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The Mechanobiology and Mechanophysiology of Military-Related Injuries

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Preface

Combat soldiers are being rigorously prepared to perform at extreme conditions, which expose them to various types of potentially-serious injuries. Life-threatening situations are understandable as part of combat scenarios, which motivates considerable efforts in developing guidelines and training protocols, as well as protective measures that are technologically-based (e.g. specialized clothing and footwear, protective gear and devices). These measures are all aimed at reducing the incidence of injuries and lessening their short-term and long-term effects, while minimally compromising performance in battle. Besides the aforementioned combat injuries, many of the other injuries reported among recruits and soldiers are a consequence of training under extreme physiological and environmental stressful conditions, and are in many cases, related with insufficient adaptation of the body and tissues to the requirements of training programs.

While epidemiological surveys highlight a clinical picture, the etiology is much more complicated to investigate, and more difficult to understand. Understanding the biophysics and biomechanics of injuries is a fundamental part in enhancing the survivability and performance of soldiers. Ultimately, better understanding contributes to development of improved guidelines, training protocols and programs as well as materials and other technologies for better body and tissue protection.

The present book in the series *Studies in Mechanobiology, Tissue Engineering and Biomaterials* focuses on the biomechanics, mechanobiology and bio-thermodynamics underlying common injuries which are encountered by recruits and soldiers during their service, either in training or at war.

In the various chapters of this multi-disciplinary book, we focus mostly on musculoskeletal, neurological and heat stress-related injuries. The book summarizes the efforts of an international group of expert authors, from military research institutes and universities in the United States, the United Kingdom, Germany, Australia, and Israel, countries that are all well-known for their massive collective experience in military medicine, military ergonomics and military biomechanics. Within the 11 chapters of the present book, the interested reader will find the most up-to-date knowledge with regard to military injuries and their prevention.

Importantly, this volume makes a unique contribution to the literature, being the first of its kind to describe the specific biomechanics and thermodynamics aspects of common military injuries. The authors, all leaders in their fields, together provide the state of science in understanding military-related injuries. We, the editors, are certain that military medicine experts, scientists and engineers working in the fields of injury biomechanics and development of soldier-oriented protective equipment, will find this book a primary source of knowledge. Finally, the relevant decision-makers in government bodies and militaries should make this book their primary reference, particularly for prioritizing research and scientific-based selection of technologies.

Amit Gefen
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Reuben H. Kraft, Rebecca A. Fielding, Kevin Lister, Allen Shirley, Tim Marler, Andrew C. Merkle, Andrzej J. Przekwas, X.G. Tan and Xianlian Zhou

Abstract In this chapter, a review of the current state-of-the-art in techniques, efforts and ideas in the area of modeling skeletal injuries in military scenarios is provided. The review includes detailed discussions of the head, neck, spine, upper and lower extremity body regions. Each section begins with a description of the injury taxonomy reported for military scenarios for a particular body region and then a review of the computational modeling follows. In addition, a brief classification of modeling methods, tools and codes typically employed is provided and the processes and strategies for validation of models are discussed. Finally, we conclude with a short list of recommendations and observations for future work in this area. In summary, much work has been completed, however, there remains much to do in this research area. With continued efforts, modeling and simulation will continue to provide insight and understanding into the progression and time course of skeletal injuries in military scenarios with a high degree of spatial and

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temporal resolution. However, more work is needed to improve mechanistic-based modeling of injury mechanisms, such as fracture, and increase the inclusion of bio-variability into simulation frameworks.

Keywords Computational biomechanics · Blast injuries · Skeletal injuries · Military injuries · Injury · Finite element modeling

1 Introduction

In this chapter, we provide a review of the current state-of-the-art in techniques, efforts and ideas in the area of modeling skeletal injuries in military scenarios. There are a number of reasons modeling and simulation is pursued, including: (1) the ability to understand the progression and time course of injury with high spatial and temporal resolution, (2) modeling provides a mechanistic-based platform to study injuries, (3) modeling can be a cost effective tool in the design and assessment of solider protection capabilities, and (4) modeling enables the study of bio-variability.

Modeling and simulation is many times employed to understand the time course of an injurious event. In many cases, modeling serves to supplement injury databases collected from the battlefield or test range. These databases[1] track and classify events and can be quite detailed. They also help to focus and motivate computational studies based on priorities identified from the data. Some of this data is used in the medical community, and some of the data is used by analysts of varying types (i.e., intelligence and military personnel, protection developers, scientists, engineers, etc.). In this process of information dissemination, modeling and simulation is sometimes used to understand the cause and effect relationships. For this reason, there is interest to improve the resolution, validity and predictive capabilities of modeling of skeletal injuries in military scenarios.

Skeletal injuries can occur in both non-battlefield and battlefield scenarios. There are numerous studies that examine the occurrence of musculoskeletal injuries during physical training and other non-battlefield scenarios [4–6]. Among non-deployed, active duty service members in 2006, there were 743,547 musculoskeletal injuries, or 628 injuries per 1000 person-years (i.e. the number of years multiplied by the number of people affected) [7]. Musculoskeletal injuries include overuse, joint derangement, stress fracture, sprain/strain/rupture, and dislocation. These injuries were distributed with 40.3 % occurring in the spine and back, 14.1 % in the upper extremity, 39 % in the lower extremity, and 6.5 % were unclassified by site. Combining these musculoskeletal injuries with acute traumatic injuries gives a total of 1.6 million injuries per year [7].

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[1]For example, the United States Joint Theater Trauma Registry [1], United Kingdom Joint Theatre Trauma Registry [2] and the United States Army's Total Army Injury and Health Outcomes Database [3].

In battlefield scenarios, explosions and blasts are a significant source of skeletal injuries [8]. In response the injuries sustained in the most recent conflicts, the United States established a Blast Injury Research Coordinating Office, which is a useful source of information concerning blast-related injuries [9]. In studying battlefield injuries, it is useful to categorize the scenarios as "mounted" or "dismounted". The mounted case pertains to situations where the solider is confined within a vehicle or some other protective structure. The dismounted case pertains to situations when the soldier is fully exposed. Blast and fragment injuries of the musculoskeletal system are the most frequently encountered wounds in modern warfare [10]. Injuries due to blast events can be classified as primary, secondary, or tertiary blast injuries [8]. Primary blast injuries describe the effect of the shockwave and the "blast wind" [11]. Skeletal injuries from primary blast result from changes in atmospheric pressure and may present as lacerations, fractures, and limb avulsions [10]. Although injuries to the musculoskeletal system can occur by any of the blast mechanisms, secondary blast injuries are predominate [11]. Secondary blast injuries describe penetrative trauma by fragments of materials accelerated by the blast [8]. These penetrative impacts may fracture bones and cause other complications [8]. Tertiary blast injuries result from bodily displacement or impact as a result of the explosion [8].

The detonation of an improvised explosive device beneath a vehicle, also known as an underbody blast, has been a significant problem for military personnel. Forces during such an event are estimated to reach up to 2000 Gs with a rise time of 1.5 ms [12] (note that a typical healthy reaction time for a human is approximately in the range of 200–300 ms [13], thus, there is no time to react to this sort of dynamic event; muscles are assumed to be passive). Such an event results in so-called "local effects", such as deformation of the vehicle floorplate, and "global effects" which pertain to the global motion of the vehicle and its occupants [14]. While under vehicle blast may create an initial blast exposure, the occupant loading is influenced by a variety of conditions including: seat mounting variations (e.g., stroking/energy mitigating, floor, wall, ceiling, structural variations, energy mitigating flooring, elevated foot rests/foot-pan, stirrups, etc.), occupant operational position (e.g., drivers and crew, seating facing anterior or posterior, seating facing laterally towards vehicle center, standing gunner), and variations in hip, knee and ankle angles (e.g., operational preload, location of blast relative to occupant, creates numerous loading vectors).

The organization of this chapter is divided into sections for each of the major body regions. Each section begins with a description of the injury taxonomy for that particular body region and then a review of the computational modeling follows. In some body regions more work has been done than in others. We begin with Sect. 2, which encompasses head, neck and spine injury in military scenarios. As the focus is on skeletal injury, brain trauma is not discussed. Section 3 discusses upper extremity injuries. Section 4 reviews modeling musculoskeletal injuries to the lower extremities. Section 5 provides an overview of a few computational models currently under development specifically for assessing skeletal injuries sustained in military scenarios. Section 6 provides a brief classification of modeling methods, tools and codes typically employed. Section 7 describes processes and strategies for validation of models. Finally, we conclude with a short list of recommendations for future work in this area.

2 Head, Neck and Spine Body Region

2.1 Injury Taxonomy

2.1.1 Head and Neck Injuries

The head and neck represent about 30 % of injuries during Operation Iraqi Freedom and Operation Enduring Freedom [15]. Of these head and neck injuries, 27 % are fractures [15]. The majority of head and neck injuries, 84 %, are caused by explosions, followed by 8 % caused by gunshot wounds, 2 % caused by motor vehicle accidents, 1 % caused by a helicopter or plane crash, 1 % caused by fragment penetration, and 3 % due to miscellaneous or undocumented causes [15]. Schoenfeld et al. [16] reviewed 7877 wounded as recorded in the U.S. Department of Defense Trauma Registry, 11.1 % (872) had suffered spinal injuries. Of these injuries, 74.5 % were due to blast and 14.8 % due to gunshots. Fracture was the most common injury mechanism in the spine, representing 83 % of spinal injuries and 91 % of fatalities [16].

The incidence of craniofacial injuries in the US military personnel has increased in recent military conflicts [17]. In the United States Joint Theater Trauma Registry [1], between 2001 and 2011, 618 mandibular fractures were identified in 391 patients, not including non-battle or fatal injuries. Among these mandibular fractures, 61.3 % were due to blast, and 12.5 % were due to ballistic injury. Compared to civilian mandibular injury, military injuries were more likely to be open, comminuted, and segmental [17]. The Joint Theatre Trauma Registry of British service personnel also identified mandibular fracture as the most common military maxillofacial injury, representing 33–40 % of maxillofacial fractures [18]. There were 74 mandibular fractures were reported among 60 patients with mandibular injury, 38 of whom died from wounds. It was noted that blast wave injuries generally caused localized fractures, whereas injuries due to blast fragmentation presented with comminution in multiple locations [18]. The high rate of mandibular fracture is likely due to its poor coverage by personal protection equipment [17–19].

2.2 Modeling and Simulation Review

2.2.1 Skull Fractures

Skull fracture is a major concern in military scenarios. There have been numerous modeling studies conducted on the ballistic impacts to the combat helmet [20–22], with some focus on assessing the stress or strain levels in the skull transferred from the helmet [20, 23]. Modeling techniques to understand the dynamic response of combat helmets are fairly advanced [22, 24–30] and have been validated in some cases. However, there are limited studies which examine the fracture mechanisms of the skull in detail. It should be noted, there is existing work regarding skull fracture modeling for non-military scenarios, not reviewed here. For more information see, for example, the work of Loyd et al. [31].

With regard to modeling in military scenarios, Aare and Kleiven [20] examined how different helmet shell stiffness and different ballistic impact angles affects the load levels in the human head during an impact using detailed finite element models of the human head and ballistic helmet. The focus of the study was aimed on getting a realistic response of the coupling between the helmet and the head and not on modeling the helmet in detail. It was concluded that dynamic helmet shell deflections (also known as back face deformation or rear effect [21]) larger than the initial distance between the shell and the skull should be avoided in order to protect the head from the most injurious threat levels. Furthermore, Aare and Kleiven [20] report that it is more likely that a fracture of the skull bone occurs if the inside of the helmet shell strikes the skull. Pintar et al. [23] also examined helmet-to-head contact forces.

Back face deformation was further studied using the Head Injury Criterion when the ballistic helmeted headform is impacted by a bullet with different impact angles and at various impact positions using LS-DYNA[2] [21], a commercial finite element package. Note that Sect. 6 includes a discussion of common software employed in modeling. The Head Injury Criterion deals with intracranial accelerations of the brain, but does not address skull fracture explicitly. A high fidelity headform model including detailed skull and brain has been used for the simulation purpose. Helmet and bullet are modeled according to the real shapes, but the head was represented using a headform model. The study concluded that based on the Head Injury Criterion scores obtained from the impact simulations at various impact positions, the bullet from back is the most dangerous position to the wearer [21]. Tan et al. [22] further examined back face deformation using experiments and simulations of the advanced combat helmet with different interior cushioning systems in ballistic impact.

As mentioned earlier, this review does not include military-related brain injury, however, it should be mentioned that there was a study involving numerical hydrodynamic simulations, which discovered that nonlethal blasts can induce

[2]Official name, not an acronym [32].

sufficient skull flexure to generate potentially damaging loads in the brain, even without a head impact [33]. The flexural ripples in the skull resultant from the blast cause regions of tension and large pressure gradients in the brain.

2.2.2 Craniofacial Fractures

As discussed craniofacial fractures are observed in military scenarios. Some modeling and simulation has been initiated to help understand the processes and mechanisms associated with this injury. One finite element mandible model was developed to study gunshot wounds to the human mandible [34]. The software LS-DYNA was used to simulate bullet and steel ball impacts to the mandible at various speeds and entry angles. A piecewise linear plastic material model was used for the cortical and trabecular bone of the mandible, and a failure strain was defined. This model allowed the researchers to assess injury severity in the mandible [34].

Animal models are frequently used to understand injury thresholds and mechanisms, thus, it is not surprising that there exist computational models of animal injury as well. In fact, one of the recommendations for future research is to continue to seek appropriate animal models for understanding skeletal injures in military scenarios. Lei et al. [35] developed a finite element model of a pig mandible based on a computed tomography scan of an undamaged porcine mandible. To generate experimental data, fifteen pig mandibles were exposed to a detonation. A hexahedral mesh was generated of the mandible geometry and LS-DYNA was used to simulate the detonation. The mandible was modeled using a piecewise linear plastic material with a failure strain criterion. The simulated results showed similar data and injury to the experimental results [35]. An attempt was made to use finite element software to model the human mandible during an explosion of 600 mg of trinitrotoluene (TNT) at 3, 5, and 10 cm distance from the mandible [19]. These simulations were completed using LS-DYNA. The mandible was modeled based on a computed tomography scan of a human female mandible. The mandible was divided into trabecular and cortical bone, with each modeled as a piecewise linear plastic material. Fracture was represented by implementing a failure strain for bone. This model allowed for the biomechanical estimation of stress, strain, and fracture patterns in the mandible due to a blast event [19].

2.2.3 Spine Injuries

Finite element modeling continues to be popular in spinal research. In general, research groups started with relatively simple models of individual components of the spine and eventually developed models that represented larger sections of the spine and surrounding anatomy. Latham [36] is considered the first to have developed a mathematical model of the spine; focused on studying pilot ejection, he used a one-degree-of-freedom model consisting of two masses, representing the body and ejection seat, interconnected by a spring to represent the response of the spine to vertical acceleration [36–39]. In Dimnet [37]'s subsequent model, the head and upper torso were modeled as a rigid body and the spine was represented by a spring-dashpot in a parallel system. Liu [40] contributed to the modeling approach by proposing a mathematical model that consisted of a uniform rod and a rigid end-mass, representing the spinal column and head, respectively.

It wasn't until the late 1960s that segmentation of vertebrae and intervertebral discs were incorporated into models. Toth [41] developed an eight-degree-of-freedom model in which vertebrae were modeled as rigid masses interconnected by springs and dampers, simulating the action of discs. A mathematical model developed by Orne and Liu [42] took into account the curvature of the spine as well as inertial properties of the vertebral bodies and axial shear and bending stiffness were introduced. The interaction between the spine and the torso, however, were not considered. Prasad and King [43] further developed this model to include the interactions of the articular facets.

Up until this point, models did not take the rib cage or viscera into account. Belytschko et al. [44] developed the first three-dimensional representation of the entire spine and several other models with increasing complexity. These finite element models consisted of the head, vertebrae (lumbar, thoracic, and cervical), ribs from T1 to T10, intervertebral discs, articular facets, ligaments, muscle, connective tissue, upper and lower viscera, and the pelvis. Skeletal components were modeled as rigid masses, intervertebral discs as beam elements, ligaments as spring elements, and articular facet interactions as hydrodynamic elements [37, 39, 45].

When constructing a finite element model of the spine, three primary areas need to be addressed: geometric representation including segmentation of spinal components; constitutive models and corresponding material properties of the components; and boundary conditions used to apply loading and constraints [46]. More recently, model components are developed with geometries that match that of a specific in vitro specimen or that represent an average vertebra. It is important to define the geometry of the structure as accurately as possible because it greatly influences deformation [47–49]. Generic models are developed from anatomical measurements while specific geometries can be formed from computed tomography and magnetic resonance imaging scans. Software packages convert data obtained from these images into a format that can be interpreted by solid modeling software and then input into finite element software. Wilcox et al. [47] found that finite element results demonstrated a greater agreement with previous experimental results when the models were morphologically accurate as opposed to generic models. It should be noted, however, that the accuracy of image-based models is affected by the original resolution of the medical image [50]. To date, a wide range of material values have been used in finite element models of the spine due to the large assortment of measurement techniques, testing strain rates, and sample preparation (see Table 1). The verification of mesh and boundary conditions and the sensitivity of results to material properties, however, has been limited [47].

The incidence of spinal injury among military personnel has encouraged an effort to develop a statistical shape model based finite element of the cervical spine. Five cadaver spine specimens representative of a 50th percentile male soldier were

Table 1 Compilation of various constitutive models and parameters found in recent computational models of the spine

Anatomic component		Material model material properties		References
Vertebra	Trabecular bone	Plastic with kinematic yield, poroelastic	E = 0.1–2.2 GPa ν = 0.2 σ_Y = 42–110 MPa	[59–65]
	Cortical bone	Plastic with kinematic yield, elastic	E = 5–22 GPa ν = 0.3 σ_Y = 132 MPa	[59, 61–64]
	Posterior elements	Elastic	E = 1000–3500 MPa ν = 0.25	[61–64, 66]
	Cartilaginous endplates	Poroelastic	E = 20–25 MPa ν = 0.1–0.4	[59, 62, 63]
	Facet cartilage	Elastic	E = 11 MPa ν = 0.4	[62]
Intervertebral disc	Nucleus pulposus	Poroelastic, hyperelastic	K = 1.667 GPa E = 0.15–4 MPa ν = 0.4999	[59, 61–64, 66, 67]
	Annulus fibrosis	Orthotropic elastic, hyperelastic, poroelastic	E = 2–450 MPa ν = 0.3–0.45	[60, 62–64, 67, 68]
Ligaments	Anterior longitudinal ligament	Hyperelastic	σ_Y = 22.5 MPa ε_Y = 0.2 E = 7.8-20 MPa ν = 0.3	[63, 69, 70]
	Posterior longitudinal ligament	Piecewise linear, hyperelastic	σ_Y = 16–20 MPa ε_Y = 0.3–0.45 E = 10–20 MPa ν = 0.3	[59, 61, 63, 69, 70]
	Supraspinous ligament	Hyperelastic, nonlinear elastic	E = 3.4–15 MPa ν = 0.3	[63, 64, 70]
Spinal cord		Elastic	E = 0.26–1.3 MPa ν = 0.35–0.49	[59, 71]
Dura mater		Anisotropic elastic	$E_{rr} = E_{\theta\theta}$ = 142 MPa E_{zz} = 0.7 MPa	[59]

Not all parameters are listed for each constitutive model. Furthermore, not all spinal components are listed. E is the Young's Modulus, ν is Poisson's ratio, σ_Y is the yield strength, ε_Y is the yield strain, and K is the bulk modulus. The symbols rr, $\theta\theta$, and zz denote directions in the polar coordinate system

scanned using a computed tomography scanner. Surfaces for each vertebra were developed using MATLAB. Vertebrae were positioned by visual observation to represent the cervical spine and tetrahedral volumetric meshes were developed. A statistical shape model was developed to describe variability among the spine samples. Dynamic loading was simulated using LS-DYNA and validated against dynamic loading data [51]. The Naval Air Systems Command (NAVAIR) spinal model [52] is in development to model spinal injury risk during military flight maneuvers and during repeated shock, ejection, and crash conditions. Geometry was obtained using computed tomography scans and material properties were experimentally determined. Model analysis was conducted using LS-DYNA [52]. A review of computational spinal models [46] suggests that for the spine, trabecular bone may be represented as an elastic material or plastic with kinematic yield, and that trabecular bone may be represented as a poroelastic material or plastic with kinematic yield. Posterior elements and facet cartilage can be represented as elastic, and cartilaginous endplates as poroelastic. The nucleus pulposus and annulus fibrosis of the intervertebral disk may be represented as hyperelastic or poroelastic, and the annulus fibrosis can additionally be represented as orthotropic elastic [46]. Recently, Spurrier et al. [53] conducted an extensive literature survey focused on blast injury in the spine they found that the dynamic response index is not an appropriate model for predicting injury. Recently, Zhou et al. [54] developed a new musculoskeletal fatigue model for the prediction of neck musculoskeletal loading and fatigue of fighter pilots during high-G aerial combat maneuvering. The model included active muscles to investigate fatigue during two common flight postures. While on the topic of aerial-related injuries, the recent work of Aggromito et al. [55, 56] focused on elucidating the effects of body-borne equipment on occupant forces during a simulated helicopter crash, should also be mentioned. This work is somewhat related to reduced-order models for assessing injuries [57].

In either battlefield or non-battlefield scenarios, a number of activities conducted by soldiers subject the spine to various loading conditions that may lead to spinal disc degeneration from prolonged extreme loading. Recently, Makwana et al. [58] have initiated computational research focused on understanding of the micromechanics and mechanisms of intervertebral disc degeneration. The focus of the research is to develop a computational model of the intervertebral disc degeneration that attempts to capture the initiation and the progression of the damage mechanism under fatigue. The goal of the model is to capture the effects on disc under short or long period biomechanical loading in combat environments [58].

3 Upper Extremity Body Region

3.1 Injury Taxonomy

In a survey of battle wounded soldiers from Operation Iraqi Freedom 1281 military personnel suffered extremity injuries, 75 % of which were a result of explosions [72]. Of these injuries 53 % were soft tissue injuries and 26 % were fractures. About 50 % of extremity injuries were sustained in the upper extremity [72]. The most commonly fractured site in the upper extremity was the hand, where 36 % of upper extremity fractures were sustained [72]. In another analysis of military injuries, 14 % of all musculoskeletal injuries were sustained in the upper extremity [73]. Shoulder injuries were most common among upper extremity injuries, comprising

63 % of those injuries and 9 % of all musculoskeletal injuries [73]. Upper extremity injuries may result in amputations, 3.2 % of which include late amputations (less than 90 days after the injury) due to complications [74]. Upper extremity injuries due to blast events are most frequently classified as secondary blast injuries (i.e. penetrative trauma), which can cause bone fractures and other complications [8].

3.2 Modeling and Simulation Review

Computational modeling of upper extremity injuries with a specific military application is presently limited. Work has been completed in computational modeling of the upper extremity for automotive applications, especially for airbag impact [75, 76]. In the absence of military targeted computational models, automotive models may provide a baseline of modeling insight. The first full finite element model of the human upper extremity, including the humerus, radius, ulna, and all bones of the hand, was developed for the prediction of side airbag interaction [75]. The anthropometry, mechanical properties, and impact response for this model were generated based on cadaveric studies [75]. A later automotive upper extremity model was developed for MADYMO, a vehicle occupant safety modeling software. Its geometry was based on a 50th percentile cadaveric sample, and it was validated against PMHS testing, including static and dynamic three point bending tests and airbag impacts [76]. The material models used in that study were chosen based on literature, suggesting that a linear elastic model may be used for cortical bone, with yielding modeled by elastic-plastic behavior. The upper extremity model showed fracture when the plastic strain exceeded a set value [76]. A similar study for automotive side impact to the upper extremity also used cadaveric geometry to develop a finite element upper extremity for use with Radioss [77]. This model was validated against postmortem human subjects bending and impactor tests. Material models used include an elastic-plastic model for bone, an elastic model for skin, and a viscoelastic model for soft tissue and ligaments [77]. One of the most fully developed full body human models in development is the Global Human Body Models Consortium (GHBMC), which was sponsored by seven automobile manufacturers. As a side note, many of the automotive human models are not permitted for use to study military events, based on requests from consortium members. The Global Human Body Models Consortium includes a mid-size male model [78]. Extensive efforts have been devoted for validating these models for automotive applications, but less work has focused on validation for military loading scenarios. Geometry for this mid-sized male model is based on magnetic resonance imaging and computed tomography scans of a volunteer [79]. The Consortium models are intended for use with LS-DYNA, PamCrash [80] and Radioss [81]. The upper extremity portion of the 50-th percentile model is classified as having sufficient detail, but lacking sufficient validation work and test data [82].

The techniques for obtaining model geometry, like magnetic resonance imaging and computed tomography scans [79] or cadaveric samples [75–77] can be employed for military applications, or existing geometry for these models may be employed. Like for automotive studies, experimental data against which models can be validated can be generated via PMHS bending and impact testing [76, 77]. Similar material models as have been presented, such as elastic plastic for bone [76, 77] and viscoelastic for soft tissue [77] may be used. Fracture can be modeled using a plastic strain failure criterion [76]. The same commercial software, like Radioss and LS-DYNA [77, 82] may be sufficient for military use as well. However, some important differences between automotive and military applications should be kept in mind. Loading conditions are likely to vary between automotive and military applications. Military injuries are likely to occur at higher strain rates than civilian injuries. Furthermore, while much of automotive computational work for the upper extremity studies impact, upper extremity injuries due to blast events are most commonly classified secondary blast injuries, meaning they are due to penetrative trauma. Secondary blast injuries describe penetrative trauma by fragments of materials accelerated by the blast. These penetrative impacts may fracture bones and cause other complications [8, 83]. Mechanical behavior is very dependent on material parameters, and high strain rate behavior, which would be present in military loading scenarios, is not adequately understood [83]. The behavior of both bone and soft tissue varies with higher loading rates [83], but this is not taken into account in existing automotive impact studies [75–77].

