ensure that rehabilitation goals/exercises were integrated into the activity directed by the Battle Back staff. Whilst there is now a separate Battle Back centre for wider adventurous training access to wounded, injured, and sick personnel, there is still a specific rehabilitation-led Battle Back service within the DMRC. It remains a central component in our arsenal, particularly supporting the psychosocial element of rehabilitation. The list of activities on offer from this and other charitable organisations is endless. Their success appears to centre on restoring identity and self-belief, however it is unclear if this occurs as a by-product of the activity, or the community formed through shared experience [75].

40.4.3 Prosthetics: Practical Lessons Learned

On transfer from QEHB to DMRC, amputee rehabilitation progressed rapidly. Early residual limb shaping had commenced with compression socks supplied at QEHB within days of final closure of the amputation and this practice was continued at DMRC. The PPAM-aid (pre-prosthetic amputee mobility aid) was also available at both QEHB and DMRC if wounds could tolerate it. However, as most patients were generally cast and fitted with a primary prosthesis within 1 week of arriving at DMRC, the team preferred to wait for the cast socket rather than to use the PPAM-aid and risk disruption to the wound or skin grafts. As the conflict progressed, unilateral amputees were by far the exception, further precluding its widespread use. If prosthetic fitting was delayed, then the PPAM-aid would be considered.

The management of the health of the residual limb was addressed, with joint range and proximal control also being a constant focus. Prosthetic sockets for traumatic amputees must offer a closer fit to improve control during high-level activities [78]. It was found that this was important from the very beginning, otherwise patients experienced problems with pressure marks and friction from loose fitting sockets. Socket comfort in the initial stages was therefore a relative term. Education and management of expectation with regard to socket use and fit were vital. Understanding the importance of skin integrity, use of socks, management of blisters, sweat and pressure areas were regularly reinforced. Peer support from fellow soldiers was critical, and over time more experienced prosthetic users also supported the clinical team by offering their own experience and tips to those still learning core skills.

This service built upon robust principles of rehabilitation, but framed within the context of our operational commitments, attracted enormous scrutiny from all sectors of society. This scrutiny placed a burden of expectation upon the shoulders of both patients and clinicians. To be able to walk following such catastrophic injury was the achievement many sought. It symbolised the bravery as the soldier faced up to the challenge with grim determination. It also elevated the skill of the clinician and the military to look after their own.

Ambulation rested upon the challenge of successful fitting of the prosthetic limb. Extensive soft tissue loss and fragmentation wounding meant that the surgeon had to be creative in their attempt to salvage a functional length of residuum frequently resulting in a residual limb with an unusual or unconventional shape. This was complicated by soft tissue loss, skin grafting (from split to full thickness), heterotopic ossification, painful neuromas, or recurrent pockets of infection. Avoiding pressure points and shearing forces in such areas meant that the prosthetic team could rarely use a ‘standard’ prosthetic prescription. Creative use of both old and new technology and materials resulted in a hybrid prosthetic solution in which carbon-fibre, silicon and endoskeletal modular components might sit alongside leather and even knee joint side steels. Supply of trial parts was critical in supporting creative solutions. Funding was made available to enable the trial and use of newly developed componentry. Manufacturers partnered in this creative provision.

Whilst the concept of an ‘ideal residual limb’ is questionable; traditionally, a trans-tibial residual limb should be 12–16 cm when measured from the medial joint line of the knee; and a trans-femoral residual limb should be 23–30 cm from the tip of the greater trochanter [1, 78]. However, one should also consider the proportional stature/height of the individual concerned. The ‘ideal’ length in someone with short legs may not leave enough room to accommodate high-activity prosthetic componentry. With regard to trans-femoral amputation, one of the key issues is the importance of muscle balance and attachments, either residual or through myodesis rather than myoplasty [78]. Ultimately, alongside the length and shape of the residuum, the more joints that remain and the more mechanically effective the muscle attachments are that cross the distal joint of the residuum, the more the rehabilitation process will be able to maximise body control and movement patterns to enable gait and function. Irrespective of this, in the blast injured amputee, the surgical team’s primary function was lifesaving after which they would seek to create the best residual limb with what was left. Experience has taught us that almost all casualties required revision surgery to improve the residual limb. For some, this issue arose early, and the prosthetist simply could not fit an effective prosthesis, or ongoing pain issues prevented progress. For others, 5–10 years post-injury, this cycle of revision is only now becoming apparent.

The socket is the key element of the prosthesis transferring forces between the residual limb and the prosthesis. These forces travel in all directions; perpendicular to each aspect of the residuum to enable the close fit of the socket and at oblique angles due to weight-bearing, suspension methods, and movement patterns. Whilst the perpendicular forces need to accommodate certain elements, e.g. bony points on the residuum, it is the oblique or shear forces that cause the most problems for amputees and prosthetists, especially where split skin grafts are present. These forces are most likely to cause rubs and skin breakdown, which can prevent the user wearing their prostheses and therefore delay rehabilitation. Whilst a larger surface area spreads load, the socket trim lines need to allow joint range of movement. The tendency for trans-tibial amputees is to start with a patella tendon bearing type socket and progress towards a total surface bearing type socket as symptoms/tissue tolerances allow. For trans-femoral amputations whilst sockets may start with a more quadrilateral style, they may move onto ischial containment, anatomical, and other styles that have been developed. There is a fine balance between comfort and function. Different materials work in different ways to maintain this balance. Carbon-fibre is light-weight and strong enough to accommodate cut-outs. Polypropylene is heat-mouldable and will allow frequent changes to fit. Foam and gel pads offer cushioning, or insertion of pads can increase loading in one area in order to off-load other areas. Since rehabilitation is a dynamic process, the choice of materials must allow for adjustment throughout the rehabilitation process as activity levels change and the soft tissues heal and adapt.

Suspension can be achieved by a variety of methods such as gel or silicon sleeves and liners, passive or active vacuums, pin-locks, straps, and belts. These can all be appropriate at different stages of healing and rehabilitation. Most commonly with the trans-tibial population, where wound/scar healing was ongoing we would start with a sock and Keasy® liner fit with a gel suspension sleeve. Once wounds were fully healed, gel or silicon liners could accommodate changes in limb volume, with the addition of terry or cotton socks. Suspension would be achieved by a passive vacuum (one-way valve and seal on liner) or pin-lock. These options generally allowed the greatest range of movement and offer cushioning whilst still enabling good control of the prosthesis. Silicon liners also had the added benefit of giving ongoing surface conditioning of the scar tissue [79]. Sweating of the limb within a liner can be another issue. Most individuals found that this settled over time, or they learned to use a variety of management strategies such as spare socks, sweat wicking socks, antiperspirants, or just regular removal and cleaning of the liner after exercise.

There are a huge number of prosthetic joints and feet/hands on the market (see Chap. 41). With this generation of technologically aware individuals, we found our patients were keen to try everything. Many components could simply not cope with the loading and intensity of use, they were put through. When assessing componentry, the term 'squaddie-proof' reflected our need to source components that were capable of moderate to high impact/activity.

Microprocessor knees (MPK) were used with the bilateral lower-limb amputees from the very start of the complex trauma prosthetic service as the programming and stability they offer greatly improved the amputee's stability and decreased their risk of falling. The confidence they gave users, enabled them to focus upon achieving optimal gait patterns through greater control and physical conditioning [35, 80]. Upper limb prosthetic options included body powered and myoelectric systems as well as high-definition cosmetic limbs. Again, choice often came down to function and durability. Resilient body powered prosthetic limbs were preferred due to their ease and functionality when compared with early myoelectric systems.

Once fitted with prostheses, an individual's prosthetic rehabilitation would move to the next stage. Firstly, we sought to remove the fear of falling by teaching and practicing how to fall and how to get up following a fall (referred to as 'falls training'). This built confidence and skills enabling them to progress out of the parallel bars quickly and safely. Use of elbow crutches was discouraged unless other injuries required it as crutches alter the gait pattern changing the prosthetic response. However, walking sticks were utilised and wheelchair↔standing skills were also encouraged to enable freedom from the parallel bars.

Social media was also a valuable teaching resource. This resource was discovered by the patients, who shared videos of more experienced amputees performing challenging exploits (e.g. hands-free stair descent, steep slopes). There were some negative effects if an individual was struggling either physically or psychologically. The apparent public/social assumption that military amputees would want to be Paralympians demotivated as many less confident patients as it inspired others. Managing their confidence and self-belief was important to maintain their engagement in the process. Peer support was vital providing what the military term 'buddy buddy' support. At other times, support also needed to be formalised. This might not always come from mental health professionals, instead it came from the member of the team in which the patient had greatest trust. However, it was important that this member of the team had support themselves and knew when to bring in the mental health professionals.

40.4.4 Prosthetic Rehabilitation Programme

As patients progressed through their physical rehabilitation, clinicians saw what worked and what did not. A more formal programme was developed which placed existing intervention into phases of recovery. These phases were purposely ill-defined, and progression was decided by consensus of the team rather than specific criteria. Figure 40.6 outlines the broad structure. The core element included a personalised strength and conditioning programme. One-to-one physio targeted joint range and muscular control or specific technique work, whilst the

<table>
<tr><th>Phase 1</th><th>Phase 2</th><th>Phase 3</th></tr>
<tr><td colspan="3">CORE
• One to one: Physio, ERI & OT
• Group Exercises Classes: ERI
• Individual Exercise Programme: ERI & physio
• Community Access Programme (work & leisure): OT, recreational therapist</td></tr>
<tr><td>• Stubbies Class (bilateral skills) (physio)</td><td>• Long Legs Class (beginner bilateral prosthetic knees) (physio & prosthetist)
• Boot Camp Class (advanced bilateral prosthetic knees) (physio & prosthetist)
• Unilateral Class (Unilateral skills) (physio)
• Pre-running/impact skills (physio & ERI)</td><td>Advanced Prosthetic Rehabilitation Programme
• Running Skills Class (physio & ERI)
• Boot Camp Week (Advanced bilat eral skills/endurance) (physio & prosthetist)
• High Activity Group (Advanced unilat eral sport/endurance) (physio, prosthetist & ERI)</td></tr>
</table>

Fig. 40.6 Stages of sub-acute rehabilitation provision at Headley Court

community programme focussed on work and leisure interests.

Group therapy was used to cluster patients of a similar ability or need. Each group trained a specific component of gait or function. Groups ranged from beginner and lower level skills training which aimed to build confidence and muscular control, to advanced level skills for those involved in higher load activities which demanded control and endurance. Bilateral prosthetic knee users require different skills to unilateral prosthesis users, so different pathways were created for each. The group ethos provided peer support and informal mentoring/teaching, as well as a competitive element. This worked well alongside input from the clinician. Both clinicians and patients were able to learn together and try things that were not in any textbooks. This programme was underpinned by the following common aims to:

- optimise biomechanical efficiency during gait in order to limit excessive loading upon remaining joints;
- practice skills and develop endurance whilst using prosthetic limbs to enable advanced community access over any terrain and conditions;
- provide where possible, the motor skills, physical conditioning, and educational understanding to enable higher impact forms of prosthetic use, such as running and other sports;
- educate the user on simple prosthetic self-maintenance, physical preparation and healthy living in order that the user may continue to maintain high functional levels in the longer term; and
- introduce prosthetic users to adapted sport, adventurous training or higher activity leisure pursuits which facilitate their health and well-being into the future.

40.4.4.1 Bilateral Amputees

Primary Gait Training. Bilateral amputees typically struggled when they were immediately put onto articulated knees. Stubbies (short, non-articulated prostheses) were therefore used initially (see Fig. 40.7). On Stubbies, the patient was quickly able to mobilise and achieve independence. Confidence, balance, and endurance were the result. 'Stubbies classes' practiced the specific skills needed when using this type of prosthesis. Stubbies could be lengthened as the patient grew in confidence. Before a patient could progress onto articulated knees, they must show they could spend a full rehabilitation day on stubbies. This ensured they had the socket tolerance, fit-

Fig. 40.7 Graphical examples of rehabilitation activity at Headley Court

ness and control needed to successfully manage articulated knees.

Long Legs Class. This class commenced alongside one-to-one sessions once the patient had moved onto articulated knees and progressed outside of the parallel bars. Akin to the stubbies class, it centred upon functional activity and skills (stairs, slopes, sit to stand) and it worked on the same construct of clinician—peer learning—confidence building. The class was jointly staffed with a physio and prosthetist. Control work and advice, fine-tuning of prosthetic alignment and programming could be accomplished and reviewed with a number of patients in a single hour. It also provided time for clinicians to teach participants the full capability of the microprocessor knee (Figs. 40.8 and 40.9).

Fig. 40.8 Graphical examples of rehabilitation activity at Headley Court: Learning to use the knee lock

Fig. 40.9 Rehabilitation activity at Headley Court: Practicing prosthetic control on slopes (on variable surfaces)

The Boot Camp Class. This was a more intense progression of the Long Leg Class. It utilised the skills already learnt in outdoor and challenging environments, but it added the element of problem solving, endurance, and risk. Due to its challenging nature and physical effort, those who took part needed to demonstrate independent mobility on articulated knees for at least half of a rehabilitation day.

Advanced Prosthetic Rehabilitation Programme. This phase sought to prepare patients for high load, high endurance activities. The Boot Camp Week was a central component of this phase. It was a physio-led 1-week course designed to incorporate functional skills, prosthetic, and exercise education and community-based activity classes. It aimed to prepare the amputees for life outside of DMRC and formal rehabilitation. There were no more than six individuals per course with a minimum staffing ratio of one staff member to two amputees. The clinical team included both physio and prosthetist. The majority of classes were community based to ensure a 'real-life' element to all skills work (Fig. 40.10). Functional strength, control and endurance, confidence, and problem-solving skills were practiced in various environments, on and off public transport and even around the Tower of London. Seminars were also provided on long-term health risks, nutrition optimisation, prosthetic maintenance, and training methods for running. Those who had the biomechanical control and physical capacity to run were also able to join the unilateral group during their 'High Impact and Sports training'.

Fig. 40.10 Advanced prosthetic rehabilitation: Steps without a handrail

40.4.4.2 Unilateral Amputees

Unilateral Skills Class. This class mirrored the long legs class, and embraced the same ethos, however the focus was distinct. Trans-tibial and trans-femoral prosthetic skills were trained in an effort to gain symmetrical loading through both prosthesis and their remaining limb.

The High-Activity Group. This physio-led 1-week course sought to help unilateral amputees to progress further with high-impact sports. It also included additional health, dietary advice and education alongside sign-posting access to local sports clubs and para-athletic contacts for ongoing advice and coaching opportunities as individuals left the military.

High-Impact Sports. As the patient cohort progressed, many wanted to return to sport/running. The risk of secondary injury due to the altered skeletal mechanics and muscular control patterns, especially to the spine, hip, or the sound limb, was a concern to the clinical team. A specific rehabilitation programme with pre-impact and running criteria was developed from both amputee and sports injury evidence, along with liaison with our US military rehabilitation colleagues. Training activities were carried out on everyday prosthetic feet to ensure that each individual amputee was motivated and capable of running safely prior to being fitted with a specialist running prosthesis.

40.4.4.3 Other Considerations

Hip Disarticulation/Hemipelvectomy Amputations. Few individuals experienced amputation at these levels during the conflict, but those that did were all bilateral amputees. The rehabilitation construct discussed here was still directly applicable to them, however, they required some further adaptations due to the unique challenges that they faced. In particular, different socket options were tried in order to achieve the best possible combination of comfort and function. Unsurprisingly, regular wheelchair use was the norm. Specific wheelchair training was provided to help them maximise independence as well as to preserve upper limb function.

Triple Amputees. Bilateral lower-limb amputation together with an upper limb amputation created a unique challenge for the prosthetist. Early fitting of the upper limb prosthesis was vital if it were to be used during daily activity [13]. A focus upon shoulder mobility and spinal rotation was also deemed extremely important. In all stages, upper limb sockets were designed around the demands of the rehabilitation environment. For example, in the early stages, upper limb prosthesis loading in the parallel bars or walking with an aid might be required to enable good gait patterns. 'Gym' arms were also created with various hooks and bolts attached to enable use of specific pieces of gym equipment.

40.4.4.4 Osseointegration/Direct Skeletal Fixation (DSF)

Whilst there is no NHS England clinical service available for trans-femoral osseointegration at present, there is an NHS England and MOD collaboration to carry out a clinical evaluation of the concept using LIBOR funding for both the surgery and rehabilitation process.

Afghan and Iraq conflict amputee veterans who have failed on sockets, and thus prosthetic limbs, due to soft tissue or mechanical problems are able to be referred for consideration to have DSF. At the initial assessment clinic, the following elements are covered:

- their hip range of movement is checked to ensure they have at least neutral hip extension, as flexion cannot be built into a DSF prosthesis as it can with a socket prosthesis;
- their postural control is reviewed to ensure that they have the spinal and hip ranges, strength and control to stand upright, whether on sockets or DSF, as this can sometimes be limited in those with other complications such as nerve or brain injury;
- their body mass is checked, as the weakest component of the DSF has a weight limit of 100 kg;
- their agreement to partake in a post-op rehabilitation programme is also confirmed, to ensure the greatest likelihood of success of the procedure; and

- baseline outcome measures are documented to enable monitoring of progress/problems over time.

If limitations are identified but deemed correctable an exercise programme is developed for them to work on. They are then referred to the surgeon to obtain any further recommendations regarding limiting factors or to adapt and add to the surgical plan to ensure the best possible outcome. Following agreement by the surgeon and achievement of the criteria, the surgery is planned, and the rehabilitation programme booked in.

DSF rehabilitation follows the Australian protocol [81] regarding bone healing, patient reports of bone pain should not go above <3/10 Numerical Rating Scale (NRS). Walking aids are also necessary in the early stages, but we have had to make some modifications for use with bilateral and triple amputees to limit the risk of falls and increased rotational forces during the osseointegration healing time. Stability and hip-trunk control are key to avoiding rotational forces and falls, however, when there is no sound limb to stabilise with it is almost impossible to avoid rotational forces when turning or transitioning from sit↔stand. An initial prolonged period on stubbies has therefore been added to the protocol prior to progressing to articulated MPKs. During this first stage, the patient can build confidence and proprioceptive awareness on the DSF, alongside ensuring they have sufficient trunk-hip control and dissociation skills to safely achieve tasks. Once achieved, they can then progress to taking on the challenge of increased height and decreased stability of the MPKs.

Basic Programme (subject to alteration and time delays dependent on symptoms and control):

1. Increase static weight-bearing in hospital from 20 kg to 50% body weight in daily increments (dependent on bone density and surgeon's guidance this will be approximately a 2-week period minimum).
2. Commence prosthetic partial weight-bearing for a 6-week period; progressing from parallel bars to two crutches to protect bone integration process. Training on using a step-to gait on stairs and navigating mild slopes.
3. Progress to partial weight-bearing with one crutch for a 6-week period and progress control work accordingly.
4. Progress to full weight-bearing at 12-weeks (or when signed off by the surgeon) and commence gait re-ed on slopes and trial step-over-step on stairs.

Bilateral/Triple Programme (subject to alteration and time delays dependent on symptoms and control):

1. Increase static weight-bearing in hospital from 20 kg to 50% body weight in daily increments (dependent on bone density and surgeon's guidance this will be approximately a 2-week period minimum).
2. Commence prosthetic partial weight-bearing on stubbies for a 6-week period: progressing from parallel bars to two walking sticks to protect bone integration process. Training on navigating mild slopes. To avoid stairs where possible but training carried out if this is unavoidable.
3. Progress over the next 6-weeks to partial weight-bearing with one stick on stubbies AND commence MPK fitting and gait re-education in the parallel bars, progressing to two walking aids as able (or one crutch for the triple amputee).
4. Progress to full weight-bearing at 12-weeks (or when signed off by the surgeon), ONLY on prostheses where enough control is demonstrated to be safe.

Throughout this process for all amputees, the prosthetist will regularly tighten the DSF components to specified torque settings, as the dual cone will settle over time and this can give both the sensation of looseness or 'clicking' and the increased risk of component rotation if it is not re-tightened. Once the cone has settled, there should be no further need to continue with regular tightening.

It is not recommended that anyone run or participate in high-impact sport with DSF, nor are they advised to open-water swim where there may be an increased risk of infection via the stoma. The maximum weight limit for the DSF componentry is 100 kg, therefore if an individual wants to be able to lift and carry things, for example shopping or their child, they must ensure that they stay well under this weight limit to accommodate the increase when they are carrying something.

Those who have opted for DSF, due to their inability to wear sockets, have reported being much happier and more functional after the procedure [82]. Outcome measures have improved overall, both subjectively and objectively. Whilst some bilateral and triple amputees primarily use their stubbies rather than articulated legs, DSF provides much greater stability and a sensation of safety whichever prostheses they choose. One veteran has even built his own house whilst wearing his stubbies.

However, there is a lot which is unknown, and the stakes are potentially high. The lesson from this period in our history is that we must continue to work across disciplines and in proximity with one another. Where surgical skills are dislocated from the rehabilitation space, there is little opportunity to learn and evolve practice. The outcomes are also unlikely to be optimal.

We know that for outcomes to be optimal, rehabilitation must be intensive and comprehensive. This is where the surgical link is so important, especially at a time when so much is uncertain. The surgeon will have an idea when and how to apply this load to facilitate recovery. The rehabilitation specialist will apply this knowledge and together with the patient, they will test the boundaries of the possible. Reflections from the resulting outcome must then be fed back to the surgical team so that product and process can be refined.

40.4.5 The Challenge of Transition

Rehabilitation and recovery are both a process of return to function and a discovery of a new landscape of opportunity. This return to function may not be sufficient for the individual to resume the life and social role that they once enjoyed. The process of transitioning into a new role and life is complex and multi-faceted. The broader sociological landscape and more specific social psychological research base are poorly appreciated by policy makers and clinicians. As recovery progresses, so must the function and role of the rehabilitation team. It must also reach out to those agencies equipped to support this journey. Integration of medical rehabilitation and transitional support agencies is critical if we are going to support the casualty in the final stages of their recovery.

As the question ‘what next’ became more important to patients, the vocational occupational therapy (Voc OT) service moved centre stage, supported by the social worker. Social work facilitated home, social and community access, whilst the Voc OT sought to bridge the gap between rehabilitation and the military’s transition provision. Above all, however, this was a psychological journey. Most, if not all had to confront their grief at what they had lost to see what they could now gain. Expectations and aspirations, at times needed to be adjusted. This was a process that started at the point of injury. Formally, this component of the role would fall to mental health, but in practice it rotated through the team, and rested with those most trusted by the patient.

The Voc OT service operated slightly differently to the rest of the team. The initial assessment identified both short- and long-term vocational goals. Achievement of these goals was a collaborative effort in which the patient would complete homework tasks between admissions. The degree of engagement was an indication of the patient’s readiness for vocational rehabilitation. Education reviews with the military learning development officer would help direct the individual towards potential courses that may benefit them and gave them access to funding. Military graded return to work programmes were as useful when they failed, as when they succeeded, with respect to an individual's ability to move forward and consider a life and career outside of the military. The provision of resettlement education and collaboration with associated agencies provided support with, for example, CV preparation and interview skills. Participation in work-experience placements was particularly effective in measuring performance capacity, building confidence and identifying transferrable skills in preparation for future employment. The promotion of educational opportunities that included vocational courses as well as academic opportunities provided the breadth required.

As the conflict continued, Personnel Recovery Units (PRU) were developed to improve the support to wounded, injured and sick individuals. Each patient had access to their own Personnel Recovery Officer (PRO) and a clinical facilitator. These roles supported the administrative process of recovery. As commented earlier, close liaison between these individuals and the rehabilitation team at DMRC maximised the benefit the patients could gain from this collective input.

Moving from soldier to civilian is an enormous cultural leap [83]. In the CBIS study cited throughout this chapter, veterans repeatedly referred to their experience of transition as a 'cliff edge'. They cite the personal challenge of adapting to 'normal life' from which military service has largely shielded them, such as the demands of paying bills, renting, or buying property, and finding and holding down a job. More significantly, they lost their military community and the clarity provided by the system of rank and daily routine. Above all, the uncertainty of their future healthcare provision and difficulty accessing what they feel they need remains their principal concern and prevents some from achieving the quality of life they seek.

Mental health challenges and complications were expected given the nature of these conflicts, with the potential of psychological trauma from injury, death of colleagues, near-death experience, disfigurement, and perceived disability. During the rehabilitation process, there was a low incidence of reported psychological morbidity [1, 34]. Casualties were, however, on a rapid and time limited road to recovery. There is growing evidence that mental health issues may lie dormant and only later start to manifest. It is also widely recognised that complex PTSD requires a unique approach that differs from other forms of counselling [1].

It is also clear that rehabilitation does not cease when a casualty is medically discharged. Whilst the requirement is less, ongoing surgical and prosthetic intervention may be required, and associated rehabilitation to manage these interventions is expected. In addition, as many age and report that their mobility and walking proficiency is declining, top up rehabilitation would seem appropriate, however, its provision is less forthcoming reflecting issues of access, capacity, and capability of civilian services to meet the needs of this population. Not surprisingly healthcare provision for veterans remains a source of great debate.

40.5 Conclusion

The functional outcomes for the Afghanistan blast and combat injured amputee cohort were extraordinary [2, 35]. Carbon sockets, microprocessor knees, and running blades are images often used to celebrate this achievement. It is therefore assumed this progress was driven by the technological advance in prosthetics. However, comparison of these outcomes with age and prosthetically matched groups supports the proposition that it was the comprehensive and intensive nature of the rehabilitation process, and not prosthetic technology, which was the significant factor [35].

For the rehabilitation to have been 'comprehensive', the process must have embraced the physical, psychological, and social impact of such injuries and reflected these in the composition of the clinical team. For the rehabilitation to have been intensive, it must have been scaled and resourced to ensure sufficient quantity of provision, load, and momentum. For the rehabilitation provision to have been both comprehensive and intensive, it must have been intricately planned and coordinated.

The description and analysis within this chapter have shown how this rehabilitation provision evolved in the hospital and military rehabilitation setting. It provides many lessons and practical tips for clinicians. Above all, however, it shows that rehabilitation, which seeks to restore a life saved, to enable this life to be lived, is about planning, coordination, and resourcing of this moral undertaking.

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41 Prosthetics and Innovation

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Abstract

Primary and secondary limb amputations are a common consequence of blast-induced extremity injuries. The resulting limb loss can lead to a severe decrease in quality of life, affecting both the social and mental status. The severity of this problem is historically well recognised, prompting the development of more primitive prosthetics. The last decade has seen rapid technological advancements, with significant impact in the dexterity of bionic limbs. The current market offers highly functional upper and lower limb systems with a variety of design options tailored towards different needs. However, the improvements in the functionality of bionic hardware are not yet matched with the quality of control offered to the users. This results in high device abandonment among the amputees. Here we provide a technical overview of the prosthetic solutions, reflect upon the clinical and user challenges, and indicate the most recent clinical and technical advancements made in the field. We also identify the most promising technologies that are likely to have a strong clinical impact and constitute the basis for the next generation of bionic limbs.

41.1 Introduction

A missing limb is a dramatic impairment that leads to a staggering decrease in quality of life. Amputees face a decreased capacity to move and to interact with the environment. This deficiency is associated with both the functional loss of a body part and loss of sensation.

In the latest conflicts in Iraq and Afghanistan, blast-induced extremity injuries account for over a half of combat wounds [1, 2] and a number of them resulted in either primary or secondary amputations. Similar outcomes have been observed in civilian population following terrorist bombings [3]. Overall, there is an estimated 1.6 million people living with a limb impairment [4]. The impact of the limb loss was recognised centuries ago [5], and the notion of a practical artificial substitution has been pursued ever since. However, actuated alternatives to rigid and passive systems with simple hinge joints appeared only at the beginning of the twentieth century. The Bowden cable-driven hook featured such an efficient design that it is still widely present on the market [6]. Following the Second World War, electrically actuated hand prostheses were introduced, however, they remained heavy in com-

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parison to simpler solutions [7]. In 1960, the Russian Hand was the first system to fully leverage the emerging transistor technology [7] and in 1965, the Belgrade Hand [8] was developed as the first multi-articulated fully wearable hand prosthesis. Since then, prosthetic design has gradually evolved following the advances made in manufacturing and material technology. Recently, the first fully 3D printed bionic hand has demonstrated conformity with relevant European regulations (CE marked) [9].

While the advancements in bionic limb hardware have been substantial, the dexterity and complexity of these systems contrast with the relatively poor control currently offered to the users [10]. Most of the commercial hand prosthetics rely on control strategies that utilise demanding switching algorithms in order to deliver more than a single function and offer minimal sensory feedback [11]. At the same time, the majority of the fitted bionic legs operate using mechanical sensors [12, 13] and as such do not include volitional control.

Because of the limits in control capacity, prosthetic rejection rates range from 25% up to 50% [14, 15] of all upper limb users and from 4% up to 11% [16] for lower limb users depending on the age and the level of disability. To bridge the gap, a significant body of research has been conducted and recent advances in man/machine interfacing involving advanced signal processing [17, 18] and invasive neuromuscular recordings [19–21] show promising results.

In this chapter, we will provide an overview of the prosthetic hardware and will introduce a number of control approaches ranging from commercially available ones to current academic state-of-the-art systems. We will also discuss sensory feedback and surgical techniques that offer improved prosthetic solutions. Finally, we will summarise the challenges and provide an overview of the most promising technologies in future bionic limbs.

41.2 Prosthetic Hardware

41.2.1 Upper Limb Prostheses

The most elaborate bionic arms, similar to natural ones, feature three major joints (shoulder, elbow, and wrist), and an end effector that aims to substitute the function and appearance of a biological hand. The design of these prosthetic parts is usually a compromise between the provided dexterity and the achieved aesthetics, and it depends on a number of factors such as price, weight, size, user needs, and looks.

Cosmetic prostheses are usually a choice of those users who are in need of a highly aesthetic solution. Because they do not replicate function, these prostheses can replicate to a remarkable extent the appearance of biological limbs. Moreover, they can feature non-biological traits and thus serve as a fashion accessory or even a personal expression. They are also lighter in weight and thus can often be fitted to children or users of small stature who are otherwise unable to manage heavy active prostheses. However, state-of-the art cosmetic systems, featuring top of the line materials, are often individualised works of art and, accordingly, they have considerably higher costs than their functional counterparts.

Body-powered systems usually require a wearable garment, such as a harness, in order to pick up the auxiliary body motions used for operating a prosthesis. By relying on the voluntary activation of the joints proximal to the amputation, the exerted movement is picked up by the harness and converted into the translational sliding of the Bowden or similar cables [22]. The sheeted cables then act through a set of pulleys to eventually activate the targeted prosthetic joint, effectively allowing control of a remote degree of freedom by a more proximal one. In the case of upper limb systems, the prehensor would commonly feature a spring-loaded mechanism that assists force delivery. The body-powered devices suffer from more frequent breakages than other systems [23]. These include teardowns of cable sheets and orthopaedic cable clamps, as well as rupture of the cables themselves. Although these breakages are mostly minor and cheap to repair, the downtime per incident is still a few days on average, which is a source of dissatisfaction for frequent users [24].

The market of electrically actuated upper limb prostheses has expanded its offer in the last two decades. An increased number of actuated degrees of freedom (DoFs) and lighter design options mainly contributed to this. Over the last 5 years, all major prosthetic manufacturers included into their portfolio multi-actuated prosthetic hands capable of performing various grips or even moving individual digits. Single DoF grippers are still the most commonly sold devices and they essentially rely on a single motor that adjusts the aperture of the prehensor according to a pair of input signals (for opening/closing). Prosthetic wrists can be divided into various systems; purely passive ones require manual positioning or work in fully compliant mode, active motorised solutions primarily targeting pronation and supination, and a combination of the two. The higher level prosthetic joints occupy a rather small portion of the market. Prosthetics elbow joints are available in both passive and active mode while the DEKA Arm HC (DEKA Research & Development Corp, Manchester, NH, USA) is currently the only approved motorised shoulder [25]. Most commercial elbows and shoulders feature a free swing mode that allows the users to have a spontaneous arm swing during walking. Electrically powered systems experience fewer malfunctions per year then their body-powered counterparts [23], although the repairs tend to be costly and take considerably longer to complete.

41.2.2 Lower Limb Prosthetics

Passive and body-powered lower limbs have been available for centuries. Since the first commercial microprocessor knees have come to the market, these systems have become the clinical standard for transfemoral amputees. In these prosthetics, the microprocessor is used to continuously monitor knee position, time, velocity, forces, and moments, and accordingly adjust the resistance and/or drive of the joint. In non-motorised solutions, the resistance for swing or stance is adjusted using different cadence control mediums (pneumatics, hydraulic fluid, or magnetorheological fluid). In contrast, powered knees offer active flexion and extension by precisely controlling the knee angle.