4 Lower Extremity Body Region

4.1 Injury Taxonomy

Here, we can classify the detonation of an explosive device beneath a vehicle as an underbody blast. Injuries to the lower extremity during underbody blast events are predominantly tertiary injuries, and with fractures characteristic of severe axial loading [8, 83]. In recent military conflicts, the incidence of UBBs has led to severe types of injuries, specifically in the lower extremities. The lower extremity accounted for 26 % of all combat injuries during Operation Iraqi Freedom, 34 % of which were inflicted by explosive devices [84]. A review of wounded in action who were exposed to underbody blast showed that, of 500, almost 200 (40 %) had foot and ankle injures. The injury review cited 57 instances of talus fracture, 56 instances of tibia fracture, and at least 100 calcaneus fractures. Malleolar fractures were also noted [85].

4.2 Modeling and Simulation Review

A computational approach to studying military blast injury is a cost-effective alternative to experimental studies, and allows for the modification of input parameters in order to represent a variety of scenarios [83]. Explicit dynamic finite element codes are employed for modeling lower limb impact. LS-DYNA, a commercial finite element code, is popularly used for biomechanical modeling [83, 86–90], though modeling is also complted with ANSYS [91], Pam-Crash [80], and Radioss [83]. Human model geometry is typically based on medical imaging data, including MRI and CT scans [78, 86, 87, 92], though computational models have also been developed using anthropomorphic testing device geometry [91, 93]. In computational models, skeletal tissue is approximated with material models including linearly elastic, linearly plastic, isotropic, and transversely isotropic materials [83]. Elastic-plastic definitions of skeletal tissue, such as the "plastic-kinematic" material available in LS-DYNA [86, 90, 94], are especially common [86, 87, 89–91]. In LS-DYNA, this material is defined by a density, Young's modulus, Poisson's ratio, yield stress, and tangent modulus [94]. Soft tissue formulations for human body models include both hyperlelastic material models, such as a Mooney-Rivlin formulation [95–99], and viscoelastic material models [88, 89]. LS-DYNA offers material models targeted to soft tissue usage with options for both hyperelastic and viscoelastic behavior [94]. As computational model development continues, it is important to recognize that mechanical behavior is very dependent on material parameters, but existing models show significant variability in material parameters, and high strain rate behavior is not adequately understood [83]. While it is generally seen that bone is brittle at strain rates above values of about 20/s, and that bone modulus and strength increase with strain rate in compression, exact quantitative data varies among experimental studies, and reports of observed behavior in tension are inconsistent [83]. Soft tissue behavior, too, varies with strain rate. Stiffness may be insensitive to increasing strain rates, but strength increases at higher rates. Expanded material testing at high rates of loading would contribute to a better understanding of human tissue material behavior and improve the accuracy of computational models [83]. Recently, Guo et al. [100] used computations to assess injury to occupants subjected to a mine blast within a military vehicle.

As with experimental testing, much of the work done in modeling of the lower extremity was completed with applications to automotive impacts. However, this work provides a baseline for progress in modeling military injury to the lower extremity. A computational study of automotive impact loading to the lower limb also suggested that incorporation of strain rate dependent material behavior could allow models developed for automotive applications to be extended to blast loading scenarios [86]. Perhaps the most rigorously developed computational human is the model GHBMC, which is fully validated for a mid-size male mode [76]. One study investigated the potential of modifying a Consortium model of lower limb for use in blast impact modeling. It was found that some modifications to the material models, such as the use of a piecewise linear plastic material model rather than a plastic kinematic model, improved the biofidelity of the limb response under loading [89]. There have also been studies of models developed specifically to study blast injuries. One validated model used ANSYS to simulate vertical impulse to the lower extremity in two seated configurations [91]. Other modeling efforts [90, 101] used LS-DYNA to simulate impact to the below knee portion of the lower extremity at 5, 10, and 12 m/s. Force data in the tibia was compared to experimental data at those loading rates. This study also modeled fracture in the extremity [90].

Just as anthropomorphic testing devices have been used for experimental blast studies, computational models of anthropomorphic test devices have also been used in simulations. One study, in an effort to determine the validity of using an anthropomorphic test device model to represent blast scenarios, used a numerical model of a HYBRID III[3] anthropomorphic test devices. The impact response and injury threshold of this model were compared to experimental data [93]. A similar computational study compared the response of a computational anthropomorphic test device model to the response of a validated computational lower extremity model [91]. An additional consideration that has been taken into account during computational studies is initial position of the lower extremity. Computational studies which have examined the effect of the lower limb pose have found that initial position does have a significant effect on impact response [91, 93].

It is also critical to examine the effect of protective footwear, which can reduce peak tibia forces and mitigate fracture in the calcaneus [83]. Limited computational work has been completed with protective footwear considered. A coupled foot and boot model was developed with goal of modeling various injury-inducing scenarios, but as yet this coupled model has only been studied for balance standing conditions [103]. The effect of a polyurethane boot sole of various levels of hardness was examined in a separate finite element study, showing a reduction in peak tibia force as well as reduced fracture in the calcaneus [90]. The fidelity over time of several existing models is illustrated in Fig. 1.

Computational approaches to fracture modeling include the extended finite element method, interelement cracks, and element deletion. The extended finite element method constructs a discontinuous function that is applied to only one element and disappears at the edges. The interelement crack approach models cracks along element edges. This can be done in two ways, with all elements separated from the beginning of the simulation, or with cracks being dynamically inserted during the simulation. The elements are joined by cohesive laws in this approach. The element deletion method simply removes elements from the calculations after these elements reach failure criteria, and the failed elements will no

[3]The Hybrid III 50th Percentile Male Crash Test Dummy is a widely used anthropomorphic test device for the evaluation of automotive safety restraint systems in frontal crash testing. Originally developed by General Motors, the Hybrid III 50th design is now maintained and developed by the Humanetics Innovative Solutions company, in conjunction with the Society of Automotive Engineers' Biomechanics Committees and the U.S. National Highway Transport and Safety Administration [102].

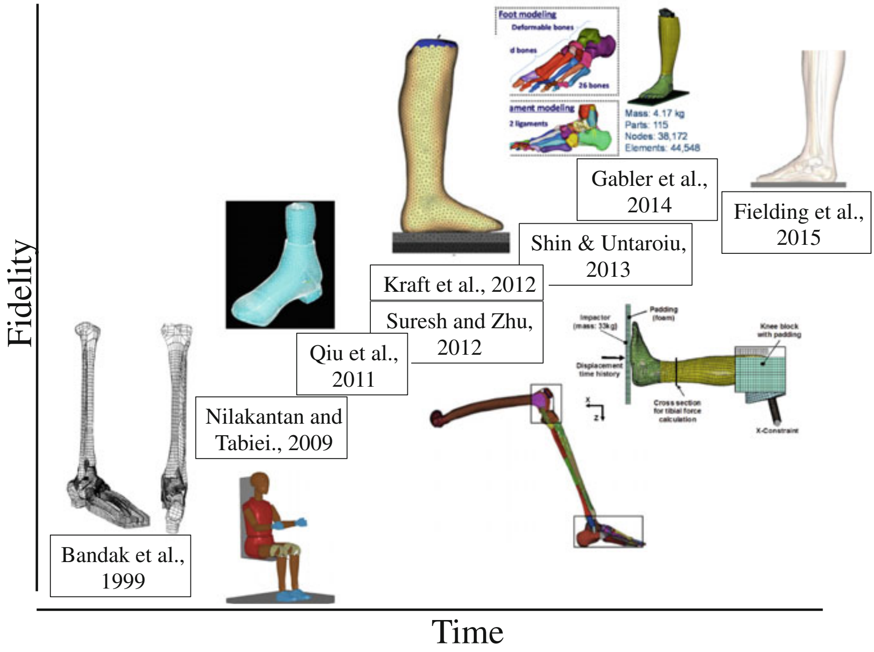

Fig. 1 Schematic showing how lower extremity models have advanced over over time

longer carry stress [104]. Though it is the least refined [104], due to its simplicity, the element deletion is often used [87, 105], using a failure criterion such as plastic strain [87].

5 System-Level Human Performance Prediction for Injury Prevention

Here, we review a few computational models currently under development specifically for assessing skeletal injuries sustained in military scenarios.

5.1 Corvid Technologies CAVEMAN™ Human Model

To support United States Marine Corps survivability efforts Corvid Technologies has developed the Computational Anthropomorphic Virtual Experiment Man (CAVEMAN™) high fidelity human system finite element model, as shown in Fig. 2. The focus is on better understanding injuries due to accelerative loading in

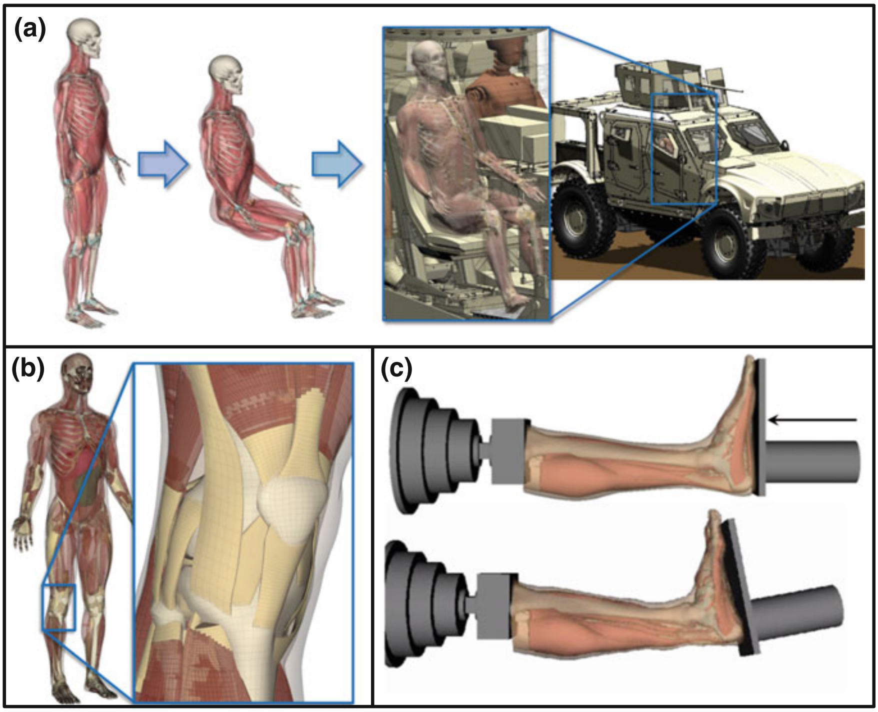

Fig. 2 Corvid technologies CAVEMAN™ human model. **a** Articulation of the CAVEMAN model for integration into military vehicles. **b** Depiction of details within the musculoskeletal system of the CAVEMAN knee. **c** Subsystem level comparison to accelerative loading experimental test data

armored vehicle improvised explosive device events. CAVEMAN is built on the Zygote human Computer-aided design model (Zygote Media Group, Inc.) of the musculoskeletal systems and internal organs including: 179 bones, 397 muscles, 342 ligaments, and 16 organs. There are approximately four million total elements incorporating this detail into the computational physics model with everything except selectively tetrahedral-meshed bones resolved with high quality hexahedral elements. Element length scales range from millimeters to tens of microns to resolve the geometric details of interest. The model is designed to leverage the high performance computing scaling and robust contact interface capability within Velodyne™ for full system runs and accurate kinematics.

The geometry was developed through a combination of magnetic resonance imaging, computed tomography and ultrasound medical scan data and has been compared to Department of Defense anthropometry metrics across the US Army anthropometry surveys (ANSUR [106], DOD-HDBK-743A [107]) and US Army Anthropometric Survey-ANSUR II. This study found that the average measurement error ranged between 4.2 and 4.8 % across all of the available metrics in each reference and at least 70 % of all measurements were within one standard deviation representing good correlation with the fiftieth percentile male warfighter. Many of the outliers were ear measurements for helmet fit, which are not the focus of CAVEMAN applications.

CAVEMAN is being used to analyze the effect of soft tissue response and connectivity on bone fracture due to accelerative loading experienced during military operations. The primary loading event that is being investigated is vehicle floor intrusion and seat acceleration resulting from improvised explosive device explosions, which can cause severe injuries to lower extremities and the thorax. The intent of this research is to develop CAVEMAN as a framework for investigating human response to impulsive loading to include bone fracture and extend beyond to soft tissue injury at a component or full system level.

5.2 CoMBIP Human Models

Supported by the U.S. Congressionally Directed Medical Research Program, Johns Hopkins University Applied Physics Laboratory led a team of researchers in developing biomedically valid human computational models for use in understanding and mitigating injury due to various military loading scenarios. The Computational Models for Blast Injury Prevention (CoMBIP) includes a suite of models developed for investigating acute loading initiated by blast events in the form of overpressure, acceleration, blunt impact, and force transference. Simulated threat scenarios include open-field and urban blast as well as vehicle-mounted underbody blast.

The Computational Models for Blast Injury Prevention program models were developed within a multi-level hierarchical framework to include tissue level characterization, component level injury mechanics, and system level model validation (Fig. 3). Tissue-level material failure thresholds determined from ex vivo postmortem human subject studies were used to develop tissue-level materials response. At the component level, post-mortem human subject testing was simulated computationally, and the mechanical response was correlated to simulation predictions. At the system level, the computational models were compared against the post-mortem human subject response. Ultimately, these models will capture mechanisms, locations, extent, and severity of predicted injuries against documented injuries from the testing phase.

Development of these models requires the inclusion of biomechanically relevant anatomical structures, rate-specific material property characterization of biological tissues, and accurate replication of loading conditions. Geometry for the Program's human model for vehicle loading scenarios was based on scaling the National Library of Medicine's Visible Human Project dataset to a targeted 50th percentile male. Each anatomical region was meshed and material parameters were taken from the biomechanical literature, when available, or customized experiments were performed. Each anatomical component was then evaluated in a simulated experiment for comparison to post mortem human subjects t results. These post-mortem

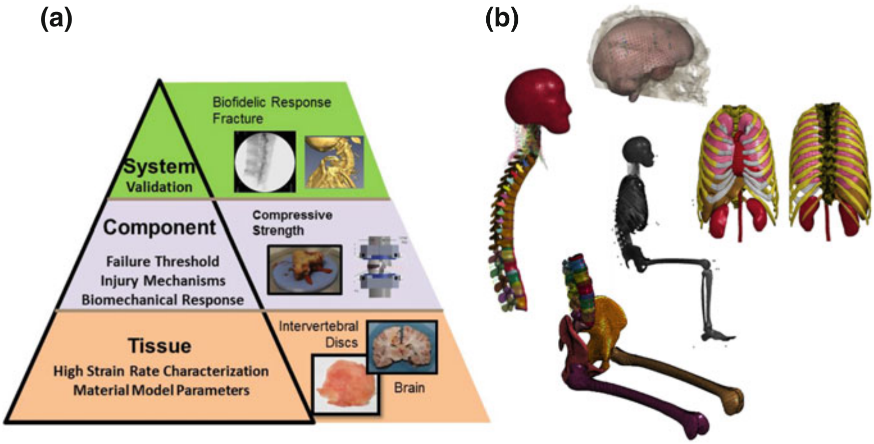

Fig. 3 **a** Multi-level hierarchical framework to include tissue level characterization, component level injury mechanics, and system level model validation. **b** Integrated system and components of the computational models for blast injury prevention (CoMBIP) model for underbody blast simulations

human subjects t experiments were designed at relevant loading rates and boundary conditions to provide the best possible source of validation. Once the components were complete, system level validation was performed.

The CoMBIP models are intended to simulate critical biomechanical responses to dynamic loading conditions. This is accomplished by developing the model within a hierarchical validation approach. As these models continue to be refined, they will be used for predicting human injury and both evaluating and designing injury risk mitigation strategies.

5.3 Santos™ Human Model

Injuries often depend not only on the failure mechanism for various musculoskeletal components or systems, but also on the orientation and environment for these systems. Injuries depend on operational position and performance. Thus, in order to predict the propensity for injury, it is necessary to predict human performance. In essence, models for musculoskeletal injuries must be integrated within a full human system, a digital human model (DHM). In addition, as one attempts to model more complex systems, depending on and growing a single model becomes less tractable. This is especially true of a system as complex as the human body. As the applications for digital human models extend beyond the ergonomics of reaching tasks, the details of gait, or the medical issues surrounding a single human organ or component, coupling multiple models becomes necessary for answering critical questions, especially in the case of injury prevention.

Digital human modeling provides a framework to assess the effect of task performance as well as various physical impairments, including weakness, fatigue, and limited range of motion. Consequently, higher fidelity and more accurate human models are becoming critical for meaningful simulations used for injury prevention, for product and process design, and for the study of human function. However, increasing the fidelity and accuracy of a model of something as complex as the human body requires a multi-disciplinary, multiscale effort.

Santos [108–111] (Fig. 4c) is a predictive digital human model that provides a platform for this kind of multi-scale model. Santos is a highly realistic, physiologically mechanistic, biomechanical computer-based human that predicts, among other things, static posture, dynamic motion, joint strength, and development of fatigue. Such capabilities can be used to predict and assess human function, providing task performance measures and analysis. Thus, in a virtual world, Santos can help design and analyze various products and processes. In addition, Santos can help study and evaluate various restrictions and impediments, such as fatigue, reduced range of motion, environmental obstacles, etc. Digital human models, like Santos, are used to help design body armor systems and to study the physiological effects of extended load carriage. He helps solve problems like reach analysis in an automobile, where Santos can indicate the optimal placement for seats and controls. And, he simulates and helps study the completion of tasks such as crawling or running when carrying heavy loads or wearing restrictive clothing.

At the crux of any virtual human is the ability to simulate human posture and motion realistically and quickly, while considering external and internal loads/forces. With respect to motion, there are traditionally two types of dynamics problems that need to be addressed. In the first problem, called forward dynamics, the external forces and torques on the system are known and the motion of the system is desired. The problem is solved by integrating the governing equations of motion forward in time using a numerical algorithm. In the second problem, called inverse dynamics, the motion of the system is known (i.e., from motion capture), and the forces and torques causing the motions are calculated using the equations of

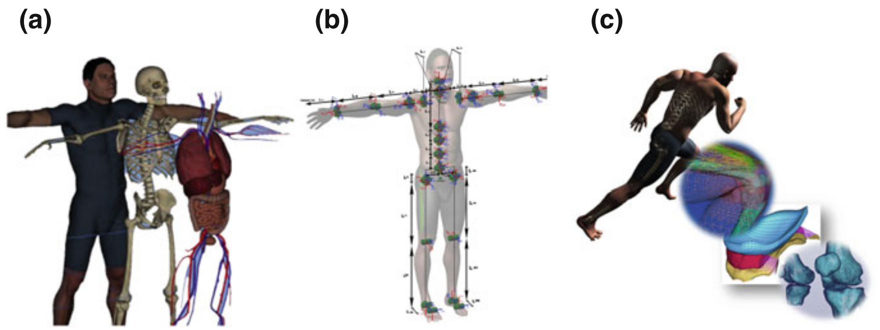

Fig. 4 **a** Santos human models which are composed of **b** a computational skeletal model, and **c** integrated multi-scale digital human model

motion. Both of these problems can be solved using traditional multi-body dynamics software. When simulating human motion for any task, both the joint torques and the motion of the joint are unknown. This problem is addressed with a novel optimization-based approach to motion prediction called predictive dynamics [112, 113]. With this approach, the joint angles—one for each degree-of-freedom (Fig. 4b) essentially provide design variables that are determined through optimization. The objective function(s) is one or more human performance measure, such as energy consumption, discomfort, and joint displacement. Finally, equations of motion are incorporated as constraints, thus incorporating laws of physics.

This optimization-based approach drives motion with an underlying mathematical model. It is physics-based, not data-based. Alternatively, traditional human models require some form of pre-recorded data—typically motion-capture data—that provides a seed to which the simulated motion gravitates. Although this approach captures the nuances of human motion well, it requires new data for new variations in problem parameters. While Santos can simply be re-run with an increased applied load, for example, and will automatically produce appropriate changes in the simulation, alternative approaches would require new motion-capture data. Consequently, Santos is afforded considerable autonomy in reacting to infinitely many scenarios. In fact, we contend that human motion is task based, so various performance measures or combinations of measures can be associated with different types of tasks, thus modeling different types of behavior. The use of this approach allows us to study how and why humans move in a particular way given particular objectives and demands. By developing and evaluating the effects of various performance measures, one can generate new hypotheses and not merely recapitulate data.

A fundamental distinguishing factor in the development of Santos is a joint-based model. The design variables are joint angles, not activation levels of individual muscles. This approach is more computationally tractable, often providing results in real-time. Furthermore, joint angles are relatively easy to validate. Finally, this approach allows for easy implementation of strength and fatigue models [114, 115]. Experimentally determined maximum joint-torques represent the strength limits for key joints (and thus major muscle groups). These limits can be compared to predicted joint torques as a post-processing technique, or implemented as constraints in the predicted motion [111]. Thus, one can alter strength characteristics, see the effects on task performance, and with the proposed development, see the effects on propensity for injury.

Although multiscale models are becoming more prevalent in the field of biology, linking the molecular scale to the whole-organism scale [116], multiscale modeling has not yet surfaced substantially in the digital human model field. To be sure, multiscale modeling has been used with some human components. For instance, researchers propose multiscale models for increasing the accuracy of bone models [117, 118] used finite element mesh of spine in connection with digital human models to simulate the effects of trauma resulting from vehicular crash testing. Fazekas et al. [119] proposed a multiscale modeling approach to assist users in the construction of a musculoskeletal system model to describe the mechanisms of the human upper limb (wrist, elbow, and shoulder). Nonetheless, little work has been completed within the context of a predictive, scalable, and open digital human model. A system level human model like Santos provides a platform for integrating a predictive joint-based system with high-fidelity joint models that simulate detailed injury mechanisms (Fig. 4c). The idea of integrating multiple models with various level of fidelity allows one to simulate the complete human more effectively and thus predict the propensity for injury more effectively.

5.4 CFDRC CoBi Models

CFD Research Corporation, in collaboration with various U.S. Department of Defense Labs, has developed a computational framework, CoBi,[4] for multiscale modeling of human body injury caused by various exposure scenarios including blast, ballistic and blunt impacts and accelerative loading [54, 120–123]. The framework integrates human body anatomical models and computational tools to simulate blast physics, human body biodynamics and injury. A database of human body anatomic geometry template models has been established based on Zygote human (Zygote Media Group, Inc.), Visible Human [124] and the extended cardiac-torso (XCAT) phantom (XCAT) [125] data. A subject specific (personalized) 3D anatomical geometry human body model can be created by morphing an anatomic template to a specific body surface scan such as in the U.S. Army Anthropometry Survey (ANSUR) [126] or Civilian American and European Surface Anthropometry Resource Project (CAESAR®) [127]. The CoBi framework provides graphical tools for interactive manipulation of the human body posture, position and interaction with the protective armor and equipment such as a combat helmet, boots and vest as well as vehicle or helicopter seat. Figure 5 presents CoBi graphical user interface framework for generation and manipulation of human body anatomical geometry models and for setup of biomechanics and injury simulations.

The CoBi computational framework enables both high fidelity and reduced order simulations of various loads scenarios, including computational fluid dynamics modeling of explosion blast waves and their interaction with the human body and finite element modeling of human body biodynamics and the interaction with structures such as a vehicle seat or a ground as well as finite element simulations of human body tissue/organ and skeletal biomechanics. The biomechanical loads to skeletal components can be used for detailed modeling of primary biomechanical injury and for mechanobiology of secondary injury mechanisms. The framework has been validated on range of experimental data of blast brain injury in animal models and on human physical surrogates and is currently adapted for fast running simulations of skeletal impact and injury biomechanics [128]. Figure 5 presents examples of blast and impact loads on a human body and selected simulation results

[4]Official name, not an acronym.

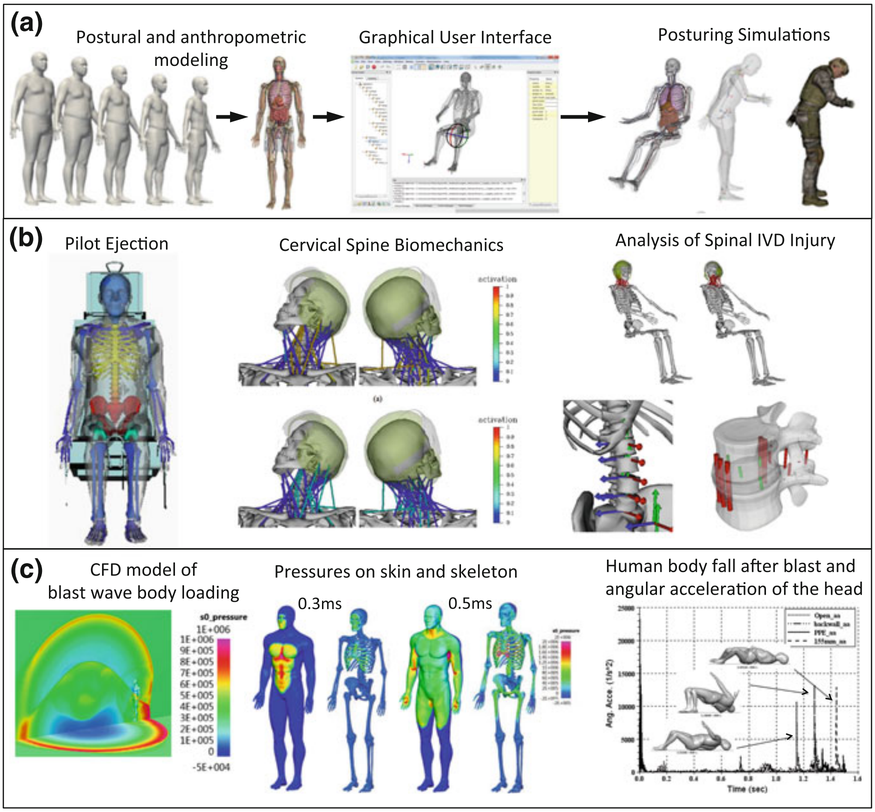

Fig. 5 CoBi framework for modeling human body injury. Generation and manipulation of human body anatomical geometry and example simulation results of human body injury biomechanics during blast and accelerative loadings, **a** shows anthropometric, posturing, and rigid body kinematics modeling, **b** displays various simulations examining loads experienced in the spine and **c** shows methods focused on understanding the stresses experienced by the skeletal system from blasts or explosions

of injury biomechanics. As discussed previously, this framework has also been applied to modeling of musculoskeletal fatigue of aviator neck maneuvering loadings [54].

6 Methods

Table 2 lists a number of common commercial, open-source or government-controlled softwares employed to model military injuries. This is not a full list but some examples. Many approaches rely on finite element or finite difference methods. These can be either Eulerian or Lagrangian based methods, however there are cases

Table 2 Overview of some computational tools used to model military injuries

Finite element software	Description	Representative studies
LS-DYNA	General purpose multiphysics simulation software package developed by the Livermore Software Technology Corporation	[19, 34, 35, 51, 52, 82, 86–89]
ANSYS	Structure analysis finite element software with non-linear dynamic capability	[91]
FEBiO	A nonlinear finite element solver that is specifically designed for biomechanical applications	[132]
Abaqus	A general purpose finite element solver that employs explicit integration scheme to solve nonlinear systems with many contacts under transient loads	[133]
Santos	Digital human modeling and simulation which includes a kinematic and susculoskeletal model	[134]
Velodyne	Non-linear, transient dynamics finite element analysis tool with both explicit and implicit multi-physics solver capability. Focus of development has been robust contact, multi-physics coupling, and optimization for massive parallel computational efficiency	[135–137]
IA-FEM	An open-source, interactive, multiblock approach to anatomic finite element model development	[138]
Uintah	Designed for the simulation of multi-scale multi-physics problems, such as those arising from combustion and fluid-structure interaction applications	[139, 140]
SIERRA	An integrated code suite which includes coupled simulation capabilities for thermal, fluid, aerodynamics, solid mechanics and structural dynamics developed at Sandia National Laboratories	[105, 141, 142]
CTH	A multi-material, Eulerian, large deformation, strong shock wave, solid mechanics code developed at Sandia National Laboratories	[143–145]
SimBios	Open-source software to create and analyze dynamic simulations of movement	[146]
ALE3D	A 2D and 3D multi-physics numerical simulation software tool using arbitrary Lagrangian-Eulerian (ALE) techniques	[147]
CoBi	A DoD open-source software created by CFDRC that integrates human body anatomical models and computational methods to simulate blast physics, human body biodynamics and injury	[54, 120–122]
MADYMO	Combines multibody, finite element and computational fluid dynamics technology integrated in one single solver	[148]
Altair® RADIOSS®	Structural analysis solver for highly non-linear problems under dynamic loadings	[81]
PAM-CRASH	Structural analysis solver for highly non-linear problems under dynamic loadings	[80]

where other methods such as rigid body models, particle-based methods and others, have been employed. As noted earlier, that in order to create a computational model, one assumes the existence of an appropriate constitutive or material model that relates the strain and stress. Since some scenarios include high rates of loading to the body (e.g. blast or ballistic impact), computational modelers typically think of the problem of assigning material parameters in terms of three broad areas: the strain rate response, the pressure response and the deviatoric response of the tissue. The development of reliable experimental methods and trustworthy reported material properties at high rates of loading (say at ≥200 s^{-1}) remain a critical gap as input into many computational models. There has been innovative and valuable research in the area of high rate biological material testing [129–131], but more research is needed.

7 Validation Data and Physical Models

Models must establish credibility before general users can use the simulations to extrapolate information and make decisions from information and results produced by the model. Specifically, an analyst must (1) assess the numerical accuracy of the underlying mathematical equations in the computation model, (2) evaluate the sensitivity of the model results to the input parameters, and (3) validate that the model results correspond to results of a real-world scenario [63, 149]. Models are often validated through one of two methods: direct or indirect validation. Direct validation involves a comparison between a model's prediction and an experimental test result. Indirect validation compares model results to a physical case where it is not possible to determine if conditions are the same. This includes comparisons between data from the literature, clinical trial results, and historical medical data [63]. Through adequate considerations of model sensitivity and validation, researchers have been able to develop more accurate and efficient simulations. Generating a precise model that incorporates verification, analysis sensitivity, and validation is a difficult task, but by doing so, the gaps between clinical medicine, experimental biology, and computational biomechanics will decrease.

Consider the case of the lower extremities. The framework for validating computational lower extremity models is typically derived from post-mortem human subject studies. Loading during impact testing to the lower extremity is typically applied by a vertical drop test [150], piston and pendulum set up [151, 152], accelerated sleds, and pneumatic or hydraulic rigs [83]. Much of the test data which exists, however, is targeted at simulating automotive impact trauma, during which the applied load is between 2 and 6 m/s in a timeframe of up to 100 ms. During underbody blast events, impact velocities can reach between 7 and 30 m/s in 6–10 ms [83]. One post-mortem human subject impact study completed impact tests at the lower end of that range, studying impacts at 7, 10, and 14 m/s over 14 ms [152]. Experimental impact tests may be limited by the experimental set-up, age of cadaveric samples, and the lack of protective footwear [83]. There have been attempts made to appropriate anthropomorphic testing devices (ATDs) developed for automotive applications to underbody blast surrogate testing. However, both the Hybrid III [102] and the Test Device for Human Occupant Restraint (THOR) [153] anthropomorphic testing devices lack biofidelity at the high impact rates resultant from underbody blasts [83]. A single frangible single use Complex Lower Leg (CLL) surrogate may offer more accuracy at these rates [154].

8 Conclusion and Future Work

In summary, a review of the current state-of-the-art in techniques, efforts and ideas in the area of modeling skeletal injuries in military scenarios has been provided. While much work has been completed there remains much to do in this research area. With continued efforts, modeling and simulation will continue to provide insight and understanding into the progression and time course of injury with high spatial and temporal resolution. In many body regions more work is needed to improve the mechanistic-based modeling of injury and inclusion of bio-variability into simulation frameworks. The following is a short list of recommendations and observations.

- **Verification and Validation**: Computational models should pursue hierarchical validation strategies spanning the component to system level. Using such an framework accommodates an "open-architecture" approach, which the government (military) may want to consider adopting. We envision a platform in which experts working in particular body regions could "insert" their models. It is not clear, if an appropriate approach would be to adopt a single "open-source" model, which the government owns and operates, or if there should be multiple open-source competing models. The latter may generate more innovate solutions, yet come at an increased cost.
- **Material Models and Parameters**: Constitutive models and their parameters are also a critical aspect of developing an accurate model. Determining the properties of hard and soft tissues, and fluids, as well as discovering their complex interactions over time are a new set of challenges currently facing researchers. Furthermore, combined computational and experimental techniques to measure material properties of biological hard and soft tissues at high rates of loading are needed.
- **Scaling Methods**: Transformation or scaling methods which describe injury thresholds from animal to human models are needed. In order to understand the injury mechanisms in a comprehensive manner, it may be beneficial to establish an experimental and computational animal model for each region of interest and injury type. For each region and injury of interest, different size animal models, ranging from small to large may be developed, starting with small animal models. Then, as modeling progresses to larger animals, scaling laws could be verified. The role of animal models is critical.

- **Inclusion of Human Biovariability**: Biovariability including distributions of anatomy, material properties, age, gender, etc., should be incorporated into computational frameworks. Statistical approaches should be used when describing injury patterns and results.
- **Numerical Methods**: For extreme loading conditions, with large displacements occurring in short time durations, new computational approaches are needed to overcome complexities of the traditional Lagrangian finite element method. Methods such as the material point method (MPM) [155], and other meshless methods should be investigated with soft materials and high rates of loading in mind. Accurate modeling of bone fracture at high loading rates remains a numerical challenge.
- **Structure-Function Relationships**: Efforts should continue to develop numerical approaches which are aimed at understanding structure-function relationships, tying certain injuries to loss of particular functions.

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Preventing Injuries Associated with Military Static-line Parachuting Landings

Julie R. Steele, Karen J. Mickle and John W. Whitting

Abstract Military static-line parachuting is a highly tactical and hazardous activity, with a well-documented injury risk. Due to the high impact forces and rapid rate of loading when a parachutist lands, injuries most frequently occur to the lower limbs and the trunk/spine, with ankle injuries accounting for between 30 and 60 % of all parachuting injuries. Although military static-line parachuting injuries can be sustained at any time between the paratrooper attempting to leave the aircraft until they have landed and removed their harness, most injuries occur on landing. Throughout the world, various landing techniques are taught to paratroopers to reduce the risk of injury, by enabling parachute landing forces to be more evenly distributed over the body. In this chapter, we review research associated with static-line military parachuting injuries, focusing on injuries that occur during high-impact landings. We summarize literature pertaining to strategies for military paratroopers to land safely upon ground contact, especially when performing the parachute fall landing technique. Recommendations for future research in this field are provided, particularly in relation to the parachute fall landing technique and training methods. Ultimately, any changes to current practice in landing technique, how it is taught, and whether protective equipment is introduced, should be monitored in well controlled, prospective studies, with the statistical design accounting for the interaction between the variables, to determine the effect of these factors on injury rates and paratrooper performance. This will ensure that evidence-based guidelines can be developed, particularly in relation to landing technique and how this is trained, in order to minimize injuries associated with landings during military static-line parachuting in subsequent training and tactical operations.

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1 Introduction

The act of parachuting involves descending through the air using apparatus to increase air resistance, thereby reducing the velocity of motion [1]. Parachuting is performed for many reasons, but chief among these are occupational and recreational pursuits. Although recreational enthusiasts constitute hundreds of thousands of the world's parachuting populace [2], most participants come from occupational groups including firefighters (known as smoke jumpers) and rescue groups [3], with the vast majority being military paratroopers [4]. One of the main purposes of military parachuting is to rapidly deliver a large contingent of armed personnel onto a battlefield, in a manner that enables the soldiers to arrive safely on the ground and ready to immediately commence operational missions [3, 5, 6]. Military parachuting was initially used during World War I by the US army as a strategy for "vertical battle engagement", which allowed for the capture of strategic objectives at the rear of an enemy [7]. Since that time, use of military parachuting has become standard practice in armies around the world to deploy combat forces.

During military operations, troops are usually deployed at low altitudes using static-line parachuting, whereby the parachutist hooks his or her parachute onto a 'static' line, which is firmly attached to a strong point on the aircraft [8, 9]. Upon exiting the aircraft, this line is designed to pull the parachute canopy and rigging free from its bag until it is fully opened [9]. In contrast, free fall parachutists, also known as skydivers, are responsible for deploying their own parachute. Although some military personnel also perform free fall jumps, static-line parachuting is the primary airborne means for mass troop deployments [8]. Due to the highly tactical nature of static-line parachuting and the type of equipment used, this form of aerial descent is generally acknowledged as the most hazardous [4, 10, 11].

Although parachuting comes with an inherently high level of injury risk, keeping injuries to a minimum is essential to the readiness, effectiveness, morale, and efficient running of any military unit [12, 13]. In an operational environment, paratroopers who sustain a severe injury are likely to be unable to continue their mission and, therefore, require both medical and evacuation resources for their management [6]. In fact, an injured paratrooper may require up to four soldiers to assist in evacuation from a drop zone [14]. This imposes a substantial operational and logistical burden for commanders [6]. Furthermore, parachuting injuries can occur in both tactical and non-tactical scenarios, with the more severe injuries potentially affecting a soldier's long-term health [13]. In fact, when parachuting goes wrong, the potential for a career-threatening or life-threatening injury is great [15]. As parachuting injuries can have a serious negative effect on the physical health of paratroopers, and in turn the combat capability of the military, analyzing factors that affect parachuting injuries is of military significance [16].

It is well recognized that landing is the most dangerous part of parachuting. That is, most military parachuting injuries occur during impact with the ground when parachutists are forced to absorb extremely high impact loads [9, 16–18]. For this reason, it is imperative that military paratroopers learn to instinctively employ a safe landing technique to effectively absorb these high impact forces, irrespective of potential distractions during the descent, such as gusting winds, uneven terrain or loose equipment [17]. In this chapter, we review research associated with static-line military parachuting injuries, focusing on injuries that occur during high-impact landings. We summarize literature pertaining to strategies for military paratroopers to land safely upon ground contact, particularly when performing the parachute fall landing technique. Recommendations for future research in this field are provided, so that evidence-based guidelines can be developed, particularly in relation to landing technique and how this is trained, in order to minimize injuries associated with landings during military static-line parachuting in subsequent training and tactical operations.

2 Military Parachuting

As stated above, one of the primary requirements of military parachuting is to enable soldiers to jump from an aircraft and land safely on the ground, ready to immediately undertake their combat duties [9]. Unlike their recreational sky diving counterparts, military paratroopers are likely to parachute in a hostile environment, which creates unique demands on how military parachuting is performed. For example, in order to avoid radar and anti-aircraft weapons systems, military aircraft fly as low as possible, usually below 2000 feet above ground level, to minimize exposure [19]. Soldiers also attempt to exit the aircraft as fast as possible to minimize the time the aircraft spends over the drop zone and to minimize troop dispersion [7, 9]. Although some special operations forces use military free fall, including tactical high altitude-low opening (HALO) airborne operations [19], the primary means of parachuting for mass troop deployment is static-line parachuting [8].

Despite substantial changes in modern warfare, principles underlying static-line military parachuting have remained relatively unchanged since the 1940s [9]. A soldier typically carries their military parachute in a bag on his or her back, with a reserve parachute attached to the front. On command from a "jumpmaster", each soldier stands up and hooks a cable from their individual parachute to a static line, which is attached to a strong point inside the aircraft and runs the length of the aircraft [3]. When the doors are opened, each soldier shuffles to an aircraft door and, on command, exits from the aircraft in quick succession [9]. As the soldiers descend, the static line pulls the parachute and rigging lines from each soldier's bag until the breaking strain of the tie holding the static line to the parachute is exceeded. At this point, the static line breaks off and each soldier drifts to the ground (see Fig. 1).

The parachute used in military static-line parachuting is typically a round non-steerable canopy. Although the parachute can be maneuvered a little by pulling on the risers (cords that connect a soldier's harness to the actual parachute) and

Fig. 1 Paratrooper drifting to the ground after a static-line parachuting exit

allowing some air to spill out [9], the jumper has little control over their vertical velocity or direction on landing [19]. If a soldier is required to carry military equipment, the equipment is usually attached below the reserve parachute and can be lowered on the end of a rope while they are descending so that the soldier is unencumbered upon landing [9]. When the parachutist contacts the ground, he or she uses a well-rehearsed landing technique in an attempt to dissipate the high impact forces that are generated at ground contact (see Sect. 5). The soldier then removes the parachute as quickly as possible so they can be ready for ground operations [3].

Despite parachuting being a relatively efficient method of deploying troops, there have been substantial injuries associated with military static-line parachuting [8]. This is not surprising, given the fast descent velocities and, therefore, high impact forces and rapid loading rates encountered when a paratrooper lands from this hazardous form of aerial descent [20]. As these injuries can have substantial negative consequences on the physical and emotional health of parachutists, as well as impact the combat capabilities of an entire military unit, it is imperative that mechanisms of parachuting injuries are understood in order to develop evidence-based strategies to minimize their occurrence.

3 Static-line Military Parachuting Injuries

3.1 Parachuting Injury Risk

In the 1930s, injury rates recorded for military parachuting were as high as 680 per 1000 jumps [12] while during World War II (January–November, 1944) an injury incidence of 21 per 1000 jumps has been reported based on 20,777 jumps undertaken by trained parachutists [21]. However, with major improvements in parachute design and technology, aircraft exit procedures, landing techniques and training, in combination with a better practical understanding of the risk factors involved, these rates have substantially declined to an average of approximately 6–11 per 1000 jumps [3, 5, 9, 22].

When interpreting injuries rates, it is important to remember that military parachuting involves a wide variety of activities, varying from controlled training jumps over flat terrain in daylight to highly tactical jumps under combat conditions over unknown terrain at night [18]. The injury rates during training courses are, of course, much lower than those reported during airborne operations or actual combat. For example, in a study of 59,932 static-line parachute descents performed by Chinese People's Liberation Army cadet pilots during basic training, the overall injury incidence was 2.6/1000 jumps [15]. In contrast, the total injury incidence during a combined United States (US) and United Kingdom (UK) parachuting mass tactical operation, involving combat equipment conducted at night during inclement weather, was 28.8 injuries per 1000 aircraft exits with 24.6 injured soldiers per 1000 aircraft exits (from a total of 4754 aircraft exists; [13]). A comprehensive table of injury rates for military parachuting (injuries per thousand descents) between 1941 until 1998 is provided by Bricknell and Craig [9]. A relatively recent tabulation of the injury incidence in military parachuting, including events associated with injury, and a quantitative assessment of injury risk factors and their interactions during military parachuting is provided by Knapik et al. [5]. It is acknowledged that between-study variations in injury incidences are frequent. These variations can be attributed to differences in injury definitions and study design, how the data were collected, as well as variability between parachute schools, jump conditions, national affiliations of the soldiers, training experience, whether the data were collected during training or operations, and risk factors present during jumps [2, 3, 5]. For example, injury definitions can vary from any medical treatment administered on a drop zone, no matter how minor, to only major casualties receiving attention at hospitals [3, 9]. Furthermore, it is speculated that the frequency of minor parachuting injuries may be underestimated in many military studies, particularly in retrospective studies, because of the desire of military personnel to complete a parachute-training course, and their consequent reluctance to report injuries for fear of medical disqualification [19, 23, 24].

3.2 Parachuting Injury Characteristics

Irrespective of between-study differences in injury incidence, research shows a common trend in terms of the type of injury incurred during military static-line parachuting [9]. That is, injuries to the lower limbs [8, 13, 15] and the trunk and spine are the most prevalent [7, 23, 25]. In fact, lower limb injuries have been reported to be as high as 70–80 % of the total parachuting injuries incurred [7, 15, 18]. Ankle injuries, particularly ankle sprains and fractures, are largely responsible for this latter statistic, accounting for between 30 and 60 % of all parachuting injuries [15, 23, 24, 26, 27]. The high incidence of lower limb and trunk/spine injuries is not surprising, given that these injuries are typically associated with the high impact forces generated at ground contact during a parachute landing [3] and the rapid rate of loading sustained when a paratrooper lands [28], with the lower extremities taking the initial impact of a parachutist's full body weight [7, 17].

A notable exception to most injury profiles was a study of 23,031 jumps taken by members of an airborne infantry unit [5]. In this study the most common injury/anatomical location combination was closed head injuries/concussions (30.6 %), although ankle fractures and ankle sprains were the second most frequently occurring injury (16.9 %). Closed head injuries are of concern as they reflect the vulnerability of the brain to impact [9]. Interestingly, the injury rate for severe injuries during parachuting is relatively low for military operations. For example, Ekeland [18] reported that the risk for fracture or knee ligament rupture was only 2.0 per 1000 jumps during basic courses for paratroopers and 1.2 per 1000 jumps during training exercises. The overall risk of incurring a severe injury in their prospective study was 1.6 injuries per 1000 jumps [18]. More recently, Guo et al. [15] reported a similar low rate of 1.2 per 1000 jumps for severe injuries, whereby fractures, dislocations and ligament ruptures were classified as severe injuries. As such, military static-line parachuting is considered a relatively safe method of troop transportation [18], although it is acknowledged that the consequences of severe injuries can be catastrophic.

Military static-line parachuting injuries can be sustained at any time between the paratrooper attempting to leave the aircraft until they have landed and removed their harness. Injuries typically occur during one of four main phases during the sequence of a parachute jump: (i) when exiting from the aircraft, (ii) during opening of the parachute, (iii) during the descent, and (iv) on landing [16]. Most injuries, however, occur on landing [17]. For example, in an analysis of 23,031 jumps performed by an Army airborne infantry unit, with an injury incidence of 10.5 per 1000 jumps [5], 75 % of those injuries in which an event associated with the injury could be determined involved problems associated with ground impact. Static line problems (11 %), tree landings (4 %), entanglements (4 %), and aircraft exits (3 %) accounted for most of the other injuries [5]. Other studies have reported that 85–90 % of all parachuting injuries occurred during the landing [8, 16, 18], supporting the belief that landing is considered the most dangerous phase of a parachute jump. Given that most military static-line injuries are associated with ground contact, the remainder of this review will focus on injuries occurring during this phase of the parachuting movement. A comprehensive overview of the mechanisms of parachuting injuries occurring before landing is provided by Bricknell and Craig [9].

3.3 Risk Factors for Military Static-line Parachuting Injuries

Despite wide variations amongst studies investigating military static-line parachuting injuries, the data are relatively consistent with respect to whether or not a particular factor increases the likelihood of incurring an injury [3]. For example, studies have consistently shown that the risk of injury increases when parachuting is conducted while there are higher wind speeds [9] and higher dry bulb temperatures, during night jumps compared to day jumps, jumping with combat or heavy loads compared to unloaded jumps, jumps without wearing ankle braces, jumping onto uneven or rough terrain where obstacles are present (e.g. trees, rocks, fences and power lines), and when there have been entanglements [3, 5, 23]. Purpose built sand drop zones used by the US military for paratrooper training have been shown to reduce the injury rate 3.2 fold [3]. Numerous other factors that have been implicated in affecting the risk of sustaining a military parachuting injury are aircraft and parachute type-specific (i.e. model of aircraft, location of exit doors, number and sequencing of soldiers exiting the aircraft, and parachute canopy size and shape; [3, 5]) or factors that reduce ground visibility during descent like low cloud cover [13]. Knapik et al. [3] has provided a comprehensive review of studies examining risk factors for injuries during military parachuting. Some of these risk factors can be controlled during training (e.g. imposing wind speed limits on training jump days). Others (e.g. night descents, multiple parachutists leaving the aircraft and the carriage of equipment), however, are fundamental to the operational capabilities of parachute forces and, therefore, cannot be avoided during military operations (see Table 1).

Table 1 Factors associated with military parachuting injuries

Factor	Effect on injury rate
Wind speed	Increasing wind speeds increase rate
Multiple parachutists leaving aircraft	Increase
Night descent	Increase
Carriage of equipment	Increase
Nature of dropping zone	Hard, uneven or hazards increase rate
Balloon descents	Decreased relative to aircraft descents
Design of parachute	Decreased with modern parachutes
Height and weight of parachutist	Increase
Inexperience of parachutist	Increase

Adapted from Bricknell and Craig [9]

Although intrinsic injury risk factors have also been explored, there tends to be less consistency with respect to which variables are associated with an increased injury risk. Higher injury risk, however, has been associated with intrinsic risk factors such as greater body weight, older age and/or longer time in service ([29–32], less upper body muscular endurance, lower aerobic fitness, and prior injuries [22]; taller stature, fewer push-ups, and slower 2-min run times [31]. Jaffrey and Steele [23] noted that 25 % of Basic Parachute Course trainees with the slowest 2.4 km run times (>10:00 min) incurred 47 % of injuries. The authors cautioned, however, that the interaction between injury, 2.4 km run time and other parameters such as body mass requires investigation, given that parachuting does not demand high aerobic fitness and that better run times are usually achieved by lighter individuals [23].

In an investigation of parachuting injuries sustained by Chinese Air Force Cadet Pilots, the intrinsic risk factors associated with reduced injury rates were excellent mental qualities and parachuting technique, and being a female cadet pilot [16]. In contrast, Knapik et al. [3] stated that, compared with men, women appeared to be at greater risk for injuries overall and, more specifically, to have more fractures, with more injuries caused by improper landing technique. There are, however, many confounding factors when making gender comparisons in military parachuting injury rates (as discussed by Knapik et al. [3] and Guo et al. [15]), such that these comparisons should be interpreted with caution.

4 The Importance of Landing

It is well documented that, by far, most military static-line parachuting injuries occur during landing (see Sect. 3.2). For example, in an analysis of injuries sustained during a series of 51,828 military training parachute descents single injuries were the result of a hard or awkward landing on 95 out of 104 (91 %) occasions [24]. Essex-Lopresti [21] remarked, "that the euphoria accompanying the glorious sense of isolation whilst floating down is tempered by anticipation of the technical difficulties of meeting the ground" (p. 3). Landing is therefore considered the most dangerous phase of parachuting, a time when the parachutist's body is subject to the interaction of gravitational forces in the vertical plane and natural forces of wind in the horizontal plane [19]. The hazards associated with any high impact landing are increased during parachuting by factors such as poor drop zone conditions and obstacles such as trees, fences, and power lines [19]. The single largest cause of these lower limb and trunk/spine injuries at landing, however, is poor landing technique [20] (Fig. 2).

Because most parachuting injuries occur on landing it is pivotal in avoiding injuries that parachutists learn to use a proper landing technique [8]. Training has to ensure that parachutists automatically use this technique irrespective of distractions occurring during the descent, such as gusting winds, uneven or rough terrain, loose equipment or any improper function of the parachute [17, 18]. The increased injury

Fig. 2 Military static-line parachuting trainee landing on the drop zone, with medical treatment facilities available in case of injury, which is usually caused by poor landing technique

risk associated with military static-line parachuting has led to the continuous study and development of military jumping techniques so that parachutists can absorb the high impact forces generated at ground contact [7].

4.1 Impact Absorption During Landing

When a parachutist makes contact with the ground he or she is subjected to a combination of forces including the downward gravitational force, lateral forces from both wind and from oscillation and, possibly, a rotational force if the parachutist is spinning, although this is rare [9]. The lateral force caused by wind is a crucial factor that determines injury rate, as the horizontal speed component caused by wind can dramatically increase the resultant velocity with which a paratrooper impacts the ground [6]. Consequently, although steering capacity is limited (see Sect. 2), parachutists are trained to steer into the wind during the last phase of a descent in order to reduce their lateral velocity to a minimum [9].