Unlike the prosthetic knees, prosthetic ankles and feet are predominantly passive and are based on the energy storing and return design principle [26, 27]. For this purpose, these systems incorporate a version of the J-shaped ankle combined with the heel-to-toe footplates. Commonly, they are built out of carbon graphite materials in order to increase energy return. Moreover, most of them feature a split toe or a similar mechanism for ankle mobility. Commercial motorised ankles rely on a microprocessor to sense specific events of the gait cycle in order to actively adjust ankle position. Moreover, they are able to provide additional energy when walking on inclined and uneven terrains [28].

41.3 Prosthetic Control

Control of prosthetic limbs can be achieved using a variety of solutions depending on the level of amputation and the available prosthetic functions. As previously mentioned, functional prosthetics can be divided into body-powered and motorised systems. For the later ones, the human neuromuscular system can be probed directly by interfacing either the brain, nerves, or muscles or indirectly by sensing the kinematics of available anatomical structures. Depending on the sensing method, the recorded signals are further processed in order to identify their prominent characteristics (features) so that a set of control signals can be mapped to the targeted prosthetic joints. Most of the commercially available upper limb motorised systems are based on muscle interfacing through electromyography (EMG). For the lower limbs, control mostly relies on kinetic and kinematic sensing.

41.3.1 Body-Powered Systems

Body-powered prostheses rely on the user's own body motions to operate available functions. These systems translate peripheral or residual motion into a dedicated prosthetic action through a set of pulleys, cables, and/or hinges. The amputee wears a harness that converts a dedicated intact joint motion (e.g. elbow flexion/extension) into a prosthetic joint movement (e.g. prehensor open/close) through a system of pulleys and Bowden cables (Fig. 41.1a) [29]. Body-powered lower limb prostheses rely on hinge-like artificial joints that enable free swing once sufficiently powerful and properly articulated motions are produced by the residual joints. These are simple yet effective devices, which offer reliable support in a straightforward manner. Users conducting intense manual activities on a daily basis currently mostly rely on body-powered grippers [24, 30].

Depending on the nature of the limb impairment, a surgical intervention in conjunction with a suitable body-bowered prosthetic attachment might be offered. As an example, rotationplasty (Fig. 41.1b) allows for the proximal transfer of the ankle joint, which following a rotation can serve as a knee substitute and thus allow voluntary gait restoration [31]. Similarly, although currently abandoned due to a number of complications, cineplasty relies on a transmuscularly implanted ivory rod that links contractions of proximal muscles to the prehensor aperture [32].

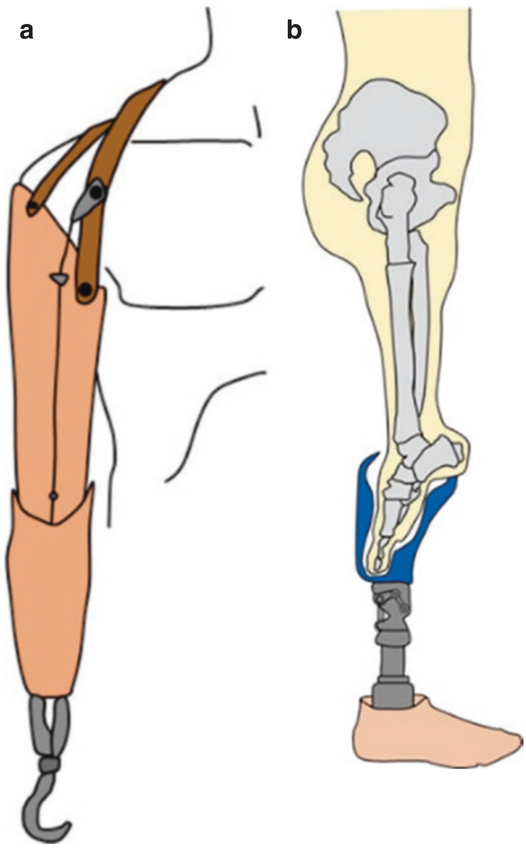

Fig. 41.1 Body-powered prosthetic systems. (**a**) Cable-driven hook opens and closes when the user shrugs the shoulders as the motion is being picked up by the harness and transferred through a set of Bowden cables to the prehensor. (**b**) Rotationplasty allows for the transfer of the ankle joint, which, following a rotation, acts as a knee and enables voluntary gait restoration

While the cost of body-powered prostheses is lower than that of externally powered systems, the level of comfort as well as potential functionality is somewhat limited in comparison. Similarly, body-powered systems tend to be lighter in weight but, at the same time, they require more effort to operate and exertion of larger forces due to the mechanical disadvantage. This also further reduces the loading capabilities of body-actuated prosthetics and as such limits their usability.

41.3.2 Commercial Actuated Prosthesis Control

The most common powered commercial hand prosthesis control methods are based on sensing electrical activity of the underlying musculature [11]. During muscle contraction, electric potentials in the range of 50 μV–10 mV are generated [33]. These EMG signals comprise the superimposition of motor unit action potentials (MUAPs) that have been evoked by action potentials of the motor axons innervating the muscle [34]. For the control of motorised prostheses, the bipolar derivation with two active electrodes placed on the surface of the skin is commonly used for sensing the underlying EMG activity. The recorded signals are then further processed in order to extract the desired control signals.

Commonly, myoelectric hand prosthesis relies on two bipolar EMG derivations from a pair of antagonistic muscles or muscle groups available at the residuum. In transradial amputees, the wrist flexors and extensors are typically monitored, while biceps and triceps brachii are targeted in transhumeral cases. In the former case, the EMG activity of the flexor muscles is mapped to a closing function of the prosthetic hand, while the activity on the extensor side is associated with hand opening. The majority of the devices provide a proportional control, i.e. the stronger the muscles contract, the faster the prosthesis moves, or a greater grip force is delivered. This control approach, often referred to as direct control, was already present in the first commercial myoelectric hand back in the 1960s [35]. To control more than one DoF with only two EMG channels, several heuristics have been developed and are in commercial use [36]. The most common switching method between DoFs is the co-contraction-based triggering—the user contracts both muscle groups at the same time, and the active DoF changes to another function of the prosthesis, e.g. from grasping to wrist rotation. In cases when only a single EMG channel can be obtained (e.g. because of excessive scarring, muscle damage), the direction of a DoF is selected either by the initial slope of the signal or by its absolute level. In extreme cases, only a single DoF direction is actively controlled (e.g. hand open), while the other one is automatic (e.g. hand closes in absence of EMG activity).

In the past few years, two companies have introduced pattern recognition-based upper limb myocontrol to the market. Using a number of EMG electrodes, this controller aims to associate natural patterns of muscle activation to specific actions of the prosthesis through supervised learning [37]. Previous laboratory research has shown that high levels of accuracy can be achieved in this way across a number of motions [38–40]. However, translation of this approach to routine use has been complicated by the challenges of surface EMG detection. EMG signals vary greatly because of a number of influencing factors, such as electrode shifts caused by socket donning and doffing, skin conditions, and muscle fatigue [41]. These factors of variability decrease the performance of pattern recognition systems. Nevertheless, current pattern recognition systems have shown acceptable robustness, resulting in some cases in acceptance rates >70%, with most rejections not being related to the control itself [42].

In contrast to predominantly EMG-controlled commercial prosthetic hands, commercial lower limb prosthetic systems primarily rely on other sensing modalities in order to provide active control of the available joints. These include measurements of inertia, joint position, pressure, and torque. Based on all or some of these parameters, the damping or the swing phase of the active joints is automatically adjusted. At the knee level, the resistance for swing or stance phase is maintained using different cadence control mediums (pneumatics, hydraulic fluid, or magnetorheological fluid) [43]. The sensors within the knee unit or pylon observe the applied loads and forces, so that the processing unit can adjust the flow of fluid for the required knee stability. Similarly, powered knees use the same information in order to actively adjust the knee position and prescribe a trajectory for the knee joint of the prosthesis [44]. In the same way, active ankles and feet adjust their properties based on the accelerometer derived gait cycle events. Moreover, using the same sensors, they can derive the terrain features, and thus permit active plantarflexion or dorsiflexion so as to increase the single limb balance [45].

41.4 Research Oriented Actuated Prosthesis Control

41.4.1 Regression-Based Continuous Control

To overcome some of the limitations of EMG classification approaches, regression techniques have been investigated [46]. Like classification, regression is a machine learning technique and needs some calibration data to be trained. However, regressors do not estimate a specific class (movement), but rather provide a continuous physical value (e.g. speed, position, force) for each DoF individually. In this way, the proportion of activation can be controlled independently for each DoF, allowing to, for instance, close the hand quickly while simultaneously slowly flexing the wrist.

A number of regression techniques has been investigated demonstrating successful control of multiple DoFs using artificial neural networks [47], autoencoders [48], non-linear kernel-based support vector regression [49], and matrix factorisation techniques [50, 51]. However, most of these approaches have been tested only in laboratory conditions, so that currently there are no commercial systems employing regression algorithms for prosthesis control.

The discrete nature of classification approaches leaves users unaware of why the misclassification has occurred. In contrast, regression approaches, with their unbounded solution space, allow for implicit adaptation of the user [52]. They can therefore actively compensate for incorrect estimates in non-ideal conditions [53, 54]. Recently, it has been shown that this allows for an increased robustness against non-stationarities, such as changing arm position or donning and doffing the prosthesis [55]. However, the simultaneous and proportional control obtained by regression increases the risk for unwanted DoF activations.

41.4.2 Control Systems Based on Invasive EMG Systems

In parallel with signal driven machine learning approaches, it has been shown that simultaneous and proportional control can also be achieved through musculoskeletal modelling [56–58]. The model-based estimation of joint moments and joint torques using muscle activations has been successfully demonstrated in both upper and lower limb prostheses [59, 60]. Forward musculoskeletal models incorporate the physiological and biomechanical constraints in order to estimate natural limb motions.

Alternatively, surface EMG can be decomposed into individual motor unit action potentials by blind source separation (BSS) algorithms [61]. The behaviour of motor units can then be mapped into control signals. This approach has been demonstrated to deliver superior control in the laboratory environment in a number of upper limb amputees [17, 62] although its real-time application is yet to be evaluated.

The previously underlined challenges of surface EMG signal acquisition can be partly overcome by relying on sensors implanted into or over the surface of the muscles [63–66]. Implantable systems, such as the MyoPlant [67], MIRA (Ripple LLC, Salt Lake City, UT, USA) [68], iSens [69] and the IMES® (Alfred Mann Foundation, Valencia, CA, USA) [70, 71] have been shown to deliver superior EMG data quality in comparison to surface recordings [19]. In a recent clinical trial, IMES has shown to be effective in establishing robust control in both transradial and transfemoral amputees [65, 70, 72, 73]. Moreover, the same implants delivered high-quality direct control even 2 years after the implantation in transhumeral patients (Fig. 41.2) [20].

41.4.3 Alternative Control Approaches

In order to access the neural drive and decipher motor intention needed for prosthetic control, researchers have explored a number of probing methods targeting different neural structures. While the brain is the origin of the high-level control information, and some impressive results in severely affected patients have been reported [75], the risks and the challenges involved in its direct interfacing are high. At the same time, wearable non-invasive approaches for monitoring brain activity are limited in their transmission bandwidth, resulting in sub-optimal control by bionic limb standards [75].

Interfacing of peripheral nerves for decoding of the volitional movements has been, to an extent, achieved in both upper and lower limb amputees [76–78]. The initial findings indicate a possibility of decoding motor information from neural recordings [79–81]. However, electroneurographic (ENG) signals suffer from low signal-to-noise ratio and limited stability [79]. This poses a substantial challenge for extracting the activity of efferent fibres with intrafascicular nerve implants in real world conditions [82].

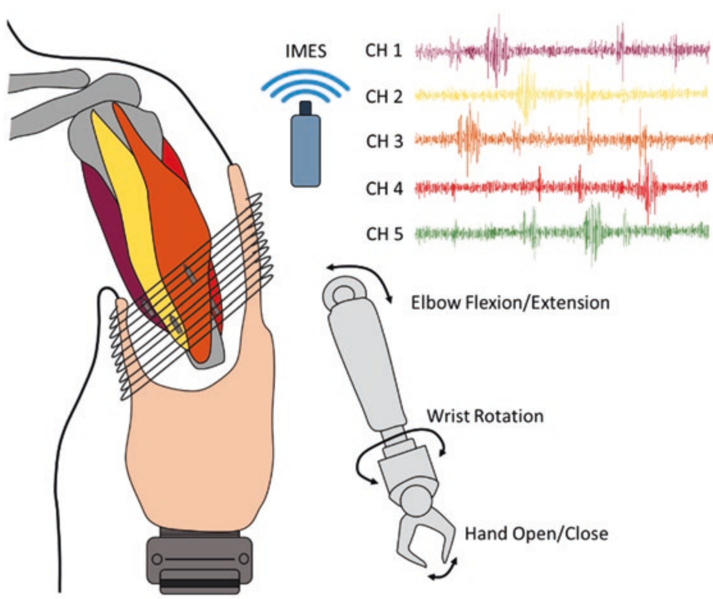

Fig. 41.2 Upper limb prosthetic direct control using implantable EMG system. IMES sensors have been successfully implanted for over 2 years in three transhumeral patients who have additionally undergone TMR surgery in order to provide at least five independent signals [74]. Each of these signals has been then mapped into a respective prosthetic function enabling robust simultaneous and proportional control

Going beyond the neural interfacing, a number of sensor types and sensor fusion approaches have been considered for prosthetic control. Among others, these include ultrasound imaging-based controllers [83], vision-based systems [84], optical motion tracking [85], and force and tactile myography [86, 87]. However, the only systems that went beyond laboratory user cases were quick response (QR) codes [88] and radio-frequency identification (RFID) tags [89] used for switching grip patterns of prosthetic hands, and gyroscope-based mode selection widely accepted in active lower limb systems and available in some prosthetic hand solutions [90].

41.5 Sensory Feedback

Human motor control is highly dependent on the congruent sensory-motor integration. Therefore, the substitution of the motor function is effective only if sensory input is sufficiently restored. However, integration of an effective sensory feedback in prosthetics has proven to be very challenging. When evaluating the state of the environment, we tend to account for the information coming from multiple feedback sources. Furthermore, we then include our previous experience as we make the final judgement. This means that the artificially delivered sensory information needs to be compatible within this complex internal framework [91].

The presence of sensory feedback can have a substantial psychological and therapeutic impact on the prosthesis user. The sense of embodiment, a subjective feeling that something belongs to someone's body, almost solely depends on the received sensory information. By adjusting the provided feedback, a user can be tricked into integrating an artificial object into the body Scheme [92]. This phenomenon is known as the 'rubber hand illusion'. Moreover, sensory stimulation can lead to a decrease of the phantom limb pain by reversing the maladaptive plastic changes in the brain [93].

Restoration of the missing sensory input in prosthetic users has received a substantial attention from the research community. However, the achieved impact has remained limited to one simple commercial system [94]. Commonly, the feedback to the users is provided through the sensors embedded into the prosthetic system tasked to monitor joint positions and interaction forces. These variables are then communicated back using mechanical or electrical stimulations via the residual anatomy [95].

41.5.1 Non-invasive Sensory Feedback

Sensory feedback can be conveyed non-invasively using electro- and vibrotactile stimulation of the surface of the skin [96, 97] or by using acoustic [98] or visual stimuli [99, 100].

The electrotactile stimulation delivers subtle electrical pulses to the surface of the skin, and so it depolarises cutaneous afferents producing tactile sensation. The quality and the intensity of the sensation can be adjusted by modulating frequency and the amplitude of the stimuli. A single pulse feels like a discrete tap and as the frequency increases a perception of vibration, tingling or constant pressure at the surface of the skin is achieved [101, 102]. Using an array of electrodes, the information can also be conveyed through spatial coding by creating specific stimulation patterns [103].

Mechanoreceptors of the skin can be stimulated through direct skin vibrations [104]. The delivered oscillations can be perpendicular or tangential to the surface and again different sensations can be achieved by varying the amplitude and the frequency of the stimuli, or by generating distinct spatial patterns by grouping several vibration elements together [105]. The advantage of vibrotactile stimulation is in the ease of application, as the stimulation is comfortable and does not produce pain as can happen in electrotactile stimulation.

Sensory feedback can also be delivered by applying subtle forces and/or torques to the residual limb. Various actuators can be used to push into the skin [106], apply torque around the targeted joint [107], stretch the skin rotationally and/or longitudinally [108], or deliver three-dimensional force feedback [109]. Alternatively, electrically actuated braces and pneumatic cuffs [110] can be used to squeeze the residuum and apply pressure around it. The advantage of mechanotactile devices is that they can deliver an intuitive modality-matched feedback, so that the information on the force acting upon the prosthesis is conveyed to the user by applying the force to the residual limb.

Finally, hybrid systems can be used in order to deliver multiple information streams in parallel through the same area of the skin. For instance, vibro-electrotactile stimulation can be achieved by placing a vibration motor on the top of an electrode to deliver electrical pulses and mechanical vibrations to the same area of the skin [111]. Alternatively, vibrotactile motors and mechanotactile stimulators can be combined in order to deliver vibration and force stimulation simultaneously [112].

41.5.2 Invasive Sensory Feedback

Akin to motor interfacing, sensory feedback can benefit greatly from invasive approaches. The implanted electrodes can be used to deliver stimuli directly to the sensory neural structures, such as peripheral nerves [19, 113], spinal ganglions [114], brain stem [115], or sensory cortex [116]. The interface has been successfully achieved using both extraneural (non-penetrating) systems, such as cuff [113] and flat interface nerve electrodes (FINE) [117], as well as intraneural (penetrating) systems, such as longitudinal (LIFE) and transversal (TIME) intrafascicular electrodes [118], or Utah slanted electrode array [119].

Using implantable interfaces, there is an advantage of directly stimulating the structures that are responsible for transmitting the sensory stimuli. Moreover, the positioning should be exact and close to permanent in comparison to non-invasive sensory systems. This allows for an increased quality and robustness of sensory feedback, both of which are essential for restoration of natural sensation.

Comparable to surface stimulation, adjustments of intensity and frequency of stimulation lead to changes in the provided sensation. This is due to the number of recruited nerve fibres and their frequency of firing. In addition, the multi-contact electrodes can activate different fibre types and elicit qualitatively different tactile sensations. Therefore, the implanted feedback interfaces can modulate location, intensity, and quality of the elicited tactile sensations. However, this research is still in a preliminary phase, and the first prototypes have only been presented to demonstrate their feasibility, and only very basic sensations were restored. Implantable sensory feedback systems have been successfully implemented in upper limb prosthesis users [19, 113] demonstrating successful discrimination of up to three different force levels. Similar approaches have recently allowed transfemoral amputees to feel the knee motion and sense the sole of the foot touching the ground [120].

41.6 Surgical Techniques for Improved Prosthetic Experience

Adequate residual limb management can have a substantial influence on the prosthetic fitting, control, and acceptance [121]. Therefore, a number of surgical techniques such as tendon transfers [122, 123], free flaps [124–126], digit and toe transfers [127–129], and skin grafting [130, 131] have been devised specifically for this purpose. Over the last decade, a surgical rerouting of residual nerves that have lost their original targeted muscles to alternative muscle sites has become a clinical state-of the-art when dealing with high-level amputations and severe phantom limb pain [132]. This procedure, termed targeted muscle reinnervation (TMR), treats muscles as biological amplifiers of the signals transmitted through nerves [133]. TMR thus allows creation of several well-separated sources of intuitive EMG signals that can be used for multi-articulated control of the prosthesis [134].

With respect to TMR, bionic reconstruction additionally utilises free functioning muscle transfers in order to generate additional EMG sources in patients with non-functioning and insensate limbs [135]. This technique allows for functional restoration following lower root brachial plexus injuries [136] or critical soft tissue injuries like those observed in blast injuries [137].

Finally, a new concept for direct interfacing of the individual peripheral nerve activity has been showcased, focusing on culturing of myoblasts directly onto the ends of transected nerves. Together with an electroconductive polymer scaffolding, these regenerative peripheral nerve interfaces (RPNI) aim to provide a number of discrete signal sources that can be used for prosthetic control [138]. This concept focused on redirecting individual peripheral nerves to small devascularised, denervated pieces of muscle. These groups of contracting muscles can therefore produce motor control signals with high signal-to-noise ratios [139, 140].

41.7 Conclusion and Future Outlook

The current state of the prosthetic limbs market shows a rapid development, which is a direct consequence of advances in technology, surgical techniques, and increased knowledge of human anatomy and physiology. The accessibility and low cost of rapid prototyping as well as increasingly affordable high-tech manufacturing have seen a number of new vendors entering the world of prosthetic hardware in the past 10 years. However, the consistently high prosthetic abandonment rates (especially for upper limb devices) and general dissatisfaction of the prosthetic users are a strong indicator that advances in the interface design are still required.

Here, we have presented some of the academic advances in the fields of upper and lower bionic limbs. We foresee that the next generation of these devices will include a robust bi-directional control in combination with the use of chronically implanted sensors. Most likely, these interfaces will target muscles for extracting motor commands and nerves will be used to convey the sensory information. However, due to the primary clinical need for control and the plethora of open challenges in artificial sensory feedback, we anticipate that the advances in control will have an earlier clinical impact, followed at a later stage by the inclusion of natural sensory components.

These solutions are likely to impact each user group differently and with varying levels of benefit. Patients with multiple impaired limbs, who rely more heavily on their prosthetic devices, are expected to have greater benefits of the new technologies. However, the impact of new technologies will be highly influenced by the support offered by the available health care services. Therefore, it is critical to ensure proper and timely education of the health care professionals in the field, and to make sure that global health policies are in place to guarantee the satisfactory implementation of the technology and to maximise its reach.

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Orthotics

42

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Abstract

Continued advances are required to address mobility limitations caused by lower extremity blast injury. Individuals who experience persistent deficits following trauma may benefit from external support and/or offloading provided by ankle foot orthoses (AFOs). Currently available AFOs vary widely in their design and potential benefit. Carbon fibre custom dynamic ankle foot orthoses (CDOs) have been increasingly used to improve mobility after traumatic injury. CDOs are made predominantly from carbon fibre and are intended to restore function across a range of daily and high-energy activities. Patient-reported outcomes, physical performance measures, and biomechanics data from studies focusing on CDO use have demonstrated positive outcomes. CDOs consist of a proximal cuff, posterior carbon fibre strut, and footplate, which can be tuned to meet the needs of the patient. Available literature provides guidance related to key design considerations during the fitting process. Further, intensive training when combined with the CDO has been found to enhance clinical outcomes and facilitate successful return to high-energy activity. A majority of available data related to CDO use following limb trauma is focused on a subset of military personnel, and available civilian data is limited.

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42.1 Introduction

Continued advances are required to more effectively address the mobility limitations caused by lower extremity blast injury. Although surgical and rehabilitative advances have helped improve lower limb function, mobility limitations are common following limb salvage procedures due to pain, decreased tolerance of limb loading, scarring, volumetric muscle loss, nerve injury, and associated reductions in muscle and joint function. Individuals with intact lower extremities and functional deficits may benefit from the external support and/or offloading provided by ankle foot orthoses (AFOs). Multiple types of AFOs are used to compensate for deficits in the injured limb, and appropriate AFO prescription results from an appropriate match between the impairments in the injured limb and a sound understanding of the relative

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benefits and limitations of available AFOs. For example, when the anterior compartment of the lower limb is injured, individuals may experience an inability to raise their foot (dorsiflex the ankle), resulting in difficulty clearing the foot during swing and an uncontrolled lowering of the foot (foot slap) when initially loading the limb. Rather than providing a large bulky device, this deficit can often be addressed using a lightweight carbon fibre device due to the low weight of the foot and small forces required to keep the foot in a neutral position. In contrast, an individual who has lost all ankle plantarflexor function, or has pain with movement or loading of the foot and ankle, will require a more robust device. The more robust the device, the greater the need for the patient to have device-specific training to maximise mobility, with the goal of a full return to activities of daily living and high-energy activities, even potentially active military service.

42.2 Device Type

Dorsiflexion assist AFOs are primarily used following blast injury to compensate for weakness of the tibialis anterior muscle, which is the primary ankle dorsiflexor. Dorsiflexion assist AFOs often consist of a lightweight carbon fibre shell, medial or lateral strut, and lightweight compliant footplate (Fig. 42.1a). The primary role of the device is to prevent foot drop, facilitating toe clearance during the swing phase of gait to reduce the risk of tripping, to help pre-position the foot for initial contact, and/or to allow for controlled lowering of the forefoot to the ground as the limb is initially loaded.

Fig. 42.1 (**a**) Dorsiflexion assist AFO (off the shelf, Allard BlueRocker), (**b**) posterior leaf spring AFO (off the shelf), (**c**) ground reaction AFO (custom), (**d**) solid ankle AFO (custom), (**e**) carbon fibre custom dynamic ankle foot orthosis (CDO)

Posterior leaf spring AFOs are typically used by individuals with decreased strength and/or control of their plantar and dorsiflexors. Posterior leaf spring AFOs are typically made of plastic and, although effective at maintaining a neutral ankle alignment during swing phase, their relatively light weight and limited stiffness provide limited support in response to the higher forces during the stance phase (Fig. 42.1b).

Ground reaction AFOs help to address both plantar and dorsiflexor weakness. Ground reaction AFOs are typically custom moulded from plastic (Fig. 42.1c), and the more robust devices control tibial progression and prevent excess dorsiflexion during stance phase. The anteriorly located proximal cuff is used to apply force to the tibia below the knee, acting to control forward progression of the tibia while the foot is pressed to the ground by body weight during the stance phase. The footplate can be custom moulded to support the foot and prevents plantarflexion during swing phase. Together, this device helps control the knee and ankle during gait.

Solid ankle AFOs are typically custom made for each patient and limit motion at the ankle joint due to their high stiffness. Solid ankle AFOs can be used in individuals with limited strength below the knee or in cases where motion is painful. This type of device is commonly made from plastic, and the thickness is crucial in determining the amount of deformation and resulting motion at the ankle joint during load-bearing activities (Fig. 42.1d).

In the last 10 years, *carbon fibre custom dynamic ankle foot orthoses* (CDOs) have been increasingly used to improve mobility for individuals who have undergone limb salvage procedures after traumatic lower extremity injury (Fig. 42.1e). CDOs are stiff devices made predominantly from carbon fibre that are intended to restore function across a range of daily activities, as well as high-energy activities including running [1]. Two types of CDOs, the Intrepid Dynamic Exoskeletal Orthosis (IDEO) and the British Off-loading Brace (BOB) [2], have undergone clinical testing in military populations from both the United States and the United Kingdom. Patient-reported outcomes, physical performance measures, and biomechanics data from these studies have demonstrated positive outcomes from both populations.

42.3 Custom Dynamic Orthoses

CDOs have a proximal cuff (Fig. 42.2) that contacts the proximal tibia just below the knee. It is the primary location where force is applied to the tibia, and when highly customised can be used to suspend and support the limb in a manner similar to that of a transtibial prosthetic socket. By applying force through the proximal cuff, the forces on the foot and ankle can be reduced, with force transferred to the ground through the posterior strut and footplate rather than through the injured limb. In addition to helping suspend the limb, the posteriorly located carbon fibre strut also stores energy when it is deformed during mid-stance loading and returns it during pre-swing providing push-off power. The footplate of a CDO like the IDEO is customised to the patient to protect and support the foot during gait. This feature is particularly important in individuals who have sustained traumatic foot and ankle injuries. Painful areas on the foot are marked during the casting process, and relief is provided during fabrication to minimise pressure or rubbing in those areas. The nearly rigid footplate also acts as a lever, working in conjunction with the posterior strut to control limb forward progression while walking. The rigidity of the posterior strut and the footplate work together to limit movement in the ankle joint, helping keep patients in their pain-free range of motion. The toe-down (plantarflexed) position of the footplate can effectively reduce pain in individuals who experience pain with dorsiflexion and improve the ability to effectively load the toe and deform the posterior strut. CDOs like the IDEO are designed to include a heel wedge between the device and the shoe to provide shock absorption during loading and accommodate for the plantarflexed alignment.

Fig. 42.2 Carbon fibre custom dynamic ankle foot orthosis (IDEO) consisting of a proximal cuff, carbon fibre posterior strut for energy storage and return, carbon fibre footplate to protect the foot and act as a lever, and heel wedge for alignment and cushioning

42.4 Design Considerations

Each element of a CDO can be tuned to meet the needs of the patient depending on their functional deficits, pain, muscle activity, and their desired subsequent mobility using the device. Some elements have greater impact on running and walking biomechanics than others, making desired activity levels an important consideration when tuning the device. Reducing loading of the foot and ankle by transmitting forces through the posterior strut of the CDO rather than through the injured limb can reduce pain associated with calcaneal fractures or other traumatic foot injuries. The ability to successfully offload depends on factors such as limb geometry and skin health. A well-defined large gastrocnemius muscle with healthy intact skin is optimal when attempting to suspend and offload the injured limb. The muscle belly of the gastrocnemius works as a shelf to counteract the downward force and subsequent motion of the limb. An intimate fit on the anterior medial tibial flair, below the tibial plateau, provides vertical support and a counteracting force opposite the posterior cuff to aid in suspension. Further, a patellar tendon bar can be incorporated akin to a patellar tendon bearing prosthetic socket. Patients with decreased muscle mass are able to offload with a cuff that has been fitted correctly; however, they may need to adjust or tighten the cuff more frequently. Patients with decreased sensation of the lower limb due to nerve injury, or poor skin health due to injury, should be closely monitored to avoid skin breakdown or irritation. The rigid footplate replaces the natural lever arm of the foot, while the geometry, which includes a more compliant or curved toe section, facilitates a proper rollover shape and centre-of-pressure progression when combined with the correct alignment and heel cushion. The stability and support from the carbon fibre footplate and posterior strut help alleviate foot and ankle pain during walking or high-energy activities, such as running.

The CDO *posterior strut* can be either fixed and permanently laminated into the cuff and footplate to minimise weight, or modular to allow changes in stiffness and alignment. Adjustability is most desirable in individuals who are rapidly changing following injury or when practitioners are first learning to fit CDOs. Posterior strut stiffness influences energy storage and return during gait [3], as well as patient comfort and perception of the device [4]. Preferred strut stiffness depends on a number of factors, including the patient's range of motion, activity level, body mass, and load carriage [3]. More compliant struts can allow greater ankle motion during gait [4], and stiffer struts increase ankle joint stiffness, which may benefit with limited pain-free range of motion [3]. Stiffness can also affect gastrocnemius muscle activity, with more compliant struts providing less body support, resulting in greater gastrocnemius activity [4]. Studies have found that strut stiffness has a limited effect on walking and running kinematics and kinetics [3, 4].

The *bending axis* of the posterior strut can vary with strut geometry, total length, or vertical positioning on the posterior aspect of the leg. CDOs are typically intended for use during everyday activities like walking, but also aim to allow patients to return to high-energy activities like running. The bending point was found to have little effect on patient biomechanics during overground walking [5], but was found to influence biomechanics while running [6]. Running requires increased power generation at the ankle, and a more distal bending point, closer to the anatomic ankle joint position, resulted in greater ankle range of motion, ankle power, and propulsive ground reaction forces [6]. The increased power is beneficial to running, but the increased range of ankle motion can be problematic for patients who have limited pain-free range of motion. Varying bending axis location had a limited effect on proximal joints [6].

The *heel wedge* between the CDO and the shoe can vary in durometer (compliance) and height and may be made from a variety of materials. The height and durometer of the wedge have a direct effect on limb loading, perceived function, and comfort of the device [7]. Heel wedge durometer affects shock absorption characteristics and timing of forefoot engagement during gait. Taller and firmer heel wedges produce generally similar changes in walking biomechanics. The firmer heel wedges take longer to compress and have a higher compressed height when loaded during gait, requiring greater forward rotation of the device before the forefoot is loaded. Higher heel wedges, which lower the heel of the shoe relative to the foot, increase the shank (tibia) to vertical angle during standing, with the added thickness under the heel of the CDO forcing the tibia into a more forward-leaning position when the shoe is flat on the floor (Fig. 42.3). As a result, tall wedges, independent of durometer, result in greater time spent on the heel resulting in larger and longer peak dorsiflexion moments, later peak knee internal extension moments, and larger radii in rollover shapes [7]. Further, tall firm wedges result in a delay in peak dorsiflexion moments, the largest magnitude peak internal knee extension moments to control tibial forward rotation, and a more posterior position for the centre of curvature in rollover shape [7]. In contrast, short wedges result in early and rapid engagement of the forefoot. Earlier forefoot loading results in a higher peak velocity of the centre of pressure as it rapidly moves from the heel to the toe. The abrupt transition from the heel to the toe, and greatest centre-of-pressure velocity, is observed with short soft wedges and is generally disliked by patients [7]. Overall, patients tend to prefer heel wedges that replicate the centre of pressure and ankle moment trajectories seen in able-bodied individuals [7].