Not surprisingly, despite variations and unpredictability in lateral and rotational forces, the primary ground reaction force generated at ground contact is in the vertical direction. It is affected by parachute design and load, including the weight of the parachutist, as well as the equipment carried during the descent [9], which may be up to an additional 45 kg [33]. Irrespective of the relative loads from the paratrooper or their equipment, vertical descent velocity under standard static-line parachutes used by the Australian and US military, for instance, may increase to as much as 6.7 m s^{-1} when the total load approaches the upper load limit of these parachutes (approximately 163 kg; [23]). The overall load (weight) of the paratrooper and their equipment affects the shape of the canopy and, in turn, the air resistance of the parachute. Basic physical principles then dictate that a system with a relatively large mass (paratrooper dressed in military fatigues and carrying equipment) and a vertical descent velocity of at least 4.6 m s^{-1}, but more likely closer to 6 m s^{-1} [17], will have a substantial amount of downward momentum at the moment of ground contact. This substantial downward momentum will, in turn, result in a relatively large impact energy that must be absorbed by the body during the landing [5]. It is beyond the scope of this chapter to discuss military parachute design in further detail, suffice to say that changes in the shape, size and materials used to construct military parachutes have been associated with a reduction in parachuting injuries (see Bricknell and Craig [9] for more details).

Irrespective of parachute design, it has been estimated that the impact forces from military static-line parachute landings are equivalent to those sustained by jumping from a 2.7–3.6 m (9–12 foot) high wall [3, 9]. By applying basic physical principles to a scenario where a parachutist must completely arrest a large amount of downward momentum in a short period of time, it is evident why these impact forces are high. A laboratory-based investigation of simulated parachute landings at three different vertical descent velocities revealed that trained military paratroopers took, on average, only 70–90 ms to reach maximum knee flexion after contacting the ground during the initial energy absorption phase of landing [20]. It is important to note that the peak resultant ground reaction force absorbed during the rapid change in momentum experienced during these landings occurred in less than half of this time [20]. These paratroopers were using the parachute landing fall (PLF) technique that is described in Sect. 5 of this chapter. The paratroopers also only took approximately 730 ms to achieve maximum body-ground contact and up to 1.4 s to complete the entire roll-over during landing (see Sect. 5; [20]). Furthermore, this research demonstrated that landing absorption times were inversely proportional to descent velocity, an effect that can potentially magnify impact forces when load carrying paratroopers are landing at higher descent velocities. As static-line parachute landings involve a feet-first initial ground contact, necessitating rapid energy absorption by the lower limbs, it is not surprising that repeated high velocity parachute landings and their associated impact forces place the lower limbs at increased risk of injury [34]. Therefore, it is imperative that appropriate landing techniques are used to adequately absorb these high impact forces.

During the initial impact phase of a landing, the lower limb joints rotate with a pre-determined degree of neuromuscular control to attenuate the vertical ground reaction forces [35]. That is, lower limb eccentric muscle contractions allow hip, knee and ankle flexion to occur in an anticipated, pre-programmed manner to absorb the forces generated at impact. Forces experienced by the musculoskeletal system are then determined by how stiff or compliant the leg is in response to loading [36] due to the effect of joint range of motion and compliance on impulse time and energy absorption [37]. A selection of pertinent data reported in the literature related to the ground reaction forces generated and absorbed during landing movements, with the lower limbs being the primary source of force attenuation, is provided in Table 2.

As landing movements are highly complex, they require a multi-joint solution to effectively absorb the impact forces generated at ground contact [34, 38]. In essence, the lower limbs act largely as springs to rapidly and eccentrically absorb external loads imposed by landings. A low joint range of motion, particularly in the ankle, knee, and hip during impact absorption, is generally associated with a stiff-legged landing [37, 39]. Butler et al. [39] stated that an optimal level of leg stiffness was required to avoid injuries during landing activities. These researchers suggested that too much leg stiffness may be associated with bony injuries, because of the rapid rate of loading during the impulse, which results in higher forces. Conversely too little leg stiffness may be associated with a larger excursion of the joints, leading to soft tissue injuries [39]. Although lower limb loading is often considered in terms of simplified leg spring models [39–41], it is important to consider that energy absorption by the lower limb involves biological tissues that display viscoelastic characteristics and, therefore, influence the behavior of the system accordingly. For this reason, more complex mass-spring-damper models may more accurately represent aspects of lower limb loading responses during foot-ground impacts [42, 43]. Nonetheless, Alexander [44] concluded a thorough discussion of modeling approaches in biomechanics by stating that "Even the most complex of the models that I have discussed are simplified representations of reality" (p. 1434), acknowledging that no matter how considered a model is, one cannot account for every eventuality. Whittlesey and Hamill [45], also noted that the more complex mass-spring-damper models were not easy to apply or interpret and, although the lower limb is not a perfect spring, simple mass-spring models can be useful in understanding gross loading responses to foot-ground impacts. As such, for the sake of simplicity and often acknowledged as the best approach in understanding the major features of a model [44], leg-spring models are frequently applied two dimensionally by examining the effects of ground reaction forces on linear displacement in a vertical plane [40].

Farley and Morgenroth [40] demonstrated that leg stiffness during human hopping was directly proportional to ankle joint stiffness and far less influenced by knee stiffness. This is not surprising during a foot-first vertical ground impact where the ankle is the first major joint to encounter the vertical reaction force. Nonetheless, greater contributions to energy absorption are required by larger more proximal joints, such as the knee and hip, as landing velocities increase [46]. Research has shown that the rate of ankle injuries sustained during parachute landings is most sensitive to increases in vertical descent velocity [47]. This most likely reflects the limited capacity of the ankle, as the first major link in the lower limb chain, to cope with the high impact forces and fast loading rates typically

Table 2 Examples of studies that have provided ground reaction forces generated during landing movements

Landing movement	Relevant results	Reference
Barefoot single leg drop landings (6 female gymnasts). 2 techniques (stiff and soft). 3 types of gym mats used. 2 heights: 80 cm (4.0 m s^{-1}), 115 cm (4.8 m s^{-1})	Peak ground reaction force (GRF) ranged from 2.7 to 4.2 times body weight (BW) for 1 foot	[64]
Two-foot shod drop landings (3 trials per condition @ 50 cm height) onto 2 force platforms with and without vision (139 male US air assault soldiers)	Peak vertical GRF ranged from 3.4 to 3.8 BW per leg in vision and non-vision conditions	[65]
30 male US special operations forces soldiers ×15 simulated PLF landings. Performed during vertical drop landings in 5 different footwear/knee and ankle brace conditions onto FP from 3 heights of 1.07, 1.37 and 1.71 m (~4.58, 5.18 and 5.79 m s^{-1}, respectively)	Effect of height on peak vertical GRF —8.9 to 17.3 BW for simulated parachuting landings from 1.07 to 1.71 m	[33]
2 foot landings from 3 heights and distances (n = 3, 81 trials each). Heights of 40, 60 and 100 cm (2.8, 3.4 and 4.4 m s^{-1}, respectively). 3 landing stiffness techniques used. All landings toe to heel	Peak GRF in 1 leg was 3.0 BW at 40 cm–6.6 BW for stiff leg landing at 100 cm	[66]
Drop landings with 2 feet onto 1 force platform. 5 expert male parachutists (>100 jumps each). 4 trials × 6 heights × 3 gaze directions. Heights of ~20, 40, 60, 80, 100, 120 cm (2.0, 2.8, 3.4, 4.0, 4.4, 4.9 m s^{-1}, respectively)	Average peak GRF = 3.7 BW at 20 cm to 5.9 BW at 100 cm	[67]
Drop landings by well-trained female gymnasts (n = 9). Competition style 2 foot landings. 3 heights × 2 mats. Heights of 69, 125 and 182 cm (3.7, 5.0 and 6.0 m s^{-1}, respectively)	Peak GRF from 4 BW at 69 cm up to 9 BW at 182 cm	[34]
Drop landings (6 elite male gymnasts and 6 recreational male athletes). 3 heights of 32, 72 and 128 cm (2.5, 3.8 and 5.0 m s^{-1}, respectively)	Peak vertical GRF for gymnasts at 3 heights = 3.9, 6.3 and 11.0 BW; for recreational athletes = 4.2, 6.4 and 9.1 BW	[38]
Double back somersaults performed by elite gymnasts	Peak vertical GRF up to 18 BW	[38]
2 foot competition style (gymnastic) drop landing (10 female, 4 male intercollegiate gymnasts). 1 drop height of 69 cm (3.7 m s^{-1}) × 3 landing surfaces (no mat, stiff and soft mat)	Peak vertical GRF lowest for no mat condition and highest for stiff mat. Forces ranged from 3 to 6 BW	[68]

(continued)

Table 2 (continued)

Landing movement	Relevant results	Reference
6 elite male gymnasts performed 3 landing types—drop landing, front salto, back salto. From platform at height of 72 cm (3.8 m s^{-1})	Peak vertical GRF ranged from 3 BW (drop) to 5 BW (back salto)	[69]
Gymnast dismounting from the horizontal bar	Impact forces (GRF) up to 11.6 BW	[57]
Barefoot drop landings (6 males). 10 trials × 5 heights. Heights were 20, 40, 60, 80 and 100 cm (2.0, 2.8, 3.4, 4.0 and 4.4 m s^{-1}, respectively)	Mean peak vertical GRF for each height were 3.9, 4.7, 5.6, 6.9 and 7.9 BW	[35]
Barefoot drop landings (7 males, 1 female). 10 trials × 4 heights × 2 conditions (vision vs. no vision). Heights were 20, 40, 60 and 80 cm (2.0, 2.8, 3.4 and 4.0 m s^{-1}, respectively)	Vertical GRF occasionally reached 12 times BW for no vision trials and up to 8 BW with vision	[70]
Drop landings (10 female competitive gymnasts and 10 female recreational athletes). 10 trials × 3 heights—30, 60 and 90 cm (2.4, 3.4 and 4.2 m s^{-1}, respectively)	Peak GRF for gymnasts = 2.8, 4.1 and 5.7 BW; for recreational athletes = 2.2, 2.8 and 3.8 BW	[59]
20 paratroopers (1 female). 5 trials × 3 descent velocities. Heights were 32, 74 and 133 cm (2.1, 3.3 and 4.6 m s^{-1}, respectively)	Mean peak vertical GRF for each height was 5.8, 9.3, and 13.1 BW. Some participants averaged up to 18 BW, with one sustaining 24 BW on a single fast velocity trial	[20]
Barefoot (single leg) drop landings (33 males). 5 trials × 2 heights—32 and 72 cm (measured at 2.25 and 3.21 m s^{-1}, respectively). Low versus high dorsiflexion range of motion (ROM) comparison	Peak vertical GRF on single leg was 4.2 BW @ 32 cm and 6.9 BW @ 72 cm. No effect of dorsiflexion ROM	[71]
Drop landings (9 males). 5 trials × 3 heights × 3 techniques. Heights of 32, 62 and 103 cm (2.5, 3.5 and 4.5 m s^{-1}, respectively)	Mean peak GRF were 2.6, 3.3 and 4.7 BW for the 3 drop heights	[46]

experienced during static-line parachute landings. Conversely, while excessive horizontal wind drift, responsible for the horizontal component of descent velocity, can lead to trunk, head and upper limb injuries during a parachute landing, ankle injuries have been shown to be unaffected by horizontal wind speed [47, 48]. Consequently, while it is unclear precisely how much of the landing impulse and resulting impact force is absorbed by lower limb joints immediately following initial ground contact, as ankle injuries are the most prevalent injury in parachute landings, the notion of a leg spring model becomes central to any investigation of parachute landing techniques. During parachuting, the ground reaction forces generated at landing are also influenced by factors such as gravity, pendulum-like oscillations of the paratrooper and the hardness of the drop zone [9]. Any other variables that can increase a paratrooper's descent velocity, such as temperature, humidity and type of parachute, add to the total ground reaction force generated upon landing [3]. As most of these variables are beyond a paratrooper's control while descending, it is critical that the paratrooper maintains good body position on the landing approach and that the correct landing technique is employed [49]. Almost without exception, poor landing technique has been identified as the largest cause of landing injury in parachuting [9]. In fact, Ekeland [18] claimed that 70 % of all parachute-landing injuries resulted from poor landing technique.

5 The Parachute Landing Fall Technique

During the early days of parachute training, German and American parachute trainees were taught to land with their feet apart and then perform a forward roll across an outstretched arm to dissipate the kinetic energy existing at landing [9]. German parachutists were restricted to rolling forward upon landing because their parachute harnesses were attached at a single point at the centre of the upper back, causing the parachutist to land in a forward facing position [9]. British parachutists, however, descended in an upright position because the rigging lines of their parachutes merged onto risers (cords that connect the paratrooper's harness to the actual parachute), which attached to the parachute harness on the top of each shoulder [9]. This allowed the British to develop a landing technique in which the parachutist landed with their feet and knees together, and then rolled sideways. As this technique, known as the Parachute Landing Fall (PLF), was associated with significantly fewer injuries than the forward landing roll, the US Army adopted it in 1943 [5, 9].

Today, the PLF is a widely accepted method used by most military parachutists worldwide to reduce the risk of injury at ground contact ([9]; see Fig. 3). To perform a PLF, during the final stage of an aerial descent, the parachutist assumes a relaxed pre-landing posture. This posture is characterized by holding the feet and knees together, with slight flexion of the hips and knees [4, 49–51] to allow for optimal absorption of the initial ground reaction forces [28]. The chin should be placed on his or her chest, with the elbows tightly tucked in and hands clasping the risers [6], in readiness for ground contact.

On impact with the ground, the PLF involves a simultaneous touchdown of both feet with the lower limbs locked tightly together. It has been speculated that landing with the feet apart is likely to cause one foot to strike the ground before the other [9], such that the majority of landing forces are absorbed by one limb, with a subsequent increase in injury risk. Dual-limb ground contact is immediately followed by the paratrooper turning side on to the direction of landing to perform a

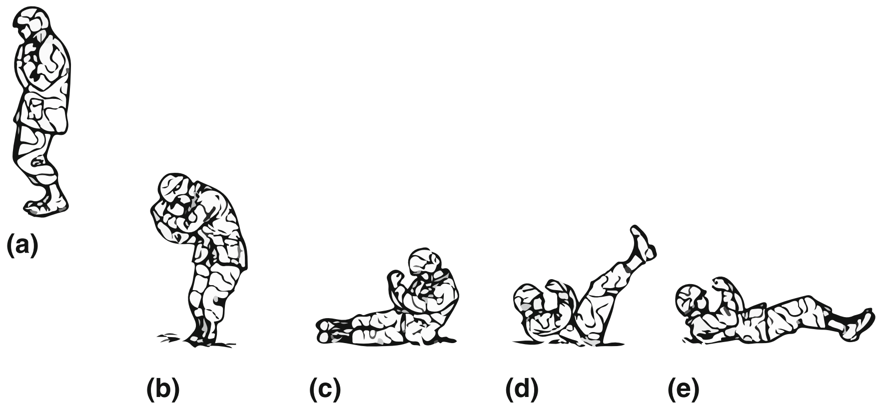

Fig. 3 Sequence of the correct parachute landing fall technique with side drift: **a** preparing to land, **b** initial ground impact with both knees and feet locked tightly together, **c** absorbing the landing forces over the lateral aspect of the calf, thigh and buttocks, **d** rolling across the latissimus dorsi region of the back, and **e** coming to rest at the end of the PLF (adapted from http://commons.wikimedia.org/wiki/File:Parachute_landing_fall.jpg)

sideways roll, involving five points of contact between the ground and the parachutist's body. The five points of body-ground contact begin with the feet, followed by the lateral aspects of the calf, thigh, and buttocks of the lead lower limb, with the latissimus dorsi region of the back on the same side as the lead limb being the final point of impact before the paratrooper dissipates the residual momentum by rolling over and coming to rest on the opposite side of the body ([20]; see Fig. 3). In theory, this sideways roll with multiple points of contact should enable the initial impact forces to be distributed across as much of the body as possible to disperse the kinetic energy carried into the landing by the paratrooper [6]. In turn, this should limit the need to dissipate all of the ground reaction forces via the lower limbs, thereby reducing the chance of injury [4, 9]. However, as stated above (see Sect. 4.1), it is important to acknowledge that the peak ground reaction forces occur at less than 100 ms [20] after initial ground contact, at a time when the paratrooper has still only made the first point of body-ground contact with their feet. Therefore, it is reasonable to conclude that a substantial portion of energy dissipation actually occurs through the lower limbs prior to the parachutists rolling through the other four points of body-ground contact. Once the roll has been completed and the parachutist is lying on the ground, he or she activates the canopy release so any remaining air is released from the canopy so the parachute can collapse and be removed, and the soldier is ready for ground operations [3].

Failure to perform the PLF correctly can lead to injury. For example, holding the feet too far forward can lead to the paratrooper absorbing the initial impact on their heels and then falling backwards to 'sit down' [23]. This causes the impact forces to be transmitted directly through the coccyx and lumbar spine, which can, in turn, cause compression-type fractures [6, 9]. Jaffrey and Steele [23] conducted a qualitative analysis of video clips, which depicted landings performed onto a drop zone by trainees during basic parachute training courses conducted throughout 2004. In almost 70 % of the assessed landings, trainees held their feet too far forward in preparation to land, rather than having them vertically aligned with the body [23]. In the same cohort, although most (88.4 %) held their chin on their chest correctly, ~35 % held their knees too flexed in preparation to land, 65 % did not 'turn their feet off' appropriately in order for the long axes of their feet to be perpendicular to the line of drift, 47.5 % had their feet apart rather than together at impact, and a PLF roll was absent in 78.7 % of the assessed landings [23].

Even when the PLF is performed properly, the injury risk can be high. For example, during the PLF, the paratrooper's ankle that is furthest from the direction of the PLF roll is subjected to a pronation/eversion moment, which can result in an ankle fracture [9]. Conversely, irrespective of the feet being held tightly together, substantial sideways drift, such as that caused by high winds, can force the ankle closest to the direction of the PLF roll into excessive inversion/supination and, in turn, injure the ankle ligament complex, fracture the tibia or fracture the tibia and fibula [9].

Although the concept of distributing the landing forces to reduce injury potential is simplistic, when the PLF is broken into phases, neuromuscular and mechanical strategies required to achieve effective force distribution are extremely complex. From Sect. 4.1, it is apparent that the ground contact phase of landing, occurring immediately after impact, determines lower limb loading via ankle and leg stiffness and that the stiffness of the leg spring is an appropriate representation of the average stiffness of the entire musculoskeletal system during this phase [40]. Furthermore, it is widely accepted that lower limb loading increases with increases in both landing velocity and leg stiffness [46] and that at higher landing velocities, greater knee and hip flexion are likely to be required to mitigate the effects of the vertical ground reaction forces generated at foot-ground contact in drop landings [34, 46]. Unlike isolated drop landings, however, the PLF involves continued dissipation of momentum and force during the rolling phase and requires that paratroopers change the direction of force from an axial direction, very quickly into a rotational and somewhat torsional direction, prior to rolling onto the additional four points of the body [52]. Not only does the initial ground contact phase of the PLF determine lower limb loading, but it also determines how quickly the parachutist goes into the roll, the magnitude of reaction forces encountered in subsequent phases, as well as the overall force encountered from the landing. Therefore, research pertaining to isolated drop landings has only limited relevance to understanding factors affecting performance of the PLF.

Paratroopers absorb the initial impact of a landing by eccentrically dorsiflexing their ankles and flexing their knees and hips [28]. A parachute landing field study conducted by Henderson et al. [49] revealed that for best PLF performance, maximum knee flexion had to occur early, as this allowed for an extended execution of the entire roll. Although this suggests a stiff landing during the initial ground contact phase, overall, forces imposed on the body may be minimized relative to a less stiff PLF landing. Henderson et al. [49] also postulated that minimizing knee flexion, although increasing the vertical ground reaction forces, may help to reduce patellar tendon loading. Hoffman et al. [51] found that more experienced paratroopers displayed stiffer landings during drop landings compared to their less experienced counterparts. The authors speculated that this increased stiffness might be necessary to more efficiently dissipate the overall landing force by providing a more expedient transition to the rolling phase of the task. It should be noted, however, that this was not tested with a full PLF movement task and therefore their conclusions need to be considered with some caution. In light of the fact that lower limb injuries account for most injuries sustained during parachute landings (see Sect. 3.2), the correct amount of leg stiffness poses a real dilemma to PLF instructors. Controlling the rate of roll-over by regulating leg stiffness may safeguard some body parts while exposing others to increased injury risk.

5.1 International Variations in the Parachute Landing Fall

Previous studies have shown that injury rates and types differ between nations. For example, Craig et al. [13] showed that in a joint military operation, UK soldiers sustained more injuries overall (3.8 injuries per 1000 aircraft exits) than the US forces (2.9 injuries per 1000 aircraft exits). In addition to differences in overall injury incidence, there were different injury patterns whereby the UK soldiers had significantly more lower extremity and closed head injuries and significantly more multiple injuries than their US counterparts [13]. Although the authors noted it was difficult to determine the cause of these differences in injury statistics, it is interesting to note that subtle variations of the PLF exist between national military organizations, particularly with regard to foot pitch angle at initial ground contact [28]. For example, the Australian Defence Force paratroops are taught to hold their feet slightly dorsiflexed to land flat-footed upon ground contact [6], whereas US paratroopers today use a plantar flexed, toes-first foot pitch and make ground contact first with the balls of the feet [5].

It has been speculated that landing with the feet plantar flexed is likely to transmit the landing forces through the metatarsal bones (which may fracture) toward the articular surface of the tibia, possibly causing the posterior lip to fracture [9]. Only one study, however, has systematically investigated the effects of variations in foot pitch at initial ground contact on PLF technique under realistic ground training conditions of vertical and horizontal descent velocities. Whitting et al. [28] investigated whether differences in foot pitch affected parachute landing technique by monitoring kinematic, ground reaction force and muscle activity data for 28 (mean age = 30 ± 7 years; 1 female) skilled paratroopers who performed parachute landings (descent velocity 3.3 ± 0.2 m s^{-1} with a constant horizontal drift of 2.3 ± 0.01 m s^{-1}) from a custom-designed monorail system. During the 134 trials analyzed for the study, 69 % of the total landings involved initial ground contact with the ball of the foot, which is in direct contrast with the neutral flat-footed posture at ground contact taught to these Australian paratroopers. Foot-pitch at ground contact significantly affected PLF technique, whereby each foot pitch group used an entirely different biomechanical strategy to absorb the initial impact forces during landing. That is, those participants who used a ball of the foot foot-pitch displayed significantly greater knee extension and ankle plantar flexion and a larger range of knee and ankle motion during impact absorption [28]. Furthermore, the ball of the foot foot-pitch group displayed a significantly lower vertical ground reaction force at ground contact (8.4 ± 1.4 BW vs. 10.8 ± 0.8 BW; $p > 0.001$), and a less rapid rate of loading than the flat-foot foot-pitch group (37.4 ± 5.9 ms vs. 20.1 ± 2.1 ms from initial ground contact until the peak resultant force time; $p > 0.001$). Interestingly, there was no statistical between-group difference in the time taken to make the standard five points of body-ground contact and both groups used a reasonably consistent neuromuscular recruitment strategy [28]. It was postulated that using the ball of the foot first technique was a protective adaptation to reduce excessive lower limb loading during initial ground contact. That is, it allows the paratroopers to use an extra segment during the impact absorption phase of landing, which can assist in force absorption by increasing the impulse time and, in turn, reducing the absolute load in this phase of the landing [28]. This notion is further discussed in Sect. 5.2.2 (Fig.4).

Chinese paratroopers are trained to use a half-squat parachute landing and not the PLF technique [53]. Similar to the PLF, half-squat landing involves contacting the ground simultaneously with both feet to diminish the energy of falling [54]. Instead of rolling, however, the paratroopers contact the ground with both hips, knees and ankles all flexed, keeping these joints flexed until their trunk regains balance, and they can resume a neutral stance position [54]. Despite differences in

Fig. 4 Paratrooper performing the parachute landing fall technique onto a force platform to assess the ground reaction forces generated at landing (taken during the study of Whitting et al. [28])

techniques, a prospective study of all military parachuting injuries recorded for Chinese Air Force cadet pilots (n = 168 injuries recorded in 153 cadet pilots) during basic static-line parachuting training between 1988 and 2008 [15, 16], showed injury profiles and mechanisms that were similar to their Western counterparts. However, for reasons discussed previously, it is difficult to make comparisons between studies due to a multitude of differences in data collection and treatment techniques with regard to injury definitions.

5.2 Training the Parachute Landing Fall

Irrespective of discrepancies pertaining to the exact technique, learning to land safely is critical as appropriate training and attention to detail can substantially reduce the risk of sustaining a parachuting injury [15, 24]. Training should be structured so that performance of the correct technique becomes automatic, even in a crisis or when performing a hazardous mass tactical, combat equipment assault at night during inclement weather [9, 13, 17].

5.2.1 Training Program Structure

Training for military parachutists is typically longer and more intensive than that required for civilian parachutists because the extreme demands and skill involved in being able to participate in a massed parachute assault at night are substantially greater than those required for a novice sport descent [4, 9, 24]. In addition, military training needs to ultimately lead to imprinting the biomechanical skills involved in executing a safe landing so that the basic drills become automatic, irrespective of external distractions [9]. For this reason, substantial time during training should be devoted to demonstrating landing techniques [26], as well as ensuring sufficient actual descents are experienced to ensure paratroopers can perform correct PLF technique in real descents. Although all paratroopers require experience jumping out of aircraft onto the drop zone, performing a high number of actual jumps in the field to improve technique is not feasible due to both high costs and the fact that it could result in more chronic injuries [7]. For this reason, military static-line parachute training incorporates a substantial component of ground training to develop the landing technique of trainees [7]. It has also been advocated that after basic parachute training is completed, that paratroopers maintain their skills by completing frequent military static-line parachuting training activities throughout a year (>three descents/year) and/or extensive ground training prior to conducting military static-line parachuting over land [6].

Adaptation to increases in descent velocity and training experience involve the development of mechanical and neuromuscular strategies, which enable parachutists to optimally perform the PLF technique [55]. As stated by Santello and McDonagh [35], during landing movements the motor control system must be able to predict the vertical ground reaction forces likely to be encountered on ground contact and activate the absorption system appropriately, providing the optimum degree of limb compliance. In a high-speed collision with the ground, reflex patterns are not quick enough to contend with force dissipation [35]. This leaves little doubt that experience through effective training is necessary to equip paratroopers with the ability to absorb loads imposed by a parachute landing.

Military parachute training schools the world over teach trainee paratroopers the PLF technique, albeit with slight between-school modifications [2, 3, 9, 56]. The Australian Military Parachute Training School (PTS) conducts several 3-week courses in static-line parachuting, for novices, each year. The goal of this training is to provide trainees with competency in the operational aspects of military static-line parachuting and approximately 60 soldiers are trained per course [23]. The obvious implication from this example is that a substantial number of military personnel, both in Australia and worldwide, are exposed to the risks associated with static-line parachuting during training and beyond as they engage in operational activities. The potential burden on personnel, military organizations and health care systems for preventable catastrophic injuries would likely be substantial should training programs be inadequate in preparing personnel.

Using the Australian Military PTS as an example of training progression, training begins in a gymnasium with floor exercises designed to teach trainees the 5-point PLF roll (see Fig. 5a). Trainees then progress to a slightly more complex scenario, such as the wheel trainer displayed in Fig. 5b, where they are suspended below a swinging wheel. This type of progression enables trainees to learn to adopt the correct 'prepared to land' posture, with their feet less than 1 m above padded mats on the gymnasium floor (see Fig. 5b). The swinging wheel trainer also acts to impart a small degree of horizontal velocity before the trainees allow themselves to drop onto the mats and practice using the PLF roll that they learned during the initial floor exercises. Due to their close proximity to the floor on release, vertical descent velocities experienced by trainees on the wheel trainer, estimated at between 2.1 and 3.4 m s^{-1} [23], are considerably lower than those experienced under a canopy in the field. Other progressions before attempting an actual aerial parachute landing onto the drop zone, can involve descents from various apparatus, such as a 30 m tower used at the Australian Military PTS (see Fig. 5c), whereby paratroopers are fixed to a cable that controls the trainee's vertical descent rate (usually between 2.7 and 3.9 m s^{-1}), with a small component of horizontal velocity [23]. The progression to an apparatus such as the tower, allows paratroopers to experience a controlled descent, albeit at relatively low velocities, while being exposed to a sense of height from the ground.

Jaffrey and Steele [23] investigated training methods used at the Australian Military PTS to determine whether improvements could be made to either the PLF technique itself or the methods used to teach the current PLF technique. Investigation by these researchers also included observation and analysis of equipment used during PLF training in order to ascertain whether equipment modifications could assist in decreasing the risk of parachute training injuries. The results of this study are detailed in a report titled "Landings during 2004 basic

Fig. 5 Ground training progressions at the Australian Military Parachute Training School, Nowra, New South Wales: **a** floor exercises on gymnasium mats; **b** wheel trainer exercises in the gymnasium; and **c** training tower landing while suspended in a harness

parachute courses: Common technique faults and injuries" [23]. One of the key findings of this report was a distinct delineation between vertical velocities achieved during ground training activities and those experienced under canopy on the first real aerial descent. It was reported that ground training apparatus restricted the vertical descent velocities achieved by trainees to between 0 and 3.4 m s^{-1}, whereas aerial parachute descents onto the Drop Zone resulted in vertical velocities of between 4.6 and 6.7 m s^{-1}. A study involving US paratroopers has shown that simulated parachute landings with vertical descent velocities within this range can result in ground reaction forces in excess of 17 times body weight being generated [33]. Vertical velocities experienced by trainees during ground training at the Australian Military PTS at the time this report was commissioned, therefore, were considerably less than those encountered when descending under a canopy for the first time [23]. Since this report, further research has been conducted to determine the effects of varying vertical descent velocity on PLF technique ([20]; see Sect. 5.2.2). Based on recommendations arising from this research, ground apparatus that can vary horizontal and vertical velocities to simulate realistic descent velocities was designed. Consequently, paratroopers at the Australian PTS can now experience an additional ground training progression that involves simulated PLF landings at more realistic descent velocities. This practice is consistent with the advice of Salai et al. [52], who stated that the best way to prevent parachute landing injuries was to ensure that novices perform landings during ground training, focusing on correct technique while accurately replicating real aerial scenarios, particularly with respect to descent rates. Clearly, it is essential that paratroopers are properly trained to handle landing impacts associated with realistic vertical descent velocities.

Aside from learning to land safely by correctly employing the PLF roll, during the ground-training phase of a basic course, military trainees will also learn the necessary basics regarding the parachute harness and rigging. This type of training allows paratroopers to engage with use of the harness and to experience being suspended below the rigging that extends to the canopy. Furthermore, while experiencing suspension in the parachute harness during ground training, paratroopers are drilled in other important tasks, such as operating the rigging and ‘risers’ that are used to provide limited control of velocity and steerage (see Sect. 2), and how to maintain the correct descent posture while positioned in the harness.

5.2.2 Vertical Descent Velocities and Landing Technique

It is widely accepted that ground reaction forces, and therefore lower limb loading, increase during parachuting landing tasks with increases in landing velocity [20, 33]. In static-line parachuting there is little a paratrooper can do to reduce their descent velocity, making efficient lower limb biomechanics crucial to moderate the high impact forces [28]. It is therefore imperative that military static-line parachute training involves activities that prepare paratroopers to be able to perform the PLF under realistic decent velocities.

Although there is limited research pertaining to parachute landings, the biomechanics of drop landings under different conditions of height and velocity has been studied extensively [35, 38, 46, 51, 55, 57, 58]. Several of these studies have shown that, although athletes display consistency in the temporal aspects of muscle sequencing as landing velocities increase, they use different kinematic landing strategies for absorbing loads imposed by the higher velocities [34, 38, 59]. Despite this wealth of landing-related research, only limited research has systematically investigated the effects of velocity on performance of the PLF, but with most landing studies having used subjects from non-parachuting populations or, where parachutists were used, the movement task did not replicate the PLF [51]. One of the most comprehensive biomechanical studies of parachute landings [33] reported results based on PLF rolls that were performed from drop boxes without a standardized or realistic horizontal velocity as experienced by paratroopers in the field. Furthermore, although one study observed parachute landings during a military parachute course, descent rates and modes of drift were not standardized [49].

Therefore, particularly with respect to differences in vertical descent velocity, there was little understanding of the biomechanics of the PLF technique until the research of Whitting et al. [20].

Whitting et al. [20] collected kinematic, ground reaction force and muscle activity data for 20 Basic Parachute Course trained personnel (19 men, 1 woman; mean age = 32 ± 8 year; height = 176 ± 67.3 cm; mass = 83.0 ± 10.2 kg) while they performed simulated parachute landings using the PLF technique. The landings were performed using a custom-designed monorail apparatus, with a constant horizontal drift velocity (2.3 m s^{-1}) and at three realistic vertical descent velocities: slow (2.1 m s^{-1}), medium (3.3 m s^{-1}), and fast (4.6 m s^{-1}). Descent velocity significantly affected most of the biomechanical variables characterizing the PLF technique. That is, as descent velocity increased, the participants activated their antigravity muscles significantly earlier prior to initial ground contact and then contacted the ground with their knees more extended and their feet more plantar flexed. This strategy likely allowed the paratroopers to absorb the significantly higher impact forces (see Table 2) generated during conditions of greater descent velocity, over a longer period than they otherwise would have experienced, by using greater ankle and knee motion during impact absorption [20].

As these technique modifications were more evident as descent velocity increased, it is postulated that the modifications were protective adaptations used to reduce excessive lower limb loading during initial ground impact [20]. Interestingly, although the participants tended to land flat-footed during the slow velocity trials, reflecting the technique taught during Australian Defence Force Basic Parachute Courses, they tended to neglect their trained neutral foot position at landing, and instead plantar flexed the foot to contact the ground with the balls of the feet progressively more as descent velocity increased [20]. In fact, 78 % of landings in the fast descent condition involved plantar flexed feet. This technique modification added an extra segment to the initial impact phase and it was postulated that this would have influenced absorption of the ground reaction forces as discussed previously.

In a follow up study Whitting et al. [28] compared the landing biomechanics displayed by individuals who consistently used a flat-footed technique and individuals who adopted a plantar flexed foot position, in a cohort of 28 trained paratroopers performing simulated PLF landings at a moderate vertical descent velocity (3.4 m s^{-1}). Peak ground reaction forces occurred significantly earlier and were significantly higher in the paratroopers who used the flat-footed technique because they performed the initial absorption phase of the landings with significantly reduced knee and ankle ranges of motion, resulting in a significantly shorter period of time for the impulse during this phase of landing. The landing kinematics displayed by the flat-footed group suggested these paratroopers used a stiffer landing with faster loading rates. Henderson et al. [49] and Hoffman et al. [51] speculated that reducing knee flexion early in the PLF to expedite the PLF roll may be beneficial to reducing overall landing forces, but this notion was not supported by the experimental findings of Whitting et al. [28]. As stated earlier, injury statistics demonstrate that lower limb injuries occur most frequently during PLF landings and that the ankle and knee joints are the most vulnerable (see Sect. 3). Furthermore, due to mechanisms discussed in Sect. 5, ankle injuries are likely to occur during the initial impact phase when ground reaction forces are high. It seems intuitive that strategies that can safely reduce ground reaction forces during the early impact absorption phase of the PLF may be advisable. Nonetheless, these conclusions remain speculative and require further investigation.

Irrespective of potential injury implications for varying foot pitch during initial impact in PLF landings, as paratroopers display a significantly different PLF technique with increasing descent velocity [20], it is recommended that PLF training programs should include ground training activities with realistic vertical descent velocities. This will better prepare trainees to withstand the impact forces associated with initial aerial descents onto the drop zone and, ultimately, minimize the potential for injury.

5.2.3 Parachuting Experience and Landing Technique

Aside from vertical descent rates, parachuting experience has also been shown to influence landing strategies employed by paratroopers. For example, Hoffman et al. [51] showed that experienced and novice paratroopers displayed significant differences in both the ground reaction forces generated at impact and the time to peak force when performing drop landings from varied take-off heights. Although participants in this study were not subjected to horizontal drifts characteristic of parachute landings and did not perform a typical PLF, results demonstrated possible differences in landing strategy with variations in training experience. The experienced paratroopers displayed a much stiffer landing strategy, even though each group had statistically similar results for leg strength and power. Hoffman et al. [51] postulated that the experienced participants used leg power by optimally tensing their leg extensor muscles prior to landing, thereby reducing the time to peak force during the initial ground contact phase of the landing. In apparent contrast, Henderson et al. [49] found that a period of myoelectric silence in the lower limb muscles following initial landing impact was exaggerated with experience, as well as increased descent velocity. However, due to the dearth of similar studies in the area of parachute landing biomechanics, further research is required to investigate the effects of different vertical descent velocities on landing technique and how this is influenced by jump experience.

6 Preventative Strategies to Reduce Parachuting Injuries

6.1 Ankle Braces and Parachute Landing Fall

As highlighted in Sect. 3.2, the ankle is the most commonly injured anatomical site during military static-line parachuting. Ankle injuries can also have a negative impact on a soldier's well-being and, possibly, on their career progression [60]. In response to the frequency and negative impact of ankle injuries, an outside-the-boot parachute ankle brace (PAB) was developed to reduce the incidence of ankle injuries. The PAB (Aircast, NJ, USA) consisted of a hard plastic outer shell lined with air bladders, which padded the medial and lateral malleoli, to prevent extreme ankle inversion and eversion while allowing plantar flexion and dorsiflexion [14]. Trials have consistently shown that the risk of ankle injury is reduced when the soldiers wear a PAB during parachuting training, with no accompanying increase in the risk of other traumatic injuries [14, 22, 60]. For example, a randomized trial conducted at a US Army Airborne School demonstrated that the PAB reduced ankle sprains among 745 trainees by 85 % [27]. Schmidt et al. [14] compared the hospitalization rates for ankle, musculoskeletal, and other traumatic injuries incurred by 223,172 American soldiers who were trained during 1985–2002 in distinct time periods during which PAB use was mandatory or not. The researchers found that, of the 939 parachuting-related hospitalizations during the study period, the odds of experiencing an ankle injury were twice as high when training occurred during intervals when the use of an outside-the-boot ankle brace was not required [14]. A summary of studies examining the parachute ankle brace is provided by Knapik et al. [61]. In fact, in their systematic review, Knapik et al. [61] found that risk of ankle injury or ankle sprain was more than twice as high among individuals not wearing the ankle brace while risk of ankle fracture was about 1.8 times higher among those not wearing the brace. Importantly, the risk of other lower limb injuries was not significantly elevated among those who wore the PAB [60, 61]. These and other consistent findings clearly establish that using the PAB is a cost-effective intervention that reduces the incidence of ankle injuries, ankle sprains, and ankle fractures during military parachuting by about one half and, in turn, can reduce individual soldier morbidity and financial costs to the military [60, 61]. In fact, Schmidt et al. [14] estimated that the ratio of expenditure ($30,000 per year USD) for braces to hospital care and rehabilitation costs saved ($835,000 per year) was 1:29. Interestingly, despite this evidence, use of the PAB in some sectors of the military has not been adopted or has been discontinued due to anecdotal reports claiming increases in risk of other types of injury, and the cost of obtaining and periodically replacing the PAB [60].

Although the impact of brace wearing on injury incidence has been well documented, less evidence has examined the mechanisms by which braces work. It is likely that the PAB reduces ankle injuries, particularly fractures and sprains, by strengthening the carrying capacity and stability of the ankle joint and associated bones, preventing excessive ankle inversion on ground impact, as well as by increasing sensory awareness or proprioception [3, 15]. Knapik et al. [61] speculated that ankle braces acted as a splint, providing stiff medial and lateral support to the ankle. The authors postulated that this splinting likely reduced the velocity and/or extent of ankle inversion or eversion at ground contact, thereby preventing excessive ankle motion that often leads to injury [31, 61]. It is possible that a brace also transfers some of the force that would be transmitted from the ankle joint to the lower calf, which can absorb force with much less risk of injury than the ankle joint [61]. Schmidt et al. [14] concluded that approximately 40 % of ankle injury hospitalizations experienced by US Army Airborne School trainees could be avoided if wearing a PAB during training was mandated. Despite evidence that the PAB is a safe, effective and cost effective prophylaxis, wearers of the brace have made negative comments related to heel strap slippage and a lack of comfort, necessitating design modifications, particularly to hold the PAB in place during PLF [31].

6.2 Shoes and Injuries

One of the main equipment items at the interface between the drop zone and the parachutists at landing is their footwear. Despite this, very little research has examined the effects of footwear on military static-line parachuting injuries. Guo et al. [15] noted that protective boots with wide soles, cushioning insoles, and ankle braces, were better than an early generation boot in reducing parachuting injury rates. They speculated that wide soles could stabilize the landing, while cushioning insoles could reduce the landing impact force. However, these findings were for Chinese paratroopers, who are trained to land on their feet, using the half squat technique. The effect of boots on landing injury rates and PLF technique warrants further investigation.

7 Recommendations for Future Research

Military static-line parachuting is a highly tactical and hazardous activity, with a well-documented injury risk. Due to the high impact forces encountered when a parachutist lands, the lower limbs carry the highest risk of injury, particularly when poor landing technique is used. Although numerous studies have investigated factors associated with the occurrence of injuries during training and operations, these investigations have predominantly focused on injury surveillance and statistical analysis of injury data, with few well-controlled studies having examined the injury mechanisms or rationale behind preventative strategies. In fact, there is a paucity of experimental studies that have investigated the biomechanical demands of parachuting, upon which to develop evidence-based guidelines for landing technique or training. Although one study was located which used instrumentation (two triaxial accelerometers plus surface electromyography) to quantify parameters during actual skydives [62], this study was related to evaluating decelerations and muscular responses during parachute opening shock. Niu and Fan [53] used instrumentation to examine the biomechanical effects of terrain stiffness on the half-squat parachute landing (HSPL). Ground reaction forces, lower limb muscle activity and motion were quantified for 16 participants as they performed landing tasks. The participants, however, were university postgraduate students rather than military personnel, who landed on variations of an ethylene-vinyl acetate insole mat rather than a surface that reflected the characteristics of a drop zone, and the peak vertical ground reaction forces generated at landing were relatively low (5.75 ± 2.21 BW for men and 4.78 ± 1.92 for women when landing on a 0 mm thick surface). Similarly, Li et al. [54] quantified foot plantar pressure distributions, using an in-shoe pressure measurement system, displayed by 20 elite male paratroopers (22.6 ± 5 years of age) as they landed on a hard surface versus a soft surface in a half-squat posture. However, the protocol required the participants to jump off a platform that was only 60 cm high and therefore did not realistically replicate the demands of parachuting (see Fig. 6).

Fig. 6 Jumping off a 60 cm box will not replicate the demands associated with trainees learning military static-line parachuting onto a drop zone

Based on this review and the paucity of experimental studies that have investigated the biomechanical demands of parachuting, the following recommendations are made for further research into the PLF technique and training methods:

1. *Parachuting landing technique*: A detailed biomechanical investigation of parachute landing techniques, including a comparison of the forces involved in the various parachuting landing techniques (e.g. PLF vs. the half squat landing technique), is recommended. Such information is required to be able to provide systematic evidence upon which parachuting landing technique guidelines can be developed.
2. *Foot pitch*: Due to the variations in foot posture recommended by military personnel around the world, it is recommended that a detailed biomechanical investigation of the effects of foot pitch on ground reaction force attenuation, as well as implications for its use in minimizing injury risk is conducted. Such investigations should include substantial longitudinal assessments of injury risk. As landing on the ball of the foot with a plantar flexed foot pitch may expose paratroopers to increased pressure in the metatarsal bones, as well as destabilize the ankle joint relative to a flat-footed landing, this should also be addressed by further research. With implications for ankle stability in such investigations, this research should also address the efficacy of ankle braces and alternative footwear to be used in conjunction with altered foot postures [20].
3. *Characteristics of the paratroopers*: As the impact forces generated at ground contact are extremely high, it is recommended that research is performed to determine the body morphology, strength and flexibility characteristics required by potential paratroopers to avoid injury. This would enable appropriate screening methods to be developed by the military to reduce injury risk. For example, it has been suggested that parachute landing casualty rates may be reduced if a bodyweight restriction on paratroopers was established, given the association between increased bodyweight and higher landing forces [6]. However, the interaction between other factors, such as muscularity and strength, would need to be factored into any such study, as the relationship between body weight and military parachuting injuries has been inconsistent [3]. Furthermore, given that static-line parachutes have a load limit, there is a potential for paratroopers with lower body weights to carry greater relative equipment loads than heavier paratroopers. Therefore, the effect of relative equipment and paratrooper loads on injury potential should be investigated.
4. *Training*: A more extensive investigation of current PLF training apparatus and methods should be performed, with the intention of providing trainees with exposure to the most appropriate training activities and progressions to ensure they are adequately prepared for real descents onto the drop zone. Factors such as the specificity, intensity and frequency of ground training, and the progression in exposure to variations in factors such as vertical descent velocities, should be further investigated.

5. *Operational equipment*: Detailed analysis of carrying techniques for operational equipment, such as rucksacks, and how this affects PLF techniques is also recommended [8].
6. *Protective equipment*: Dedicated studies focused on improving landing safety using different protective equipment (such as ankle braces) is recommended and likely to yield high returns [8]. Protective equipment should ultimately be examined in an operational setting, as well as in the laboratory, to provide a comprehensive assessment of the equipment's effectiveness in terms of both performance and injury prevention under realistic conditions [63].
7. *Head injuries*: Recent studies have shown an increase in the incidence of closed head injuries/concussions during military static-line parachuting [5]. The prevention of head injury in parachuting has not been thoroughly explored and therefore research investigating traumatic brain injury prevention and treatment is recommended [8]. For example, although jumpers wear combat helmets during all jump operations, these helmets were designed primarily for ballistic protection and not specifically for protecting the head during ground impacts [5]. It is recommended that helmet design modifications be investigated to improve head protection for military static-line parachuting.

Ultimately, any changes to current practice in landing technique, how it is taught, and whether protective equipment is introduced, should ideally be monitored to determine the effect of these factors on injury rates and paratrooper performance. These studies should be well controlled, prospective rather than retrospective in nature, with the statistical design accounting for the interaction between the variables. It is also strongly advocated that international research conform to a consistent injury definition and method of describing associated details, such as anatomical distribution and type, to allow better between-study comparisons [9]. This will ensure that evidence-based guidelines can be developed, particularly in relation to landing technique and how this is trained, in order to minimize injuries associated with landings during military static-line parachuting in subsequent operations.

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Part II
Load Carriage-Related Injuries

Biomechanics of Load Carriage

Joseph F. Seay

Abstract Loads carried by the Warfighter have increased substantially throughout recorded history, with the typical U.S. ground Soldier carrying external loads averaging 45 kg during recent conflicts. Carrying heavy loads is one potential source of injury that has been researched from a performance and epidemiological perspective. This chapter focuses on the biomechanics of military load carriage, primarily focusing on lower extremity joint stresses and potential overuse injury mechanisms that may be associated with carrying a load. Studies into the biomechanics of load carriage have documented motion-related differences such as increased step rate, decreased stride length, and more trunk lean with increases in pack-borne loads. Ground reaction forces have been found to increase proportionately with loads up to 40 kg. However, there is a paucity of literature on the relationship between load carriage and biomechanical mechanisms of overuse injury. Findings of recent studies will be presented which add mechanistic information to increased stresses on the lower extremity. Efforts to model injury mechanisms require continued biomechanical measurements in humans while carrying occupationally-relevant loads in order to be validated. In addition to lab-based biomechanics data needed to further explore the mechanistic relationship between load magnitude and injury, technologies should be exploited to accurately quantify stresses related to load carriage the field.

1 Overview

Loads carried by the Warfighter have increased substantially throughout recorded history, with the typical U.S. ground Soldier carrying external loads averaging 45 kg during recent conflicts [1]. Research has established that carrying heavy loads

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can substantially increase energy expenditure during the most ideal foot march conditions [2], which can lead to decreased Soldier performance and can adversely affect the ability of the Soldier to accomplish the mission at the end of a tactical movement. While the "rule of thumb" recommendations of limiting carried loads to 33 % of body weight has maintained a presence in the literature [3], the Soldier load has historically been driven more by what is operationally essential than what is physiologically sound [4]. Literature has established that carrying heavy loads can lead to decreases in Soldier performance [5] and the physical symptoms associated with load carriage (soreness, aches and pains, feelings of fatigue) can affect the ability of the Soldier to accomplish the mission [6]. It is also well documented that incidence of disability in the U.S. Army has increased 6-fold over the past 25 years, with the occurrences of musculoskeletal injuries rising faster than other causes of disability [7].

Carrying heavy loads is one potential source of injury that has been researched from a performance and epidemiological perspective [8], and has received media attention [9]. Biomechanics is the study of human movement, and incorporates techniques which analyze movements in terms of the motions that are observed and the forces that generate them. This chapter will focus on the biomechanics of military load carriage, with the primary focus on lower extremity biomechanics and potential overuse injury mechanisms that may be associated with carrying a load. Load carriage will be discussed in terms of biomechanical adaptations to load carriage, to include temporal-spatial, kinematic (motion-related), kinetic (force-related) and energy expenditure adaptations.

2 Biomechanics Tools

In the early years of biomechanics, photographs were used to document body position during various activities. Since that time, technology has advanced and electronic data capture methods have been developed which allow for capture with motion-type cameras (cinefilm, electronic data capture) for more precise motion measurements at higher sampling rates.

The treadmill is a useful tool to simulate marching in the laboratory, as it allows the researcher to more tightly control other things that may affect march performance such as the environment and/or terrain [10–14]. Force-sensing treadmills have been developed which allow for the collection of Ground Reaction Force data while walking; these treadmills have been studied extensively, and researchers agree that from both kinematic and kinetic perspectives, walking mechanics on force-sensing treadmills simulate over ground walking mechanics with reasonable fidelity [15, 16]. In terms of experimental design, treadmills can also allow a researcher to control a desired condition, such as a constant walking speed between loads [14, 17]. Treadmills also allow researchers to simulate perturbations that are difficult to produce in the field, such as prolonged exposure to non-level (i.e., uphill or downhill) terrain [11]. In terms of fatigue, treadmills allow the researcher to maintain conditions for prolonged periods of time whereas the local physical environment may force breaks in activity.

Laboratory-based studies also allow researchers to collect information in addition to motion and force. Two examples of this are expired gases, which allow for determination of energy expenditure (VO_2), and electromyography (EMG), which allows for insight into muscle activity. Both types of information inform on how external load is affecting the carrier beyond changes in motion and/or force parameters, and therefore provide valuable information on how load carriage affects the Soldier.

3 Gait Changes

The ultimate objectives of biomechanical changes observed during load carriage are to manage the load and to conserve energy. The load is managed by maintaining the load + body center of mass (COM) over the base of support, and energy cost is optimized by minimizing the vertical excursion of the pack COM. When the Soldier dons a loaded backpack, the addition of the center of mass of the pack (COM_{PACK}) shifts the center of mass of the soldier-pack system (COM_{SYS}) superiorly and posteriorly relative to the center of mass of the soldier. This COM_{SYS} migration becomes more pronounced as load magnitude increases. The primary goals of all of the biomechanical responses to the addition of a load are to (1) control the load as best as possible and (2) minimize the energy cost of carrying the load.

3.1 Trunk Lean

In the most visible response to adding pack-borne load, studies have documented that the soldier leans forward while walking and carrying a load (Fig. 4.1). One early study documented trunk lean from photographs in the sagittal plane (side view) while Soldiers walked with a backpack on a treadmill while the participants were at the extremes of their trunk leans [18]. Since that time, many studies have documented trunk lean during loaded walking [19–30]. From this work we know that forward trunk lean serves to counterbalance the torque that the load causes at the pelvis and hips [26, 31], and serves to stabilize the COM_{SYS} [19, 20, 23, 27, 32]. This response has been observed when walking with pack-borne loads as light as 8–9 kg [20, 23]. The forward lean during locomotion places stress on the lower back and abdominal musculature, which has been observed as different muscular activation patterns using surface electromyography (EMG) amplitude [22, 24, 33, 34].