Fig. 42.3 Alignment is critical to patient comfort, device function, and control of motion in a pain-free range. Alignment can be affected by casting position and may be accommodated for with heel wedges. Increasing the height of the heel wedge increases the shank to vertical angle

CDO alignment is critical to patient comfort, device function, and the ability to keep patients in their pain-free range of motion. A range of alignments may be clinically acceptable for each patient [8]. Alignment is usually determined by positioning during the initial casting process and the accompanying modifications made to the device during fabrication and fitting. The patient's natural post-injury foot position and pain-free range of motion should be considered during the casting process. Some patients have limited range of motion after blast injury due to joint and soft-tissue injury and scarring, leaving some positions unsuitable for casting. The high stiffness of CDOs typically results in less than 10° of motion into dorsiflexion relative to its unloaded alignment [4, 5]. Accordingly, CDOs are typically cast in a plantarflexed pain-free position and ideally greater than 10° from the dorsiflexion limit of their pain-free range. If the foot and ankle positions are aggressively corrected during casting, skin irritation and increased pressure can be expected during more dynamic activities.

The alignment of the CDO affects walking and running biomechanics. A plantarflexed alignment results in earlier engagement of the forefoot, slowing progression of the tibia, and acts in a manner similar to a shorter softer heel wedge while walking. A more plantarflexed alignment results in decreased peak knee flexion, internal knee extensor moment, and quadriceps activity during loading response of level-ground walking, and lowers peak ankle power absorption, reduces internal knee extensor moments, and creates a greater plantarflexion foot contact angle at foot strike in running [8].

42.5 Training, Evaluation, and Outcomes

The US military personnel prescribed with an IDEO after a limb salvage procedure typically participate in an intensive rehabilitation programme referred to as the return-to-run (RTR) clinical pathway. The programme was developed to facilitate successful return to running as well as other military-relevant high-energy activities, including participation in sports and return to active duty. The RTR pathway includes a sports medicine-based progressive training programme and occupation-specific training to emphasise the skills needed for return to duty [9, 10]. The RTR pathway focuses on strength, agility, and speed to help patients return to high-energy activities [9] and high-level physical function while decreasing the likelihood for late amputation [11]. The RTR pathway results in up to 30% improvements in multiple objective measures of physical performance including the four square step test, self-selected walking velocity, timed stair ascent, Illinois agility test, sit to stand five times, and shuttle run [11, 12]. Each of these measures has demonstrated reliability and is responsive to changes in physical function [13]. It is important to note that significant improvements are observed in individuals greater than 2 years after injury [11]. Patients also report high levels of satisfaction with the IDEO after completing the RTR pathway, with some attenuation after 6 and 12 months [12]. In one study, participants were asked if they were considering late amputation before and after completing the RTR pathway; 82% of individuals initially interested in late amputation favoured limb salvage after completion of RTR pathway [11]. Other studies evaluated the percentage of patients who returned to duty and found that more individuals returned to duty after completing the RTR pathway than not [9, 10, 14]. Studies have shown that CDOs, accompanied by intensive physical therapy, may be beneficial for patients who have undergone limb salvage procedures.

Performance measures, previously validated in able-bodied populations, are a valuable tool for evaluating physical ability in patients who use CDOs [13]. Commonly used measures, as mentioned previously, include the four square step test (4SST) and the Illinois agility test to evaluate agility, sit-to-stand (STS) and timed stair ascent (TSA) tests to assess strength and power, and self-selected walking velocity (SSWV), 10 m shuttle run, and 40-yard dash to evaluate speed and general physical capability [9, 12, 15]. Each test is timed, and, with the exception of the self-selected walking velocity test, participants are instructed to complete the test as quickly as possible.

The 4SST is completed by moving around a Maltese cross laid out on the ground using 1 in. obstacles [15, 16]. Subjects begin by standing in the lower left corner, and moving forward, right, backward, and left and retracing that pathway by then moving right, forward, left, and backward to finish the trial in the same position that they began (Fig. 42.4). They cannot touch the obstacle and must keep one foot in contact with the floor at all times [15]. In military populations, results were significantly faster with the IDEO than without bracing [12, 15, 16]. Further, when compared to other commercially available AFOs, times were significantly better with the IDEO [15]. Those whose results improved with the use of the IDEO still had slower times than able-bodied control subjects [16]. Improved results in the 4SST likely result from limb stabilisation provided by the device.

During the STS test, participants begin seated in a chair with their back touching the vertical backrest. They are then instructed to stand up, fully extending their knees, and sit down, touching the chair, five times as quickly as possible [15]. Studies including the STS test with the use of the IDEO have shown differing results. One study found that the IDEO, and completion of the RTR pathway, improved STS times [12]. Others reported that the IDEO had no significant effects on STS test outcomes [15, 16]. The STS test evaluates strength and power but is also reliant on lower limb mobility. CDOs limit the foot and ankle range of motion and ultimately foot position while sitting, which can affect the ability to get up and out of a chair.

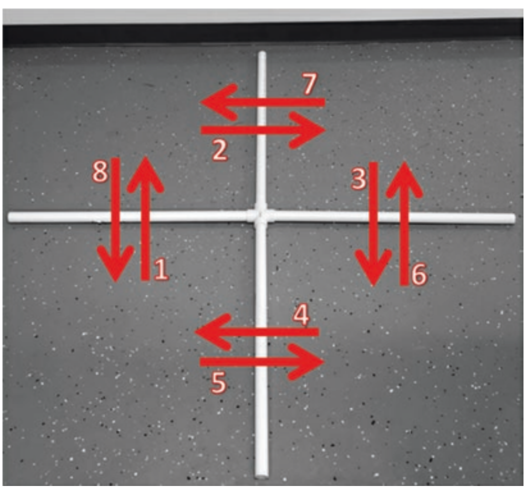

Fig. 42.4 For the four square step test (4SST), a Maltese cross is laid out on the floor using 1 in. obstacles, and participants move between each quadrant in the order shown as quickly as possible while keeping one foot in contact with the ground at all times. Time starts with initial movement and ends when both feet are in the starting quadrant of the cross

During the TSA, participants ascend a set of steps (12) as quickly as possible, contacting each step and without using the railing [15]. The IDEO significantly improved TSA times compared to other commercially available braces [15] and no brace conditions [12, 16]. Similarly, results were still slower than those of able-bodied control subjects [16]. Improvements associated with the IDEO can be attributed to energy storage and return from the posterior strut.

For the SSWV test, participants are instructed to walk at a comfortable pace down either a flat surface or an uneven terrain (Fig. 42.5). The time required to cross 15 m on level ground, or 6 m on rocky terrain, is recorded, and an average velocity is calculated after completion of five trials [15]. The test is completed in an area with light foot traffic, so subjects are not distracted. Results, over level and uneven terrain, were faster in the IDEO than in other commercially available braces [15]. Faster self-selected walking speeds can be attributed to the additional support provided by the brace.

The 10 m shuttle run includes two cones, set 10 m apart, and two wooden blocks. Participants are instructed to line up at cone 1, run to cone 2, pick up one wooden block, and run to place it on the ground behind cone 1. They then run back to cone 2 for the second block turn around and run past cone 1 to complete the test. This test combines running and the need to pivot and change directions quickly. CDOs are beneficial because of the added support for the injured limb and

Fig. 42.5 Self-selected walking velocity can be measured using both flat level ground and uneven terrain as seen here. The time required to cross a specified distance is used to calculate the walking velocity

energy storage and return from the posterior strut, but limited foot and ankle motion requires changes in cutting technique during the quick changes in direction. Shuttle times improved with the use of the IDEO and completion of the RTR pathway [12].

The 40-yard dash is completed on a straight level surface. The time to run 40 yards is recorded, with time beginning when the patient starts moving and ending when they cross the 40-yard mark [15]. IDEO use resulted in the fastest times when compared with other clinically available options [15]. Patients benefit from the energy storage and return of the posterior strut during this test.

Questionnaires are useful when evaluating patient-perceived outcomes related to mobility, satisfaction, and function related to CDO use. The Paffenbarger Activity Scale (PAS) reports frequency and duration of participation in sports or recreation activity in the last week [9], which can indicate overall function with or without the CDO. The Orthotics and Prosthetics Users' Survey (OPUS) assesses patient use and satisfaction with an orthotic device and with the related services [9, 12]. Device satisfaction questions include topics such as weight, comfort, pain when using the device, ease of use, cosmetic appearance, fit, durability, and effects the device has on clothing [9, 12]. Questions concerning satisfaction with service include topics like wait time, respect, communication between involved parties, patient input, use of a team approach, and device training [9]. The OPUS is a freely available questionnaire that applies to all prosthetic and orthotic devices. Patients can also complete the Short Musculoskeletal Function Assessment (SMFA) to self-report their physical function [12] and the Lower Extremity Functional Scale (LEFS) to report their ability to complete both everyday and high-impact activities [17]. Perceived Global Rating of Change (GRC) can effectively compare differences between activities when using different devices or different device conditions.

The IDEO has generally been reported as comfortable, easy to take on and off, easy to clean, durable, and minimally abrasive to skin [15, 17]. Patients have also reported less pain when walking with the IDEO than without it [17]. Patients who used other types of AFOs before the IDEO reported higher satisfaction with the IDEO than the other AFOs [12], and 94% of those who compared multiple devices in one study preferred the IDEO over the other options [15]. Use of the IDEO also decreased the number of patients who were considering late amputation [14, 15]. SMFA is responsive and able to demonstrate improvements in domains related to completion of the RTR pathway with the IDEO [12]. Similarly, LEFS scores with the use of the IDEO were significantly greater than scores reported without the use of the IDEO [17]. Further, patient-reported outcomes indicate that it is easier for patients to perform everyday and high-impact activities with the IDEO than without [17].

Biomechanical evaluation of gait using motion capture and force plates can guide both individual patient fittings and general device design approaches (Fig. 42.6). Joint angles, moments, and powers, when combined with electromyography, have been used to provide insight into the effects of specific design changes on resulting walking and running gait.

Some individuals ultimately proceed to delayed amputation following CDOs and intensive rehabilitation due to persistent pain or functional deficits [15]. However, data comparing individuals with CDOs and individuals who have had a transtibial amputation are limited. In a small study comparing individuals with amputation and limb salvage, higher resting pain levels were reported following limb salvage. [18]. Following transtibial amputation, gait deviations are primarily at the knee, with attenuated knee extensor moments and power [18]. In contrast, individuals using an IDEO demonstrate greater deficits at the ankle, with decreased range of motion and power production [18]. Both groups show a decrease in ankle power output compared to able-bodied controls, with the transtibial amputation group having a 43% deficit and the IDEO group showing a 63% deficit [18]. Objective tests of physical mobility have shown deficits in mobility in individuals with limb salvage with or without CDO use, and in individuals with transtibial amputation [16]. Able-bodied individuals have faster times on the 4SST, STS five times, and TSA tests, and individuals with limb salvage without any bracing had the slowest times [16]. Participants who had undergone either an amputation or a limb salvage procedure were limited in at least one of the performance measures [16]. Mobility did not differ between the group with below-the-knee amputation and the group who used an IDEO after limb salvage [16]. Functional deficits, gait deviations, and compensation mechanisms can be seen with severe trauma affecting the lower limbs.

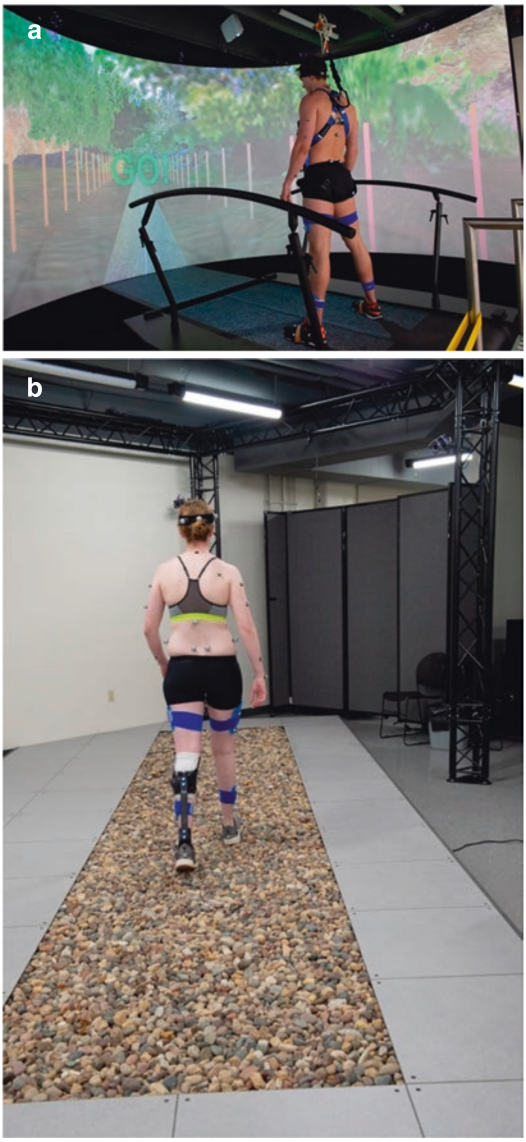

Fig. 42.6 Evaluation of gait biomechanics using motion capture and force plates in (**a**) virtual reality environments or (**b**) on level ground can provide insight into the effect of design on walking and running gait

42.6 Summary

A majority of available data related to AFO use following limb trauma, and blast injury in particular, is focused on a subset of military personnel who have undergone limb salvage procedures and completed intensive rehabilitation. Although many types of AFOs, including CDOS, are now commercially available and in use by civilians, available data is limited. Advances in the civilian setting are expected to benefit injured service members. Ongoing efforts to use CDOs to prevent the development of post-traumatic osteoarthritis and further refine device design, fabrication, and fitting have the potential to further improve the already beneficial effects of AFOs on function following blast injury.

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Sockets and Residuum Health

43

Matthew Hopkins, Louise McMenemy, Shruti Turner, and Alison H. McGregor

Abstract

Major limb amputation affects millions of people worldwide and became a common procedure in the latter years of the conflicts in Afghanistan and Iraq.

Prosthetic devices are commonly introduced to lower-limb amputees to assist with mobility and functionality. Despite vast leaps in the technology of lower-limb prostheses, acceptance of a prosthesis is limited by the quality of interfacing between the residuum and the device. No single design fits all, and production of a socket is a complex task, requiring significant skill and expertise from the prosthetist.

The fit and pressure distribution of the socket can significantly impact the health of the soft tissue, leading to conditions such as pressure sores if not properly managed. The residuum-socket interface is further complicated by natural volume fluctuations of the limb, issues with thermoregulation, infections and musculoskeletal pathologies.

Researchers, clinicians and industry have attempted to better understand and inform optimal socket fit through computational and hardware methods, and surgical techniques such as direct skeletal fixation have been proposed to bypass the socket altogether.

This chapter explores the nature of the prosthetic socket, common issues users present and methods to improve the interface between the residual limb and the prosthesis.

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43.1 Introduction

The conflicts in Afghanistan and Iraq were associated with a high number of casualties surviving complex military trauma [1–3] with multiple injuries [2, 4].

The weapon of choice for insurgent groups against coalition forces was the improvised explosive device (IED) [2] and was the dominant cause of fatal battlefield injury [3, 5]. These devices are associated with an alternate injury profile in comparison to conventional munitions [3, 4] involving blast and fragmentation trauma to the extremities of the body in the majority of instances [1, 5]. The improvements in personal protection equipment (PPE), modern resuscitative strategies, shorter casualty evacuation times [2, 6], prehospital enroute care, receiving field

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hospital management [5] and in-hospital damage control resuscitation protocols [3] led to an unprecedented improvement in survival rate [4] leading to an increase in the numbers of military personnel surviving with long-term disability [3, 5]. Between 2003 and October 2014, a total of 265 survivors sustained 416 amputations either as a direct result of wounding or delayed due to failed limb salvage [2] producing a mean number of limbs lost of 1.6 (SD ± 0.68) [2]. Of this cohort, transfemoral amputations were the most common (153 patients, frequently bilateral), followed by transtibial amputation (143 patients) [2]. By the end of the conflict, there were 323 surviving UK amputees, 113 of whom had multiple significant amputations [6, 7]. Due to the young age [6] and high levels of physical fitness prior to injury of military amputees, their expectations of function post-amputation are high [8].

As lower-limb amputations occur with greater frequency than upper limb amputations [9] and the impact of lower-limb injury on quality of life, mobility, health, and social engagement is significant and is associated with a high incidence of pain [8, 10], the focus of this chapter is traumatic and delayed lower-limb amputation due to previous traumatic injury.

43.2 Prosthetic Devices

Prosthetic devices are used to restore some level of functionality to an amputee [11–14], allowing them to take part in normal daily activities [10, 15, 16]. Between 68% and 88% of individuals with amputation wear a prosthesis for a minimum of 7 h each day to enable performance in everyday activities [10]. Lower-limb prostheses usually consist of several essential components, namely the socket, foot, pylon, ankle and in above-knee cases, the knee joint [11, 15]. Despite considerable advances in prosthetic technology, particularly with the advent of robotic and microprocessor-controlled joints [17] (see Chap. 41), there is a high incidence of prosthesis abandonment and low satisfaction level [11]. Patient dissatisfaction is often attributed to socket related issues including poor comfort and control, reduced biomechanical function [11] and resultant injuries from poor fit [16].

The prosthetic socket provides the interface or mechanical coupling between the user's residual limb and the prosthetic device [8, 11, 14, 15, 18–22]. In this capacity, the socket is responsible for load transfer [20] in static and dynamic conditions [8, 19, 23] from the prosthetic foot to the soft tissue of the residual limb [13]. The efficiency of this interface is highly dependent on minimal movement between the residuum and the prosthesis (the coupling stiffness) [18]. The socket also provides proprioceptive feedback [8], stability in the form of vertical support and mediolateral balance [24] and control of the prosthesis [11, 19].

The socket is regularly cited as a priority for lower-limb amputees [8, 12, 18, 23, 25, 26] and can have enormous effect on user comfort [11, 23, 27], the length for which patients can use their limbs [17], and acceptance of the prosthesis [10, 11]. Up to 55% of lower-limb amputees report dissatisfaction with socket comfort, pain in their residual limb and issues with skin breakdown [19].

The tissue of the residual limb, particularly the skin, is not intended to tolerate the forces applied by the socket [18, 27] and is subject to unnatural levels of moisture and exposure to the chemical compounds of the prosthesis [28]. Furthermore, the unique nature of an amputee's residual limb, particularly following blast trauma, has complex physiology and poses a significant challenge to the fitting process [15]. Poor fit and inadequate suspension can increase the forces distributed over the residual limb, causing discomfort, which may impact mobility leading to gait alterations increasing the likelihood of other physical complications [16, 18, 29]. Prolonged exposure to repetitive, excessive load can cause skin irritation, ulcers and in some cases, a need to surgically revise the residuum to resolve these complications [15].

There is an intricate relationship between factors affecting comfort and performance of a socket, hampering efforts to produce an ideal solution [11]. Socket shape, fit [18], the suspension mechanism and the materials used to construct the socket are all important for skin health and perceived comfort [18, 23]. Variability in amputee anatomy, the size and length of their residual limbs, and levels of voluntary control demand differing socket designs [8, 30]. Additionally, sockets suffer issues with excess heat and moisture retention and are unable to account for daily volume changes in the residual limb [8].

Socket design and choice of suspension method are highly reliant on the prosthetist's experience and skill [13, 17] and are based on the amputee's physiology, lifestyle and activity level [11]. At the time of fitting there is limited information available to inform socket design [31], and feedback from the amputee is subjective in nature and may be misleading [32]. A lack of consistency between clinical standards or practice for prosthetic device creation, socket fitting [18], suspension, alignment [30] and the reliance on the prosthetist's experience [11] when producing a socket further complicates the procedure. Additionally, there are no common guidelines for continuation of prosthetic care beyond prescription of a prosthesis through to physical and occupational therapy [8]. Lower-limb amputations are categorised into below-knee (transtibial), through-knee (knee disarticulation) and above-knee (transfemoral) [33, 34] (see Fig. 43.1).

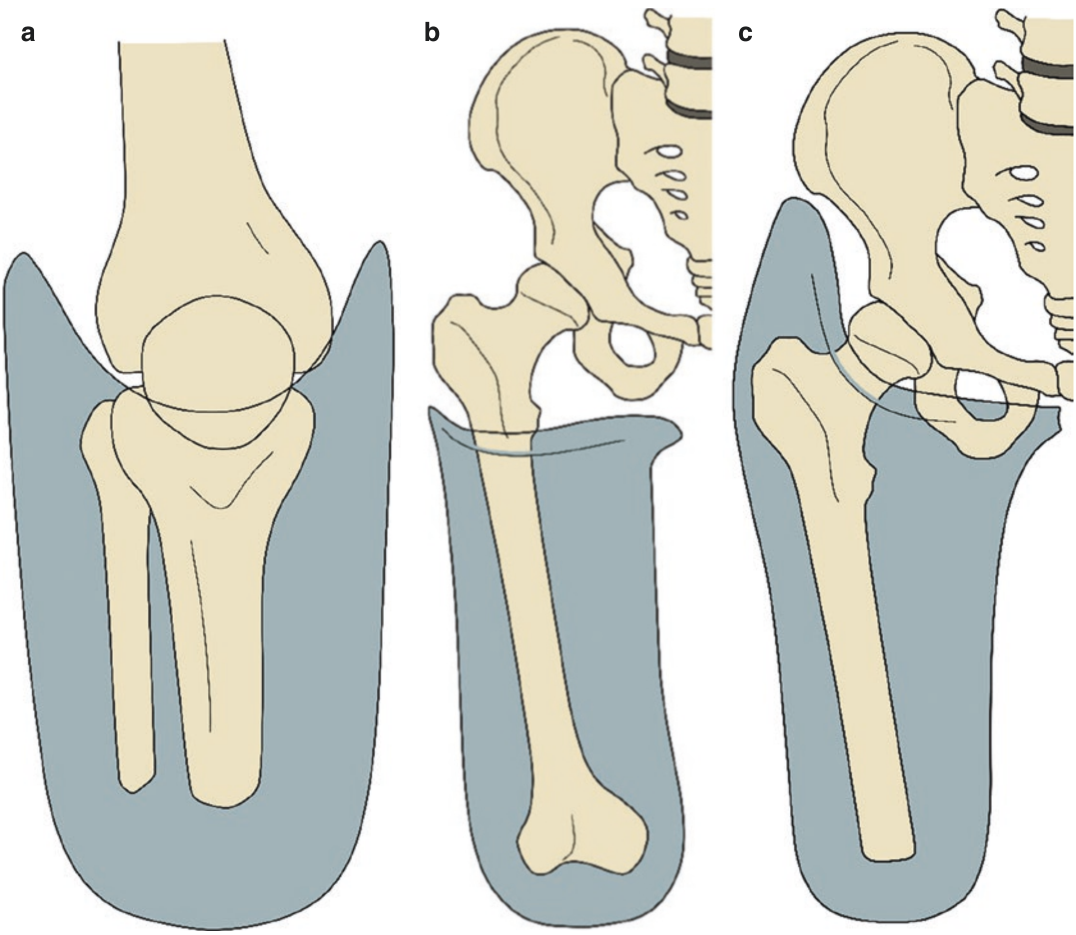

Fig. 43.1 Socket structures for (**a**) transtibial, (**b**) through-knee and (**c**) transfemoral amputees

43.2.1 Transtibial

Below-knee or transtibial amputations account for almost 50% of all new amputations [29]. The conventional below-knee prosthesis, a design dating back to 1696, utilised a thigh corset with side bars (TC-SB) and an open-ended wooden socket [29] in which load was split between the socket and the corset [23]. Modern-day transtibial sockets tend to be either patellar tendon bearing (PTB) or total-surface-bearing (TSB) [11, 18, 35, 36] (see Fig. 43.2).

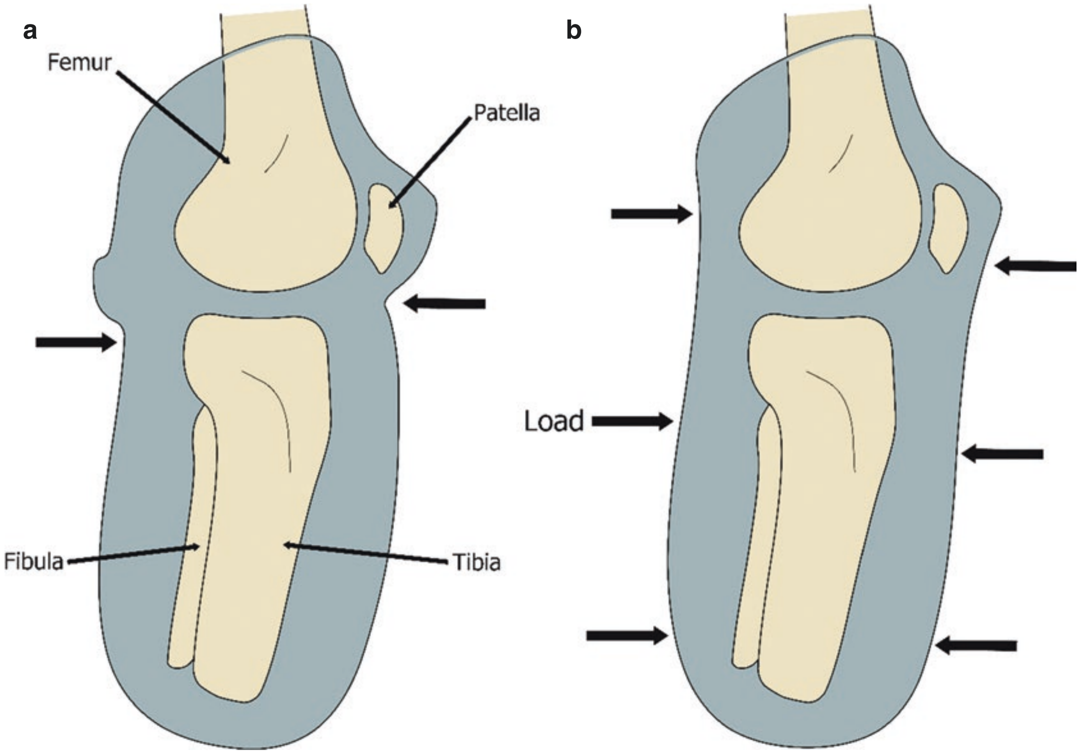

Fig. 43.2 Common transtibial socket structures (**a**) patellar tendon bearing (PTB), (**b**) total-surface bearing (TSB)

PTB sockets were introduced by Radcliffe in the 1950s [15, 23] and are a form of specific-surface-bearing (SSB) socket. SSB sockets apply loads to specific, typically more pressure tolerant anatomical regions of the residual limb [11]. In PTB sockets, load is applied primarily to the patellar tendon, anterior medial tibia flare, anterior muscular compartment and the popliteal area [15, 18]. The design aims to relieve loads on the fibular head, anterior tibial crest and anterior distal aspect of the tibia [15, 18], however, their ability to achieve this has not been assessed objectively. The close fit of the socket to the residual limb enabling more medial placement of the prosthetic foot under the socket is postulated to enable narrower stance-phase support, similar to that of able-bodied individuals [37].

PTB designs feature mediolateral grip on the femoral condyles [11] for increased knee stability through extension of the socket walls to the level of the adductor tubercle of the femur [15]. There is enclosure of the distal third of the patella to share body weight bearing, and the proximal posterior rim is flared out to allow comfortable knee flexion by prevention of hamstring tendon loading [15]. The socket is generally aligned with the knee in a flexed position, located over the middle third of the foot to encourage patellar tendon weight bearing [37]. The distal end of the socket may provide a small amount of contact pressure to the residual limb, useful in prevention of oedema [37].

PTB sockets presented notable enhancement over prior conventional socket designs in their absence of side bars and mechanical hinges, allowing for free motion of the knee [37]. PTB sockets are produced from a cast of the amputee's residual limb, that is modified to direct weight to load-bearing regions and away from bony prominences [35]. The socket is manufactured using laminated woven materials in combination with acrylic resins or from moulded thermoplastic sheets [15]. Polyethylene foam liners [15, 35] may be introduced to reduce interface pressure between the residual limb and the socket [15].

PTB sockets are often reported to provide good fit, however, they suffer with suspension issues and inherently subject the residuum to high concentrated pressures on loaded areas by reduction of the overall loaded anatomical area [11, 15]. These high loads lead to skin stretching which is a significant factor in many residual limb injuries such as skin ulcers [11, 15]. Furthermore, the design demands a skilled prosthetist in order to produce a form with adequate load distribution, a task that is often time consuming [15].

PTB sockets are not suitable for amputees with sharp bony prominences or particularly sensitive residual limbs [15]. Even with a well-fitting PTB socket, the amputee will experience constantly changing forces in the anteroposterior and mediolateral directions [24]. It has been suggested that PTB sockets are suitable for very short transtibial limbs with at least 2 inches of tibia below the tibial tubercle, particularly if the amputation is through cancellous bone, where high pressures can be tolerated [37].

Alternative TSB sockets were introduced by Kristinsson in 1986 [15] and aim to distribute pressure more evenly across the residual limb than their PTB counterparts [11, 18, 23]. The altered distribution is believed to assist in avoiding high localised stresses, stabilises bony areas [15] and may provide greater comfort and enhanced proprioception [11]. The TSB design has a significantly different shape from PTB designs [15] and relies heavily on the mechanical properties of its accompanying liner [18, 23]. The accompanying silicone or elastomeric liners are sticky, pliable and wrap tightly around the residual limb surface [15]. These liners tend to utilise one of two major suspension methods, namely a pin attached to the distal end of the liner or circumferential seals that produce a vacuum for suction [15].

Considerable advantages of this design include improved suspension and protection of the residuum due to the adhesion of the liner to the residual limb [15]. There is suggestion of potential benefit of TSB sockets over PTB designs in improvement of gait symmetry [11] and better suitability for more active persons [23], younger users, and those with traumatic injury, skin problems, skin grafts or skin tenderness [18]. Additionally, viscoelastic interfaces are associated with decreased dependency on walking aids, improved suspension, better load distribution and improved comfort [23]. In general, amputees state a preference for TSB sockets including silicone liners as a suspension mechanism [15].

TSB sockets are, however, associated with notable problems in relation to skin health [15]. The liner covers skin pores, limits air circulation [38] and thermally insulates the limb [39, 40] leading to increased perspiration [15] which may cause discomfort [38], itching and lead to infection [41]. Additionally, the liner increases bulk around the knee, which may cause discomfort, particularly when sitting. Elastomeric liners must be rolled on and off, inherently increasing the difficulty of donning and doffing [15]. This can be a particular challenge for blast injury patients due to the poly-trauma nature of their injuries [30]. Concomitant partial hand loss and major upper limb amputation decrease dexterity increasing the frequency of liner creases and subsequent skin ulceration [42, 43]. TSB liners may cause volume changes of the residual limb during daily activities [15]. Persons with longer residual limbs or an excess of soft tissue may not be suitable for TSB sockets [18] due to problems with thermoregulation and volume loss, respectively.

43.2.2 Transfemoral

Transfemoral sockets are designed for application on above-knee amputations. The primary goals of a transfemoral prosthesis are to provide comfort in weight bearing, a narrow base of support during standing/walking, and a swing phase close to normal [30]. Transfemoral sockets may be categorised into ischial-ramal-containment (IRC) or sub-ischial (SI) designs depending on whether the socket encompasses the ischial tuberosity or not [30]. Examples of IRC and SI designs may be seen in Fig. 43.3.

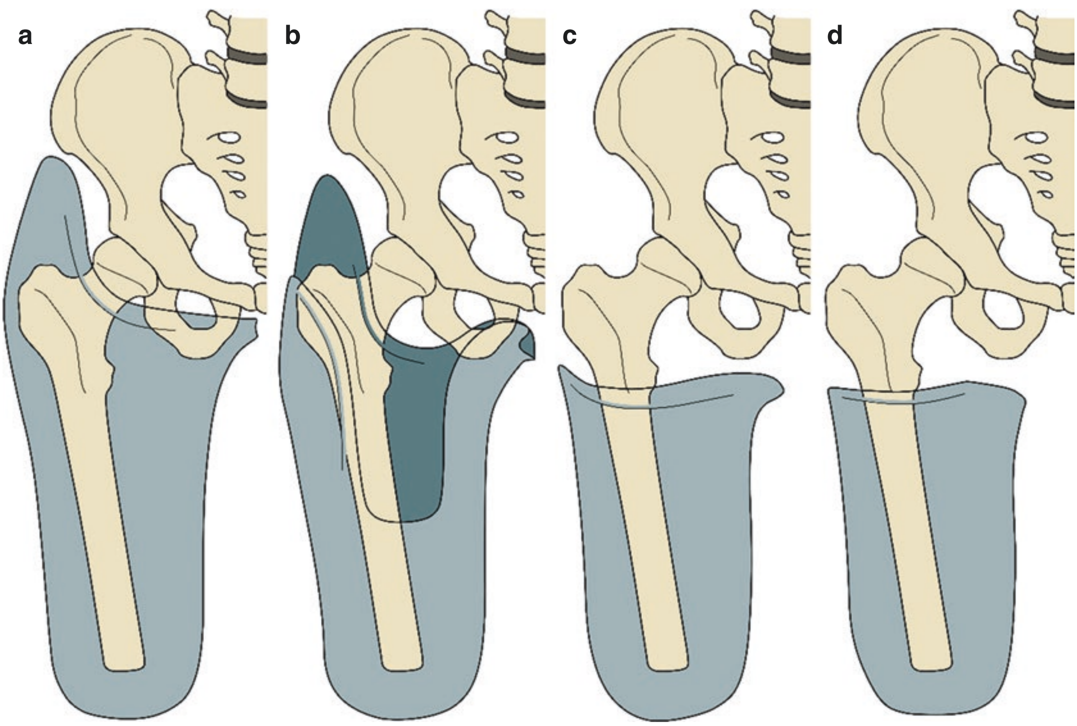

Fig. 43.3 Common transfemoral socket types (**a**) ischial containment, (**b**) ramal containment, (**c**) sub-ischial quadrilateral, (**d**) sub-ischial vacuum assisted suspension

From the 1950s [30, 44] until the 1980s, the most prevalent transfemoral socket design was the sub-ischial German quadrilateral socket [11] using skin suction suspension [30]. The quadrilateral socket featured ischial and gluteal muscle weight bearing [44] on the posteromedial brim [27] and an element of hydrostatic loading [30]. The socket was named for its distinct four-walled shape [44].

During the 1980s, the first ischial containment design was produced [30], variations of which have mostly replaced the quadrilateral design [11]. Ischial-ramal-containment sockets may be sub-categorised into ischial containment (IC) and ramal containment (RC) designs [30]. Most IRC designs are total-surface-bearing and use bony containment (or skeletal lock [45]), through extended medial-lateral brim lines, to improve alignment of the femur and medial-lateral stability [11, 27].

IC sockets tend to be the most commonly used and involve an intimate fit with a narrow mediolateral dimension, in combination with enclosure of the ischial tuberosity and ramus [30]. The majority of IC sockets utilise hydrostatic weight bearing, as opposed to ischial weight bearing, using the ischium for enhanced mediolateral stability of the limb during single-limb support [30]. A common challenge for prosthetists in production of IC sockets is in determining the amount of bony support that the amputee can tolerate [30].

IC sockets are produced in numerous configurations including the Normal Shape-Normal Alignment (NSNA) [46], the Contoured Adducted Trochanteric-Controlled Alignment Method (CAT-CAM) [47] and the Marlo Anatomical Socket (MAS) [11, 27, 30]. The NSNA design includes a narrowing of the mediolateral dimension of the socket to improve support of the femur and to align the prosthesis directly under the femoral head, matching the natural orientation of the femur [46]. The CAT-CAM design introduces undercutting of the trochanter and a fossa in which the ischial tuberosity and descending ramus are supported [47]. The Marlo Anatomical Socket (Ortiz International) limits bony containment to ascending ischial ramus and lowers other proximal trim lines to achieve coronal stability and allow a greater range of hip motion [27, 30].

SI designs have had a resurgence with two recent solutions, namely the SI Northwestern design and the High Fidelity (Hi-Fi) socket [11]. These sockets are advantageous for medium-long stumps and provide lower proximal brim lines [11], increased comfort and improved stability [11]. The Northwestern University Flexible Sub-ischial Vacuum (NU-FlexSIV) socket is formed of a fabric-covered cylindrical liner, a flexible inner socket, a shorter rigid exterior socket and a sealing sleeve that forms a vacuum with the liner [27]. The socket is suggested to improve user comfort without compromising gait characteristics [48]. The Hi-Fi socket is constructed of 3–4 longitudinal struts used to compress the residual limb [49]. The fenestrations between struts stabilise the bone through selective compression, allow the skin to protrude from the socket, and reduce the mobility of the soft tissues [11, 49]. The Hi-Fi socket has shown improvements in walking capacity, activity-specific balance, and mean distance walked in comparison to ICS sockets [50].

Both SI and IRC socket designs are used due to their suitability for different amputations. SI sockets tend to be preferred for their low proximal trim lines and comfort; however, IC designs provide improved mediolateral stability and are beneficial for shorter limbs or those with poor or no voluntary control of the adductor muscles [30].

43.2.3 Through-Knee (Knee Disarticulation)

An alternative to below-knee and above-knee amputations is the knee disarticulation or through-knee amputation. With a lower-limb amputation, surgeons aim to retain as much length as possible [51] to preserve a longer lever arm for ambulation. Below-knee amputations are preferred for their greater functional outcomes and reduced energy expenditure as compared to above-knee amputations [33, 34, 51]. Between 30% and 50% more energy is required to ambulate with a prosthesis following an above-knee amputation compared with a below-knee amputation, resulting in fewer ambulatory amputees [34].

Through-knee amputations provide an end weight bearing residuum [33, 34, 52–54], loaded along normal proprioceptive pathways [55]. These amputations are beneficial in their facilitation of enhanced proprioception, more effective use of the remaining muscles leading to reduced muscle atrophy [34, 53], improved balance, mobility [52, 54], reduced componentry space requirements [55] and longer lever arm [54] leading to lower metabolic costs of ambulation in comparison with ischial weight bearing above-knee limbs [33, 34].

Despite these benefits, through-knee amputations account for less than 2% of all lower-limb amputations in the US annually [34]. This low frequency is attributed to the perception of a high incidence of healing complications associated with the long tissue flaps required to close the wound around the femoral condyles [33, 34]. Consequently, the skin of through-knee residual limbs may be subject to breakdown [52] leaving it unable to bear weight [55].

Through-knee residuum also produce issues with socket fitting [52] and donning/doffing [55] over the bulbous residual limb [33, 34, 53, 55, 56] formed as a result of femoral condyle retention [52]. Furthermore, there is difficulty in positioning of the artificial knee joint [54] due to the extra length of the residuum [55] leading to unequal joint placement in which the prosthesis joint is distal to the intact contralateral joint [34, 53]. Thus, the resulting socket may bear little resemblance to above-knee sockets [56].

A surgical method developed by Mazet, involving shaving of the femoral condyles and excision of the patella to produce a conically shaped residuum, may alleviate some of the issues associated with socket fit [34, 54]. Mazet residual limbs enable use of suction suspension prosthesis (see Sect. 43.2.6 for further information), providing a simpler method of attachment in comparison to typical above-knee sockets [33].

43.2.4 Socket Construction

Prosthetic sockets are generally constructed either as a hard, rigid structure, a soft, flexible material or a combination of hard exterior frame and soft, flexible interior [30, 37]. The material choice can have significant impact on the type of socket fitting [37]. Conventional sockets are produced from a positive model of the amputee's residual limb [30]. The shape of the residual limb is obtained by wrapping a cast around it to produce a negative mould either whilst loaded or unloaded, depending on the prosthetist's preference [17, 49]. A positive mould is then produced from this cast, onto which material may be added or removed to relieve or increase pressure at specific sites of interest [17]. This work is labour and cost intensive, time consuming [8] and requires skill and experience from the prosthetist [19].

Hard sockets are simple, single-walled constructions that provide direct contact with either the skin, a roll-on gel liner or a sock [30]. Hard sockets tend to be fabricated from carbon fibre or rigid thermoplastics [30, 57]. Hard sockets are intended for limbs with firm tissue, stable volume and good skin sensation [30]. They do not absorb shear and do not allow for residual limbs that fluctuate considerably in limb size [30]. Hard sockets are not suitable for individuals with sensitive bony prominences, scar tissue or invaginations [30].

Designs utilising flexible inner sockets offer an interface capable of elastic movements around sensitive bony regions that allow for volumetric accommodations and lower proximal trim lines [30]. The inner socket contains the proximal tissue and is produced from silicone which is beneficial for user comfort [30]. The inner socket is either in direct contact with the user's skin, a roll-on gel liner or a sock [30]. A rigid outer frame is added for support and stability [30]. The outer frame may be fenestrated to increase proprioceptive feedback by exposing the compliant inner socket to any surfaces that come into contact with the leg [30].

An alternative design structure employs panels whose positions can be adjusted through attached tensioning cords [30]. This design allows the user to manually adjust the shape of the socket allowing flexibility, greater reported comfort, minimal trim lines and perceived improvement with respect to control of the prosthesis [30]. However, these sockets are heavier, more difficult to fabricate and can suffer issues with durability [30].

43.2.5 Liners

Liners form the primary interface between residual limb and prosthesis in many socket designs [11, 16] and aid in cushioning the transfer of loads to the soft tissue, which is unaccustomed to weight bearing [15]. Liners aid in absorption of impact and shear forces, cushion the residual limb [16], help to stabilise the soft tissue of the residual limb and assist in compensating with volume fluctuations [30].

Early liners were made of open and closed cell foams [32] such as Pelite [11], with common usage up until the early 1990s, particularly in the transtibial population [15]. During the mid-1990s, roll-on elastomeric liners (see Fig. 43.4) were introduced boasting improved cushioning, suspension, durability, comfort, the reduction of shear forces, more uniform pressure distribution [23, 32], protection against abrasion and the ability to adhere to the skin [11, 15, 58]. Elastomeric liners are usually based on silicone or urethane elastomers [11, 23].

Fig. 43.4 Example of an elastomeric socket liner (image reproduced courtesy of Blatchford)

43.2.6 Socket Suspension

Suspension methods form the basis for attachment of the residual limb to the prosthesis [29, 44] and determine the stability of the stump within the socket [11]. The suspension system has an enormous effect on the function [16, 32], design and fit of a socket [11], however, there is currently no evidence to support a clinical standard [30]. Suspension is also one of the primary factors affecting user satisfaction [11], although little information is available to inform liner choice [18]. The suspension system is chosen by the prosthetist [16] and there is a variety of suspension methods available that may be categorised into cuff mechanisms [44], harness systems and sub-atmospheric systems [23, 30].

43.2.6.1 Cuff Mechanisms

For transtibial amputees, supracondylar cuff suspension is an option that was popular in the 1970s and 1980s [44]. This method of suspension relies on bony lock using the medial condyle of the femur and the supra-patellar aspect of the knee [44]. In order to achieve this form off suspension, the socket must be altered to be supracondylar (SC) in which the femoral condyles are enclosed or supracondylar and supra-patellar (SCSP), enveloping both of the femoral condyles and the patella [11, 15, 29, 35]. PTB-SC sockets improve mediolateral stability and contribute to self-suspension of the prosthesis, PTB-SCSP sockets additionally stiffen the mediolateral walls and provide greater proprioception through force applied proximal to the patella [35].

43.2.6.2 Harness Suspension

Belt-type or harness systems were used historically to secure the prosthesis to the user with a harness comprised of rigid or elastic belts [11, 30]. Prior to the introduction of PTB sockets, conventional transtibial sockets tended to employ artificial joints and a thigh corset [29], sometimes with inclusion of a waist-belt as a suspension method [15]. These systems are now used largely as secondary suspension mechanisms [11] to provide joint stability through reduction of rotation or coronal stabilisation [30].

Early below-knee thigh corset and side bar (TC-SB) suspension methods consisted of a leather or thermoplastic thigh corset and two metal side bars connecting to the socket through bilateral hinges [29]. In modern applications, this form of suspension is limited to amputees with fragile skin, unstable knee joints, short stump length or inability to bear weight on the patellar tendon [29]. The TC-SB suspension adds considerable weight to the prosthesis, increasing overall weight by up to 31% [29].

There are three primary varieties of belt system, namely elastic-belt suspension, the Silesian belt and the hip joint & pelvic belt [30] (see Fig. 43.5). Whilst belt-type suspension offers convenience in their ease of donning, they provide minimal primary suspension [30] and poor stability [11]. Their combination with other suspension systems can be useful [11], for example with individuals with very short stumps [24].

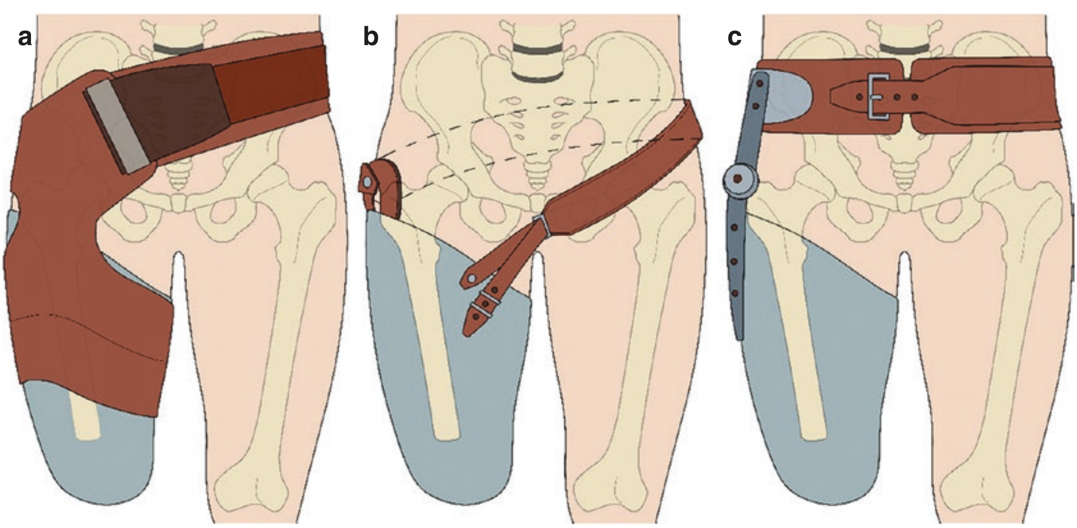

Fig. 43.5 Typical belt suspension systems (**a**) the total elastic suspension (TES) belt, (**b**) the Silesian belt, (**c**) the hip joint and pelvic belt

Elastic-belt suspension and Silesian belts constitute soft belt systems [30] and are comprised of neoprene or cotton [44]. They attach to the proximal end of the prosthesis and encircle the user's waist [44] to reduce rotation, but provide minimal suspension and are only suitable as a primary suspension mechanism for amputees requiring a socket that is easy to don, a socket that demands free movement of air or those who move at minimal cadence [30]. Similarly, hip joint and pelvic belt systems offer stabilisation in the coronal plane that is particularly useful for shorter residual limbs or those with poor abductor muscle control [30].

43.2.6.3 Sub-Atmospheric Pressure Suspension

Sub-atmospheric pressure suspension systems use a negative pressure differential [11] in combination with friction and surface tension to secure the residual limb to the prosthesis [30, 44]. These systems are the most prevalent form of transfemoral suspension [30]. Subclasses of sub-atmospheric systems include suction (using skin suction, hypobaric suction or liners with mechanical locking mechanisms (see Fig. 43.6)) and vacuum-assisted-suspension (VAS) methods [23, 30, 44].

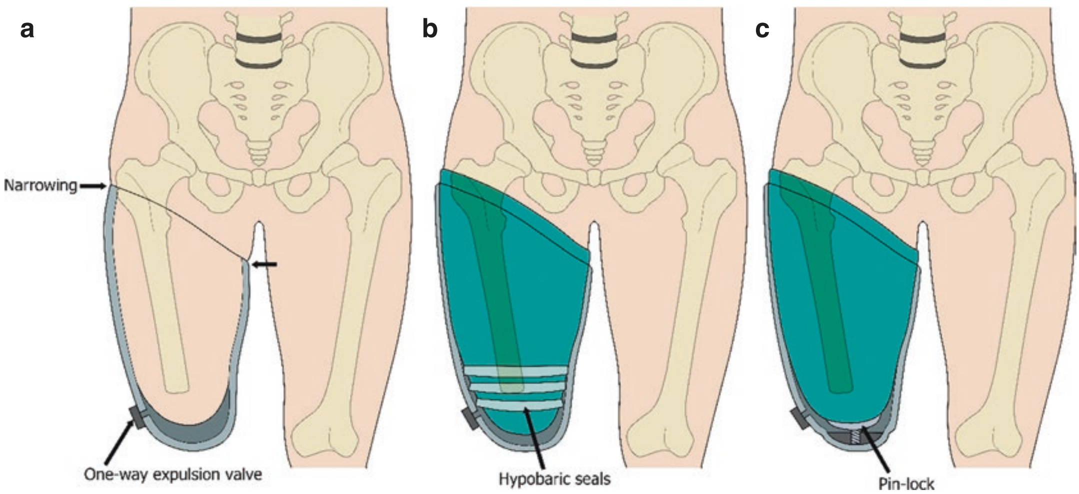

Fig. 43.6 Suction suspension methods (**a**) skin-fit suction, (**b**) suction suspension with a hypobaric liner, (**c**) suction suspension with a pin-lock liner

Suction

Suction suspension is a subclass of sub-atmospheric devices in which the internal pressure of the socket is not actively regulated, however, air is expelled from the socket and prevented from entering [30] usually by means of a one-way expulsion valve [23, 59]. Suction may either be created between the skin and a hard socket [59] or between the skin and a roll-on gel liner and between the gel liner and the socket [30]. Suction sockets are popular among transfemoral amputees, however, are often avoided for transtibial amputees due to bony characteristics compromising the formation of a tight seal [44].

Original suction methods were skin-fit suction [30, 60]. Skin-fit suction is a simple form of suspension in which proximal circumferential reductions provide an air-tight seal against the skin [30]. Typically, a one-way expulsion valve on the distal end of the socket is used to expel air during weight bearing [11, 61], and surface tension is created along the inner socket through contact with skin [30]. Internal pressures are reported to reach approximately −8 inHg during swing phase [30].

Advantages of this suspension method include high levels of proprioception and muscular control [44] due to direct contact with the skin [11] and movement of skin with the socket [30]. Despite this there are numerous disadvantages including poor mitigation of shear stresses which is problematic for individuals with scar tissue or invaginations, inability to account for residual limb volume fluctuations which may compromise suspension, difficulties donning [30] and an increased risk of skin problems [11].

Suction may also be achieved via contact of an elastomeric roll-on liner with the socket wall or through use of internal or external hypobaric sealing sleeves or rings [23, 30, 61]. Suction is induced in the air space between the stump and the distal end of the socket which prevents movement of the socket during swing phase [15]. A sleeve is usually used to seal the air space at the proximal end of the socket and a one-way valve allows air expulsion at the distal end of the socket [15, 61]. Suspension sleeves are made of neoprene, latex or elastomers [44]. One end of the sleeve is fitted over the proximal end of the prosthesis, whilst the other end is rolled over the residual limb [44]. The smooth, flexible and non-porous structure of gel liners enables superior air sealing in comparison to the irregular and porous surface of the skin [30]. An isopropyl alcohol is usually applied to reduce surface tension when donning the liner [30].

Alternatively, liners may be affixed to the socket through distal pin-lock (shuttle lock) systems [23, 44], magnetic-locks or lanyard straps [11, 30, 62]. Pin-lock systems use liners with a metal pin affixed to the distal end used to connect the liner to the distal end of the socket [15, 23, 44]. This suspension method is common for amputees with gel liners and undersized total-surface-bearing sockets [15]. Benefits of this method of suspension include cushioning, torque control and shock absorption resulting from liner use [44]. Shuttle-lock systems are used for both transtibial and transfemoral amputees [44]. Suspension is achieved through suction of the liner and load transfer through the locking mechanism [23]. Mechanical locking liners experience greater socket rotation than hypobaric sealing liners [30].

Vacuum-Assisted Suspension

Vacuum-assisted suspension mechanisms differ from suction systems in that a negative pressure is sustained within the socket at all times [63], as opposed to negative pressures generated only during swing phase [23, 30]. VAS systems may produce pressures as low as −25 inHg through mechanical/electrical activation [11, 63], typically using an external mechanism to create the vacuum and a one-way expulsion valve to maintain it (see Fig. 43.7) [30]. The lower the pressure, the greater the suspension force [30].

VAS suspension reduces pistoning motion between the residual limb and the socket [23], improves proprioception [18], increases control, reduces volume loss of the limb [18, 23], improves wound healing and allows for lower proximal socket brims [11]. VAS is reported to include lower interface pressures during stance phase [23], enable improved walking ability and greater gait symmetry [11, 18]. Despite these strengths, VAS systems have not garnered widespread acceptance due to difficulties in fabrication, maintenance of the proximal vacuum seal, donning, doffing, increased socket weight and difficulty maintaining high pressure differentials over time [11, 30].

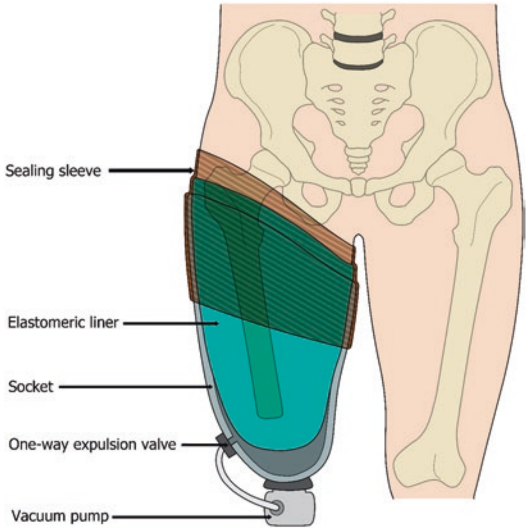

Fig. 43.7 Typical VAS suspension format

43.2.7 Socks

Socks are inexpensive, easy to maintain [64] compressible textiles that may be infused with thermoplastic elastomer gels [23]. They are available in numerous sizes, thicknesses and materials, and may be worn individually or layered [64]. Socks may be used to protect the skin of the residuum [65] or alter the fit of the socket, an approach usually taken when the socket does not fit well due to limb volume changes [16, 23, 25, 30, 64, 66–69]. However, this is inconvenient [67] and the impact of this is not well understood [68]. Socks are commonly used in sub-ischial transfemoral sockets [30] to allow for the transfer of air molecules between the socket and the liner to provide a uniform internal socket pressure [30]. Occasionally, they are used as the primary interface between the limb and the socket [23].

43.2.8 Orientation/Alignment

The alignment of a prosthesis may be defined as the relative position and orientation of its constituent components [10]. Prosthesis alignment is a further factor impacting the load transfer between the socket and residual limb [18]. A poorly aligned prosthesis may negatively impact socket fit, altering the pressure distribution and thus, potentially leading to discomfort, pain and tissue breakdown [10]. Alignment has implications for optimal function of the prosthesis [8] impacting walking ability [10], gait stability, fluency and energy expenditure [11]. Alignment also influences the stresses observed in the contralateral limb [10].

Alignment of the components must be optimised with regard to the rotation and translation of each prosthesis component in each plane [11]. As with the socket fitting process, alignment of the prosthesis lacks consistent practises and protocols [18]. Acceptable alignment appears to range greatly [10], and daily alignment is problematic in its inconsistency [8].

43.2.9 Components

The function of the socket is also affected by the remaining prosthetic components which impact factors such as metabolic cost, muscle activation and gait symmetry [11]. These components include items such as the replacement ankle and knee joints, connective pylons and prosthetic feet [11].

43.3 Overview of Amputee Issues

The levels of amputation experienced greatly affect quality of life and perceived ‘normality’ [4]. Whilst unilateral transtibial amputations enable a largely ‘normal’ life [4], there is an association with diminished physical health and reduced capability to work [4]. Bilateral transfemoral amputation is associated with significant mobility issues leaving many individuals unable to work following injury [4].

Issues with prosthetic sockets evolve from a complex interplay of socket shape, materials and suspension mechanism [11]. These interconnecting factors can each influence normal and shear stress load distribution, internal temperature of the socket and volume fluctuations resulting from exposure to such conditions [11]. The skin of the residual limb is not adapted for exposure to the heat, humidity and stresses it is subjected to [70]. Resulting issues are further complicated by an ageing amputee population [30] and implementation barriers such as small market size, high costs and intellectual property restrictions [8]. The following section explores common issues reported by amputees and the factors that may impact their experience of lower-limb prosthesis use.

43.3.1 Fit and Pressure Distribution

The fit of a socket is of paramount importance to the success of the prosthesis and the overall rehabilitation process [11, 19]. Socket fit has significant implications on local stresses skin health and the potentially, the requirement for re-amputation [15]. Poor fitting sockets are common [69] and can encourage displacement of the residual limb which may reduce the areas upon which force is transferred leading to abrasion, blisters, cysts and infections [27, 37] and ultimately revision surgery [13]. Additionally, prosthetic fit has implications for gait, activity levels and loading of the contralateral limb in unilateral amputees [10], as well as tolerance of the prosthesis.

The capacity to produce a well-fitting socket is heavily dependent on the skill and experience of the prosthetist [18, 24, 69]. To date, there are no universal clinical standards in relation to the fitting of prosthetic sockets, and no single socket design can accommodate all amputees [30]. The fitting of the socket should resolve forces to ensure stabilisation and support are present throughout the walking cycle, without compromising comfort [24]. There is a suggestion that the socket should intimately fit the contours of the residual limb to accomplish this aim [30]. However, the concept of optimal fit is poorly defined and has not been standardised [30, 71].

Improper fit can lead to pistoning [15, 30], the relative vertical displacements between the residual limb and the socket [11] can take place at the skin/socket, bone/socket or liner/socket interface [11]. Pistoning can lead to loss of prosthetic control, gait deviations, discomfort and skin irritation [30]. The degree of pistoning present is dependent upon the friction coefficient between the skin and socket, the skin and liner or the liner and socket [11]. If the friction coefficient is too low, then the surfaces will slip leading to abrasion [11]. If the friction coefficient is too high, then shear stresses will be generated, leading to tissue distortion and risking tissue breakdown [11].

The transfer of load between the residual limb and the prosthesis is a crucial factor for socket quality, function and user comfort [11, 15, 17, 23]. The residual limb is cyclically subjected to high stresses during the gait cycle [11]. Excessive stresses applied to the limb over a long period of time in a non-uniform fashion can lead to pressure sores [17], sensitive skin, irritations, pain and partial or total vascular occlusions, which may impact thermal regulation of the limb and encourage tissue maceration [11].

The stresses acting on the residual limb are a combination of normal and shear forces [13, 15, 69, 71]. The contact pressures or normal stresses are influenced by a combination of factors [24] such as socket features and material properties, in addition to the form of the amputee's residual tissues [11]. The stresses can vary greatly between amputees [19] and alter with time due to changes in residual limb shape and tissue properties [11]. Friction between the residual limb and the socket produces shear stresses which lead to tissue deformation and increase the risk of injuries [11]. Shear stresses may lead to reduced blood flow within the skin [72] which can lead to lesions on the skin [15].

The capacity of tissue to bear these loads is related to the condition of the skin, its vascular state, the strength of the bony tissue and joint function [37]. Additionally, certain regions of the residual limb are more able to tolerate higher pressures than others [11]. The stress thresholds deemed tolerable for each subject are highly subjective [69] and vary depending on factors such as their bodyweight, residual limb musculature and levels of activity [11].

Prosthetists attempt to control interface stresses through alteration of the socket shape, choice of socket and liner materials, componentry and alignment of the limb [13]. Pressures must be balanced across the socket to prevent socket migration and compromise prosthesis control [30]. Care must also be taken to avoid tissue bulging over proximal trim lines as this may lead to skin breakdown, blisters, subdermal cysts and discomfort [30]. The socket fitting process is an iterative and labour-intensive process [69].

43.3.2 Volume Fluctuation

Fluctuation of the residual limb volume is an issue that may affect socket fit [63] and thereby, alter the distribution of normal and shear loads across the stump [11, 16, 23], exposing it to risk of injury [16]. A loss of volume can lead to displacement of the residual limb, an increase in pistoning [11] and abrasion of skin overlying bony prominences and tendons [37]. Conversely, an increase in volume may exert higher pressures and shear stresses on the skin [11]. Secondary effects may include alterations to thermoregulation, increased sweating, formation of skin irritation and odour [11]. Management of the socket pressure distribution under the effect of these volume changes is an on-going problem [16].

Volume changes may occur over a period of hours to months [11, 25, 32, 63]. This issue afflicts both new and mature residual limbs [63]. Volume reduction may result from liquid movement [11] due to loss of oedema, loss of body weight [12] or atrophy of the muscle tissue [37, 63, 69]. These changes encourage the residual limb to settle deeper into the socket [37]. Volume increase may result from increased oedema [37], increase in body weight [12] or hydration [69]. Women have also reported temporary swelling of the residual limb during their menstrual cycle [12, 63].

Daily volume fluctuations are reported to occur in the range of −11–7% and may be accentuated by the absence of a liner [11, 25]. A volume increase of 3–5% is enough to cause discomfort and issues with donning and doffing the socket [63]. The rate of volume change due to fluid movement is affected by the socket size and method of suspension and can be up to 0.10–0.12 mL/min when standing and 0.20–0.30 mL/min after motion [11]. Studies suggest that negative pressure socket systems tend to induce a gain in volume, whilst non-vacuum systems mainly cause volume losses [11].

43.3.3 Temperature and Thermoregulation

Excessive heat within the socket is a common problem for lower-limb amputees [41], reported to affect up to 53% of prosthesis users [73]. Heat and perspiration are frequently cited as significant contributory factors for reduced quality of life [20, 40, 41]. The body's ability to regulate temperature is affected by the decreased skin surface area post-amputation [39], atypical loading conditions [74] and heat generation of the limb [40]. Patient factors also play a role; factors such as age, sex, activity level [20] and vascular insufficiencies may further impact the thermoregulatory process [11, 39].

The issue of heat management is compounded by the thermoregulatory system's tendency to increase perspiration in full-contact sockets due to the thermal barrier created by liners and sockets [11, 39, 74]. The materials commonly used to construct prosthetic sockets and liners have poor thermal conductivity and moisture permeability properties [11, 20, 38–41]. Additionally, the closed environment of the prosthesis prevents the evaporation of sweat [40]. Temperature related issues may be exacerbated by the use of roll-on gel liners which further reduce ventilation leading to greater amounts of sweating [15, 30].

Elevated temperature affects the mechanical stiffness and strength of the skin [69]. An increase of 1–2 °C is sufficient to produce skin problems including discomfort, skin maceration, bacterial invasion and infection [11, 20, 41], allergic contact dermatitis, verrucous hyperplasia [74] and skin irritation [20, 40]. Slight moisture on the skin has been associated with increased susceptibility to blistering in comparison to wet or dry skin [20, 39, 40] that may elevate shear forces through alteration of the skin's coefficient of friction, encouraging greater cellular permeability providing greater susceptibility to skin irritation [69].

Temperature increases may also affect the fit of the socket and encourage slippage of the socket through increased perspiration, altering the stress distribution [11, 12] and causing loss of suspension [20, 41]. Consequently, daily use of a prosthesis and participation in recreational activities may be compromised by excessive heat [20]. Disruptions to suspension demand that the amputee halt their activity remove the socket and remove the moisture on the residual limb and socket interface [20].

There is currently little information pertaining to temperature values desirable at the skin level of prosthesis users [11] and it is difficult to quantify what 'cool' or 'warm' means to individual amputees [74]. Skin temperatures are affected by muscle metabolism and perfusion [11]. Temperature is generally lower in regions of low perfusion and is higher in regions of high perfusion and in close proximity to large muscles [11].

43.3.4 Skin Conditions and Infection

Complaints of discomfort and skin breakdown are common amongst both civilian and military amputee cohorts [10, 70, 75, 76]. Dermatological problems affect approximately one-third of the amputee population using prostheses. These include allergic contact dermatitis (ACD), infection and constitutional skin diseases such as psoriasis [70]. In a survey by Meulenbelt et al. 63% of the 507 participants reported one or more skin problems in the month prior to receiving the questionnaire and 25% of these reported less frequent prosthesis use as a consequence [75].

The residuum of a prosthesis user is particularly prone to skin problems due to its exposure to unnatural normal and shear stresses, increased humidity and prolonged contact with the prosthesis chemical compounds [77]. The presence of humidity and occlusion of the residuum may increase sensitivity to the prosthesis, moisturising creams and medicaments [70]. Skin irritation and discomfort may also be encouraged due to the poor classification of liner chemical compounds in which different chemical structures may be branded under the same product name [18].