Mechanically, this forward lean response serves to center COM_{SYS} within the base of support as much as possible and lower COM_{SYS} closer to the ground [19, 22, 23, 26]. Ideally this adjustment would center COM_{SYS} directly over the hips to optimize mechanical efficiency, as depicted in Fig. 4.1c; however, this does not

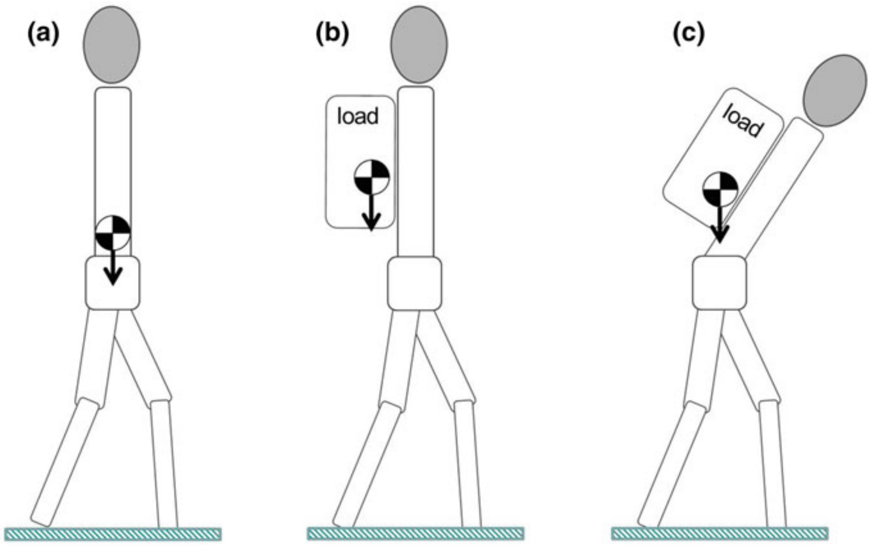

Fig. 4.1 Visual representation of the effect of adding a pack-borne load to an unloaded individual during walking. **a** Unloaded individual walking. **b** Adding mass shifts the pack + soldier system center of mass (COM_{SYS}) posteriorly and superiorly. **c** In response to the added load, the carrier increases trunk lean in an attempt to bring COM_{SYS} over the base of support

happen in reality [22]. The maximum and minimum trunk lean values become more pronounced as the pack load increases; in one study the minimum trunk lean value (defined as maximum angular displacement from vertical) increased from 6° to 15.5° as pack load increased from 6 to 47 kg [22], with other authors reporting similar responses [18, 23, 25, 26, 30, 35]. Trunk lean excursion changes only slightly; significant increases in trunk lean excursion have been reported which were less than 1° between loads [22, 32, 36], which suggests that trunk lean ROM acts to optimize energy expenditure regardless of how heavy the load is.

Load position also plays a role in how much the load carrier leans forward. Some studies have reported changes in trunk lean with different pack systems when loaded to similar weights [29, 35], and others have reported increases in trunk lean when COM location is changed within the same pack system [30]. The primary mechanism behind both of these changes appears to be the magnitude of the pack COM change between conditions. For instance, in some studies, trunk lean did not change with kit configuration or COM_{pack} adjustment, such as when the loads were increased while carriers wore double packs [19] or vest-borne loads [37, 38] when compared to backpacks [19, 22]. In both of these cases, these were expected results which elucidated other aspects of gait which changed with added load.

When the pack loads are heavier, the carrier also involves the neck and head to act as an additional counterweight. This phenomenon has been observed primarily when carrying pack-borne loads above 35 kg [22, 23]. In some instances, the head has been placed farther forward with pack-borne loads as light as 15 % body weight

(BW) [24, 39] while others have not reported similar changes [22, 23]. In further support of these findings, one group of investigators [40] noted an increased variability in head accelerations and force transmission ratios as load increased and suggested a potentially destabilizing effect of load carriage on head trajectory.

One potential effect of increased trunk lean with increased pack-borne load magnitude is to maintain control of the new COM_{SYS} by keeping it as low to the ground as possible [26, 29, 41, 42]. Early studies have demonstrated that while peak and average vertical COM_{SYS} values decrease as pack-borne loads increase, the vertical *excursion* of COM_{SYS} does not change, even at relatively high load magnitudes [22, 26]. This observation has been consistent across different backpack-type military load carriage systems [26]. Studies have reported that maintaining the vertical excursion of COM_{SYS} allows for optimization of the interplay between overall gait mechanics and energy expenditure, where the 'bobbing' effect observed during human walking serves to conserve energy and also allows the lower extremity musculature to work more efficiently [43]. This is consistent with other reports that minimizing ("flattening") the COM_{SYS} vertical excursion is less energy efficient in unloaded walking [44]. As will be discussed later in the chapter, the vertical excursion of COM is ultimately modulated by lower extremity stiffness [41, 42], which is largely controlled by the lower extremity joint moments [41, 45].

3.2 Temporal-Spatial Parameters

The responses described above also accompany changes in spatiotemporal variables that occur as a result of carrying a load. When walking speed is not controlled, walking velocity typically decreases as load increases [21, 46, 47]. However, in order to systematically control for variability in effects of biomechanical responses to carrying heavier loads, walking speed is often kept constant across loads during load carriage studies. When walking speed is held constant, and the load is primarily pack-borne, individuals respond to heavier loads (typically greater than 30 kg) by decreasing step length [20, 46, 48] and increasing step rate [20, 22, 46, 48].

The gait cycle can be divided into two primary phases: the *stance phase*, during which the foot is on the ground and propelling the body forward, and the *swing phase*, during which the foot is off the ground and moving forward to take the next step (Fig. 4.2). Stance time increases with heavier loads [46–50], as well as decreased time spent in the swing phase [20, 22, 47, 51, 52], likely in an effort to stabilize the load (Fig. 4.2).

Within the stance phase in walking there are two periods of double support, where both feet are on the ground (Fig. 4.2). Whether walking speed is controlled or not, the most notable spatiotemporal change in stride parameters in response to carrying a heavy load is an increase in double support time, usually expressed as a percentage of the entire gait cycle [19, 20, 22, 26, 46–48]. This adjustment,

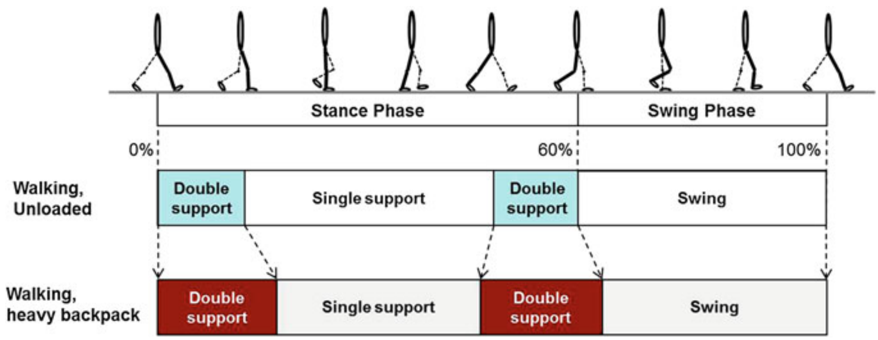

Fig. 4.2 Changes in temporal-spatial variables when walking with a heavy pack-borne load. Increases in double-support time and overall stance time are thought to increase stability

depicted in Fig. 4.2, occurs with a corresponding decrease in percentage of time spent in single support [19, 47] and serves to increase gait stability while carrying heavy loads. The increase in time spent in double support also assists with controlling the vertical excursion of the COM_{SYS} as well, which minimizes energy expenditure as discussed earlier in the chapter.

Studies have been conducted where the aforementioned spatiotemporal changes did not occur. In some cases, the load may have been too light (e.g., 15 % BW or less) [24, 36]. In other cases, load weight was increased via carrying a heavier simulated weapon, which would increase the weight carried by the Soldier, however there would be an anterior shift in COM_{SYS} [23, 25]. Other investigators have added loaded vests to their volunteers [37, 38]; these studies reported an increase in double support time at lighter loads [38] and additionally reported increased stride frequency and decreased stride length when the loads increased above 30 % of BW [37].

4 Kinematics

Lower extremity motion is also affected by increasing the weight of a carried load. Most studies to date have reported on the effects of increasing load in the sagittal plane (side view) of motion for the lower extremity (hip, knee and ankle) throughout the entire gait cycle (Fig. 4.3).

4.1 Hip Angle

The hip angle is generally defined in earlier biomechanics studies on load carriage as the angle measured between the trunk and the thigh [22]; in more recent

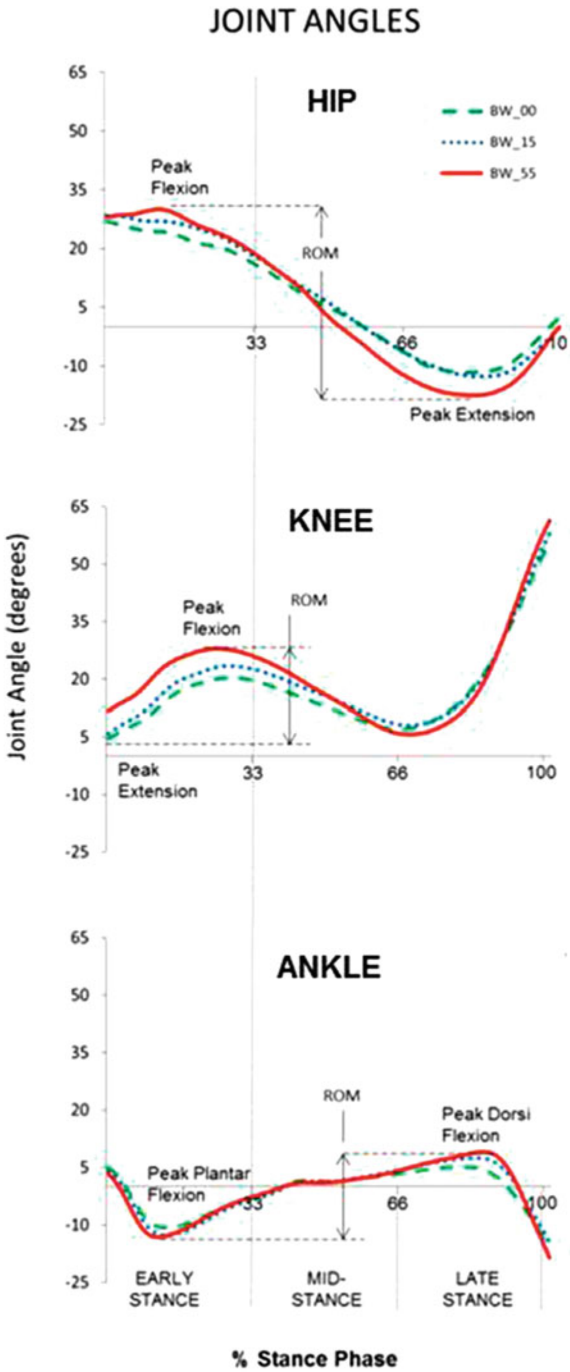

Fig. 4.3 Examplar plots of lower extremity joint angle data for one participant while walking unloaded (BW_00), and with 15 kg (BW_15) and 55 kg loads (BW_55). Peak values analyzed are indicated on their respective plots [17]. Reprinted with permission from [17] (Military Medicine: International Journal of AMSUS)

publications this definition has modified to measure the relationship between the pelvis and the thigh. Several studies have shown that increasing pack-born load increases peak angles at the hip. Studies have reported increases in peak hip flexion angle as loads increase [22, 23, 25, 39, 53]. In one study conducted by Harman and colleagues, the hip flexion angle increased by 11° as the backpack load increased from 6 to 47 kg [22]. This is likely attributable to increased trunk lean in backpack loads during walking [22, 25, 39]. This phenomenon has not been observed in studies where the load is more evenly distributed [19, 38], although an increase in peak hip flexion angle has been observed with increased load when the analysis has been limited to the stance phase only [17, 37], albeit with loads heavier than 30 % BW.

Peak hip extension angle also increases with load [22, 35, 38, 54], which agrees with findings of load carriers spending a larger portion of the gait cycle in stance with increased load (i.e., longer time with their foot on the ground during each stride, expressed as a percentage of stride time) in order to increase stability during walking.

Studies have shown that sagittal hip range of motion (ROM) increases with added load [22, 25, 26, 36, 55–57]. While this ROM tends to be larger for backpack loads (4° from [22]), similar findings with smaller magnitudes of change (around 2° with loads up to 30 kg) have been reported when examining vest-borne loads [17, 37, 38].

Other studies have demonstrated that peak hip angles also increased as a result of changes in walking speed [55] and decreases in cadence (i.e., longer steps) at the same walking speed [14], both while keeping the carried load constant. Using a military cohort, Harman and colleagues [55] demonstrated that there was no interaction between walking speed and backpack loads ranging from 6 to 47 kg; both were affected by load, but hip angle was not affected moreso by a combination of, for example, the fastest speed combined with the heaviest load. However, increasing walking speed from 1.17 to 1.50 m/s did increase hip flexion and extension angles by 2° each, as well as increased hip ROM by about 4°. Using civilians with load carriage experience, Seay and colleagues demonstrated hip ROM increased by 10° when step rate was decreased by 10 % from preferred while walking at a constant velocity of 1.3 m/s. However, hip ROM did not change when step rate was increased by the same amount [14], indicating that hip angle did not change until the step rate perturbation had reached some threshold (likely related to increased step length) substantial enough to warrant an adjustment.

4.2 Knee Angle

Knee angle is defined as the posterior angle between the thigh and the shank, and is typically expressed in degrees of flexion during gait, where lower flexion values indicate that the knee is more extended. However, knee ROM in the sagittal plane over the entire gait cycle (stance phase + swing phase) does not respond consistently with the addition of a load; this response depends on the load configuration, walking speed, and load-carrying experience of the volunteers. Some studies report increases in knee ROM as carried load increases [17, 23, 33], some report no differences [22, 25, 26], and some report significant decreases in knee ROM [26, 29, 56]. Despite conflicting statistical results, the ROM differences between the lightest and heaviest loads carried (range 6–48 kg in Harman et al. 2000, and 12–40 kg in Polcyn et al.) were no greater than 5°. Also worth noting are recent observations which suggest that knee ROM increases do not respond linearly with added load during the stance phase [17]. Specifically, knee ROM increased with the addition of a 15 kg vest-borne load as compared to no load; however, there was no further increase in knee ROM with the addition of loads up to and including 55 kg [17].

Polcyn et al. [26] reported controversial results with regard to knee ROM within the same report. The authors reported that increases in the mass of pack-borne loads may or may not alter sagittal plane knee range of motion. In one study comparing the Modular Lightweight Load-Carrying Equipment (MOLLE) and All-Purpose Lightweight Individual Carrying Equipment (ALICE) packs during walking at 1.3 m/s, using masses that represented fighting loads (mean: 13.1 kg MOLLE, 11.8 kg ALICE), approach loads (26.8 kg MOLLE, 24.1 kg ALICE), and sustainment loads (40.2 kg MOLLE, 38.4 kg ALICE), there were no differences in knee ROM between load configurations (MOLLE or ALICE) or load magnitudes. However, other studies comparing similar loads at different speeds using the same pack systems found decreases in sagittal knee ROM as load increased [26].

Studies have also shown that knee ROM increases in response to increases in speed, and that this response is not affected by the load being carried up to ~47 kg [55]. Another study has shown that knee ROM increases in response to a 15 % decrease in cadence relative to preferred step rate; however, there was no change in knee ROM in response to the same increase in step rate [14].

4.3 Ankle Angle

Ankle angle is defined as the angle formed between the shank (lower leg) and the foot. In the sagittal plane, peak dorsiflexion angle is defined as the minimum angle between the shank and the foot segments (i.e., toes up), and peak plantar flexion angle (toes down, or pointed) is defined as the maximum angle between the same segments.

Results are controversial as to how adding load changes peak ankle angles, or if it does at all. In studies where load was been primarily borne via backpack, adding load did not affect peak plantar or dorsiflexion angle [12, 22, 56]. However, when load was added using more evenly distributed paradigms (i.e., a mix of vest/weapon/pack changes), investigators reported increases in peak ankle angle on the order of 2–3° as load increases for peak dorsiflexion angles, both at touchdown and during mid-stance [23, 25, 38]. Despite these differences, most authors reported no differences in ankle ROM with added load up to 50 kg [12, 22, 23, 46, 55, 56]. The common theme among the few authors that have reported differences in ankle ROM [17, 25, 37] seems to be volunteers are carrying a more evenly distributed load. It is likely that the more evenly distributed load allows for more upright posture, which requires the load carrier to go through greater ankle ROM in order to increase the amount of time spent in double support for increased stability while walking with the load.

5 Kinetics

Since biomechanics is the study of human movement, every action contains two primary components: a motion and a force produced to generate that motion. Up until now motion adaptations have been discussed, without discussing the causes of that motion beyond being a reaction to carrying an external load. For the next section, we will delve into the forces which cause and accompany these motions, and how they relate to stress on the lower extremity and low back, and may lend insight into the etiology of overuse injury.

5.1 Ground Reaction Force (GRF)

The Ground Reaction Force (GRF) is a contact force which represents the interface between the load carrier and the ground during locomotion [58]. The resultant GRF vector is a representation of the combined force of the person plus the load being carried, as are any changes in magnitude, direction, and point of application as the person walks with the load. It can be broken down into its three major components of mediolateral (GRF_{ML}), anterior-posterior (GRF_{AP}) and vertical (GRF_{VERT}) (Fig. 4.4). The GRF_{VERT} during walking is characterized by a bi-modal curve, with one maximum each in early and late stance (termed “loading response” and “push-off,” respectively), and one minimum occurring around mid-stance. The anterior-posterior component of the GRF (GRF_{AP}) is characterized by a conventionally negative peak called the braking peak (GRF_{brak}), followed by a positive peak during the second half of stance phase called the propulsive phase (GRF_{prop}).

As GRF_{VERT} and GRF_{AP} are the components with the most consistent profiles, and are the most affected by changes in load carried [22, 55], this chapter will focus on these two components (Fig. 4.4). Previous researchers have noted increased ground reaction forces with increasing loads [23, 41, 50].

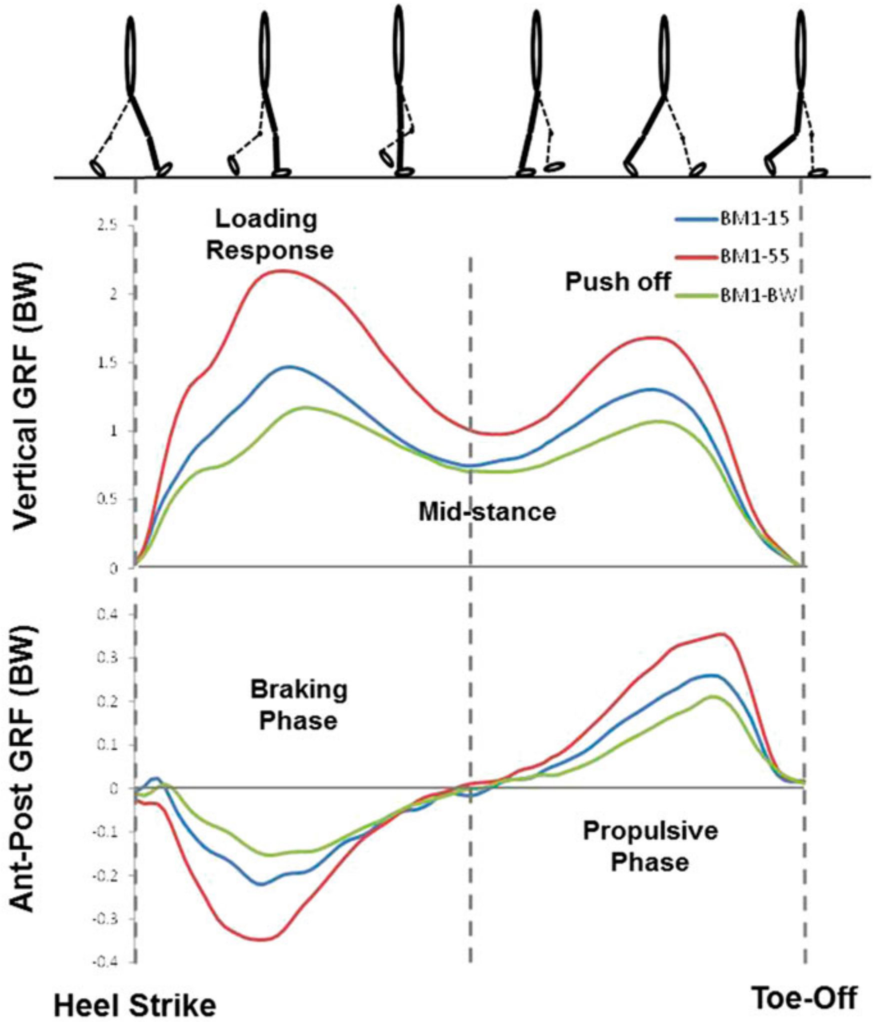

Fig. 4.4 Examples of vertical and anterior-posterior GRF data from a 73.4-kg male while walking with no load (*green line*), with a 15 kg (*blue line*) and a 55-kg vest-borne load (*red line*)

5.1.1 Vertical GRF

The vertical GRF is the largest component of the GRF force vector during walking. During unloaded walking at a preferred walking speed, both peaks register at approximately 1.2 times body weight (i.e., 1.2 BW), and tend to be symmetric if the person is walking at a steady speed. The literature suggests that at a constant walking speed, the two vertical GRF peaks increase in direct proportion with load magnitude ranging up to 50 kg, regardless of load distribution [19, 22, 26, 29, 35, 38, 50, 59, 60]. Further, changes in GRF_{VERT} are not as sensitive to changes in load

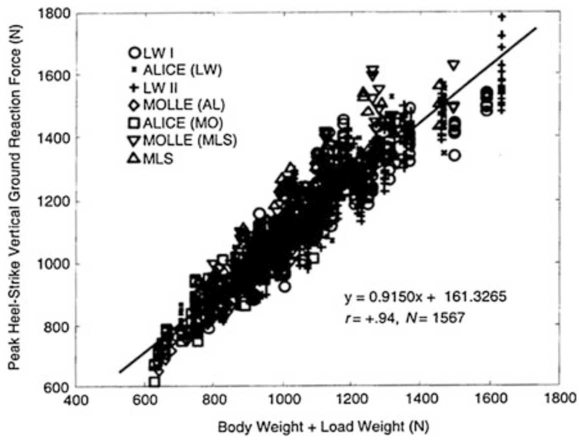

Fig. 4.5 Scatter plot and diagram of simple linear regression for peak vertical ground reaction force during breaking phase (weight acceptance) as a function of body-plus-load weight for several different load carriage systems and configurations (*Source* Polcyn et al. [26], Natick TR02-10). Legend and abbreviations represent different pack-borne load carriage systems worn by volunteers

configuration as other kinetic measures [29, 35, 49, 53]. Increasing walking speed also increased GRF_{VERT} peaks [49, 55]. However, adjusting step rate ±15 % from preferred while walking with a 20 kg load did not change peak GRF_{VERT} in a group of experienced load carriers [14].

In a study which combined data from over 1500 Soldiers wearing many different backpack types, there was a strong correlation between both loading response and push-off vertical GRF peaks and load magnitude, with over 85 % of the variance being explained by the relationship between external load magnitude and peak vertical GRF (r^2 = 0.88–0.9), regardless of which pack system was worn [26] (Fig. 4.5). Other studies suggest that the vertical GRF increases approximately 10 N for every 1 kg of external load added [50]. Taken together, these data and other data from the literature suggest that the vertical GRF is affected primarily by the static effects of the load being added, and not changes in attributes associated with the dynamics of walking, such as acceleration [50].

5.1.2 Anterior/Posterior GRF

During unloaded walking at a preferred walking speed, both GRF_{AP} peaks register at approximately ±0.15 BW (Fig. 4.4). These peaks tend to be symmetric, with the force value crossing zero at almost exactly half way through the support phase if the person is walking at a steady speed. In addition to being affected by increased load magnitude [19, 26, 29, 46, 49, 50, 53, 59, 61, 62], the GRF_{AP} is more affected than GRF_{VERT} by changes in cadence [14], and occasionally is affected by load distribution at a given velocity [49]. Concern has also been expressed in the literature

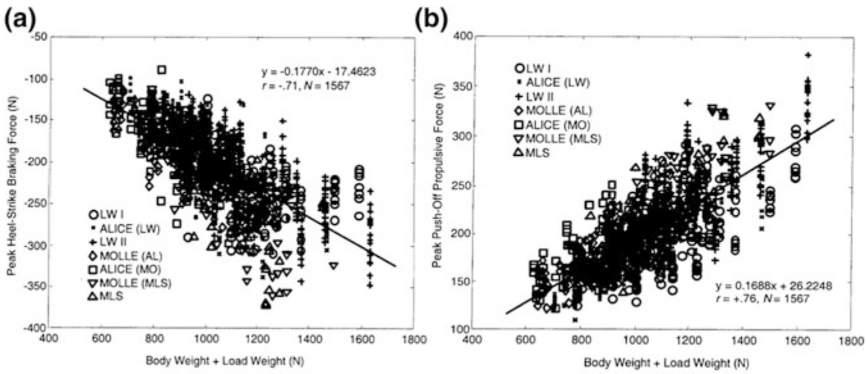

Fig. 4.6 Scatter plot and diagram of simple linear regression for peak antero-posterior ground reaction force during (**a**) breaking phase and (**b**) propulsive phase as a function of body-plus-load weight for several different load carriage systems and configurations (*Source* Polcyn et al. [26], Natick TR02-10). Legend and abbreviations represent different pack-borne load carriage systems worn by volunteers

that increased friction forces at the foot-boot interface, most directly expressed using GRF_{AP}, may increase the risk of blisters in load carriers [5, 33, 49]. Results from another study suggest that shifting the COM posteriorly by carrying load solely in a backpack significantly reduced the force produced at toe-off, whilst also decreasing stance time at the heavier loads. Conversely, distributing load evenly on the trunk significantly decreased the maximum braking force by 10 % [49].