Common skin problems include allergic profuse sweating, sensitive skin [75], contact dermatitis, bullous diseases, epidermal hyperplasia, hyperhidrosis, infections, malignancies and ulcerations [77]. Skin problems resulting from poor socket fit include development of conditions such as calluses, lichenification, erythema, acroangiodermatitis [77] and follicular hyperkeratosis [70]. The residual limb is prone to surface infection through superficial bacterial infections, most of which are temporary, or allergic reactions, which may require a change of socket materials [37]. Poor residuum hygiene may encourage bacterial infection; however, infection may occur regardless of the level of hygiene [37]. Roll-on liners often increase difficulty with regard to maintaining hygiene [30] and thus may leave users more prone to infection. Infections may also be encouraged through development of oedema in which the skin of the residuum becomes weakened and may break down [37].

Oedema is a notable issue for lower-limb amputees in which fluid accumulation in the residual limb tissues produces a hard swelling at the end of the limb [37]. This condition is problematic in that it may weaken the skin, leading to breakdown and infection [37]. Persistent residuum oedema has implications for the formation of ulcers and development of verrucous hyperplasia [70]. Prosthetists may ensure that contact is made across all areas of the residuum [16] to prevent onset of oedema [30].

The implications of skin conditions such as skin rashes, blisters [17] and ingrown hairs can significantly impact an amputee's use of their prosthesis [12, 70, 75] by reducing the length of time for which the prosthesis is worn [27, 75]. This may then impact the ability to perform household tasks, engage in social activities or participate in sports [27]. The greater the number of skin problems reported by lower-limb amputees, the greater the reported influence of these issues on the use of the prosthesis [75]. In some cases, the clinician treating the skin condition may instruct the amputee to avoid use of their prosthesis to aid in healing [75]. Skin conditions may also progress into chronic infections which in the worst cases can demand re-amputation [11]. Good skin health on the residual limb is crucial for normal use of the prosthesis [70]. In blast amputees, skin health has further issues as a result of burns [78, 79] and the widespread use of skin grafts [80].

43.3.5 Other Musculoskeletal Pathologies Related to the Socket

Musculoskeletal pathologies are common secondary complications for lower-limb amputees [10] that may have significant long-term effects as activity increases or is sustained over time [10]. Most prosthetic users present with at least one gait deviation as a result of poor socket fit, poor gait habits or compensation for physical limitations [10]. Resultant alterations to gait have implications for the stress and strain profiles throughout the skeletal and soft tissues of the body and over time can lead to increased pain and disabling degenerative joint diseases [10]. Poor socket fit is related to:

- The high risk of back pain in amputees with 52–63% experiencing persistent, problematic back pain shortly after amputation [81, 82] and
- A higher level of energy consumption [11].

43.4 Understanding and Optimising Socket Fit

There is a complex interplay between residuum physiology, the socket and gait characteristics. Ensuring optimal socket fit requires a good understanding of the pressure distribution within the socket and identification of acceptable pressure thresholds [11, 15]. The monitoring of normal and shear stresses is one of the main challenges [11] for prosthetic sockets and is becoming increasingly important [15]. Many different measurement techniques have been applied to lower-limb sockets over the past 50 years for determination of stress profiles [15]. Prior to this, quantitative measurement techniques did not exist, and pressure was assessed through skin colour observed through clear check socket during static weight bearing [15]. Blanching of the skin is indicative of high contact pressures, however, it does not provide information on the level of loading and requires a highly skilled prosthetist for interpretation [15].

This section illustrates some of the solutions posed to date by various research, clinical and commercial groups. Proposed solutions have taken the form of measurement devices, technological modifications and innovations and surgical interventions.

43.4.1 Qualitative Assessment

A survey of 94 participants, 50 major lower-limb amputees and 44 rehabilitation team clinicians, investigated the impact and frustrations of socket fit during prosthetic rehabilitation [83]. When asked in a free text question what the biggest impactor on rehabilitation was for the individual, 62% identified socket fit or a related issue [83]. When asked to specify the largest impact from their own list, 56% identified a socket fit or socket related issue [83]. In detailing their frustrations with prosthetic sockets, 56% of amputees and 52% clinicians identified socket fit over options such as weight of the socket and sweat from high socket temperature as problematic [83]. An extended telephone interview with a subset of 18 amputee and clinician participants explored the impact on their function and experience when describing their issues with socket fit [83], highlighting pain, pressure sores and volume fluctuation of the residual limb. One participant stated, 'the nearest analogy I can think of is wearing a shoe that doesn't fit. Never mind if it's a Jimmy Choo or whatever brand name. If the shoe is too small or doesn't fit, it hurts. You can't walk properly. It's very similar with socket fit'. Comfort was a theme that was established from the prosthetists' interviews also: 'each component of the limb is equally important, but the socket is the only one that gives an absolute "no I can't use the prosthesis"'. The clinicians were divided on the specifics of the socket fit issues they referred to: physiotherapists were focussed more on gait re-education and rehabilitation, whereas the prosthetists named socket fit as the issue directly [83].

The interviews demonstrated that despite all groups using the same term, more specifically 'good socket fit', the interpretation of the phrase is different for each of the groups and the individuals themselves. This makes it difficult to solve the problem of ill-fitting sockets, because addressing issues described by one group may not solve the issues for another [83]. The use of the same term does not guarantee that communication is clear, which may result in delays or inhibit the solving of socket issues. There was a perception that there was not enough focus on socket fit [83]. One amputee participant said, 'socket fit is something that's really important, but, of course, not as glamourous and therefore gets forgotten'.

43.4.2 Computational Methods and Socket Fit

Computational methods such as finite element analysis (FEA—see Chap. 5) and artificial neural networks (ANN) have been used to quantify load distributions in prosthetic sockets [11, 16]. This has been applied to quantify pressure, normal stress and shear stress at the interface as well as within tissues [69, 84], both statically and dynamically [15]. These methods may offer value with regard to applications such as evidence-based socket fitting, analysis of prosthetic limb options, identification of risk factors, translation of interface loads to underlying tissues, changes in soft tissue compliance and the effects of atrophy on socket fit [69]. As FEA techniques can use parametric analysis [84], a number of studies have explored the combination of FEA techniques with computer aided design (CAD) and computer aided manufacturing (CAM) processes to aid in socket design [69, 84–86]. Additionally, FEA has been explored in combination with sensors to act as a portable stress monitor [87] or in the investigation of internal mechanical conditions of the residual limb [88, 89].

Combination with ANN-based approaches has allowed researchers to form general transfer functions that allow characterisation of the complex internal socket environment [90], predict residuum/prosthesis boundary conditions from external strain gauges [26] or identify rectification zones within sockets [22]. These frameworks aim to expedite the iterative process of socket design and production.

Like all uses of FEA, the quality of the predictions is highly reliant on geometrical and material assumptions, loading conditions [91], choice of temperature and friction models [69]. Difficulties exist in the description of objects such as the residual limb and prosthetic interface [84]: biological soft tissue is irregular in geometry, non-homogenous, anisotropic, viscoelastic and time-dependent in its response [84], meaning that many assumptions need to be made using FEA; another assumption is that prosthetic sockets are mostly modelled as infinitely stiff [69, 91]. Very few approaches consider full dynamic loading conditions and the time dependency of soft tissue response or the implications of thermal conditions and moisture [69].

Also, computational analysis has a tendency to focus on common transtibial and transfemoral techniques and often does not factor in interfaces such as compression socks [69]. Additionally, these models require extensive evaluation through comparison with experimental measurements before they can be considered reliable [13]. A significant drawback of ANN techniques are the vast amount of training data and large number of input devices [15] required to achieve reliable results [11].

Whilst computational methods do not offer a complete, continuous picture of stresses within the socket, they do provide benefit in terms of estimating the implications of loading structures such as bony tissue. More information regarding such applications can be found in Chap. 44.

43.4.3 Sensing Methods

There is a clear demand for enhancing our understanding of the relationship between residual limb shape, the socket structure and the method of suspension [8, 18]. There are numerous measurement techniques for socket load mapping, including strain gauges, piezoresistive transducers, capacitive sensors and optical sensors [11, 15], yet none are in common use.

Development of suitable sensors for quantification of fit, comfort and stability is a challenging problem with few guidelines [8, 15]. Such systems must not interfere with socket function [92] or negatively impact user comfort [11] and sensor mounting should be easy to accomplish [15]. As such, sensors must be compact, flexible and able to support application to curved surfaces [92]. They must also be capable of providing high accuracy and sensitivity, present low hysteresis and drift errors and sufficient monitoring intervals [15]. The inclusion of shear measurement is also of great importance to fully understanding the conditions within the socket [15]. For application in clinical scenarios, technology must not negatively impact consultation time, process or cost [69].

The development and challenges associated with in-socket stress sensors are illustrated in the following section.

43.4.3.1 Strain Gauges

Strain gauges (SG) are small silicone or metal devices that produce a change in electrical resistance when subjected to any mechanical strains [15, 93]. As such, they measure the changes in length. They have high accuracy and high sensitivity [11, 15]. They are typically applied in one of two formats to detect stresses within prosthetic sockets, namely attached to a diaphragm or connected to a detection beam displaced by a cylindrical plunger piston [11, 15, 71, 94].

Diaphragm SG transducers are beneficial in their small size, low profile, good range sensitivity [15, 71, 94]. However, they are only capable of measuring normal stresses [15], are unable to accept a large pressure gradient across the sensor face [94], demand extensive wiring in array format and experience issues with crosstalk due to the rigidity of the backing material [71]. Boundary issues are particularly pronounced when applied to curved surfaces [11].

Piston-type SG transducers enable measurement of normal stresses, shear stresses or a combination of the two based on their configuration [13, 94, 95]. The piston structure is beneficial in its reduction of crosstalk and edge effects; however, it adds considerable bulk to the socket and requires significant, cumbersome instrumentation [13, 94] that may restrict motion and distort local stresses [15].

Regardless of format, strain gauges are sensitive to humidity and heat changes, often demanding use of a Wheatstone bridge [15]. Placement usually requires extensive modifications to the socket in the form of hole drilling and the attachment of external diaphragm sensors [94] or cylindrical pistons [15]. This task is laborious, permanently modifies the socket and may alter the pressure distribution within the socket [15]. The sensors also tend to demand large amounts of power [15]. As such, SG transducers are limited to a research tool rather than a clinical tool [15].

43.4.3.2 Piezoresistive Sensors

Piezoresistive sensors typically utilise a material in which application of pressure results in a change in electrical resistance [15]. Like strain gauges, this change in resistance may be converted to a voltage through an instrumentation circuit such as a Wheatstone bridge [15]. These sensors are characteristically constructed as force-sensitive-resistors (FSRs) in which the pressure sensitive material, an elastomer, conductive ink, conductive rubber or carbon fibre, is sandwiched between two flexible film layers [15] with a flexible circuit (see Fig. 43.8).

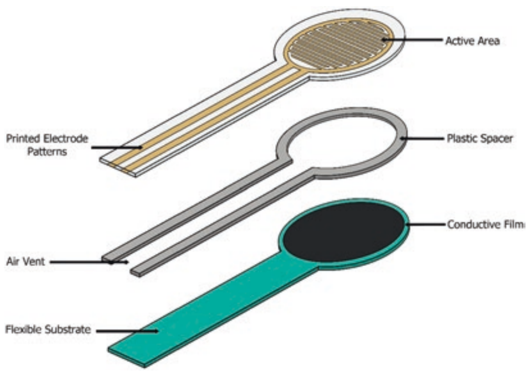

Fig. 43.8 Typical FSR construction

Piezoresistive transducers are advantageous in that they afford the construction of thin, flexible structures that offer good sensitivity [96], are easy to use and able to map large areas of the residual limb [11, 15, 97, 98]. They can be positioned within sockets without modification to the socket or the requirement for sophisticated electronics [15].

However, piezoresistive sensors cannot measure shear stresses [11, 99], they require large arrays of sensors to map significant regions of the residual limb [15] and they are susceptible to high errors when applied to a curved surface [96, 97]. Commercial devices suffer from sensor drift, hysteresis, temperature sensitivity, exhibit low frequency response in comparison to capacitive sensors and have accuracy issues due to creasing of the senstive elements [15, 96, 97, 99].

43.4.3.3 Piezoelectric Sensors

Piezoelectric sensors utilise the piezoelectric effect in which specific materials, subjected to mechanical deformation, produce an electric potential [100–102] (see Fig. 43.9). The voltage generated by the material is directly proportional to the applied force [100]. Materials that exhibit piezoelectric properties include quartz, berlinite, tourmaline crystals, ceramics such as zirconate titanate (PZT) and polymers such as polyvinylidene fluoride (PVDF) [101, 102]. Quartz crystals and ceramics generally display better piezoelectric properties; however, polymer options are popular due to flexibility, dimensional stability, low weight, workability and chemical inertness [101, 102].

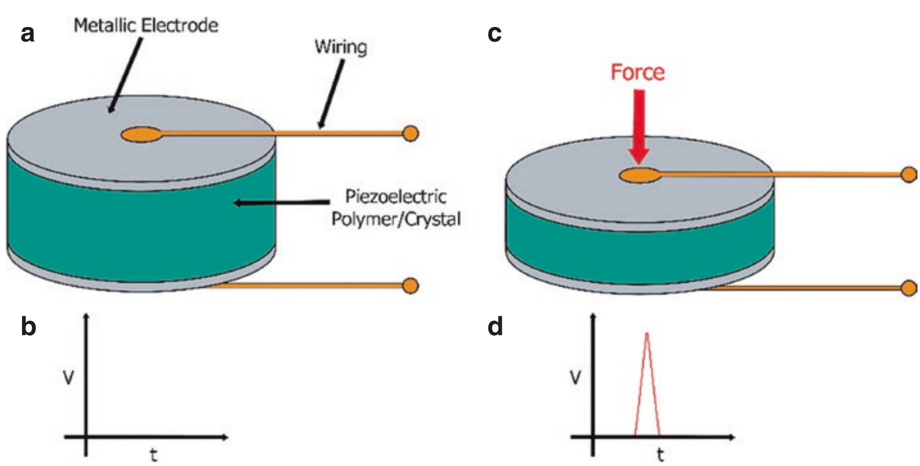

Fig. 43.9 Piezoelectric sensor working principle (**a**) unloaded polymer/crystal, (**b**) unloaded voltage response, (**c**) loaded polymer/crystal, (**d**) loaded voltage response

Piezoelectric sensors produce a sizable voltage when stressed, demonstrate high sensitivity, high bandwidth, high reliability and as a passive sensor, do not require a power source [102]. They also have a good high-frequency response that makes them ideal for measuring vibrations, however, due to their large internal resistance causing developed charge to rapidly decay, they are unable to observe static forces [100, 101]. Piezoelectric sensors are temperature sensitive [102].

These have been used in lower-limb prosthetic sockets [103, 104], however, it is unclear what their static response is and they often display inferior linearity and poor operating range [105].

43.4.3.4 Capacitive Sensors

Capacitive sensors are formed of a dielectric material sandwiched between parallel conductive surfaces [15] and can measure both normal and shear forces [11] which may be accomplished through exploitation of two displacement principles [15, 101] (see Fig. 43.10). The first principle is based on a change in the overlapping surface area of the plates [15, 101]. This technique provides high precision and constant sensitivity [101]. The second principle relies on the change in distance between the plates as the transducer is compressed [15]. Novel GmbH are the only commercial suppliers of capacitive sensors for prosthetic applications, providing normal stress measurement only [11, 14, 106, 107]. A number of tri-axial stress sensors have been reported for use in lower-limb sockets, enabling measurement of both normal and shear stresses [108–110].

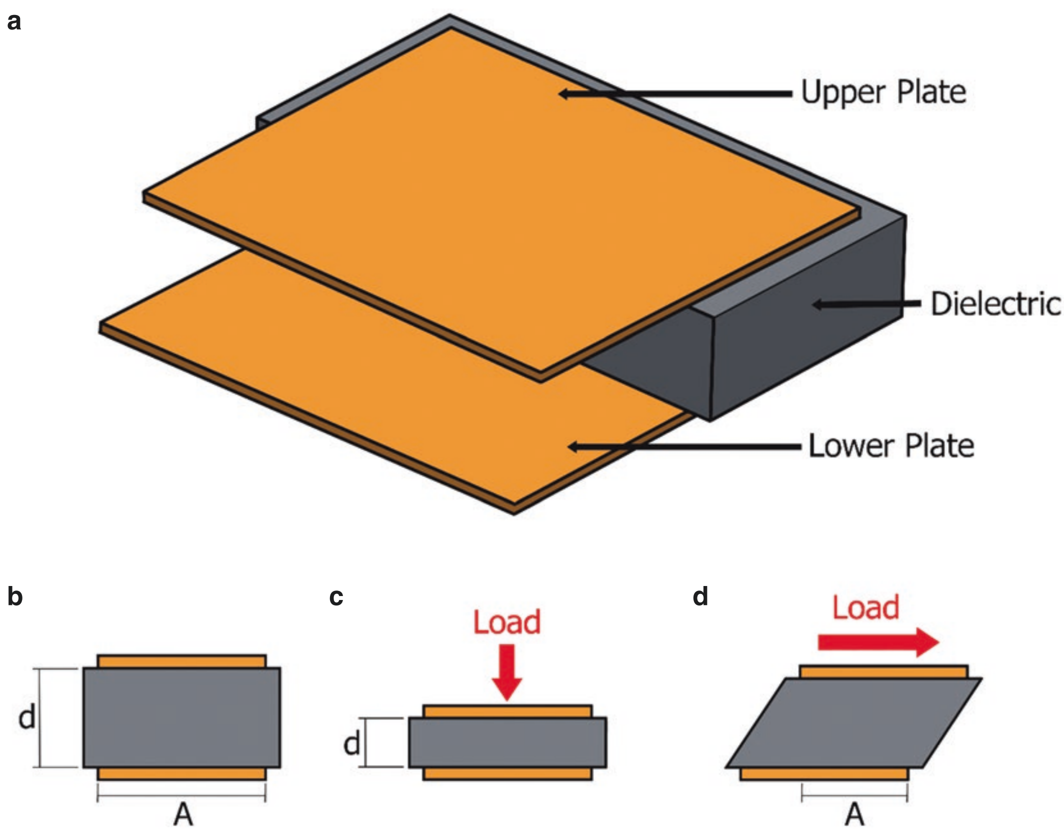

Fig. 43.10 Capacitive sensor construction (**a**) structure, (**b**) unloaded state profile view, (**c**) axially loaded profile view, (**d**) shear loaded profile view

Capacitive sensors offer a high resolution, large dynamic range [101], high sensitivity, good frequency response [15] and good linearity [108]. They tend to exhibit superior drift and hysteresis characteristics to alternative piezoresistive sensors [11] and are less affected by temperature [15].

Significant drawbacks of capacitive sensors include non-negligible crosstalk noise when arranged in an array, the need for complex compensatory electronics and their high fabrication costs [11, 15]. Exploration of rapid prototyping techniques, however, offers a potential solution to the production of complex shapes at low cost [108]. Generally, these devices are also hindered by their rigid substrates [109] which limit their application to contoured socket structures [15].

43.4.3.5 Optical Fibre Sensors

Fibre-optical sensors have been used for numerous medical applications since the 1960s [15, 93, 111–123] due to their small size, high spatial resolution, ability to be multiplexed [124], high sensitivity, resistance to water and chemicals, biocompatibility, ability to be applied to curved surfaces and immunity to electromagnetic interference [92, 111, 112, 115, 116, 125–127].

Fibre-optic sensors tend to consist of three core components, namely a light source, a light modulator and a detector [128]. The sensing techniques utilise properties such as intensity, phase, frequency, polarisation [126] and wavelength modulation [111, 128]. Fibre Bragg Gratings (FBGs) are a variety of fibre-optical sensor that have gained popularity for pressure sensing in the field of prosthetics [93]. FBGs utilise gratings in the optical fibre core at a fixed spacing or period [93, 111] (see Fig. 43.11). Mechanical stimulation of the fibre alters the pitch of the grating, thus causing a single wavelength to be selectively reflected [93, 111, 115, 124, 127, 129].

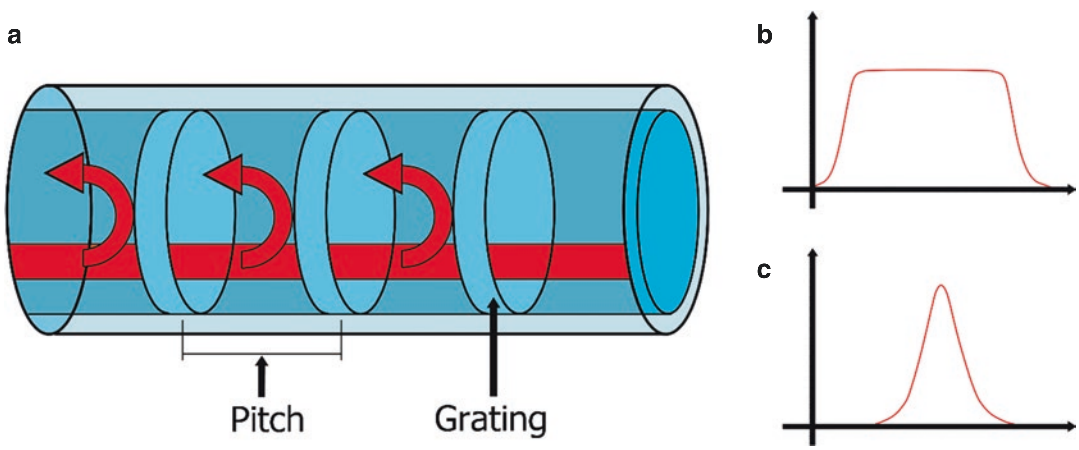

Fig. 43.11 FBG operational principle (**a**) structure of the grating, (**b**) input light signal, (**c**) output light signal. Gratings reflect light based on their pitch

Several research groups have developed uniaxial [112, 125] and tri-axial [130] FBG sensing pads for detecting stress in prosthetic sockets following preliminary work by researchers from the University of Malaya Centre for Applied Biomechanics in 2013 [129]. These sensors tend to consist of a FBG embedded within a polymeric sensing pad [15, 129, 130]. The polymer pad is usually produced using polydimethylsiloxane (PDMS) and allows the sensor to mimic skin behaviour due to its flexibility and elasticity [112], enables attachment to irregular shapes and geometries, provides a biocompatible protective layer [112, 114] and increases the sensitivity [120].

FBGs have a number of distinct advantages over alternative sensing methods. There is a suggestion that FBG sensors exhibit a higher accuracy than intensity modulated optical sensors [111]. FBGs may be manufactured from silica glass, a dielectric material with inherent immunity to electromagnetic interference (EMI) [93], providing biocompatibility, non-toxicity and making them chemically inert [127]. FBGs can be produced as small devices [127] with a high sensitivity to strain [93]. Temperature and pressure dependencies are both linear, unlike many electronic or mechanical sensors, allowing for simple compensation methods for these factors [93].

The wavelength encoding of data allows for self-referencing, advantageous for multiplexing, however, this also requires extensive multidisciplinary expertise to properly extract data from the sensors [69, 126, 127]. This process is yet to receive a standardised procedure as depending on the application, wavelength shifts may represent different parameters due to cross-sensitivity to strain, temperature and refractive index [127]. External connections may be problematic [111] and there is currently a demand for detector miniaturisation as existing devices are cumbersome and limit practicality outside a clinical setting [127]. Additionally, many applications require very high resolutions which demand either high-cost spectrometers or specifications that are not currently available [127].

43.4.3.6 Optoelectronic Sensors

Optoelectronic sensors are an alternative approach to fibre-optical sensors and tend to incorporate a printed circuit board (PCB) with a light emitting diode (LED) emitter and photodiode [15] or phototransistor [131] receiver with a deformable structure that alters the light intensity received by the detector [15]. Examples of optoelectronic devices for the measurement shear stress and a combination of normal stress and shear stress have been developed by a number of research groups in relation to applications such as lower-limb prosthetic sockets [92, 131, 132]. Optoelectronic sensors are advantageous in their ability to measure both normal and shear stresses and their reliance on surface-mount components that may easily be integrated into printed circuit boards [131, 132]. However, whilst FBGs are immune to electromagnetic interference, optoelectronic devices are still susceptible [15] and there is also a risk of damaging the electronic components [11, 15].

43.4.3.7 Sensor Mounting Techniques

Sensors are positioned within the socket using a variety of mounting techniques including placement on the socket wall, insertion into the socket or embedding into the socket wall [11, 15].

Mounting on a socket wall has typically been used in conjunction with cylindrical piston-type transducers housing sensors such as strain gauges, attached via holes drilled in the socket wall [94]. This technique may be used for normal and shear stress measurement; however, it introduces considerable bulk to the socket which may impede motion, distort stress distributions and limit the spatial resolution [13, 15, 71, 94, 95].

Transducers may be inserted between the residual limb and the socket, between the residual limb and the liner or between the liner and the socket [15]. This mounting technique is advantageous in that in does not require permanent modification to the socket, however, sensors must be thin enough to accommodate insertion [15]. Strain gauges, piezoresistive, piezoelectric, capacitive and optical sensors have been mounted in sockets using this approach [15, 71, 104, 133, 134].

Researchers have also explored the potential for embedding sensors directly into the socket wall [15] or into a liner [132]. The intention of this mounting method is to apply internal strains within the socket wall to the transducers [15]. It is unclear, however, what effect this mounting method has on the load distribution within the socket.

43.4.4 Optimisation of Interfacial Stresses

A solution to the issue of interfacial stress distribution within lower-limb sockets is to apply variable stiffness or heterogeneity to the socket in which stiffer materials are placed near soft portions of the residual limb and compliant materials are positioned near bony prominences [11, 69]. Researchers have explored the use of rapid prototyping techniques to introduce variable impedance to the socket [17] and overcome some of the issues associated with their required manufacture [15].

Rapid prototyping, in combination with CAD methods [16], may reduce socket fabrication time from days to hours through elimination of traditional fabrication steps such as development of negative and positive moulds, hand lamination and finishing procedures [19]. Consequently, the costs of manufacture may be lowered [15]. Socket forms can be generated directly from subject-specific digital morphology data with direct integration of additional socket components [19]. Additionally, complex geometry can be introduced to the design without incurring difficulties and additional costs in manufacturing [19].

Fabrication frameworks have been proposed by researchers using selective laser sintering (SLS) [19] and 3D printing techniques to produce variable-impedance sockets [17]. These methodologies were advantageous in their allowance for controlled modifications to shape and structure providing variable compliance and reduced interface pressures [17, 19]. Significant drawbacks included the requirement for numerous CAD and finite element method steps, clinical practicalities limiting feasibility [19] and material limitations resulting in bulky and heavy devices [17].

43.4.5 Accounting for Volume Fluctuation

There is a desire from users for more adaptable sockets capable of accommodating minor changes in residual limb size [12]. Currently, there are no accepted solutions with regards to residual limb volume fluctuation [11]. Users may attempt to accommodate volume changes through addition of residuum socks [67, 69] and pads [25], however, these restrict space for the bony prominences [37] and offer discrete adjustments to volume that do not accommodate the continually changing limb volume [25]. A number of potential solutions have been proposed to address daily volume changes including flexible socket systems, inflatable bladders and pads that can be inserted into the socket [11] and closed-loop suspension systems [16, 135].

Inflatable systems using either air or liquid have been proposed by multiple groups [25, 63, 136]. Bladders may be manually inflated [63] using mechanisms such as external needles [137] or controlled via electronic units in combination with pressure sensors [25, 138, 139]. Variations of the bladder have included an artificial muscle system in which inextensible braided fibres cause contraction of an inflatable balloon, allowing the artificial muscle to support higher loads in comparison to regular bladders [140].

The combination of sensory techniques with closed-loop adaptable suspension systems has been proposed by a number of research groups [11]. The aim of these systems is to adjust the internal shape or stiffness of the socket to optimise the residuum-socket stress distribution [11]. A closed-loop pneumatic suspension system was developed by Pirouzi et al. comprised of an air cuff attached to the interior of the socket, air pumps, pressure regulation valves and a microcontroller unit [16]. Systems have also been proposed based on magnetism and the use of magnetorheological fluids [135, 141]. Through alteration of magnetic fields, the stiffness and volume of magnetorheological fluids can be controlled, enabling adjustments to socket fit [141]. The proposed devices exhibit improvements in interface pressures across the residuum and may be more easily donned and doffed in comparison to traditional sockets, however, the systems are often large, bulky and have high-power consumption requirements [11, 16, 141].

A considerable disadvantage of bladders, pads or inserts is that they introduce non-uniform deformation across the socket which can negatively impact the stress distribution [11]. Pneumatic inserts exhibit poor support of the residuum due to over or under-inflation and localised tissue compression [25]. Liquid-filled inserts tend to perform better than air-filled inserts due to the incompressibility of the fluid and the greater permissible pressure range [63, 136].

Flexible socket systems are suggested to accommodate residuum volume changes and consist of a polyethylene or silicone elastomer inner socket and a hard outer thermoplastic structure, usually with fenestrations [142]. Notable examples of flexible socket systems include the Ipos system, the Icelandic-Swedish-New York (ISNY) above-knee prosthesis and the Scandinavian Flexible Socket [142–144]. The soft and flexible inner socket is used for tissue containment, whilst the outer rigid frame provides weight transmission [143]. In addition to accommodation of volume changes, benefits include lighter weight than conventional sockets, improved comfort and greater heat dissipation [143, 145], however, fenestrations may increase localised loading [30, 49].

Sockets have been produced with movable panels attached via straps, clamps or lacing systems that may be manually adjusted by the amputee [146–148]. Solutions such as the infinite socket adopt a similar structure in which 3–4 struts encompass the residual limb using pivoting and sliding joints and are secured with a lacing system and clamps [148]. These solutions offers adaptability, however, there is a danger of overtightening which may be detrimental to long-term residual limb health [11].

Alternatives to inserts and socket structure modifications include the use of materials such as auxetic foams for fabrication of liners [66]. These materials may accommodate volume changes due to their ability to expand and contract as the residual limb reduces and increases in size [66].

43.4.6 Heat Management

A number of solutions have been proposed for heat management and the prevention of associated discomfort [11]. Proposed solutions include antiperspirant sprays, botulinum toxin injections, medication, electrical stimulation, surgery or temperature and moisture regulation technologies [20]. To date, few of these options are commercially available to amputees [11].

Available commercial solutions tend to be passive and include socks with silver-coated fibres providing antimicrobial and anti-odour benefits [11], and breathable liners introduced by Endolite and the Ohio Willow Wood Company [20, 41]. The Endolite Silcare Breathe Liner uses laser-drilled perforations to allow ventilation of the limb [20] (see Fig. 43.12). The Ohio Willow Wood Company SmartTemp Liner uses phase change materials (PCM) incorporated into a traditional silicone liner [41]. PCM store and release thermal energy via state change from a solid to a liquid and a liquid to a solid, respectively [41]. Non-commercial passive systems have been proposed such as the use of a porous wicking material attached to a hypobaric vacuum-assisted liner [149].

Fig. 43.12 Endolite Silcare liners (**a**) without a pin-lock, (**b**) with a pin-lock (image reproduced courtesy of Blatchford)

A number of active thermoregulation devices have been proposed based on air cooling, liquid cooling or thermoelectric heat management [40], however, these remain unavailable to the public. These systems tend to employ conductive pipes or channels within the socket or liner combined with a heat sink or fluid pump to draw or drive heat through the system [40, 74, 150, 151]. Variations of this format have explored different structures of pipe or channel, different working fluids and the integration of vacuum systems to alter the boiling point of the working fluid [40, 74, 150, 151]. Additionally, researchers have looked at active devices utilising thermoelectric cooling (TEC) using the Peltier effect to produce a compact, controllable solution embedded into the socket wall [152] or liner [38, 39, 66].

To date, many promising solutions have been explored, however, there are currently significant drawbacks with regard to bulk, weight, efficiency and energy consumption. Passive devices such as the SmartTemp liner are suggested to reduce mean skin temperature [20], however, it is unclear what impact they have on residuum-socket pressure profile. Active devices driven by air cooling systems may help reduce temperature, however, they often provide ineffective heat transfer [40]. Liquid cooling systems by comparison provide a high coefficient of convective heat transfer at the requirement of high-power consumption [40]. Thermoelectric-based devices provide good temperature control, are compact, contain no moving parts, however, they suffer with a low coefficient of performance (COP) [40].

43.4.7 Surgical Methods

43.4.7.1 Transtibial

Surgical methods have been investigated to aid in prosthetic fit including methods to increase the load-bearing surface of the residuum. The Ertl technique is an osteomyoplastic transtibial amputation which uses a tibiofibular bonebridge to increase the load-bearing surface and encourage distal weight bearing. Despite over 80 years of experience, there is no high-quality level I or II evidence to support the use of a bonebridge over traditional below-knee amputation techniques [153]. Traditional techniques encourage maintenance of 10 cm of residual limb distal to the tibial tubercle to aid in post-operative prosthetic fitting and rehabilitation. A posterior myocutaneous flap was described by Ernest M. Burgess in the 1960s [154] and continues to be used to provide adequate wound cover and enable prosthetic mobilisation. In blast injury, this is not always possible due to vascular injury, soft tissue loss and traumatic amputation dictating the required level.