Polcyn and colleagues reported a strong correlation between both braking and propulsive GRF peaks and load magnitude across several studies (Fig. 4.6) [26]. While this correlation was still clearly linear, only 50–60 % of the variance (r^2 = 0.50–0.57) was explained by the relationship between peak antero-posterior GRF and BW + load carried [26], leaving a larger portion of the relationship unexplained by load magnitude as compared to GRF_{VERT}.

5.2 Impulses and Joint Reaction Forces

The variables of joint reaction forces and GRF impulses have also been analyzed to examine the relationships between added load and lower extremity mechanical stress. The GRF impulses are defined as area under the component curve, and allow the investigator to track the changes in mechanical energy within the desired curve, and have tended to mimic results of their component GRFs with respect to increases in the magnitude of the load carried [22, 59].

Lower extremity joint reaction forces, defined as the net force acting across the joint, also tend to closely mimic GRFs [22, 26, 54, 55]. This is not surprising since the component GRF is the largest input component into the calculation of the lower extremity and low back joint reaction forces, especially during walking [58].

In addition, Goh and colleagues reported that lumbosacral (L5-S1) joint forces increased with load in a similar pattern to lower extremity joints [21].

5.3 Load Rate

The load rate of the GRF is another variable that has been used to examine differences due to load carriage experimental paradigms. Loading rate in general is calculated as the increase in force over time between foot contact and the loading response peak in the vertical ground reaction force curve; more recent papers have calculated loading rate as the middle 80 % of the curve between foot contact and the first peak of interest. This parameter has been able to discriminate between those with and without a history of lower extremity injuries in runners [63] and during walking in individuals with knee osteoarthritis [64]. Load Rate is valuable clinically in that it conveys information on how rapidly a load is being applied to a joint in question, which is a suspected component of lower extremity musculoskeletal injury. Some studies have reported no differences between load rate for loads ranging from Body Mass equivalent (BM) to an additional 20 % BM load [19, 60]. However, load rate has significantly increased when loads have increased over 40 % of BM [19, 59, 62], suggesting a potential protective effect may be mitigated when carried loads exceed approximately 36 kg [62].

Holt and colleagues [41] reported similar findings when untrained men walked unloaded (body mass, BM) and carried pack-borne loads equivalent to 40 % of BM (BM + 40 %). Tilbury-Davis and Hooper [59] had a male military cohort walk at a self-selected pace while unloaded (clothes and boots only, BM) and with pack-borne loads of 20 and 40 kg. They found that the loading response GRF_{VERT} peak remained constant between conditions (72.2 % BM for BM only, 63.4 % BM for BM + 20 kg, and 71.3 % BM for BM + 40 kg), suggesting that other adaptations were made in an attempt to keep this constant in order to mitigate stresses on the lower extremity. These authors also reported no difference in vertical loading rate between walking unloaded and carrying 20 kg, but significant increases between the 40 kg load and both of the lighter loads. The vertical loading rate data suggest that a threshold exists for discontinuity on loading and unloading rates between 20 and 40 kg (which translated to roughly 30 and 60 % of average BM for their cohort). The authors suggested that the lack of change in loading rates with increasing load was a protective response to the load, supporting earlier findings [65] which suggested gait adjustments occurred at around 34 kg in subjects who were inexperienced load carriers. It is difficult to determine the extent to which this response was protective, however, as it may have been driven by changes in walking speeds, which were not controlled between loads. Adjusting to the load by changing walking speed could have allowed the participants to adjust their loading rates, which could still be a protective strategy, just not the one the authors postulated. While the ability to adjust walking speed adds operational realism to the study design, as Warfighters are likely to choose their pace in the field, different gait speeds confound the interpretation of the effect of load on these participants.

5.4 Joint Moments

Another way to measure the response to load carriage is by examining joint torques, sometimes referred to as moments-of-force (hereafter referred to as joint moments), defined as the angular resistance to (or manifestation of) force. Using the knee joint as an example, we know that the force generated by the body's mass plus any additional carried load will generate a force called the ground reaction force (discussed earlier). The angular expression of how the body responds to this force called an external moment (M_{ext} in Fig. 4.7); this moment will act to bend (flex) the knee if unopposed. In order to counteract this external flexion moment (and remain standing), the knee musculature must generate a net internal moment (M_{ext}) which overcomes the load imparted by body mass plus the carried load and extends the

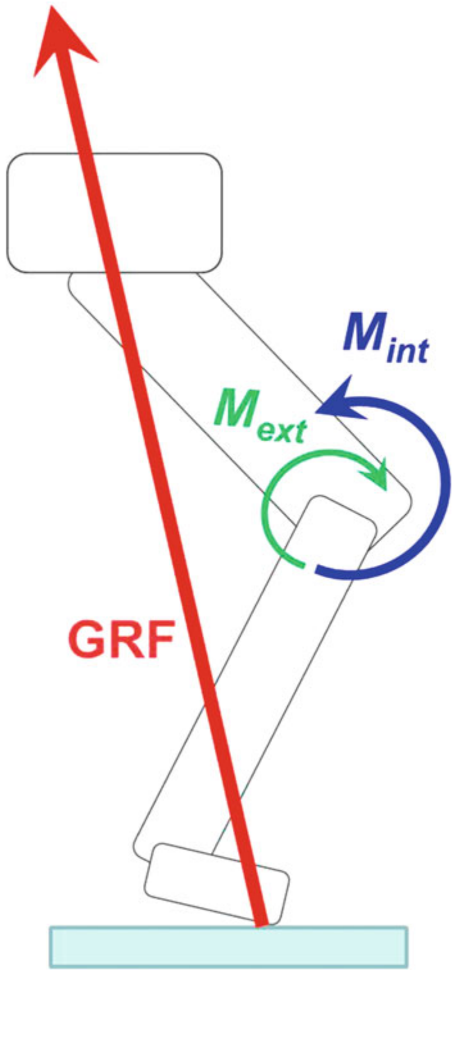

Fig. 4.7 Theoretical representation of M_{ext} (*light green arrow*) acting on the knee, and the internal moment (M_{int}, *dark blue arrow*) which is generated to counteract it. *GRF* ground reaction force

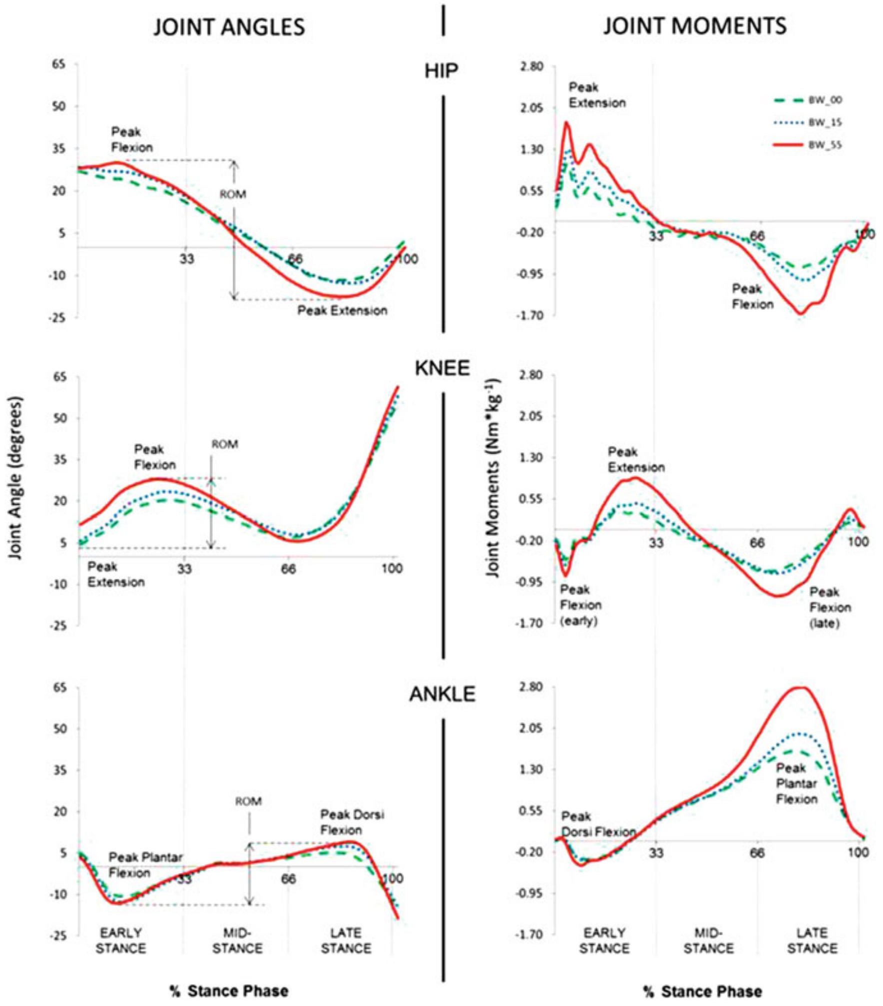

Fig. 4.8 Exemplar plots of lower extremity sagittal plane gait joint angles and joint moments for one study participant while walking unloaded (BW_00), and with 15 kg (BW_15) and 55 kg loads (BW_55). Peak values analyzed are indicated on their respective plots [17]. Reprinted with permission from [17] (Military Medicine: International Journal of AMSUS)

knee, allowing the load carrier to continue walking (Fig. 4.7). Even if the knee is still flexing, the extension moment allows the load carrier to control the descent of the load during early stance. Joint moments are critical in evaluating the potential for overuse injuries during walking as the values of joint moments represent the effort exerted by the muscles to counteract and control the external load during the first 30 % of the stance phase (GRF_{VERT} loading response or GRF_{AP} braking phase) and to propel the loaded body forward during the final 30 % of the stance phase (late stance or propulsive phase).

While there is a paucity of joint moment data available reporting how joint moments respond to added load, this body of literature is growing. The literature shows that different lower extremity joints respond differently to added load when compared to unloaded walking (Fig. 4.8).

Hip. Peak hip extension moment occurs during early stance, counters the descent of the pack mass during early stance, and increases as load increases [12, 17, 22, 38, 66]. While hip flexion moment during late stance appears not to be affected by lighter loads [17, 38], this moment has been reported to increase as loads get heavier [17, 22].

Knee. Knee extension moment during early stance, which is also associated with controlling the descent of the load after foot contact, increases as load increases [12, 17, 22, 38, 66], even with loads as light as 10 % BM [38]. Quesada et al. [12] reported that the knee extension moments that resulted from increasing a pack-borne load from body mass (BM) to BM + 15 and BM + 30 % were proportionately larger than the hip extension or ankle plantar flexion moments at the beginning of a 40-min march in college-aged Reserve Officer's Training Corps (ROTC) cadets. In another study that compared lower extremity biomechanical responses to carried loads of 15 and 55 kg [17], the knee extension was the most sensitive lower extremity parameter for modulating all loads. The responses of the peak knee extensor moment from both of these studies may lend insight as to why those who carry heavier loads in the military are more susceptible to knee injuries [67, 68]. Since this increase in knee moment occurred with minimal changes in knee ROM in both of these studies, there was increased stress placed on the knee musculature to counteract the heavier load [17].

Ankle. Studies have shown that at the ankle, peak dorsiflexion moment sometimes increases with symmetrical load [14], while when worn in a pack it does not [22, 38, 66]. This is again related to increasing double stance time with increased load in order to stabilize the load while walking; trunk lean with a backpack may inherently facilitate increased double support, while the carrier may need to increase dorsiflexion moment as an alternate mechanism to increase time in double support. Peak plantar flexion moment, however, which occurs during late stance push-off, increases as load increases [12], regardless of how the load is distributed or how small the increments may be (Silder et al. loads increased up to 30 % BM).

5.5 Stiffness

With increases in load mass carried eliciting proportional increases in vertical GRF, the lower extremity must somehow counteract the adverse effects of the increase in loads. One way to do this is by adjusting walking speed, but this does not address the entire problem. Even at slower walking speeds, the lower extremity must mitigate the increase in GRFs that result from the added load mass. Since knee

ROM does not change proportionately with the increase in load, the eccentric muscular demand must increase in the lower extremity to slow the descent of the center of mass at a similar rate, regardless of load. To this end, stiffness is a useful measure for quantifying the relationship between joint motions and the stress placed on the joint.

Quesada and colleagues interpreted the initial increase in knee extension moment with added load as an attempt to dissipate shock, a mechanism that would potentially mitigate overuse injury [12]. Another group of researchers [41] expanded on the concepts above by examining the effects of load and speed on whole body and lower extremity stiffness. Both stiffness measures are indices of how the system responds to (weight acceptance) and mitigates the effect of the loads placed upon it (loading rates, acceleration). *Whole body stiffness* was defined as the center of mass maximum vertical excursion divided by the mass of the participant (BW + pack mass), and *joint stiffness* was an angular measure defined as the peak slope of the joint angle-joint moment plot. Eleven civilian participants walked at six different speeds (range 0.6–1.6 m/s) without (BM only) and with a pack mass equal to 40 % BM (total mass = BM + 40 %). The authors reported increases in knee ROM and vertical excursion of whole-body center of mass as the result of increasing speed, but NOT as a function of increasing load at any given speed. Stiffness measures, however, increased as a function of load and speed. The authors postulated that the increased stiffness served to maintain a relatively constant whole-body COM in order to limit potential increases in metabolic energy expenditure associated with increased COM excursion. However, increased knee stiffness without an increase in knee angle (ROM) indicates a larger knee extensor moment most likely created by increased eccentric contraction during the weight acceptance phase of stance (i.e., first half of stance). One study has demonstrated that the larger knee extensor moment will likely degrade over time [12], and this would hold especially true for the inexperienced load carriers that were used in the Holt et al. [41] study.

6 Electromyography

Muscle activation is sometimes measured in conjunction with motion or force load carriage data via electromyography (EMG) in order to supplement information on how the muscles of the shoulder, lower extremity or low back are responding to different experimental conditions. One common area for collection of EMG data is the lower back and abdominals, with the lower back being primarily represented by surface electrode collection of the erector spinae (ESPIN) musculature [69]. Studies have reported either no increase in ESPIN activity [22] or a decrease in ESPIN activity with loads up to approximately 33 kg [24, 34]. However, one study reported an increase in ESPIN activity with the addition of a 47-kg backpack [22]. Authors attributed the lack of increased ESPIN activity at lower loads due to the slight forward trunk lean associated with these loads being more mechanically advantageous for this muscle group, perhaps placing stresses on other muscle groups, such as rectus abdominis and/or external obliques, as reported by Devroey [24].

For the purposes of this chapter, lower extremity muscle activity will be divided into the following functional groups: QUAD (Quadriceps, or knee extensors: consisting of vastus medialis, vastus intermedius, vastus lateralis, rectus femoris), HAMS (Hamstrings, or primary knee flexors: consists of biceps femoris (long and short heads), semimembranosus, semi tendonosus), GAS (Gastrocnemius-soleus complex: consists of medial and lateral gastroc heads, soleus) and TIBA (tibialis anterior). With increased load, activation of the knee extensors (quadriceps) increased for both backpack [22, 51, 70] and vest-borne loads [38], likely to control the descent of the load during early stance weight acceptance. The other muscle group which increased activation with increasing load was GAS [22, 38, 70], which coincides with increased ankle plantar flexion moment and likely occurred to facilitate push-off later in stance.

7 Energy Expenditure

Most existing recommendations on safe loads to carry are based on energy expenditure or the metabolic cost associated with carrying a load. The Soldier needs to carry the appropriate supplies to complete the movement and execute the mission. To this end, several investigators have researched the ability to predict the energy cost of carrying a load (e.g., Pandolf [71], Santee [72]). The main goal for this section of the chapter is to demonstrate that there are several factors that increase the energy expenditure associated with load carriage, so any biomechanical changes noted in this chapter from laboratory-based studies may be exacerbated or occur earlier depending on the environmental conditions under which the load is being carried in the field.

Research on the energy expenditure of loads carriage shows that energy cost increases in a systematic manner as the load carried increases, and also with increases in walking velocity, grade or a combination of these factors [10, 32, 73–78]. Terrain also effects load carriage: one compilation of studies has shown that for the same load weight, walking through swamp or on sand essentially doubles the energy cost of walking on a paved road, and walking in snow without snowshoes can increase this cost by 4–6 times [79–82].

Physiological research has also demonstrated that increasing load magnitude can substantially increase Soldier energy expenditure during typical foot march conditions. Epstein and colleagues [83] performed a study in which expired metabolic gasses were collected while Soldiers walked for 120 min at 1.25 m/s (2.8 mph) at a 5 % grade on a treadmill carrying pack-borne loads of 25 and 40 kg. Patton and colleagues [13] also observed that metabolic effects over time were different between lighter and heavier loads during a simulated 12-km road march at 1.34 m/s (3.0 mph) on a treadmill while carrying loads of 5.2, 31.5, and 49.1 kg. For both of these studies, physiological variables (VO_2, Heart Rate) below 30 kg reached a steady state within the first 10 min of the march, indicating that the Soldiers could walk with this load for a long time and have the energy to accomplish the mission at the end of the march. However, for loads greater than 30 kg, these physiological variables increased throughout the duration of the march, and ultimately indicated that it would be substantially more difficult for the Soldier to accomplish any tasks at the end of the march.

7.1 Load Placement

Energy Cost The addition of an external load will increase energy cost, whether that load is added via armored vest [38, 84] or placed in a backpack [3, 13, 83, 85, 86]. Some studies have reported that vertical load placement may redistribute the load in a manner that may be more comfortable; specifically, some prefer the COM of the load in the pack to be high in the pack and close to the back, especially while walking on level terrain [85, 87]. Loads placed higher in the pack may lead to

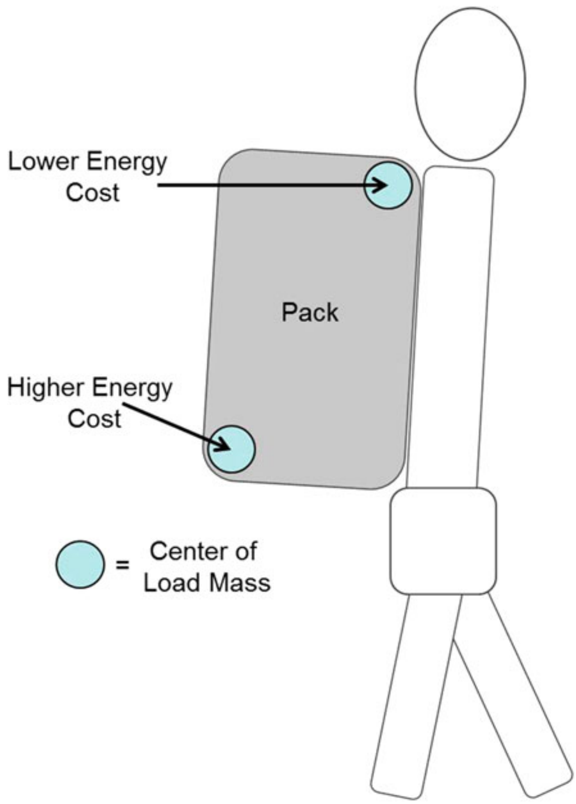

Fig. 4.9 Effects of load distribution within the pack on energy cost. A high and more anterior load results in a lower energy cost than a low and more posterior load. *Data source* Obusek et al. [88]

perceptions of reduced stability over unsteady terrain, due to greater rotational inertia associated with the higher COM placement [34]. However, the location of the load within a pack can affect the energy expenditure during load carriage, as well as the mechanics of how the load is carried (discussed earlier in the chapter). Different load COM location within the backpack has shown equivocal results. One study reported that loads packed higher and closer to the body can reduce energy cost by approximately 25 % as compared to a load that is placed low in the pack and away from the body while carrying a 35 kg backpack (Fig. 4.9) [88]. These findings were reinforced by a later study which found that higher placement of the load within a pack (approximate level of T1-6 vertebrae) was associated with significantly lower energy cost than a lower load placement in the region of the lumbar vertebrae [85]. No such differences were reported by a more recent study [24], although the differences in results may have occurred because investigators used a lighter load (15 % of body weight) and a shorter vertical distance between the highest and lowest load placement.

Sometimes Soldiers will carry loads in uniform pockets on the thighs or lower leg in an attempt to distribute the load or for a tactical advantage (i.e., quicker access to ammunition or ordinance). However, there is an increased energy cost associated with a strategy of distributing the load to the lower extremities (cargo pockets, etc.). Loads carried on the feet or ankles results in 7–10 % increased energy expenditure as compared to loads carried in the pack [89–91], with loads on the thigh increasing energy cost per kilogram carried by about 4 % [92, 93].

Loads Acting on the Spine In addition to energy expenditure, load placement also affects the forces realized on the lumbar spine. Consistent with information presented earlier in this chapter on the lower extremity, increasing the weight of the backpack increases the vertical load on the spine [21, 94, 95]. In a study where spinal loads were measured in-vivo, Rohlmann et al. [95] confirmed that a 9 kg load carried in a backpack generated less force on the lumbar spine than carrying the same load in front of the body with both hands.

When the load is borne in the pack, however, the stresses at the lumbar spine do not always increase. The addition of a backpack creates an extensor torque on the hips which actually reduces the stress on the extensor musculature if the carried load is less than 10 % BW [96]. Part of the increased stress on the spine is due to increased muscular activity, as measured by EMG. Studies utilizing EMG on the erector spinae musculature have reported that lighter carried loads elicit *decreases* in EMG activity of the erector spinae [24, 34, 55]. In fact, studies using military-relevant loads have reported that erector spinae activity does not increase beyond that of loaded walking until the pack-borne loads exceed 30 kg [33, 55]. It should be noted that these studies did not include measurements of abdominal musculature (i.e., flexors of the low back); this is an aspect of load carriage that should be further examined to complete the picture on the stresses at the low back.

8 Effects of Fatigue

While fatigue can be a nebulous term, in general it can be defined as having people exert themselves until they drop, or at least until they can no longer put forth some pre-determined required effort [97]. In the ergonomics literature, fatigue is more loosely defined as "the reduction in performance ability caused by a period of excessive activity followed by inadequate recovery time," where the decreases in performance can be either mental or physical. Muscle fatigue is accompanied by a buildup of lactic acid in the working muscle [98]. Using a combination of the definitions above for this Chapter, fatigue is most typically defined during load carriage as progressive degradation of performance or mechanics in this case due to the duration of an event or activity.

In terms of fatigue related to load carriage, several different paradigms have been used. Some have involved marching for periods of time or fixed distances in an attempt to closely simulate operations and/or training [5, 6, 83]. Some investigators have tested the effects of prolonged load carriage on walking mechanics under steady-state conditions [11–13]. In one such study, Soldiers marched for 2 h carrying a 25 kg pack on both level and downhill grades while walking at 1.8 m/s [11]. After this march, the authors reported significant decreases in isometric and isokinetic knee extensor and trunk flexor torque as measured on an isokinetic dynamometer. Neuromuscular recovery took 2–3 days, depending on the region (lower extremity or trunk). However, there were no difference in neuromuscular decrements between level and downhill walking, which was surprising since there was more eccentric knee extensor (quadriceps) work done while walking downhill [11]. On the other hand, perhaps the 25-kg pack was not enough relative weight (approximately 31 % of the 79.4 kg cohort average body mass) to elicit true muscle 'damage,' which may have caused a differential in the strength deficits and recovery time. Either way, the authors showed that neuromuscular recovery takes more than 24 h, so considerations should be made in the field if possible.

Another investigator required 12 healthy male army recruits to walk at 6 km/h (1.67 m/s) on a treadmill for 40 min while carrying no additional load (BM), and pack-borne loads of 15 and 30 % BM [12]. Increases were observed at hip, knee, and ankle moments in response to increasing load at the beginning of the march. However, the knee extensor moment during early stance was the only moment which changed at the end of the march; while knee extensor moment increased with both loads, post-march data showed a decrease in knee moment values, indicating that these initial responses to the added load were not maintained. These changes occurred despite physiological measures (VO_2, Heart Rate) remaining within recommended limits for 8-h shift work, especially for a young, fit cohort [98, 99]. The authors posited that the increase in knee extensor moment during weight acceptance served a protective effect, and that protective effect was not maintained over the duration of a 40-min march, thereby potentially increasing the participants' risk of injury [12].

Other investigators have more specifically targeted musculoskeletal fatigue, which has been defined as a decline of muscle force output relative to a maximum voluntary effort. One recent study utilized a task in which participants participated in a metered step test (modified Queens College step test [100] while wearing a 16 kg backpack) until participants could not maintain a 24 cycle per minute pace (1 cycle = step up/step down). Once they were no longer able to maintain cadence, participants then performed 20 standing heel raises on the edge of a box, and then performed a maximal effort vertical jump. This sequence was repeated until the maximal vertical jump height was less than 80 % of their baseline (fully rested) maximal vertical jump [62]. Walking mechanics were compared with and without a 25 kg pack before and after this fatiguing protocol. Wang and colleagues reported changes during early stance (weight acceptance) in the form of decreases in ankle dorsiflexion angle at foot contact and dorsiflexor moment, as well as increases in hip extensor moments and powers [66]. They also reported increases in both peak GRFs and loading rate in the vertical and anterior-posterior directions during weight acceptance [62], both of which have been associated with overuse injury in the running literature [63]. Interestingly, there was no statistical interaction between load and fatigue, indicating that load did not exacerbate the effects of musculoskeletal fatigue, which one would have thought more likely in an inexperienced cohort.

9 Shoulder Straps and Hip Belts

Carrying a backpack is one of the most efficient and preferred method of transporting a carried load in the military [82]. Packs carried using shoulder straps alone can generate high pressures at the shoulders [101], which could increase risk of rucksack paralysis (also known as rucksack palsy) [102].