43.4.7.2 Transfemoral

The alternative to socket suspended prosthetics is direct skeletal fixation (DSF) via an osseointegrated (OI) percutaneous implant [69], first developed in the mid-1990s [155]. This bypasses the soft tissue and loads the bone directly via an OI percutaneous implant, therefore conferring the advantages of easy connection of the prosthesis, protection of vulnerable soft tissues and enhanced proprioceptive feel [156–158]. Results in civilian unilateral above-knee amputees have demonstrated that DSF significantly improved functional outcomes [158–161]. Results for military blast mediated bilateral above-knee amputees followed up at minimum of 2 years demonstrate improved quality of life outcomes and improved physical performance measures. The procedure also appears to be safe with minimal superficial infection complications, no deep infection and no need for explanation.

43.5 Summary

The prosthesis–residual limb interface is most commonly cited as the cause of greatest problems for the blast injured amputee. Approaches to address this include technologies to improve fit, pressure distribution and even bypassing the soft tissue interface through direct skeletal fixation or end-bearing. However, the complexity of this interface and the many confounding factors including high functional expectations mean that this remains unsolved for most amputees.

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44 Bone Health in Lower-Limb Amputees

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Abstract

Bone mineral density (BMD) loss in lower-limb amputees has in the past been referred to as either osteopenia or osteoporosis. However, evidence and hypotheses in emerging literature are beginning to challenge this, suggesting that the use of these terms could be inappropriate due to key differences in the aetiology and mechanisms underpinning the bone loss in the younger amputee population. Computational and clinical analysis carried out within the Centre for Blast Injury Studies at Imperial College London and the ADVANCE Study has provided strong evidence to support this stance. Investigating BMD discordance in the spine and femur of 153 lower-limb amputees and a frequency-matched control population has shown that bone loss in amputees is localised to the amputated limb rather than systemic (as it manifests in age-related osteoporosis). Combined musculoskeletal and finite element modelling goes some way to explaining the cause of this. Weight bearing through a prosthetic socket offloads the distal femur, and consequently large areas of the femoral shaft and neck experience significantly reduced levels of stimulation when compared to weight bearing on a healthy limb. The long-term result of this is a phenomenon that we refer to as *unloading osteopenia.*

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44.1 Introduction to the Mechanics of Bone Structure and Prediction in Finite Element Modelling

Since the nineteenth century, it has been observed that the architecture of bone is influenced by the mechanical environment that it experiences, produced by forces acting on bone, including joint reaction and muscle forces. A discussion between Culmann, an engineer, and von Meyer, an anatomist, found that the arrangement of trabecular

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bone in a coronal slice of the proximal femur resembles the latticework that would be formed by drawing the trajectories of principal compressive and tensile stresses or strains [1, 2].

The resulting hypothesis that has its origins in the Culmann/von Meyer discussion is commonly referred to as "Wolff's law" based on his 1892 work [3]. However, while Wolff has been popularised as the first to observe the fundamental mechanical principles behind the observation that bone adapts so that its form follows its function, where form includes internal structural arrangement and external shape, this gives him undue credit. Wolff stated in the majority of his works that the arrangement of trabecular bone followed a hereditary blueprint unfolding with growth from the cortex in the proximal femur and did not acknowledge bone resorption alongside bone apposition as having a role in bone adaptation or maintenance [4, 5]. The 1885 work of Roux is of greater importance in informing how researchers might model bone adaptation in response to an altered mechanical environment caused by a change in either muscle or joint loading [6]. Hence, "Wolff's law" should neither be referred to as a "law" nor "Wolff's".

Frost introduced the notion of a "Mechanostat", performing in a similar way to a thermostat to maintain temperature within a range, as a phenomenological way of looking at the adaptation of bone in response to a mechanical stimulus such as stress or strain. The use of a Mechanostat in computational models allows adaptations to be made to either material properties or architectural geometries, representing bone resorption or apposition, driving the mechanical stimulus towards a target, with a lazy zone surrounding it [7]. The Mechanostat has been criticised as too simplistic an approach, although it has perhaps found application for the same reason.

The finite element method is the computational technique of choice for bone adaptation simulations. At the whole-bone level, macroscale analysis, where the elements used to represent trabecular bone are larger than individual trabeculae, is generally used for computational efficiency. Carter et al. [8] were amongst the first to predict the bone density distribution in a continuum 2D model of the proximal femur based on multiple load cases applied at the femoral head and greater trochanter. 3D continuum models have since confirmed the importance of including multiple load cases in any predictive analysis, as well as modelling bone as an orthotropic (with material properties that differ along three mutually orthogonal axes) rather than an isotropic material (with material properties that are the same in all directions) [9]. A structural approach, modelling trabecular bone using truss elements, and cortical bone using shell elements, has also been successful in predicting bone architecture in the femur, with loading for multiple activities taken from simulations using a musculoskeletal model developed with a focus on reproducing the hip joint contact force [10–12]. In the Phillips et al. [11] study, stair ascent and descent were found to be critical in stimulating regions of trabecular structure in the femoral neck and greater trochanter, with localised bone resorption predicted when loading due to these activities was removed.

44.2 Bone Changes in Amputees

44.2.1 General Trends of Decreased Bone Density

BMD loss in the hip and femur is associated with lower-limb amputation and is considered as a secondary complication. Reduced BMD carries an increased fracture risk, with many fractures occurring with minimal trauma [13]. The effect of fracture can have serious implications on mobility, physical dependency, and morbidity [14]. For amputees, this has a disproportionate effect on functionality as it can compromise short- and long-term prosthetic use.

Osteoporosis is typically a later-life bone degenerative disease of which the effects and prevalence increase substantially with age [15]. Approximately 7% of men and 22% of women over the age of 50 have osteoporosis based on WHO diagnostic criteria, and these figures increase to 25% of men and 45% of women over the age of 80. Women are at greater risk beyond menopause due to a decrease in oestrogen production, which can accelerate bone loss [15].

Osteoporosis is diagnosed using criteria defined by the World Health Organization (WHO) based on the T-score. This is defined by the mean BMD of an average 30-year-old adult with peak bone mass:

- Normal BMD = T-score of −1 standard deviation (SD) or above
- Osteopenia = T-score of between −1 and −2.5 SD
- Osteoporosis = T-score of −2.5 or below

Z-scores are similarly used to compare bone density to age- and sex-matched groups; however, these are subject to criticism due to the lack of matching for ethnicity.

Burke et al. [16] carried out one of the earliest reported studies on bone changes in lower-limb amputees where they began associating bone loss to osteoporosis. They noted "osteoporosis" in the amputated limb of 37 out of 42 lower-limb amputees studied. Since then, a variety of cohort studies have confirmed this trend of reduced BMD in both above-knee and below-knee amputees. Reports have comprised of both retrospective and prospective studies with population sizes varying from 7 to 200 amputees, both below and above knee. The majority measure BMD by dual-energy X-ray absorptiometry (DEXA) scan or peripheral quantitative computed tomography (pQCT) [17–20].

It has been reported that:

1. Bone density is significantly lower in the amputated limb than in the average limb of the healthy population [19, 20].
2. Bone density is lower in above-knee amputees than below- and through-knee amputees [17, 19].
3. In unilateral amputees, bone density is lower on amputated than non-amputated sides, although bone density in the non-amputated side is often lower than that in the average healthy limb [19–21].
4. The areas of lowest reported bone density occur at the femoral neck, the greater trochanter, and the pelvis [17–21].
5. By use of T-score or Z-score, high proportions of amputees have been diagnosed with osteopenia and osteoporosis [16, 22–24].

Although the majority of current available literature diagnosis osteoporosis or osteopenia with reduced BMD levels in lower-limb amputees [16, 18, 19, 22], more recently, some researchers have begun to make subtle distinctions between bone loss in amputees and bone loss in the general population. Sherk et al. [20] first highlighted the possible similarities between bone loss post-amputation and bone loss as a result of periods of prolonged unloading from scuba diving, bed rest, spinal cord injury, and space travel studies. Later studies by Bemben et al. [25], Flint et al. [17], and Ramirez et al. [21] made similar comparisons and do not equate amputee bone loss to osteoporosis or osteopenia but rather to physical phenomena such as stress shielding by the socket resulting in reduced load transmission through the skeleton. This is discussed below.

44.2.2 Mechanisms of Bone Loss

Osteoporosis is defined as a systemic disease characterised by loss of bone microarchitecture leading to an increased fracture risk [26]. Comparatively, low BMD manifests itself quite differently in the amputee population compared to the general population diagnosed with osteoporosis, with the key differences being the age of onset and localisation of bone loss. In the general population, studies have shown that bone loss begins in both men and women after peak bone mass has been attained at around 40 years old [27]. The loss of bone density with increasing years is a natural process beyond this point due to resorption exceeding formation. However, substantial bone loss has been recorded in a significant proportion of amputees well below this age. This is alarming due to the potential complications of long-term treatment of osteoporosis and the associated fracture risk with osteoporosis. In 1974, Hungerford and Cockin [23] first reported characteristic findings of osteoporosis (*Z*-score <-2.0) and osteopenia ($-1.0 >$ *Z*-score > -2.0) in 90% of investigated above-knee amputees. Since then, in more recent studies, this figure has dropped, although the proportion remains large [16, 17, 24]. A study by Flint et al. [17] found that out of 156 amputees tested, both unilateral and bilateral, the observed rate of low BMD (Z-score <−1.0) was 68% in above-knee amputees and 37% in below-knee or through-knee amputees despite the median subject age for the study being just 24 years old and 98% of participants being male. The vast majority of existing literature supports the same trend: low bone mineral density affects disproportionate amounts of the amputee population regardless of age.

Low BMD in lower-limb amputees is also found to be more commonly localised to one particular area (the amputated limb), whereas in the general population, the problem is often systemic. Generally, T-score (or Z-score) discordance is much more common in the amputated than in the non-amputated population. A study by Moayyeri et al. [28] found diagnosis concordance (between T-score at the spine and the pelvis) in 2442 of 4188 (58%) non-amputee men and women scanned. It found major T-score discordance in only 2.8% of those studied and minor T-score discordance in a further 38.9%. By comparison, in a population of 16 unilateral amputees, Rush et al. [22] found normal BMD levels in the spine and contralateral limb but significantly lower bone loss in the amputated femur of all subjects. Femoral BMD was, on average, 28% lower in amputated than non-amputated sides, which suggests that major discordance between the two areas was common. Bemben et al. [25] and Smith et al. [29] found similar results. Additionally, there are many examples of studies showing clear diagnosis discordance in unilateral amputees between the amputated and contralateral limbs. To the best of our knowledge, no studies have been published showing full concordance between multiple areas in a young lower-limb amputee cohort.

Diagnosis discordance in amputees and general population seems to occur inversely. Moayyeri et al. [28] found that of 1601 able-bodied subjects who showed diagnosis discordance, 87% represented worse BMD in the lumbar spine than in the femur. This finding is supported in earlier able-bodied studies by Woodson et al. [30] and Mounach et al. [31] as well. In amputee studies by Bemben et al. [25] and Smith et al. [29], the direct opposite is true: where diagnosis discordance occurs, BMD in the femur is lower than that in the lumbar spine.

44.2.3 Aetiology of Bone Loss in Amputees

As described above, one of the key differences in BMD loss in the amputee population compared to the general population is the age of onset. As T-scores are compared to a 30-year-old, and many young military amputees are below this age, they may have not reached peak bone mineral density and therefore the T-score is not a suitable diagnostic mechanism. Additionally, osteoporosis is diagnosed with the BMD at the femoral neck as a reference [32]; however, the localised effects, discussed above, may result in a systemic misdiagnosis. Other relevant age effects are that treatment strategies are recommended only in those aged over 50 and include bisphosphonates and denosumab, a human monoclonal antibody. There are no recommended treatment guidelines for those aged under 50; even where bisphosphonates may be used, the long-term use of these has been associated with atypical femoral fractures that would be severely debilitating for amputees. In amputees, secondary causes of osteoporosis should always be excluded including hyperthyroidism, myeloma, vitamin D deficiency, malabsorption conditions, and, particularly in polytrauma blast injury patients, hypogonadism.

Taking into account these age effects and having excluded other causes, the aetiology of bone loss in amputees seems to be a predominantly mechanical phenomenon. Ultimately, changes in the loading environment post-amputation reduce the stimulation experienced by certain regions of bones causing a net resorption of bone over the course of many remodelling cycles [7]. There are multiple theories describing how and why the loading environment changes; however, there is very little conclusive evidence showing the direct effect of these changes on bone health and bone density. Some of the theories introduced in previous studies have been outlined below.

Ambulating and weight bearing through a prosthetic socket result in vastly different loading through the femur and hip post above-knee amputation. Proximal load transfer, through ischial weight bearing, is encouraged in order to deliberately offload the distal femur where the direct effects of surgery are most prominent. Ramirez et al. [21] describe this effect as a form of stress shielding where load is redirected through the stiff carbon fibre socket. Reduced loading through the distal end of the bone means consistently lower joint contact forces through the femur and at the hip.

Bemben et al. [25] highlight the effect of dramatic unloading during bed rest immediately post-surgery. The study finds that the largest proportion of bone loss takes place in the first 6 months following amputation (up to 14.8% in the amputated hip) relative to the 6 months following where BMD stabilised or even increased slightly in some subjects. Flint et al. [17] demonstrated that amputees who begin ambulation within 2 months of amputation record significantly less bone loss than those who begin later. Reports of a positive correlation between prosthetic use and volumetric BMD in both suggest that a lack of loading during bed rest could be responsible for substantial bone loss in some amputees [17, 25]. It also highlights the potential restorative effect that increased activity, even with a prosthesis, could have on bone loss.

Ambulation becomes much more difficult post-amputation. Stepien et al. [33] showed that civilian amputees take an average of 3063 ± 1893 steps per day, less than most healthy people. Various studies suggest ill-fitting sockets and prevalence of residual limb pain in long-standing amputees as potential reasons for this [34–36]. Moreover, most amputees simply do not possess the means and mobility to partake in activities such as running, jumping, stair ascent, and stair descent. These activities are associated with some of the highest joint contact and muscle contractile forces experienced by human beings, thus playing an important role in stimulating the skeleton. In general, most amputees miss out on key loading profiles, which have been proven to contribute to the structure of healthy bones [37].

General gait changes post-amputation alter loading intensity and loading time during ambulation [17]. Research has shown that unilateral amputees spend less time on their amputated limb during gait [38, 39]. Linked to this is a suggestion by Burke et al. [16] that reduced loading on the amputated limb is a result of other secondary conditions associated with amputation such as osteoarthritis.

Finally, a study by Tugcu et al. [40] suggested that lower muscle forces due to post-amputation muscle atrophy could contribute to reduced BMD in the amputated limb as post-amputation muscle contractile forces experienced by the bone are reduced. The study found a weak correlation between the strength of the quadriceps muscle and total femur BMD.

Overall, similar to the effect of resistance training on muscles, levels of stress and stimuli that the human skeleton is subjected to play an important role in its physical structure [37]. Reduced stress and stimuli post-amputation, as a result of some or all of the theories highlighted above, are very likely to play a major role in the state of amputee bone health. Equally, this suggests that induced stress and stimuli from targeted physiotherapy and external loading could prevent or restore the bone lost in a young amputee population [41].

44.3 Clinical Study on Bone Changes in Amputees

To investigate the question of localised BMD loss versus systemic osteoporosis in military blast injury amputees, a prospective observational cohort study was conducted as part of the Armed Services Trauma Rehabilitation Outcome Study [42]. Full data are presented by McMenemy et al. [43] and are summarised here. The participants consisted of 575 male adult UK combat veterans from the UK-Afghanistan War with combat-related trauma (injured group), 153 of whom had lower-limb amputations. These were frequency-matched to 562 uninjured men by age, service, rank, regiment, deployment period, and role in theatre (uninjured group). BMD was assessed using DEXA scanning in the hips and the lumbar spine.

There was no statistical difference between the spine BMD of the injured and uninjured groups; however, femoral neck BMD was lower in the injured versus the uninjured group. Subgroup analysis revealed that BMD in the femoral neck of amputee subjects—for the amputated limb only—was significantly lower than that in injured non-amputees and uninjured participants. Also, femoral BMD was significantly higher in below-knee amputees compared to above-knee amputees. These differences could not be accounted for by alcohol consumption, smoking, nor any difference in activity levels as defined by the International Physical Activity Questionnaire (IPAQ).

The changes in bone health demonstrated in this study appear to be mechanically driven as below-knee amputees are more likely to load their proximal femur physiologically, whereas above-knee amputees have load transfer that partially bypasses the proximal femur due to shear loading in the socket as well as ischial bearing. Therefore, it is proposed that the BMD loss seen in amputees is not a consequence of regular, systemic osteoporosis but is a localised response to specific unloading by the prosthetic socket, termed *localised unloading osteopenia*.

44.4 Computational Modelling of Stress and Strain in Amputees

Through a combined musculoskeletal and finite element modelling approach, we have been able to estimate the key regions at risk of bone loss within the femur of an amputee. A subject-specific musculoskeletal model is used to generate the physiological loading scenario of the muscle and joint reaction forces experienced by an above-knee amputee subject during gait. This loading scenario is applied to a finite element model of the same amputee with MRI-derived bone and prosthetic socket geometry. Analyses can investigate exactly how forces are transmitted through the prosthetic socket boundary into the residual limb soft tissue and finally into the amputated femur.

Modelling results support the theories given by those in previous studies: changes to the musculature in the residual limb, altered gait patterns, and shielding of forces as a result of loading through a prosthetic socket all contribute to an altered loading environment within the residual limb of lower-limb amputees. Previously monoarticular muscles that passed the knee (such as the vastus muscles) no longer provide any mechanical leverage when ambulating post-amputation. The loss of muscles across the knee and ankle joints means motion is almost completely driven by the muscles of the hip. As a result, most of these muscles generate substantially more force than that in an able-bodied body-matched control condition. The ground reaction force is also substantially higher during amputee gait. This can be attributed to a loss of proprioception as well as reduced control from the knee and ankle joints. Subsequently, the accelerations and decelerations associated with amputee gait increase. This issue is particularly pronounced in bilateral amputees, where proprioception is lost in both legs. These factors contribute to an increase in overall joint reaction force experienced at the hip.

In spite of this, the results of finite element analysis show that the force being transferred through the femur of an above-knee amputee is actually largely reduced when compared to an able-bodied control. This is due to the prosthetic socket boundary. Rather than weight bearing longitudinally into the end of the bone, as would happen in an able-bodied individual, the majority of load is supported by the ischial rim of the socket or dispersed over large portions of the residual limb musculature. Assessing the subsequent risk of bone resorption, by comparing the forces experienced in an amputee femur to that of an able-bodied volunteer, shows large areas of the amputated femur at risk of bone loss due to a lack of direct stimulation (Fig. 44.1). This is most prominent in the distal femur (as found in studies by Sherk et al. [20] and Bemben et al. [25]), but also significant in the femoral neck and greater trochanter (as shown in studies by Flint et al. [17], Leclerq et al. [18], and Kulkarni et al. [19]).

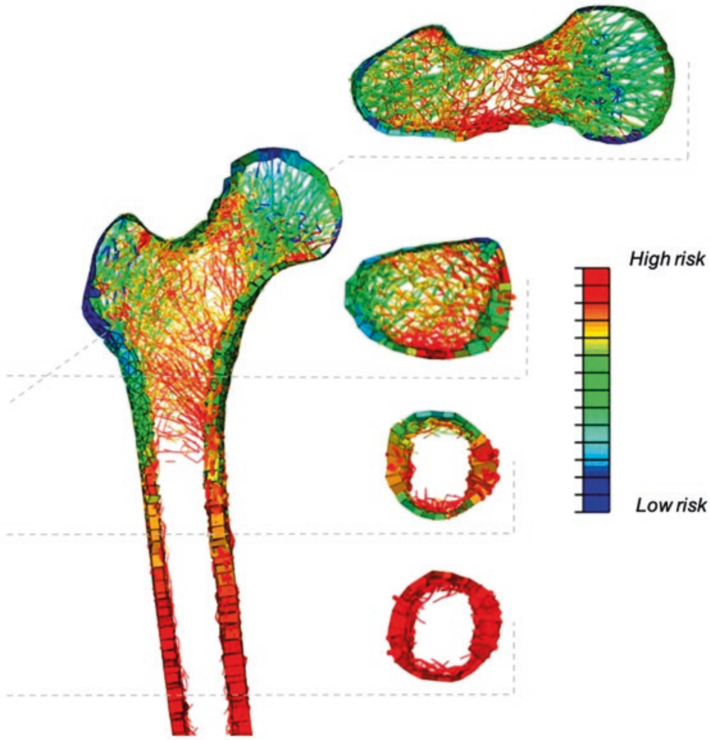

Fig. 44.1 Frontal and transverse slices showing bone resorption risk in the femur of an above-knee amputee

By adaptation of an already-verified predictive modelling pipeline, we have attempted to predict the manner in which bone loss will occur. Results show cortical thinning in the distal shaft of the femur and a trabecular density decrease of over 10% within 5 years of amputation (from purely mechanical load changes) (Fig. 44.2). The principal and secondary tensile, secondary compressive, and greater trochanter trajectories within the proximal femur begin to fade. We hope that we can use this model to identify means of preventing further bone loss (or even means of stimulating bone growth) in amputees by targeting those areas at risk of bone loss already identified with changes to the current prosthetic design or physiotherapy.

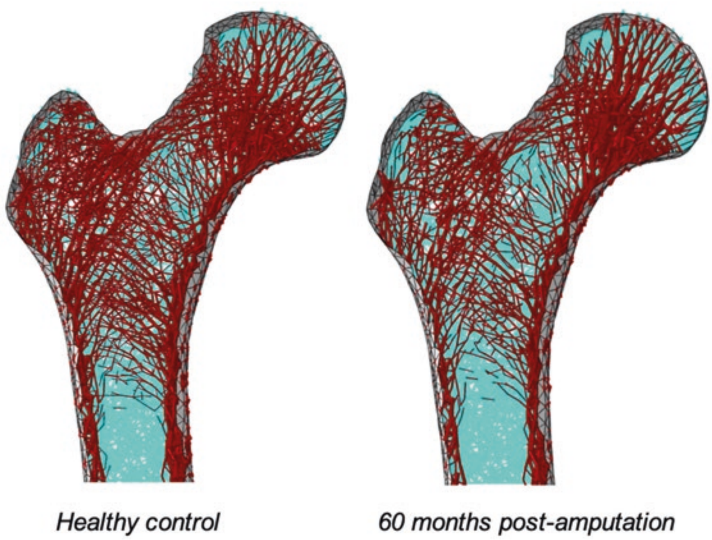

Fig. 44.2 Frontal slices showing the trabecular bone structure in a healthy control femur (left) and an amputee femur after predictive FE adaptation (right)

44.5 Discussion

With all of the current evidence at our disposal, we believe that there needs to be a distinction made between a diagnosis of osteoporosis and low BMD in, particularly, the younger amputee population. Reduced BMD might not be a sign of osteoporosis but rather a sign of a bone's adaptation to a new mechanical loading environment post-amputation. Hence, the problem could potentially be reversible with appropriate research into socket design and bone stimulation. The best evidence of this can be found in the prevalence of low BMD in the younger amputee population. Rush et al. [22] even reported an inverse correlation between patient age at amputation and bone density.

We believe that a diagnosis of osteoporosis potentially carries the implication that the problem is systemic and irreversible. This could have an important psychological impact and subsequently affect the subject's personal outlook on rehabilitation. From a treatment perspective, the difference could also be vital. Osteoporosis is typically associated with endocrine changes in the older population. As a result, it is typically treated by hormone replacement therapy or bone-preserving medication, such as bisphosphonates. Both treatments have been shown to have adverse effects elsewhere in the body [44, 45]. Hence, it would be beneficial to avoid them in a younger subject, for example an amputee, whose bone loss could be mainly attributed to unloading during periods of bed rest or changes in mechanical loading during daily life. A diagnosis of regular osteoporosis may result in prescription by family physicians where one is not needed. It could also stifle further research into the effect of end loading, ischial weight bearing, and design of future prosthetic sockets. Hence, in amputees, a more specific diagnosis of *localised unloading osteopenia* is more suitable.

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Musculoskeletal Health After Blast Injury

45

Anne K. Silverman, Brad D. Hendershot, and Alison H. Mcgregor

Abstract

Rehabilitation after blast injury is critical for regaining mobility and functional independence. In addition to immediate rehabilitation goals to facilitate activities of daily living, rehabilitation should address movement patterns and prosthetic interventions that mitigate long-term secondary (musculoskeletal) conditions. For example, after limb loss, which is a common result of blast trauma, people experience disproportionately high rates of osteoarthritis and low back pain relative to the general population. These conditions often develop and/or deteriorate over time and can have detrimental effects on mobility and quality of life. In this chapter, we describe and summarise existing knowledge of these musculoskeletal conditions secondary to blast injuries that include limb loss. Relationships between movement strategies and biomechanical outcomes are also discussed. While many musculoskeletal health conditions are multifactorial in onset and progression, these conditions are strongly related to movement biomechanics, and thus can be mitigated through rehabilitation approaches. Specifically, rehabilitation strategies that balance immediate goals of clinical outcomes and community engagement with long-term goals of healthy joint mechanics are critical for this population. In addition, rehabilitation and prosthetic interventions should be continually monitored and delivered so that they appropriately account for movement adaptations and changing mobility needs of the individual.

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45.1 Introduction

Secondary conditions, such as those related to musculoskeletal health, are an important concern after blast injury, particularly to ensure optimal long-term outcomes and quality of life. To improve and maximise outcomes after blast injury, a thorough understanding of the mechanisms of development of these long-term secondary health conditions is necessary so that appropriate prevention, intervention, and mitigation strategies can be developed.

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Given the complexity of trauma to individuals resulting from blast(s), it is difficult to generalise their rehabilitation, or to identify the causal mechanisms for onset and recurrence of long-term secondary health conditions. However, alterations in musculoskeletal loading are inevitable after blast injury, largely due to the altered/new mechanics of the neuromusculoskeletal system. Changes in loading can result in pain and injury over time in a wide variety of populations, and there are large changes in musculoskeletal dynamics in people who have recovered from a blast injury.

In this chapter, we link to existing knowledge on secondary musculoskeletal health conditions in those who have undergone a blast injury and/or traumatic lower limb amputation. While there is no "typical" blast injury, our knowledge of the mechanics of people who have a traumatic amputation allows the development of a baseline for understanding how long-term secondary conditions develop, as well as the implications of ageing in this cohort. Of particular emphasis in this chapter is overall joint health, osteoarthritis, and low back pain after blast injury. The results of epidemiological, experimental, and modelling studies have informed our understanding of conditions secondary to blast trauma. As these areas advance, there are positive implications for rehabilitation and assistive device interventions that can be tailored to individuals.

45.2 Prevalence of Joint Health and Secondary Injuries

Low back pain (LBP) is an exceedingly common secondary health condition after limb trauma and limb loss—with estimated prevalence rates as high as 52–89% [1–3]—and adversely interferes with daily social, recreational, family, and work-related activities [4]. Notably, LBP secondary to amputation has been surprisingly rated as more bothersome than phantom or residual limb pain [2], and the single most important condition contributing to a reduced quality of life [5]. While recent evidence suggests that LBP may be more prevalent among people with limb loss of dysvascular compared to trauma-related aetiologies (relative risk ratio of 1.80 and 1.14, respectively; [6]), among those with traumatic amputations, the extent of injury can also influence the risk for LBP. For example, among a sample of 192 people with traumatic lower limb amputations, 81% (62/77) with transfemoral and 62% (71/115) with transtibial limb loss reported LBP; 60% of the total sample reported the onset of LBP symptoms within 2 years since amputation. These data suggest a dose-response relationship between the extent of initial trauma/subsequent exposure to altered movement mechanics and the risk for LBP. Nevertheless, evidence surrounding the specific underlying factors/mechanisms for LBP, particularly among those with traumatic amputation, remains unclear [7].

Joint degeneration and osteoarthritis (OA) are also highly prevalent musculoskeletal conditions secondary to limb loss, particularly in the contralateral knee and hip for those with unilateral limb loss. Compared to uninjured individuals, people with (unilateral) limb loss are 17 times more likely to report pain and joint degeneration in the contralateral knee, and cross-sectional studies report prevalence rates ranging from 28% to 55% [8–10]. Similar to LBP, there is also evidence to suggest a relationship between level of injury and risk for (intact) knee OA; compared to uninjured individuals, the prevalence ratios for people with transfemoral and transtibial limb loss are 3.3 and 1.3, respectively [11]. Similarly, contralateral hip OA is more prevalent following limb loss [8, 12], with a threefold increase in the risk for hip OA in people with transfemoral vs. transtibial amputation [13]. While no specific single factor predicts the higher risk for knee OA (e.g. age, amputation level, time since amputation; [9]), OA is largely theorised as a mechanically induced (yet biologically driven/mediated) condition, and thus probably accelerated in the limb loss population due, in part, to repeated exposures to altered/asymmetric gait and movement biomechanics. However, it is important to note that lifestyle/activity level plays a critical role in modulating the risk for OA and LBP in this cohort [14].

Battlefield injuries to the limbs account for more than 50% of combat wounds sustained in recent conflicts [15]. These military personnel who have sustained blast-related trauma, up to and including limb loss, represent a unique and typically young cohort, with challenging needs for long-term care. In addition, these individuals present an opportunity to study risk factors for secondary health outcomes in ways not otherwise possible in the dysvascular or general limb loss populations [16–18]. Among U.S. Service Members, those with lower limb amputation are nearly twice as likely to develop an overuse injury within the lower extremity than those with mild combat-related injury [19]. Currently, however, the long-term outcomes within this patient population remain unclear. Understanding the mechanical environment in the joints from altered movement patterns is critical for predicting and maximising long-term health outcomes and can be investigated with musculoskeletal modelling approaches.

45.3 The Relationship Between Movement Biomechanics and Joint Health

The factors influencing musculoskeletal health following blast-related limb loss are multifactorial [19]. For example, biological, psychological, and social factors all contribute to the development of LBP. Repeated exposures to altered mechanics of movement are commonly theorised to play an integral role in LBP, and people with lower limb loss adopt aberrant movement patterns during activities of daily living [20–22]. For individuals with unilateral limb loss, preference for the intact limb results in larger and prolonged mechanical loads applied to the joints within the intact limb. These forces are influenced by push-off power of the prosthetic foot [23, 24] and, while dependent on the level of limb loss, this has been shown to be larger at both the early and later time points in the rehabilitation process [25]. Relatedly, altered motions of the trunk and pelvis following limb loss [26–29]—suggested as an active movement strategy due to larger positive joint powers [30]—influence joint loads both in the lower and upper body. Given the interplay between the upper and lower body, it is therefore critical to use movement analyses and modelling approaches that consider the entire body.

45.3.1 Lower Limb Joint Mechanics and Osteoarthritis

While altered movement strategies and biomechanics are risk factors for the development of joint disorders, it is challenging to link this development directly to specific biomechanical metrics. Many studies have evaluated joint kinetics via inverse dynamics approaches. Development of joint disorders is likely linked to internal joint contact forces, which are difficult or impossible to measure directly. Thus, many musculoskeletal modelling studies have estimated in vivo contact forces for a variety of movements and activities. While musculoskeletal modelling and simulation studies of people recovering from blast injury are limited, these techniques have been applied to investigate movement in people with lower limb amputations. Walking and rising from a seated position have been evaluated as these tasks are critical for mobility and independence.

Using an inverse dynamics approach, intersegmental knee joint forces and net joint moments while walking at a range of speeds were investigated for people with a unilateral transtibial amputation who were asymptomatic of OA [31]. No differences were found between the intact limb and the limb of people without amputations, although there was asymmetric loading between the intact and amputated limbs. These results suggest that intersegmental forces do not indicate the risk for developing OA; however, greater intact limb knee loading relative to non-amputees may develop as a result of the onset of OA. At the transfemoral level, hip and knee moment asymmetry has been evaluated during both sit to stand and stand to sit for different prosthetic knee components [32]. People with transfemoral amputation had greater asymmetry in peak hip and knee moments compared to non-amputees, highlighting that for this task, there is great reliance on the intact limb for raising the body to standing and for lowering the body to a seated position. Differences between prosthetic knees were not observed, although there may be important differences between prosthetic components at the individual level. Both sit to stand and stand to sit place high demands on the intact limb. These demands increase with higher amputation levels and are important to consider in relation to overall joint health and movement strategies that are taught during rehabilitation.

While providing important insight into lower limb joint mechanics during activities of daily living, inverse dynamics studies do not account for the compressive action of muscles crossing the joint. To estimate joint contact loads that may be linked to joint disorders, a musculoskeletal modelling approach is needed to account for muscular action. Using a musculoskeletal modelling and forward dynamics simulation approach to evaluate people with unilateral transtibial amputation, Silverman and Neptune [33] evaluated peak knee joint contact forces and impulses for the amputated limb, intact limb, and non-amputee limb during the stance phase of walking. This modelling approach incorporated the effects of musculature spanning the knee joint and also revealed contributions of muscles not crossing the knee joint as well as the prosthesis. Knee contact forces and impulses in the residual limb were found to be smaller than the intact and non-amputated limbs, which was largely a result of smaller quadriceps forces and lack of a functional gastrocnemius muscle. While impulsive loading was not found to be different between the intact and non-amputee limbs, peak loading was found to be larger in the intact limb relative to non-amputees. This study suggests that peak loading, rather than impulsive loading, may be one of the biomechanical factors that contribute to the development of OA. For people after blast injury, these results suggest that peak knee contact loading should be minimised through rehabilitation and/or prosthetic device intervention.

Additional modelling work suggests that there is the possibility of reducing knee and hip joint loading among people who have had a transtibial amputation. Using a forward dynamics simulation framework to produce walking in the sagittal plane, Koelewijn and van den Bogert [34] found that when joint moment asymmetry was penalised within a modelling framework, joint contact forces and the hip and the knee were reduced while also resulting in a greater metabolic cost of walking and abnormal kinematics. These results suggest that clinical procedures that aim to improve joint kinetic symmetry may have positive implications for long-term health outcomes and must balance this desire with metabolic cost of walking. A moderate improvement in joint moment symmetry may only have small detriments to metabolic cost and kinematics while also reducing joint contact forces, highlighting that rehabilitation should focus on multiple objectives for movement. In addition, those who have recovered from a blast injury may have substantial asymmetry in their skeletal structure and movement kinematics, and thus symmetric movement may not be possible or optimal for long-term joint health.

Also, at the transtibial amputation level, modelling and simulation of walking have suggested that prosthetic intervention has the potential to improve joint loading symmetry. Specifically, the vacuum level of vacuum suspension prosthetic sockets affects the magnitude and symmetry of knee joint contact forces, showing intact limb contact forces comparable to non-amputees at a vacuum level of 50.8 kPa [35]. Furthermore, prosthetic foot stiffness has also been found to affect intact knee contact forces, and modelling analyses suggest that a more compliant ankle and heel section with a stiffer midfoot and toe in prosthetic feet has the potential to reduce intact knee loading as well as the metabolic cost of walking [36].

Using an inverse dynamics approach combined with static optimisation, musculoskeletal modelling investigations have shown that leveraging imaging to incorporate individual anatomy can provide more details of estimated contact forces [37]. In the population with blast injuries, there is great variation in musculoskeletal structure, and thus accounting for this variation is important for predicting internal mechanical loading across individuals. In those with bilateral transfemoral and unilateral transtibial amputation, incorporating this anatomy can significantly affect muscle excitation predictions and associated joint loading at the hip and knee [38–40]. Furthermore, these studies have indicated greater loads at the hip (bilateral transfemoral) and contralateral knee (unilateral transtibial), supporting the link between joint loading and prevalence of joint disorders. Understanding the effects of altered anatomy can also inform surgical procedures to help address long-term joint health.

45.3.2 Low-Back Mechanics and Pain

In relation to the development of LBP, many biomechanical analyses of people with limb loss have found altered kinematics and kinetics in the low back. For example, in people with a unilateral transfemoral amputation, greater transverse plane rotations have been observed in people with back pain compared to those without back pain during walking [41]. Also, during walking, three-dimensional joint reaction forces and moments have been shown to increase with the level of lower limb amputation relative to non-amputees [42], which suggests greater muscular demands and risk for spinal degeneration. These results are consistent with asymmetric trunk kinematics commonly observed in populations with limb loss [43]. In addition, people with transfemoral amputation have been found to have greater joint moments and powers, computed via inverse dynamics techniques, at the L5–S1 vertebral level during both sit to stand and stand to sit movements [30]. Mediolateral joint powers at the low back are also larger in this group during walking [26].

Similar to studies of hip and knee mechanics, investigating the biomechanics of low back pain often involves a musculoskeletal modelling approach, as measuring joint and muscle forces in vivo is not feasible. Musculoskeletal modelling studies have been directed at gaining a better understanding of internal joint mechanics in those who have had a limb loss. People with unilateral transtibial amputation have been found to have greater trunk-pelvis lateral bending towards the amputated limb along with greater compressive forces at the L4–L5 vertebral level [44]. Overall, greater ranges of motion, which may be critical for balance regulation during movement tasks, combined with greater spinal loading may put the lumbar region at greater risk for degeneration and pain. Similar results have been found during sit to stand [45] using a musculoskeletal model validated for predicting low-back loads during moderate activities [46], where greater compressive loads were found shortly after the time of lift-off from the chair. Furthermore, the alterations observed in the frontal plane trunk-pelvis angles were indicative of shifting of body weight that is common for people with unilateral limb loss during this task. In rising from a chair, greater weight is placed on the intact limb from lift-off until standing, and body weight shifts between limbs as the person adjusts and repositions/balances at standing. During the sit to stand task, low-back loads are generally larger than walking or running [47], and thus this activity is important to evaluate in relation to spinal health. Rising from a seated position is critical to maintain mobility and independence, and thus instructions on how to complete this task safely may be particularly important for rehabilitation to maintain musculoskeletal health in those with limb loss due to blast injury.

People with unilateral transfemoral amputation have also been evaluated during the sit to stand task using a finite element model of the spine including low-back musculature driven by experimentally measured kinematics. Those with an amputation had greater peak and average compressive and shear forces in the spine [48]. While the estimated loads were below thresholds for injury, higher cumulative loading from activities of daily living has the potential to contribute to degeneration and pain development over time. A similar approach was used to investigate spinal loads in people with either unilateral transfemoral or unilateral transtibial amputation [49]. Not only did people with an amputation have greater spinal loads at self-selected speeds, but also as walking speed increased, these loads increased to a greater extent among people with an amputation relative to non-amputees. This result highlights how specific mobility needs of individual patients may place them at greater risk for low back pain and should be considered through the rehabilitation process. For those with blast injury, trunk-pelvis motion and mechanics should be evaluated, as many have also sustained spinal fracture in addition to limb loss. Degeneration in these joints may be exacerbated given prior fracture.

45.4 Implications for Rehabilitation and Future Directions

There are several immediate objectives of movement that are considered when selecting a movement strategy, which may or may not conflict with long-term movement objectives that promote long-term mobility and health. Current clinical practice encompasses the immediate objectives to encourage home and community engagement, with a large focus on individual feedback and clinical outcome measurements such as the 6-minute walk test and the timed up and go test. These immediate objectives are critical to achieve, but rehabilitation interventions should also incorporate metrics for long-term joint health to ensure mobility in the years after blast injury. Long-term health can be assessed through frequent follow-up appointments, adaptive rehabilitation, and utilisation of additional outcome measures. For example, as wearable sensing technology becomes more prevalent and cost effective, assessment of activity levels and fall events will improve. From a joint health perspective, biomechanical symmetry is an important target although current debates question the reality of this goal. Clearly assistive device interventions that mimic the function of biological limbs and improve human-device interaction are needed.

An important issue is to consider rehabilitation provision. Currently, rehabilitation is frequently deemed complete when an individual is mobilising with their prosthetic device(s), but many patients and therapists argue that rehabilitation should be seen as a lifetime process. Movement patterns and human function are known to change over time, and past work has suggested that humans learn an energy-efficient movement strategy that may or may not be biomechanically sound to maintain joint health. Such adaptations in people without an amputation are thought to underpin chronic musculoskeletal conditions. On top of such adaptations, function can change as a result of injury, periods of inactivity as a result of illness pain and/or ageing, and changes in prosthetic componentry.

Results from Jarvis et al. [50] indicated that military service members with an amputation, including those at the transfemoral level, achieved high levels of efficient gait and function attributed to intensive rehabilitation. However, it is less clear if these functional levels are sustainable and how they relate to the high rates of osteoarthritis and back pain described above. During rehabilitation, levels of activity were often much higher compared to levels while on home leave [51]. Changes in activity and compliance with exercise on discharge have been reported during ongoing research, with many struggling to transition to normal life while also maintaining an exercise habit. Changes in activity levels will affect movement biomechanics and associated cumulative loading, which need to be understood in the context of future injury implications. In contrast, many people with limb loss embrace sports participation after rehabilitation, including competing in the Invictus, Paralympic, and Warrior Games. High internal mechanical loading and activity levels associated with sports are likely to have both positive and negative implications for the body, which are not fully understood. Biomechanical analyses, including in vivo and musculoskeletal modelling studies, have the potential to help understand "duty factor" and exercise dosage for improved joint and bone health.

45.5 Conclusion

Long-term musculoskeletal health is an important concern for individuals who have had a blast injury, given elevated risks for development of

OA, LBP, and joint pain. Intervening early to improve long-term joint health will have substantial positive effects on mobility. To guide such interventions, quantifying underlying movement biomechanics in the blast-injured population during a variety of activities is critical in understanding the potential biomechanical effects on long-term joint health. Prior experimental and modelling studies have found asymmetric and larger loading in the lower limb joints and low back in those with lower-limb amputation, revealing potential targets for rehabilitation and prosthetic intervention. Our understanding of the link between abnormal movement biomechanics and joint health is critical to the rehabilitation process, which has multiple objectives. These objectives should incorporate the potential effects of movement on joint health in addition to energy efficiency and comfort. In addition, rehabilitation and prosthetic intervention should be an ongoing process to account for movement adaptations and changing mobility needs of the individual.

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46 Biomechanics of Blast Rehabilitation

Anthony M. J. Bull

Abstract

Blast injuries can result in significant musculoskeletal deficits due to neurological or anatomical damage. This chapter focuses on the biomechanical effects of blast injury and its impact on the ability of an individual to achieve desired tasks. Motion is driven by muscles that have small moment arms relative to the centres of the joints that are moving, causing very high muscle and joint reaction forces relative to the external load. A small disruption to the muscular anatomy can dramatically reduce the ability of the person to achieve the task, and so rehabilitation seeks to optimise the way the task is performed in terms of the path of motion as well as how the muscles activate to achieve the tasks. Motion and muscle activation can be analysed using biomechanical models. These models have been applied to reducing osteoarthritis in blast-induced traumatic amputees, optimising prosthetic parameters to reduce fatigue, and changing gait patterns to reduce muscle fatigue and lower the loads on joints.

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46.1 Loading the Musculoskeletal System

Chapter 3, Biomechanics in Blast, highlights how force transmission through the musculoskeletal system is a function of the task demands (the external loading) and the muscle forces. The task demands can be reduced by changing the acceleration required and posture (motion), but fundamentally the muscle forces drive the internal loading. These forces are higher where the muscles are closer to the joint (i.e. they have a small moment arm about the centre of rotation), the muscle stresses are higher where the muscle cross-sectional area is smaller, and the overall forces and stresses are higher when co-contraction is required to achieve joint stability (particularly for joints that are unstable or have low congruency such as the shoulder). Therefore, muscle moment arm and muscle physiological cross section area are key parameters for consideration when analysing musculoskeletal loading. Blast injury can produce an anatomical or neurological deficit. The most extreme injury examples are amputations or traumatic brain injury.

46.1.1 Musculoskeletal Capacity

Musculoskeletal capacity is defined as the physiological abilities of the musculoskeletal system. As this chapter focuses on

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biomechanics, capacity in this remit is defined in terms of the biomechanical capability. Biomechanical capacity accumulates due to genetics (some people are simply bigger and stronger than others) and environmental (including exercise, nutrition) factors up to a point in the mid-20s at which point healthy age-related decline sets in. Reserve is the difference between the individual's capacity and the requirement of the task, meaning that there is biomechanical redundancy in the musculoskeletal system [1] (Fig. 46.1). This redundancy can then be exploited to achieve the task in many different ways. A trivial example of this is where the task is getting up out of a chair; a young person with high capacity also has a large reserve and so can choose to achieve the task in many different ways, but an older person may have to exploit their capacity to its age-related reduced maximum to achieve the task without any reserve, resulting in some form of compensation such as using their arms to push off on the seat [2, 3].

To rehabilitate blast-induced musculoskeletal impairments, we need to understand capacity and task demands in order to optimise the way that tasks are achieved. This becomes critical in situations where there is little reserve, such as for amputees, or those with significant muscle damage. Musculoskeletal biomechanics as described

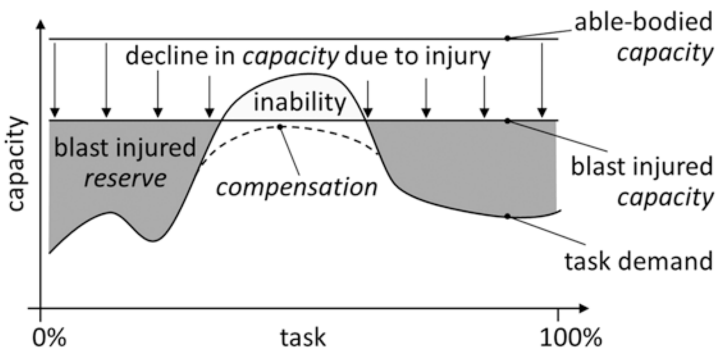

Fig. 46.1 Musculoskeletal capacity, reserve, and compensation. *Capacity* is the physiological abilities of the neuromusculoskeletal system that are available for the task. *Reserve* is the difference between the *capacity* and the task demands. If the *reserve* cannot meet the task demands, *compensation* can be applied to adjust the task while achieving the same goal. (Figure modified from van der Kruk et al., 2021 [1])

in this chapter quantifies *capacity* in terms of muscle forces/stresses and fatigue (cumulative loading of muscles over time) and relates this to *reserve* by understanding the muscle loading requirements of each task of interest.

Computational modelling combined with advanced laboratory experiments is the tool used to quantify capacity, reserve, and task demand.

46.2 Computational Modelling to Analyse Musculoskeletal Loading

Chapter 3 described how musculoskeletal dynamics can be used to quantify musculoskeletal loading. This complex methodology requires detailed computational models to calculate muscle forces, ligament forces, and joint reaction forces at the joints of interest [4]. There are three main software technologies available to conduct musculoskeletal dynamics analysis (OpenSim [5]; AnyBody [www.anybodytech.com/]; FreeBody [6, 7]). Most of these modelling technologies use similar principles to calculate the loading parameters of interest and have been validated using instrumented implants [8–13] and measurement of muscle activity using electromyography.

These have created models for all major joints of the body, including the ankle, knee, hip, elbow, wrist, and shoulder (Fig. 46.2). Spinal modelling using these tools is more complex and further from clinical application, yet examples in blast rehabilitation exist and are presented in this book.

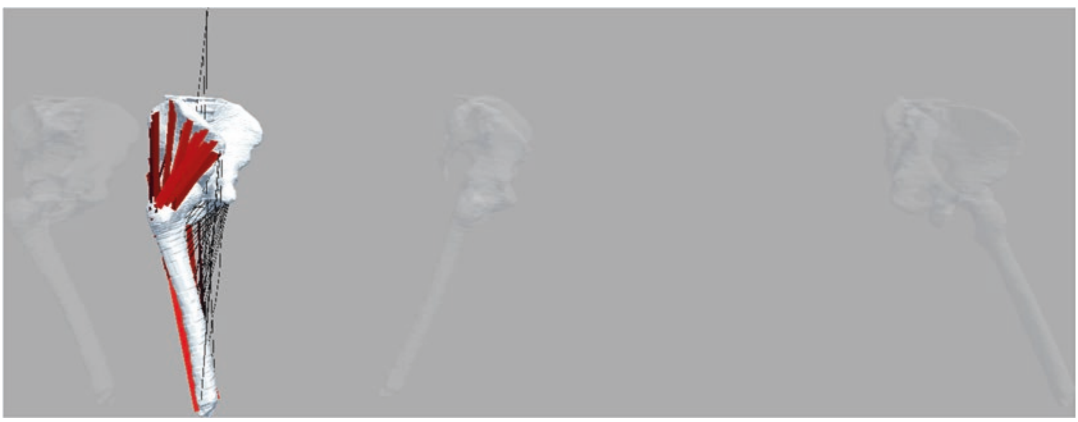

Fig. 46.2 Visualisation of muscle activations during gait, calculated for an above-knee amputee using FreeBody [7]

However, despite the demonstration of the fidelity of these models, there is very little work published that demonstrates their uses in the clinical domain. The main uses that are prevalent are those that provide an insight into the causes and prevention of osteoarthritis following injury, identify suitable candidates for hamstring-lengthening surgery, and assess exercise programmes to reduce damage to joints [14].

Musculoskeletal model outputs are highly dependent on the anatomical description of the muscles, their lines of action, cross-sectional areas, and objects over which they wrap to change their path [15]. Customisation of these improves modelling accuracy when compared to generic scaling [16–18], and there are now data sets that can be used to customise anatomy for both the upper limb [19] and lower limb [20]. These data sets can be used as atlases to select the most appropriate data set to scale to [20] or to create statistical representations to enable morphing of data sets to an individual [21].

However, all of these approaches suffer from representing healthy able-bodied anatomies (or, even, cadavers) and the anatomy of those with severe blast injuries, and anatomical deficits have greater variability than is captured in these [22]. Therefore, there is an urgent need to enable the customisation of musculoskeletal models for a specific blast-injured patient.

46.3 Examples

There is currently no published clinical use of musculoskeletal modelling for blast rehabilitation, yet there are multiple studies that have demonstrated their utility on a population basis, if not an individual basis. These are summarised below.

46.3.1 Biomechanics and Osteoarthritis in Amputees

High-functioning military transtibial amputees (TTAs) with well-fitted state-of-the-art prosthetics have gait that is indistinguishable from healthy individuals [23], yet they are more likely to develop knee osteoarthritis (OA) of their intact limbs [24]. This contrasts with the information on the knees of the amputated limbs that have been shown to be at a reduced risk of pain and OA [25, 26]; this may be due to subtle protective measures. There is a mechanically based hypothesis that these amputees have the capacity to achieve the same gait as able-bodied subjects and thus they do not need to compensate by changing or adapting the task; however, their musculoskeletal reserve is significantly reduced, thus overloading joints and muscles. As the task does not change (the kinematics are unchanged), this problem is suited to an inverse dynamics approach, where the measured kinematics and kinetics are used as inputs to quantify muscle and joint forces as outputs.

In a gait analysis study of military unilateral TTAs and matched healthy controls, muscle forces and joint contact forces at the knee were quantified using inverse dynamics-based musculoskeletal modelling [6, 8], validated using measures of muscle activation [27]. This study found that peak knee contact forces for the *intact* limbs on both the medial and lateral compartments were significantly greater than the healthy controls (Fig. 46.3) and that the intact limbs had greater peak semimembranosus and gastrocnemius muscle forces compared to the controls. The study provides robust evidence that supports a mechanically based hypothesis to explain the documented higher risk of knee OA in this patient group and thus opens the way for mitigation strategies through rehabilitation and technology such as improvements of the prosthetic foot control and socket design and strengthening of the amputated muscles.

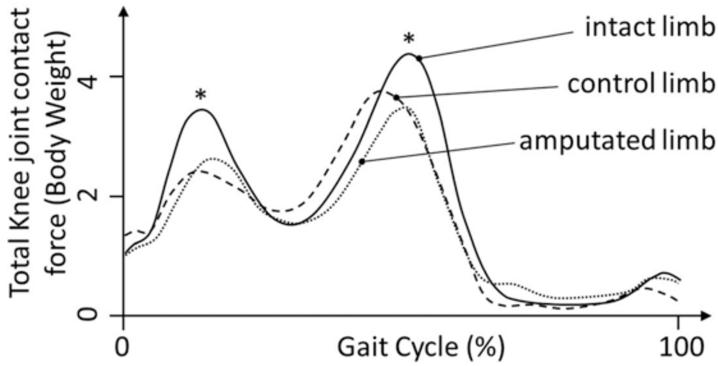

Fig. 46.3 Military unilateral transtibial amputees display increased loading on their intact limbs, compared to able-bodied controls (* represents a statistical difference) [28]

46.3.2 Optimising Prosthetic Parameters

Prostheses for amputees can be designed to optimise many different parameters. For example, the stiffness of a prosthesis can be tuned to optimise running speed in running blades, or by modifying the damping algorithm in a microprocessor-controlled prosthesis an amputee's stability can be enhanced to facilitate walking over different terrains. Powered actuation prostheses can provide a replacement for the muscle anatomical deficit due to the amputation, yet these are not in widespread use following lower limb amputation and therefore the relatively cheaper passive prostheses remain more widely used.

Passive devices for transtibial amputees typically comprise tibial and foot segments connected by a spring and damping mechanism at the ankle. This means that there are modifiable elements such as the coefficients of stiffness and damping, the rest angle of the mechanical ankle, and the masses and lengths of tibial and foot segments. These parameters can be optimised for the different uses proposed, yet the methods to optimise these typically require design, personalisation, and testing of new hardware prototypes. This can be a very lengthy, expensive, and inefficient process that frequently does not result in an optimal solution. Therefore, computational design and optimisation frameworks can be used as an alternative approach to create optimised designs [28].

Where the previous example used an inverse dynamics approach, creating an optimised kinematic output would require a forward dynamics approach. Therefore, studies that optimise parameters such as the maximum achievable running speed require the use of musculoskeletal models that use forward dynamics.

A recent study used an OpenSim 19-muscle, 14-degree-of-freedom neuromusculoskeletal model with a unilateral below-the-knee passive limb prosthesis. The model was used to optimise two key variables: minimising the muscle activation required for the motion and maximising the contralateral ankle kinematics' fit with able-bodied kinematics (Fig. 46.4). Unlike in other studies, the resulting prosthetic kinematics was not constrained to follow any particular path of motion. As a test of the feasibility of using such modelling to design prostheses, a single prosthetic variable was chosen that could be changed. In this case, the coefficient of damping of the passive ankle spring of the prosthesis was selected [29]. Damping was chosen as prior work has suggested that the manipulation of damping induces more natural ankle kinematics [30]. Minimising muscle activation was chosen specifically to test the ability of the model to reduce the muscular effort required to achieve the task; if the muscular effort is reduced, then this increases the subject's *reserve* for the task being performed and minimises fatigue.

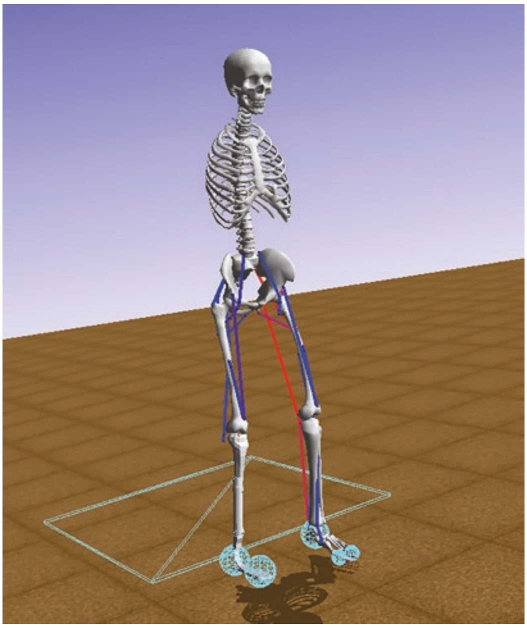

Fig. 46.4 Graphical representation of the OpenSim humanoid model with below-knee passive prosthesis in the simulated environment (from Rane, 2020 [29])

The study found that by increasing damping from its default value, the model was able to increase general adherence to able-bodied kinematic trajectories while simultaneously decreasing energy cost. These kinematic improvements were mainly found during stance phase on the prosthetic limb, and the improvements were observed throughout the kinematic chain: at the pelvis, and in both the prosthetic limb and the intact limb. This highlights the interdependence of kinematic trajectories during walking. Of note, the improvements found on the intact-side knee could relate to the compensatory changes that the previous study showed were likely associated with mechanically induced osteoarthritis.

Therefore, this study has shown for the first time how musculoskeletal models can be used to enable design features to be optimised for optimal rehabilitation and prosthetics design. Their more widespread use is constrained by computational time as these studies require huge computational resource and there is a lack of clinically relevant computational tools.

46.3.3 Functional Electrical Stimulation Interventions

As joint movement is a function of external forces and muscle forces, where these also produce forces in ligaments and at the joints, changing the muscle forces not only potentially changes the movement, but can also change joint forces and ligamentous forces. In addition, as the musculoskeletal system has a degree of redundancy and there are many different contraction patterns that can produce the same kinematic output, changing muscle contractions during motion can preferentially offload selected components of the musculoskeletal system. Rehabilitation techniques can focus on changing the motion (kinematics) or the muscle forces, and technologies can assist with this. One technology that shows promise is functional electrical stimulation (FES) and is already currently used in neuro-rehabilitation. Work has shown that activating muscles crossing the knee with FES is able to reduce anterior tibial translation [31], anterior tibial shear force, and tibial internal rotation torque [32], which are relevant for certain anatomical deficits such as anterior cruciate ligament disruption. Additionally, FES does not have to be applied continuously as FES can be used to learn a muscle contraction pattern that then, once learned, persists despite halting the use of FES [33].

Inverse dynamics-based musculoskeletal modelling can be used to simulate the effect of preferential muscle activation using FES [34] in order to create desired musculoskeletal outputs such as reduced medial knee joint loading [35]. This is a type of *compensation* where the task does not change, but the way the task is achieved by the musculoskeletal system is modified. Therefore, this is a tantalising technology that has the potential to be used in blast injury rehabilitation, to reduce knee forces on the intact limb of unilateral transtibial amputees [27], reduce the very high hip joint contact forces in bilateral above-knee amputees, or compensate to reduce fatigue in overloaded muscles such as rectus femoris in above-knee amputees [22].

46.4 Conclusion

Musculoskeletal modelling utilising advanced inverse and forward dynamics approaches has been shown to be an effective tool in blast injury rehabilitation. The models have demonstrated their utility in a number of areas, including reducing osteoarthritis in blast-induced traumatic amputees, optimising prosthetic parameters to reduce fatigue, and changing gait patterns to reduce muscle fatigue and highly loaded joints. However, despite this promise, because the technology is currently mainly only used in a research environment and has not yet reached the stage where there are clinical software packages, its use is not yet widespread.

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Pain

47

Paul Wood, Peter F. Mahoney, and Dominic J. Aldington

Abstract

Blast injury is associated with significant widespread trauma that often results in pain, both acute and chronic. Persistent pain is thought to be predominately neuropathic in nature.

This chapter starts by considering some of the probable pathophysiological processes that may lead to the development of neuropathic pain.

In the second section, techniques to manage the pains, both acute and chronic, are outlined. There is an emphasis on the biopsychosocial nature of pain, especially persistent pain.

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47.1 Pathophysiology

47.1.1 Introduction

The clinical effects of blast on the whole body are well documented. Injury may result solely from, or as a combination of, primary, secondary or tertiary blast effects as described in Chap. 9. Pain is experienced as a consequence of tissue disruption including direct and indirect injury to the nervous system. Post-injury, both acute and chronic pain is often neuropathic in nature due to damage to the nervous system and is frequently associated with a sustained systemic inflammatory response (SIRS) [1, 2].

Throughout this chapter, neuropathic pain (NeuP) is emphasised because it is distinguished by two significant factors—firstly, its tendency to chronicity, irrespective of apparent resolution of any suspected cause, and secondly, its resistance to treatment.

47.1.2 Definitions

In making sense of clinical and research pain literature, it is helpful to have clear definitions. Table 47.1 lists the key descriptors employed in this chapter. All are detailed by the International Association for the Study of Pain (IASP) [3]. Causes of neuropathic pain (NeuP) associated with blast are detailed in Table 47.2.

Table 47.1 Pain terminology. Adapted from [3]

• Pain—an unpleasant sensory and emotional experience associated with actual or potential tissue damage or perceived as such
• Nociceptor—peripheral afferent nerve receptors responding to high-threshold noxious stimuli
• Nociceptive pain—pain secondary to nociceptor stimulation by disease or injury; this includes inflammatory pain
• Neuropathic pain (NeuP)—pain resulting from disease or damage to the somatosensory nervous system
• Nociplastic pain—a recently introduced term to describe pain without apparent actual or imminent tissue injury and therefore cannot be classified as either nociceptive or NeuP
• Acute pain—usually nociceptive in nature; expected to resolve following 'cure' of the precipitating event
• Chronic pain—pain extending beyond the resolution of injury/surgery or persisting for more than 3 months; it is often neuropathic in nature
• Hyperalgesia—heightened response to a noxious stimulus; primary hyperalgesia occurs at the site of injury, whereas secondary hyperalgesia extends to adjacent uninjured tissue
• Allodynia—painful response to an ordinarily benign stimulus, e.g. light touch; postulated mechanisms include increased spinal cord (dorsal horn) excitation causing non-nociceptive inputs to be interpreted as pain
• Causalgia—sustained burning pain resulting from traumatic nerve injury; it is associated with profound allodynia, and features suggesting regional sympathetic dysfunction may coexist
• Sensitisation—increased responsiveness of nociceptive neurons to their normal input and/or recruitment of a response to normally subthreshold inputs; clinically, this presents as hyperalgesia and/or allodynia
• Wind-up—a component of spinal cord sensitisation; defined as an increased output of nociceptive neurons subject to a continuous fixed input; associated with spinal cord (dorsal horn) excitation of N-methyl-D-aspartate (NMDA) receptors; wind-up is experienced by wide dynamic range (WDR) neurons within the dorsal horn of the spinal cord

Table 47.2 Blast-related nerve injury and NeuP. Adapted from [2]

- Acute peripheral nerve injury, e.g. direct and indirect energy transfer from blast or fragmentation
- Amputation—classic consequence is phantom limb pain (PLP), but neuromas (post-injury nerve fibre swelling) also significant
- Distraction or stretching trauma, e.g. brachial or lumbosacral plexus injury
- Spinal cord injury and traumatic brain injury (TBI)
- Nerve entrapment mechanisms, e.g. heterotrophic ossification (ectopic bone), post-surgical scar or fibrosis resulting in nerve ischaemia (inadequate perfusion)
- Ischaemia from haemorrhage or secondary to prolonged tourniquet application
- Chronic regional pain syndrome following nerve injury

47.1.3 Acute Versus Chronic Pain

Modern concepts of pain as a sensory experience stress the concept of neural plasticity. This rejects the idea of 'fixed' anatomical routes of transmission in favour of a series of neuronal connections from the site of injury through to cortical centres that are capable of dynamic but potentially maladaptive communication. At the cerebral cortex, the anatomical origin of the pain is registered by the sensory cortex, but connections to the limbic system provide the emotional component to the conscious perception [4].

Fundamental to chronic pain is the principle of peripheral and central sensitisation, which results from sustained stimulation of, and neurochemical changes in, peripheral nociceptors (free nerve endings that detect damaging stimuli) and second-order spinal neurons (connections within the spinal cord from the peripheral sensory nerves before upward transmission to the brain) and their connections. The role of the spinal cord dorsal horn is crucial [5]. At this level, there is an increasing appreciation of the role of the midbrain and brainstem as part of a 'top-down' (i.e. from cortical, thalamic and subthalamic locations to the spinal cord) endogenous inhibition of nociceptive transmission [6]. The relevant neurochemistry continues to be studied, but central sensitisation is associated with brainstem responses identifiable with functional magnetic resonance imaging [7].

Clinically, sensitisation presents with hyperalgesia and allodynia. Key features of the neuronal responses contributing to sensitisation are displayed in Fig. 47.1.

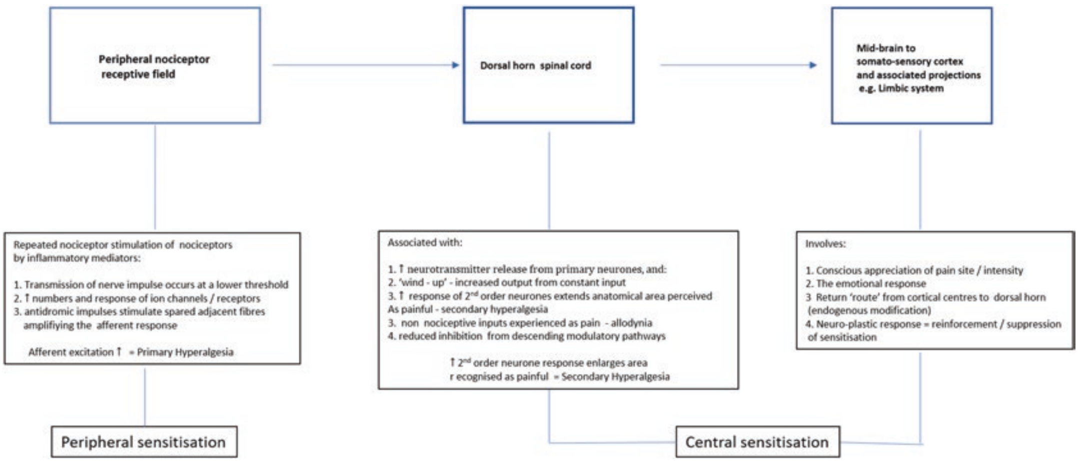

Fig. 47.1 Mechanisms of sensitisation. Adapted from [2] with permission

47.1.4 The Role of Nervous System Inflammation

There is a growing awareness that acute and chronic NeuP may be initiated and maintained by a neuro-immune reaction, involving neurones, immune cells and glia. Historically, glial cells were thought of as cellular supporting architecture, but modern studies have demonstrated a range of functional relationships with neurones [8] (Table 47.3). Like neurones and immune cells, glia release pro-algesic mediators following peripheral nerve or spinal cord injury, and the involvement of these three cell types has been described as the *neuropathic pain triad* [9].

The resulting effects are called *neurogenic inflammation* (NGI) [10], when applied to the peripheral nervous system (PNS), and *neuroinflammation* (NI) when within the central nervous system (CNS), although some authors will refer to NI in either case. The effect of NGI and NI is an increase in excitatory synaptic activity and persistent pain [11]. In humans, NI has been linked to post-traumatic stress disorder, depression, chronic headache and neuromuscular pain [12, 13]. Following exposure to blast, a correlation between NI and traumatic brain injury (TBI) is well established, resulting from both cranial and extra-cranial injury [14, 15]. Postulated precipitating mechanisms include direct trauma from the blast pressure wave, transmission of pressure via the vascular system and associated immune reaction including the infiltration of leucocytes in injured areas of brain secondary to a reduced blood-CNS barrier [16]. Animal studies have identified associated vascular changes and a sustained inflammatory response [14, 15].

Table 47.3 Major nervous system glial cells

Cell type	Location	Role
Schwann cells	Peripheral nerves	Produce the myelin sheath. Involved in conduction of nerve impulses. Interact with immune cells during nerve damage and repair
Satellite cells	Dorsal root ganglion	Nutrition and structural support. Proliferate in response to peripheral nerve injury
Microglia	CNS—dorsal horn of spinal cord/brain	Resident nervous system macrophages directing the immune response to CNS injury
Astrocytes	CNS	Influence synaptic function. Mount inflammatory response to brain and spinal cord injury
Oligodendrocytes	CNS	CNS correspondent of Schwann cells. Provide myelin covering with nutritional support to axons

47.1.5 Peripheral Nerve Injury (PNI)

Narrative accounts of the effects of PNI have been well described from the American Civil War onwards. The classic description of the neuropathic burning pain, *causalgia*, following nerve injury, is attributed to Dr. Weir Mitchell (1872) [17]. Seddon's 1942 classification of PNI accorded nerve injuries one of three categories [18] (Table 47.4). This classification was enhanced in 1951 by Sunderland [19], who described five classes by expanding Seddon's analysis of *axonotmesis* (Table 47.5).

Table 47.4 Seddon's classification [18] and clinical correlates

Classification	Anatomy	Outcome/ prognosis
1. Neuropraxia	Nerve contusion, e.g. compression/ ischaemia	Temporary conduction block with motor paralysis and sensory dysfunction Spontaneous recovery in days to weeks
2. Axonotmesis	*Tmesis* derived from Greek—'to cut'. Hence, axon 'cut' with variable degrees of damage to axon and myelin sheath but endoneurium, perineurium and epineurium intact. Causes include stretch or crush injury	Wallerian degeneration occurs in distal segment (see below) Associated with a Tinel sign (paraesthesia elicited by percussion of the nerve at the level of injury) Spontaneous recovery possible but delayed by months
3. Neurotmesis	Nerve physically severed or equivalent physiological disruption	Complete severance associated with distal Wallerian degeneration and proximal neuroma formation Recovery only possible with surgery

Table 47.5 Sunderland's classification [17]

Description	Correspondence to Seddon
First degree	Neuropraxia
Second degree	Axonotmesis (AXT)
Third degree	AXT + endoneurium (EN) injury
Fourth degree	AXT + perineurium (PN) injury
Fifth degree	Neurotmesis (AXT + EN + PN + epineurium)—complete transection

Seddon's second and third classes are significant for the presence of *Wallerian degeneration* distal to the point of injury. In these cases, axonal degeneration is followed by the breakdown of the myelin sheath and removal by Schwann cells and macrophages as a prelude to attempted nerve regeneration. Wallerian degeneration is itself sufficient to initiate NeuP [20, 21].

Schwann cells are directly involved in the immune response by releasing chemokines, which facilitate further immune cell recruitment. The resulting concentration of leucocytes releases additional inflammatory mediators sustaining the immune reaction [11, 22].

Animal models have demonstrated an associated inflammatory response in the dorsal root ganglion and spinal cord horn [22] plus related microglia activation in the midbrain suppressing descending inhibition of dorsal horn excitation [23]. Interestingly, this mechanism is also postulated to have a role in pain after traumatic brain injury [13]. Microglia activity associated with NeuP has also been observed in the thalamus of human amputees [24].

Preclinical studies have visualised histological changes in intact un-severed peripheral nerves following shear stress [25] and also after ballistic injury to a contralateral limb [26]. The relevance of these findings to the potential for NeuP in human casualties subject to corresponding injury mechanisms, in particular the primary blast wave, is not known. A further precipitant is ischaemic injury from prolonged tourniquet use [27]. While this has been recognised following haemorrhage control, the effects of systemic hypotension are also unknown.

Experience from the peripheral nerve injury clinic established between 2005 and 2010 at the UK Defence Medical Rehabilitation Centre, Headley Court, was that neuropraxic injury was the commonest form of PNI [28]. Overall, 36 of 100 consecutive patients with PNI (excluding amputations) presented with persistent NeuP, with 63% of a total of 261 nerve injuries caused by trauma from improvised explosive devices (IEDs).

Forty-four per cent of all of the nerve injuries examined did not involve division, and prolonged recovery from conduction block was usual (mean 4.17 months). The commonest indication for reoperation was continuing NeuP.

47.1.6 Phantom Limb Pain (PLP)

The conflict in Afghanistan (2006–2014) resulted in 275 UK amputee casualties, the majority suffered in explosions [29]. PLP is almost inevitable post-amputation, and modern medicine still remains largely impotent with respect to treatment options. It can be considered as the definitive example of NeuP, but its physiological basis remains disputed. Current theories favour combined peripheral and CNS inputs [30]. Experience at the University Hospitals Birmingham (the UK receiving facility for military casualties) revealed that PLP was a concern for 76% of amputees at some point [31]. This figure was consistent with previous reports, presenting significant implications for rehabilitation.

47.1.7 A Current Conceptual Model of PNI and NeuP

Figure 47.2 illustrates a current model for the relationship between PNI and NeuP. Note the directional signalling between the three involved cellular elements. It is reasonable to speculate as to how neurogenic inflammation might contribute to the systemic inflammatory response experienced by ballistic casualties. There is evidence that TBI is worsened in the presence of extracranial injuries [32], but the systemic effects, if any, of peripheral traumatic neurogenic inflammation are presently conjecture.

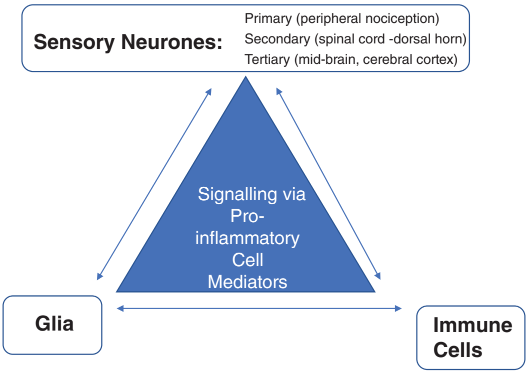

Fig. 47.2 The neuropathic pain triad

It should be noted that normally the immune system directs a compensatory anti-inflammatory response syndrome (CARS) to limit the possibility of persistent inflammation (Chap. 13). There is potential for selective inhibition and reinforcement of mediators as future therapies for NeuP [11, 22].

47.1.8 Patient Assessment

External objective assessment of what is ultimately a personal experience is difficult. A key tenet is to avoid reliance on unidimensional tools, which only measure one aspect of a patient's pain experience. The relationship of pain to function is fundamental to outcomes from subsequent rehabilitation. Specialised scoring systems are available for clinical and research activity, including for NeuP, but quantitative sensory testing is in practice confined to the research setting.

47.1.9 Clinical Correlates

Experience of clinicians during the Afghanistan conflict emphasised that severe injury scores, often of an order previously associated with unsurvivable injury [33], were predictably associated with difficult pain problems. Particular issues included:

- amputation-related PLP [31];
- post-amputation neuromas or peripheral nerve entrapments (see Table 47.2);
- excluding persistent occult infection as the pain generator;
- lumbar-sacral or pelvic nerve root/plexus injury;
- the 'acute on chronic' pain experienced during repeated surgical interventions;
- the confounding influence of TBI; and
- the effects of all of the above factors in creating delays and complications with rehabilitation.

Rehabilitation is the final stage of a treatment timeline that begins with combat casualty care immediately post-wounding. All of the mechanisms discussed above can at some stage impede the patient's eventual functional outcome. Pain management is best achieved by an 'end-to-end' approach [34], commencing at the point of injury. This mandates continual management with a proactive pain team respecting a multimodal approach. Chronic pain intervention requires the input of multidisciplinary specialists utilising a biopsychosocial model.

47.2 Treatment

47.2.1 Introduction

Probably, the most important aspect of managing any pain is to consider it as a 'biopsychosocial' condition [35]. Thus, it should not be surprising that biological techniques alone will not solve all pain issues, and neither will only psychological or social based techniques. The underlying proportions of the three components will change during the evolution of the pain, and healthcare providers must never lose sight of the separate aspects. Patients should also develop an understanding of their importance.

The rest of this chapter will consider the management of pain from the point of injury to rehabilitation and beyond. At each stage, it is assumed that all other aspects of blast casualty care are ongoing. Throughout the period of care, the options available for pain management will depend upon the environment, casualty numbers, providers' skill sets, and the equipment available. Initially, resources may be very limited, but as care proceeds, a significant increase is to be expected.

47.2.2 Immediate Response

Responders will initially concentrate on removing casualties from further danger and basic first aid. Priorities will include haemorrhage management and clearing airways [36], but concurrently acute pain will, where possible, be addressed using a methodical approach (Table 47.6), beginning with assessing severity.

Table 47.6 A structure for treating acute pain

Treating pain requires a combined (multimodal) approach:
• Psychology—attention and reassurance
• Physical methods—e.g. cooling and covering burns, elevation, splinting and distracting fractures (accept medication may be required for fracture reduction)
• Pharmacology – Systemic use of drugs to target analgesic receptors and inflammatory mechanisms – Topical with local anaesthetics to inhibit nerve conduction of pain signals
There is evidence that, humanitarian considerations aside, early attention to acute pain may influence the development of chronic pain syndromes. Using drugs in combination increases overall analgesic efficacy while limiting the unwanted effects of any single agent. Initially, severe pain may require a combination of systemic and regional analgesic techniques. As pain improves, drugs are weaned until the minimum possible are consumed.

There are many different scores and scales available, but the most effective is to simply ask the casualty how severe their pain is. A scale of 0, for no pain, to 3, the maximum, provides clinically relevant information, since each step change is probably clinically significant. The importance is to legitimise the casualty's experience while providing an indication of the severity of the pain and thus the treatment needed. Subsequent use of the same scale will provide an idea of the efficacy of the analgesic interventions and must be repeated regularly throughout the casualty's care, with the aim of pain treatment being a reduction in the level of pain experienced to the first third of any scale used [37].

Inevitably, drugs will be required at some point (Table 47.7). The medication to consider will depend on the expertise of the provider, any licensing restrictions, the environment and the comorbidities of the patients. This chapter cannot provide prescriptive rules about which agent to use within any situation, but Table 47.7 details core drugs. It is important to understand that there is no proven, evidence-based, benefit of one agent over another.

Table 47.7 Acute pain pharmacology

Drug	Mechanism of action	Comment
Systemic agents		
Paracetamol (acetaminophen)	Uncertain but CNS effects apparent	Analgesic but not anti-inflammatory. In combination with other agents tends to have a synergistic action Very safe unless taken in overdose (liver failure) [38]
Non-steroidal anti-inflammatory drugs (NSAIDs), e.g. ibuprofen, diclofenac	Inhibit cyclooxygenase enzymes, which are key in producing pro-inflammatory mediators	While proven analgesic agents, they can have significant effect on other systems such as reducing renal blood flow, which may already be critically impaired as a result of shock. A putative negative effect on bone healing has a poor evidence base [39]
Opioids (narcotics), e.g. morphine, fentanyl	Agonists of specific receptors found throughout the central and peripheral nervous system	Historically, opioids are the default analgesics for severe pain. Opiate receptors respond to a variety of narcotic formulations, particularly in respect of acute visceral pain. Narcotic analgesics are drugs of addiction and have significant side effects including respiratory depression. These considerations have required a reappraisal of their use in chronic pain conditions A number of opiates exist, with a variety of routes of administration (e.g. oral, intramuscular, intravenous) and onset and duration profiles. There is little evidence to choose one over another, but for comparison purposes, morphine is traditionally regarded as the gold standard
Ketamine	A phencyclidine (hallucinogenic) derivative which antagonises particular receptors associated with the spinal processing of pain	In larger doses, it causes anaesthesia, while in lower doses, analgesia tends to predominate albeit not without the psychomimetic actions Evidence suggests that it reduces opioid requirements in the acute phase [40]. Its use in the prehospital environment is well established [41]
Inhaled agents		
Methoxyflurane	Central nervous system depression	Historically used as a volatile anaesthetic agent but modernly excluded because of concerns over renal toxicity, which have not been reported in analgesic doses. The inherent analgesic activity has driven a resurgence in its use in the prehospital environment [42, 43]
Entonox®	Central nervous system depression	Entonox® is a trade name for a mixture of 50% oxygen and 50% nitrous oxide (N_2O), supplied in cylinders at 137 bar. It provides relatively rapid analgesia with minimal cardiovascular, respiratory or neurological side effects
Regional (local anaesthetic agents)		
e.g. lignocaine, bupivacaine	Supress excitation of nerve membranes by blocking transmission of electrical signals	In the prehospital environment [44], appropriately skilled operators may employ sighted single-shot regional anaesthetic techniques. Subsequent surgical evaluation of injuries could be made problematic if the receiving teams are unaware. For this reason, it is of paramount importance that the presence of any regional anaesthetic 'block' should be clearly recorded

47.2.3 Perioperative Care

Once the patient has been stabilised, they enter the perioperative phase of their care, which may involve multiple interventions over a protracted length of time. The fundamental principles of analgesia that were initiated in the prehospital environment are continued and developed further, including extending the utility of regional analgesia by epidural or continuous peripheral nerve blocks [45].

Patients must be regularly asked about the severity of their pain. Specific side effects of analgesic interventions, such as constipation, nausea or drowsiness, should also be addressed. At this stage, it is possible that the psychological components of pain will become more significant, especially in relation to anxiety over prognosis (Fig. 47.3).

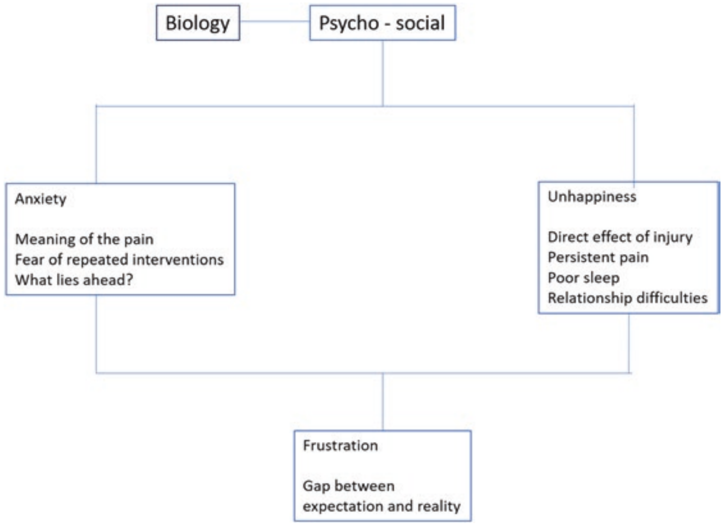

Fig. 47.3 Pain: biopsychosocial structure

Stringent efforts in reducing pain experienced during this period may potentially reduce the incidence of chronic post-trauma pain, and in particular, consideration should be given to agents that may reduce neuropathic complications, e.g. gabapentinoids (Table 47.8). It should be noted that currently there is no definitive evidence that they reduce subsequent chronic pain conditions, but their co-analgesic action does allow a reduction of other drugs.

Table 47.8 Anti-neuropathic co-analgesics

Drug	Action	Comment
Gabapentinoids, e.g. gabapentin and pregabalin	Suppress neurone membrane excitation (different mechanism than local anaesthetics)	May reduce neuropathic pains following trauma [46]; administered perioperatively, they can beneficially reduce opioid requirements [47]
Tricyclic antidepressants, e.g. amitriptyline	Multiple actions include CNS noradrenaline/serotonin reuptake inhibition. Resulting effect promotes spinal inhibition of pain signalling	Tricyclic antidepressants are a class of drug that are now principally reserved for managing NeuP. Despite variable evidence [48–50], they are frequently included in guidelines for the management of neuropathic pains. They assist sleep, which is frequently problematic in pain patients
Benzenoid analgesics, e.g. tramadol and tapentadol	Dual actions: 1. Opioid receptor agonist 2. Noradrenaline/serotonin reuptake inhibition within the CNS	Anti-neuropathic action derives principally from the second action [51]; tapentadol is also recognised to be potentially helpful in chronic musculoskeletal conditions [52]

47.2.4 Chronic Pain and Rehabilitation

47.2.4.1 Introduction

Chronic pain is defined as pain that has existed for more than 3 months. This definition has nothing to do with mechanism, nor severity, but is a chronological definition.

The prevalence of chronic pain after blast injury is unclear. Studies showing a link between traumatic brain injury and chronic pain [13] include some patients subjected to blast injuries.

47.2.4.2 Treatment

Probably, the most important single aspect of the long-term management of pain is for the patient to become the expert in their pain. This requires education of the patient to enable engagement and decision-making. However, it also needs healthcare professionals who are willing to support this model of treatment. It is an accepted methodology in the management of other conditions, such as hypertension and diabetes, and yet there is often reluctance by all parties to adopt this model for chronic pain.

When discussing persistent pain with patients, it is important to ensure that they understand that their pain is 'real' and is not fictitious. Many patients find it difficult to understand why pain should continue in the absence of further injury and hence believe that either there is more damage or the pain is not real. This thinking is incorrect as significant damage can occur with minimal pain and *vice versa*. Unless suitably educated, phantom pain can be particularly destructive for amputees who are struggling to understand why they are experiencing pain in a part of their body they have lost.

Interventions aimed at the emotional components of pain start with recognising its relevance. The next option is to talk to people; friends and family are one possibility although their input cannot be completely comprehensive and will require the support of multidisciplinary pain services. Anxiety and unhappiness may benefit from medication, but there are no drugs for frustration (Fig. 47.3).

The management of the physical components of chronic pain follows a relatively well-worn path. Exercise and continuing activity are fundamental. Indeed, if interventions for pain management do not improve rehabilitation, there is an argument to suggest that they should be withheld. Together with exercise and activity, the use of simple hot and cold packs can be helpful for musculoskeletal pains. There are also a number of topical agents available [53], with varying pharmacology. Some preparations contain a non-steroidal anti-inflammatory drug (NSAID) (Table 47.7); others are used to create heat or cooling to the area they are applied to, including capsaicin-based products that are thought to work via action on specific peripheral receptors. Finally, there are the local anaesthetic products.

As time progresses, it becomes more important to focus on the wider psychosocial components of pain (Fig. 47.3) [54], whose potentially negative presence will not be influenced by increasing analgesia. It is important for patients to ensure that their medication is actually providing the benefit they hope for, and the best way to do this is to encourage regular trials of dose reduction. If the medication does not increase their ability to work, rest or play (as the patient defines these), then its value must be questioned.

47.2.4.3 Comorbidities

Any concomitant condition affecting a patient can increase experienced pain, either via direct biological interaction or by indirect psychosocial mechanisms.

The greatest confounders are mental health issues, and within the realm of blast, the biggest problem is likely to be post-traumatic stress disorder (PTSD) [55]. Unfortunately, each condition may reinforce the other [56]. Until recently, it has been standard clinical practice to manage these concurrently rather than separately because of different healthcare provider skill sets. Some practitioners question this approach and, although not evidence based, believe that better outcomes will result from treating these conditions together.

47.2.4.4 Prognosis

Regardless of the techniques used, some survivors of blast injury will be left with pain long after their initial medical interventions.

By understanding the 'meaning' of their pain, the value of their medication and how to titrate the doses for best effect, patients are able to proactively respond to the fact that chronic pain is not static. A good prognosis seems to be associated with not having the mindset of a 'victim'. To facilitate this, a survivor should be supported (while avoiding unnecessary medicalisation) to maintain their sense of purpose in life, accept their losses and adjust their sense of self-worth accordingly.

47.3 Summary

Good pain management is not always possible to achieve but should always be attempted. The act of striving towards it can have almost as much benefit as achieving it.

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Correction to: Rehabilitation Lessons from a Decade of Conflict

Peter A. Le Feuvre, Kate E. Sherman,
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The chapter was inadvertently published with military ranks included along with author names and with an error in the running head. These errors have now been corrected.

The updated original version of this chapter can be found at
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Glossary

Acoustic impedance A physical property of a material that defines the ease with which stress waves can travel through it.

Actin, f-actin, actin dynamics The most abundant protein in most eukaryotic cells. It is a globular protein that can exist in soluble or polymerised forms. Filamentous, f-actin, is a polymerised form that assembles as microfilaments, which are present in the cytoskeleton. Soluble and polymerised forms play many cellular roles including maintaining cell shape and enabling cell motility. The processes involving transitions between soluble and polymerised forms can be referred to as actin dynamics.

Adenosine triphosphate, ATP A small molecule used by cells in many metabolic processes. Hydrolysis or release of its phosphate groups is the most common chemical reaction associated with its cellular functions that include phosphorylation and signalling.

Adiabatic A process for which there is no heat transfer between a system and its surroundings. An adiabatic process that is reversible is isentropic.

Alginate A polysaccharide that can form gelatinous structures suitable for the encapsulation of cells.

Amorphous Lacking definite form.

Angiogenesis The process involved in the growth of new blood vessels from pre-existing ones.

Anisotropic A material whose microstructure is such that its mechanical behaviour depends on the direction of loading.

Apoptosis A cell-controlled mechanism for causing cell death—sometimes referred to as "programmed cell death". A number of well-known changes occur in cells during apoptosis. These changes can include membrane blebbing (formation of bulges from the plasma membrane), cell shrinkage, chromatin condensation and DNA fragmentation. The remaining cell debris may be engulfed by other cells. A key difference between apoptotic cell death and necrotic cell death is the absence of an inflammatory response.

Aseptic necrosis Cell death arising from a loss of blood flow.

Astrocytes Also known as astroglia. Star-shaped glial cells that interact with and support the function of neural cells.

Atom The smallest component of an element, having the chemical properties of the element.

Bending A loading mode whereby the structure deforms by changing in curvature.

Brisance The shattering behaviour exhibited by a detonating explosive.

Brittle A material that undergoes negligible plastic deformation prior to rupture/failure.

Buckling A failure mode due to instability in the long axis of a slender structure.

Burning The propagation of combustion by a surface process.

Carbide A compound formed of carbon and a less electronegative element.

Cell viability The ability of live cells to survive.

Cerebral vasospasm An intense constriction of arterial vessels in the brain, reducing blood flow to surrounding tissues.

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Chemokines Cytokines that specifically attract cells, usually to sites of inflammation and/or infection.

Coagulopathy (clotting/bleeding disorder) A condition in which the blood's ability to coagulate (clot) is impaired.

Conservation equations Mathematical expressions which state that a property of a physical system (mass, momentum and energy) is conserved (i.e. do not change) as the system evolves over time. In shock physics, they are also known as the jump conditions or the Rankine–Hugoniot relations.

Constitutive relation A relationship between two physical properties (such as temperature, pressure and volume) of a material that describe its behaviour to external loads.

Constructive interference The interference of two or more waves of equal frequency and phase, resulting in their mutual reinforcement, thus producing a wave of amplitude equal to the sum of the amplitudes of the individual waves.

Cortical neurons Nerve cells located in the outermost layer of the brain, which is known as the cerebral cortex.

Cytokines A class of small proteins secreted by several cells that interact with other cells and alter cellular behaviour.

Cytoskeleton The complex set of filaments and tubules within a cell that provides mechanical support, maintaining shape and facilitating motility.

Deformable solid A solid that can change shape or volume when external loading is applied; all solids are deformable.

Detonation An extremely fast explosive decomposition in which an exothermic reaction wave follows and also maintains a shock front in the explosive.

Detonation pressure The dynamic pressure in the shock front of a detonation wave.

Discretisation A procedure when setting up a numerical simulation in which the domain is subdivided into cells or elements of finite dimensions.

Ductility The ability of a solid material to sustain plastic deformation prior to rupture/failure.

Eigenvalues A set of scalars associated with a linear system that are invariant under a linear transformation.

Electronegative The tendency of an atom to attract electrons towards itself.

Erk1/2 signalling pathway A cellular pathway triggered by extracellular substances that in turn triggers other cellular pathways to affect cellular functions such as differentiation, proliferation and survival.

Extracellular matrix A mixture of proteins and carbohydrate-containing molecules external to cells that provide structural and biochemical support to surrounding cells.

Extracellular matrix metabolism The processes involved in the production and degradation of the constituents of the extracellular matrix.

Explosion A violent expansion of gas.

Extravasation A passage or escape into the tissues, usually of blood, serum or lymph.

Glycoproteins Proteins containing glycans covalently attached to a polypeptide side chain.

Growth factors Molecules able to interact with cells and cause or "stimulate" them to grow and possibly proliferate. Growth factors can be small molecules such as vitamins or hormones or even large molecules such as proteins.

Haemorrhage (Pathology) bleeding from a ruptured blood vessel.

High (order) explosive An explosive capable of detonation under the normal conditions of use.

Homeostasis The maintenance of cellular parameters (e.g. concentrations of cellular components) within the range for proper functioning of the cell. For example, the cellular processes that maintain the appropriate concentration of cations engage in cation homeostasis.

Hydrophilic Having a tendency to mix with, dissolve in, or be wetted by water.

Hydrodynamic pressure The difference between the pressure of a fluid and the hydrostatic pressure.

Hydrostatic pressure (Fluids) The pressure exerted by a fluid at equilibrium at a given point within the fluid due to gravity.

Hydrostatic stress (Solids) The mean normal stress (= 1/3 of the sum of the three normal stresses in a three-dimensional stress system).

Inflammation; inflammatory responses or pathways Complex biological responses to cell injury that can result in a number of physiological changes such as fever, swelling, and removal of damaged cells and tissues. Inflammation is a protective response and integral part of the healing processes. Poor or dis-regulated inflammation can lead to negative physiological changes such as unwanted tissue destruction or chronic diseases such as arthritis or allergic reactions. Pro-inflammatory responses refer to the induction of biochemical processes that promote inflammation, while anti-inflammatory responses refer to the induction of biochemical processes that reduce inflammation.

Inorganic Not consisting of, or derived from, living matter.

Integrins A class of proteins that transverse cell membranes and that facilitate interactions between different cells or between a cell and the extracellular matrix.

Ischemia A restriction in blood supply to tissues.

Isotropic A material whose microstructure is such that its mechanical behaviour does not depend on the direction of loading.

Lattice A repetitive arrangement of atoms.

Lysis Disruption of a cellular membrane that results in the release of cellular contents.

Machinability The ease with which a material can be cut.

Malleable Capable of being extended or shaped (by hammering or pressure).

Mesenchymal stem cells, MSCs Cells derived from connective-tissue frameworks that have the potential to differentiate into certain cell types. As the source of these cells can vary, they are sometimes referred to as multipotent stroma cells.

Microstructure The structure of a material as revealed by a microscope above 25× magnification.

Mitochondria A cellular organelle that primarily functions to enable cellular respiration and energy production.

Mitogen-activated protein kinases, MAPKs See definition of *Erks*.

Necrosis Biochemical processes associated with cell death arising from external factors such as mechanical damage, toxic substances, or infection. Unlike apoptosis, necrotic processes are unregulated and include a loss of membrane integrity and uncontrolled digestion of cell components. Inflammation is often associated with necrosis.

Neurotransmitter A chemical that acts as a messenger for transmitting information between nerve cells.

Oedema, Oedema, Dropsy, Hydropsy A condition of abnormally large fluid volume in the circulatory system or in the tissues between cells.

Osteocytes Non-dividing cells, embedded in bone, derived from osteoblasts. The majority of bone cells are osteocytes. Osteocytes are involved in tissue remodelling and biochemical processes such as phosphate metabolism and calcium availability.

Oxidative stress The state a cell is in when there is overabundance of reactive oxidative species, which can cause cellular damage.

Parenchyma The functional parts of an organ in the body. In contrast to the *stroma*, which refers to the structural parts.

Particle velocity The velocity associated with a point of infinitesimal dimensions attached to a material as it flows through space.

Phospholipase C A class of enzymes that cleave phosphate-containing groups from phospholipid molecules. These enzymes are often part of signal transduction pathways.

Phosphorylation The biochemical addition of a phosphate-containing group to a macromolecule such as a protein or a lipid. Phosphorylation is a key biochemical step in many signal transduction pathways.

Polycrystalline Of many crystals of varying size and orientation.

Protein nitration The biochemical modification of proteins, usually at tyrosine residues, involving the addition of a nitro-containing group. Protein nitration is generally associated with stress responses and can involve loss or gain of function.

Proteoglycan Any of a group of polysaccharide-protein conjugates present in connective tissue; they form the ground substance in the extracellular matrix of connective tissue and also have lubricant and support functions.

Quasi static Infinitely slow.

Rarefaction wave A wave that reduces the normal stress (or pressure) inside a material as it propagates; the mechanism by which a material returns to ambient pressure after being shocked (the state behind the wave is at lower stress than the state in front of it). Also known as unloading, expansion, release, relief, or decompression wave.

Reperfusion The act of restoring the flow of blood to an organ or tissue.

Reperfusion Injury The tissue damage caused when blood supply returns to the tissue after a period of ischemia or lack of oxygen.

Shock impedance Defined as Z ¼ *p* U. Describes the ability of material to generate pressure under given loading conditions. Generally a function of pressure.

Shock velocity The velocity of the shock wave as it passes through the material; the velocity is pressure/stress dependent. In the limit of an infinitesimally small shock wave, it is equal to the bulk sound speed of the material.

Shock wave A wave that travels at a velocity higher than the elastic (uncompressed) sound speed of the material.

Signal transduction The transmission of extracellular signals to the interior of a cell. The process can involve a wide range of small molecules and macromolecules.

Spallation The process whereby fragments are ejected from a body due to impact or stress.

Sequelae A pathological condition resulting from disease or injury.

Stem cells Cells that have no specific function other than having the potential to develop into another cell type with a specialised function, or multiple cell types with different functions. In adults, stem cells can be found in bone marrow, blood and fat.

Stem cell differentiation The process of a stem cell changing into a cell type with a specialised function.

Strain energy The energy stored by a system undergoing deformation.

Tetrahedron A polyhedron composed of four triangular faces.

Trabeculae Fine spicules (needle like) forming a network in cancellous (porous) bone.

Trace evidence Fragments of physical evidence such as hairs, fibres and glass, transferred when two objects touch or when small particles are disbursed by an action of movement.

Tropocollagen The basic structural unit of all forms of collagen; a helical structure of three polypeptides wound around each other.

Velocity of detonation This is the speed at which a detonation wave progresses through an explosive. When, in a given system, it attains such a value that it will continue without change, it is called the stable velocity of detonation for that system.

Viscoelasticity The property of materials that exhibit both viscous and elastic characteristics when undergoing deformation.

Viscosity A measure of a fluid's resistance to flow.

Vulcanise To treat with sulphur and heat thereby imparting strength, greater elasticity and durability.

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