Hip belts and internal framed packs are used to transfer some the weight from the shoulder straps to the hips [94, 103, 104]. The shoulder-hips weight distribution has been reported at 70 % shoulders and 30 % hips for loads ranging between 12 and 40 kg [94]. Hip belts also aid in walking mechanics by assisting the shoulders with the typical rotation associated with gait: specifically, the addition of a hip belt assists in slowing the maximum rotation of the pack, and has been shown to start counter-rotation toward the next step much earlier than without the hip belt [103], which likely increases walking efficiency, or at least decreases maximum pressure experienced by the shoulders during the loaded gait cycle.

10 Rifle Carriage

Rifles are almost always carried in dismounted military operations and training. Carrying a rifle in both hands (i.e., at a “ready” position) while walking has the effect of adding weight, restricting arm swing and moving the COM more anterior with respect to the body. As such, carrying a rifle while walking can increase GRFs during early stance in both the vertical (GRF_{VERT}) and anterior-posterior (GRF_{brak}) directions, as well as increase the medio-lateral impulse [50, 105]. These effects are primarily due to the effects of decreased arm swing rather than the absolute weight of the rifle [49]. One study reported on the effects of carrying a rifle in both hands on movement patterns coordination between the pelvis and the trunk in the absence of additional load. The authors reported that running with a rifle in both hands altered pelvis-trunk coordination patterns [106], and generated similar coordination patterns to runners with a history of low back pain [107]. When conducting research, the general consensus is that the rifle should be consistently carried or not carried so as to minimize any potential confounding effects on biomechanics [23, 25, 56].

11 Men Versus Women

Over the past few decades, the percentage of females in the U.S. Army has increased by fifty percent from approximately 10 % [108] to 15 % [84]. Many studies show the incidence rate of training related injuries are up to 2.5 times greater for females [109–111], particularly with regards to musculoskeletal injuries, which can occur at even greater rates relative to male counterparts [112, 113]. Because females carry the same packs but tend to be smaller and lighter than their male counterparts, their bodies are exposed to higher relative loads. This increased relative load suggests that load carriage could be a factor in regards to increased injury incidence. This section will review and compare responses of male versus female load carriers, and discuss implications for lower extremity injury.

Some studies on load carriage have documented performance differences between men and women in terms of slower road march times for a given load and distance [82], and sprint times after a road march [114]. Data also suggest that some differences in load carriage may be attributable to the fact that equipment has been primarily designed for men, to the extent that women are more uncomfortable in the pack systems than men [115, 116].

Load carriage experiments which compare the biomechanics of men and women carrying the same absolute loads have reported biomechanical adjustments which support increased stress placed on women. The adjustments reported have been mainly temporal-spatial in nature, where women tended to increase stride frequency with increasing load significantly more than men [20, 22, 29]. In one study, stride length and swing time decreased much more dramatically with load for the females, while their time in double support increased and actually surpassed that of the males at the highest load [20]. One study did not notice significant increases in stride frequency or decreases in stride lengths as the previous two [116], perhaps attributable to the fact that these investigators padded the belt on the pack system worn in that study in order to ensure a better fit. However, the authors did report that the women in their study did hyperextend the neck and protracted their shoulders in order to maintain a more upright posture, which were different mechanics than previously observed in men [116]. Bhambhani and colleagues [117] report that while load carriage mechanics were similar when men and women carried loads up to 20 kg, women demonstrated increased heart rate and metabolic responses. These results suggest that the kinematic differences between males and females may exist even when potential anthropometric factors (women tend to be shorter and weigh less) are taken into account [20, 82, 116].

However, another study involving biomechanical and physiological comparisons revealed no differences between men (n = 17) and women (n = 13) while carrying vest-borne loads up to 30 % BW. When participants walked at their preferred speeds, the authors observed no sex-based differences for biomechanical variables (temporal spatial, joint angles, joint moments) or muscle activation patterns [38]. Further, these investigators did not observe any interactions between load and sex, which would have suggested that one sex would have increased stress or effort as the carried load increased in magnitude [38]. When analyzing differences in energy expenditure between men and women, the authors found increased metabolic cost for men versus women relative to baseline VO_2 measures; however, when normalizing these values to fat-free mass, there were no differences between men and women at any load [38]. These results were supported by those of other investigators who reported that energetic cost was similar between men and women when normalized to fat free mass [118], and supported previously reported relationships between net metabolic cost of walking and body composition [78, 119].

The findings of Tilbury-Davis and Hooper [59] discussed earlier in the chapter suggest that a threshold exists for discontinuity on loading and unloading rates between 20 and 40 kg (which translated to roughly 30 and 60 % of average BW for their sample). This was supported by earlier findings [65], which suggested gait adjustments occurred at around 34 kg in subjects who were inexperienced load carriers. The response in the Simpson et al. [46] study occurred at a lighter absolute weight ($\sim$20 kg) for experienced female hikers, as compared to the participants in Tilbury-Davis and Hooper [59] and Pierrynowski et al. [65], indicating there may be a difference between the sexes. While this difference is due, at least in part, to women being lighter on average, the pack weights carried by Soldiers in the field are based on equipment needs, so adjusting external load to body mass is not a logistical reality.

12 Imaging Studies

Another aspect of biomechanical research that should not be overlooked is imaging technology which allows investigators to record and track the effects of load carriage on participants in vivo. Imaging data have reported on the short-term impact of load carriage in military cohorts on the lumbar spine [120–122] and knee [120, 123, 124].

Imaging studies of the lumbar spine have found that the spine straightens slightly while standing with lighter loads up to 16 kg; however the individuals in that cohort demonstrated varying degrees of curvature [121]. A more recent publication reported a more dramatic straightening of the spine during both walking and standing with a 50-kg load [122]; however, these changes had resolved to baseline values after 45 min of side-lying recovery. Another study examined MRI data for the lumbar spine before and after a 14-week Navy Seal training, and reported no conclusive changes after undergoing the training [120].

The same study involving Navy Seal training reported on the knee MRI data. They reported that the knees of their cohort were affected by the training, with post-training knee images showing cases of patellar swelling and bone edema, while the spinal images remained essentially unchanged [120]. These findings are consistent with findings from the running literature, which have suggested while activity such as running will elicit changes in MRI scans (such as edema near the cartilage) immediately after the running activity [123, 124], only individuals who had pre-existing knee injuries showed residual decrements 6–8 weeks after a marathon [123].

13 Dynamical Systems

Up to this point in the chapter we have reviewed results based on traditional biomechanical assessments of gait using discrete measures (e.g., peak excursions, forces, etc.). Continuous methods have emerged within recent decades which allow for quantifying movement coordination throughout the gait cycle and can provide spatial–temporal information about the interaction between the two segments during a movement cycle [125]. From a dynamical systems perspective, variability is a hallmark feature of biological systems that express flexibility and adaptability. Coordination variability has proven useful in differentiating between healthy and pathological coordination, such as lower extremity couplings between individuals with and without knee pain [126] and iliotibial band syndrome [127], and pelvis–trunk couplings between individuals with and without low back pain during walking [128] and running [129]. As such, dynamical systems analyses may lend additional insight into different movement patterns which may be associated with pathology during load carriage [36]. This section will provide an overview of studies that have utilized dynamical systems methods.

Non-pathological walking is characterized with anti-phase rotation patterns between the pelvis and trunk in response to increasing speed, such that left shoulder and right pelvis rotate forward when stepping with the right foot (this is done, at least in part, to conserve angular momentum during walking). One group of authors reported that individuals with non-specific low back pain lacked the ability to adopt this pattern due to their low back pain [128]. LaFiandra observed similar responses between the pelvis and trunk with the addition of even a relatively light pack and an increase in walking speed, which could have been due to an increase in the inertia of the trunk + load system as compared to trunk alone [130]. As previously discussed, Seay and colleagues also reported decreased pelvis-trunk coordination and coordination variability in response to running with a rifle in both hands [106], likely due to a restriction of arm swing. While the long-term effects of these adaptations are not known, both adaptations were likely protective, yet coordination patterns were similar to individuals with low back pain [128]. One interpretation of decreased relative phase variability is that it may increase the likelihood of overuse injury due to the fact that repeatable actions are occurring anatomically within a very narrow range of potential coordination patterns [129, 131].

Others have used dynamical systems measures to assess gait stability [132–134], with the idea that a less stable gait pattern may require more energy to maintain and thus increase risk of injury. Whether using the length of the relative phase path during loaded walking [134], Pioncare' analysis [132], or the uncontrolled manifold hypothesis [133], no differences were found in gait stability while carrying loads up to 30 % of BW.

Recently a group of investigators has applied dynamical systems measures to assess the effects of load during perception-action task coupling, such as how loads on the body are affecting Soldier performance on target acquisition simulations. In one such study, coordination between the head and trunk was assessed as a potential underlying mechanism for time to establish upright posture after landing under different loaded conditions after stepping down from a 61 cm height [135]. Collectively, results demonstrated that addition of a load and distribution of that load increased time to establish upright posture, forced the volunteers to maintain a more downward head angle and contributed to a decreased field of regard for acquiring the target. Results also confirmed that head-trunk dynamics were key underlying relationships in time to establish upright posture, and reported that these underlying dynamics are affected by load distribution as well as load magnitude when establishing a ready posture for a Soldier-relevant task, such as target acquisition after jumping and landing from a height [135]. In another study, the same group related segmental coordination and adaptability under different load to degraded threat discrimination. In one aspect of that study, the authors reported that trunk-head and head-weapon coordination were also underlying mechanisms that affected dynamic target acquisition during a simulated threat discrimination and marksmanship task [136].

14 Summary and Conclusions

When the load is carried in a pack, the carrier leans forward commiserate with load magnitude and distribution in order to keep the COM over the base of support. When carrying lighter loads, minimal forward lean appears to minimize low back muscle activity, however both trunk lean and muscle activity increase as the loads get heavier. Whether loads are carried in a pack or in a vest, individuals carrying these loads attempt to stabilize the load by increasing double support time during walking and maintaining the pack + load COM over the base of support while maintaining a regular vertical COM excursion to optimize energy expenditure.

Lower extremity joint angles respond to the load in order to maintain this vertical COM excursion strategy while controlling the load and minimizing the effect of the load on the lower extremities. Due to this interplay between mechanical dampening and energy cost optimization, the knee joint excursion changes minimally, leaving the stress to be absorbed by the knee musculature. This response manifests as increased knee extensor moments and quadriceps activity during early stance, and increased ankle plantar flexion moments and gastrocnemius-soleus activity in late stance. Load rate also increases in early stance. Results from the literature also suggest that the knee is the joint that is the most sensitive to load magnitude; the knee moment responds in a proportionally greater manner than the other lower extremity joints to increases in load weight. Results from one study also imply that the initial response may not be sustainable over a longer period of load carriage, which may tie into increased likelihood of soft tissue knee injury in soldiers who are more likely to carry loads as part of their occupation (e.g., infantry vs. support).

Future studies on load carriage should consider the effects of extended periods of load carriage. More studies which incorporate nonlinear analyses will lend further insight into how joint and segment interactions change with loads and over time. Also worth considering is the fact that the results presented in this chapter are based on trials conducted in controlled laboratory environments. Future research should consider the effects of terrain on mechanics as well as energy expenditure. Additionally, modeling studies are starting to emerge which will add to the load carriage literature. These models will be designed to examine a few aspects of load carriage in more detail. Biological and musculoskeletal models are emerging which attempt to capture the responses of the cardiovascular and musculoskeletal systems to increased carried loads. Results of such models have recently examined the effects of load carriage on modeled muscular forces based on subject-specific tibia images [137], with the ultimate intent to model the probability of injury assuming specific exposure conditions over time, as has been published in the running literature [138].

In addition to modeling studies, there is also an emerging desire in biomechanics to collect high-fidelity data from the field. While laboratory-based studies will still serve to inform on mechanisms of injury and serve as input into more sophisticated models, collecting data on force and rate of force development in the field while Soldiers are performing occupationally-relevant tasks and training will provide more complete information on dose-response relationships between training and injury [139]. Emerging technology will likely facilitate biomechanics studies to be designed in a way that allows for characterization of loading parameters while carrying loads and conducting military relevant activities in the field.

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Load Carriage-Related Injury Mechanisms, Risk Factors, and Prevention

Joseph J. Knapik and Katy Reynolds

Abstract Since the beginning of recorded history, soldiers have been required to carry arms and equipment on their bodies. Recently, the weight of these loads has substantially increased due to improvements in weapons and personal protection. As soldier loads increase there are increases in energy cost, altered gait mechanics, increased stress on the musculoskeletal system, and more rapid fatigue, factors that may increase the risk of injury. Passive surveillance of injuries experienced by soldiers on load carriage missions and surveys of hikers and backpackers indicate that foot blisters, stress fractures, compression-related paresthesias (brachial plexus palsy, meralgia paresthetica, digitalgia paresthetica), metatarsalgia, knee problems, and back problems are among the most common load carriage-related maladies. This article discussed these injuries providing incidences, rates, symptoms, mechanisms, and risk factors, and provides evidence-based preventative measures to reduce injury risk. In general, lighter loads, improving load distribution, using appropriate physical training, selecting proper equipment, and using specific techniques directed at injury prevention will facilitate load carriage. An understanding of injury mechanisms and implementation of appropriate prevention strategies will provide service members a higher probability of mission success.

1 Introduction

In military operations dating back to the beginning of recorded history soldier have been required to carry arms and equipment on their bodies. Before the 18th Century, soldiers seldom carried loads that exceeded 15 kg while on unopposed marches,

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but in more recent history, beginning with the Industrial Revolution, loads have progressively increased [116, 121]. During United States (US) military operations in Afghanistan infantry soldiers carried an average ± standard deviation approach march load of 45 ± 11 kg [50]. The increase in soldier loads is presumably due to new technologies that have increased the lethality of the individual soldier while at the same time increasing the soldier's survivability. For example, infantry soldiers can now carry weapon systems that can disable and/or destroy aircraft and armored vehicles, while developments in body armor have provided enhanced individual protection from hostile fires [50, 73, 143].

Loads carried on the body have been shown to increase the energy cost of locomotion [159, 164], alter the mechanics of walking [9, 81], place stresses on the musculoskeletal system [74, 145], and lead to more rapid fatigue [60, 69]. These and other factors may be associated with high risk of injury. The purpose of this chapter is to describe the most common injuries experienced by individuals involved in load carriage along with symptoms, incidences, and/or rates of each injury. The mechanophysiology of these injuries will be explored and where available, methods for injury prevention will be provided. Other reviews have provided information on load carriage-related injuries [114, 119, 157] but this review is unique in its focus on the injury mechanisms and injury prevention techniques.

2 Surveys of Load Carriage Injuries

Three investigations have examined injuries during military load carriage missions. Table 1 shows injuries experienced by 335 male United States (US) infantry soldiers who carried about 46 kg during a maximal effort 20-km road march. Medics and a physician were present on the march course to record and treat injuries, and soldier medical records were screened for march-related injuries up to 12-days after the march [110]. Table 2 shows injuries experienced by 218 male US infantry soldiers who carried about 47 kg during a 161-km road march conducted over 5 consecutive days (about 32 km each day). Injuries were recorded at fixed medical facilities along the march route [177]. Table 3 shows the anatomical locations of injuries reported by Australian soldiers over a 2-year period, obtained from an Australian Army medical surveillance system [156]. The most common activities reported at the time of injuries included marching (66 %), patrolling (13 %), combat training (12 %), and physical training (6 %). Two laboratory investigations reported injuries among soldiers carrying loads while walking on treadmills. Table 4 shows injuries experienced by 15 male soldiers who walked on a level treadmill at three speeds (4.0, 4.9 and 5.8 km/h) and 3 load conditions (5, 32, and 49 kg) over a 7 week period [176]. In an experimental study involving 18 British soldiers who walked on a treadmill for 2 h carrying 20 kg at 5.8 km/h, 61 % (n = 11) experienced blisters [21].

Several investigations have also obtained injuries experienced by recreational hikers that carry backpacks [7, 28, 29, 68, 78, 131, 207]. While instructive, there are

Table 1 Injuries among soldiers during a 20-km maximal effort road march [110]

Injury	During march[a]		1–12 Days post-march (n)[b]	Totals	
	Soldier continued march (n)	Soldier did not continue march (n)		N	%
Foot blisters	13	0	19	32	35
Foot hot spots[c]	3	0	0	3	3
Back pain/strain	5	7	9	21	23
Metatarsalgia[d]	1	1	9	11	12
Leg strain/pain	0	0	7	7	8
Sprains (foot/knee/ankle)	1	1	4	6	7
Knee pain	0	0	4	4	4
Foot contusion	0	1	1	2	2
Other	1	2	2	5	5
Total	24	12	55	91	100[e]

[a] From medics and physician during the march
[b] From medical records of soldiers up to 12 days after the march
[c] Hot spots are erythemic areas from frictional rubbing and are blister precursors
[d] All cases of foot pain involved metatarsalgia
[e] Within rounding error

Table 2 Injuries and symptoms experienced by soldiers during a five-day, 161-km road march [177]

Injury	During march[a]		1–15 Days post-march (n)[b]	Totals	
	Soldier continued march (n)	Soldier did not continue march (n)		N	%
Foot blisters	43	3	3	49	49
Foot hot spots[c]	1	0	0	1	1
Metatarsalgia[d]	6	2	9	17	17
Ankle and knee sprains	5	0	0	5	5
Back strain	3	1	0	4	4
Knee pain	3	1	3	7	7
Hip and thigh pain	2	1	0	3	3
Ingrown toenail	0	3	0	3	3
Foot swelling	1	1	0	2	2
Hip abrasion	2	0	0	2	2
Stress fracture	0	1	0	1	1
Black toe	1	0	0	1	1
Leg numbness	1	0	0	1	1
Groin strain	0	0	1	1	1
Foot pain[d]	2	0	0	2	2
Other	1	1	0	2	2
Total	71	14	16	101	100[e]

[a] From physician's assistants at fixed medical sites along the march. A soldier could have had >1 injury
[b] From medical records of soldiers up to 15 days after the march
[c] Hot spots are erythemic areas from frictional rubbing and are blister precursors
[d] Most cases of foot pain involved metatarsalgia
[e] Within rounding error

Table 3 Anatomical locations of soldier load carriage-related injuries complied from an Australian defence medical surveillance system [156]

Anatomical location	Frequency (n)	Proportion (%)
Back	92	22.8
Ankle	66	16.3
Knee	46	11.4
Neck/shoulder	42	10.4
Foot	39	9.7
Systemic[a]	27	6.7
Shins	20	5.0
Multiple	20	5.0
Lower limb[b]	14	3.5
Hip	9	2.2
Thigh	4	1.0
Calf	4	1.0
Knee/ankle	4	1.0
Unknown	4	1.0
Hands/arms	3	0.7
Abdomen	3	0.7
Head	3	0.7
Back/knees	2	0.5
Upper torso[b]	2	0.5
Total	404	100[c]

[a] Communication with the author indicated that these were heat-related injuries
[b] General areas reported
[c] Within rounding error

differences between these studies and those involving soldier load carriage. Compared to soldiers, recreational hikers are generally older and carry lighter loads [7, 131]. All published studies involve self-report surveys and may suffer from recall bias which may tend to underestimate the actual injuries experienced. Surveys ask either about injuries experienced while hiking a specific trail system [7, 29, 68], injuries experienced by hiking club members while hiking at any time [131], or injuries during a long-term wilderness leadership training course [207].

Table 5 shows general categories of injuries reported by 155 hikers (average ± SD age = 45 ± 5 years) who had completed the 438 km Long Trail in Vermont, US, as either thru hikers (completed the entire trail in continuous, consecutive days) or section hikers (completed sections of the trail with very long breaks between sections) [68]. Table 6 shows injuries and symptoms from a survey of 280 hikers (average ± SD age = 35 ± 15 years) who were on the Appalachian Trail for at least 7 days as thru or section hikers. Average time on the trail was about 137 days and average distance was about 2,609 km. Injuries were the primary reason for not completing the hike [29]. Table 7 shows the types and anatomical locations of injuries reported by 702 New Zealand recreational hikers (aged 16 to over 50 years). About 77 % had 5 or more years of hiking experience and reported

Table 4 Types and anatomical locations of injuries during nine treadmill load carriage walks conducted over 7 weeks [176]

Injury/Symptom	Frequency (n)	Proportion (%)
Foot blisters	24	29.3
Upper back strain	19	23.2
Foot hot spots[a]	10	12.2
Shoulder tendonitis	4	4.9
Metatarsal pain	4	4.9
Shoulder/abdominal abrasion	3	3.7
Plantar fasciitis	3	3.7
Knee tendonitis	3	3.7
Toenail injury	3	3.7
Shoulder paresthesias	2	2.4
Groin strain	2	2.4
Low back strain	1	1.2
Hip pain	1	1.2
Ankle strain	1	1.2
Foot contusion	1	1.2
Metatarsal fracture	1	1.2
Total	82	100[b]
Anatomical Location	Frequency (n)	Proportion (%)
Feet	46	56.1
Back	20	24.4
Shoulders	8	9.8
Knees	3	3.7
Groin	2	2.4
Hips	1	1.2
Ankle	1	1.2
Abdomen	1	1.2

[a] Hot spots are erythemic areas from frictional rubbing and are blister precursors
[b] Within rounding error

Table 5 General categories of injuries reported by hikers of the Long Trail in Vermont, US [68]

Injury/category	Thru hikers (n = 75)		Section hikers (n = 80)	
	N	%	N	%
Musculoskeletal[a]	35	46.7	30	37.5
Blisters	25	33.3	20	25.0
Trauma[b]	7	9.3	4	5.0
Insect Bites	5	6.7	0	0

[a] Most common musculoskeletal injuries were ankle sprains and back pain but only the number of overall musculoskeletal injuries were included in article
[b] Included lower extremity fracture, ligament damage, and lacerations but only numbers of overall traumatic injuries were included in article

Table 6 Injuries and symptoms reported by long distance hikers (≥7 Days Hiking) on the Appalachian Trail[a] [29]

Injuries/symptoms	Thru hikers (n = 161)		Section hikers (n = 51)		Hikers who did not complete trail (n = 88)	
	N	%	N	%	N	%
Foot blister	110	68	30	59	40	45
Skin abrasion	91	57	21	41	31	35
Numbness/paresthesias	80	50	15	29	20	23
Acute joint pain	69	43	14	27	19	22
Lost toenails	56	35	9	18	18	20
Tick bite	42	26	6	12	20	30
Back pain	40	25	9	18	15	17
Sunburn	40	25	9	18	23	26
Muscle cramps	36	22	6	12	13	15
Achilles heel pain	36	22	8	16	11	13
Shin splints	27	17	5	10	7	8
Tendonitis	23	14	4	8	11	13
Muscle strain	23	14	6	12	7	8
Ankle sprain	21	13	3	6	8	9
Fracture[b]	4	2	3	6	4	5

[a] A hiker could have >1 injury
[b] Of the 11 fractures, 5 were stress fractures, 5 were due to falls, and one was unclassified

carrying average loads of about 14 kg over distances of 11 or more km/day on consecutive days [131]). Table 8 shows the results of a survey of 128 Appalachian Trail and Pacific Crest Trail hikers (average ± SD age = 33 ± 11 years) who had walked at least 800 km in a single hiking season. Pack weights (excluding food and water) of 5–9, 10–14 and >14 kg were reported by 42, 30 and 28 % of the sample, respectively [7]. Finally, Table 9 shows the only study to compare injury incidence among men and women while on a 23-day Outward Bound training course that involved extensive backpacking in addition to rock climbing and development of basic camping skills. One or more injuries was reported by 85 % of the men (n = 216) and 94 % of women (n = 127) (risk ratio (women/men) = 1.10, 95 % confidence interval = 1.01–1.16, p = 0.03) and participants were younger (average of about 20 years of age) than other recreational hiker investigations.

Taken together, the military and recreational data suggest that foot blisters, back problems, and metatarsalgia are among the most common load carriage-related injuries [7, 21, 29, 110, 131, 156, 176, 177, 207]. Other commonly reported injuries and symptoms include stress fractures [7, 29, 177], non-specific paresthesias [7, 29, 177], and non-specific knee pain [110, 131, 156, 176, 177].

Table 7 Injuries and symptoms reported by New Zealand hikers[a] [131]

Injury/symptom	Frequency (n)	Proportion of reported types (%)
Sprain	108	27.8
Abrasions/laceration	91	23.5
Blisters	44	11.3
Contusion	31	8.0
Fracture	31	8.0
Cramp	13	3.4
Back Pain	7	1.8
Other[b]	63	16.2
Total	388	100.0
Anatomical Location	Frequency (n)	Proportion of reported locations (%)
Knee	101	27.9
Ankle	87	24.0
Foot	47	13.0
Leg (NOS)	40	11.0
Hand/wrist	29	8.0
Shoulder	15	4.1
Back	14	3.9
Head/face	11	3.0
Arm	11	3.0
Ribs	7	1.9
Total	362	100.0

[a] A hiker could have >1 injury
[b] Included burns, sunburns, insect bites and toenail injury or loss
Abbreviation *NOS* Not Otherwise Specified

Table 8 Injuries and Symptoms Reported by Hikers Traversing ≥800 km in a Season [7]

Injuries/Symptoms	Frequency (n)	Proportion (%)[a]
Blisters	110	85.9
Buttocks Abrasion	81	63.3
Paresthesias	60	48.2
Insect Bites	53	42.1
Sprains	26	20.6
Shin Splints	24	19.4
Sunburn	23	18.1
Cellulitis	18	14.2
Abscess	10	7.9
Stress Fracture	6	4.7
Traumatic Fracture	2	1.6
Eye Injury	1	0.8

[a] Number responding ranged from 124 to 128. Respondents could report more than one injury

Table 9 Comparison of Self-Reported Injuries among Men and Women in a 36-Day Outward Bound Course Emphasizing Backpacking [207][a]

Injury	Men (n = 216)		Women (n = 127)		Risk Ratio—Women/Men (95 %CI)	Chi Square p-value
	Injured (n)	Injured (%)	Injured (n)	Injured (%)		
Laceration	117	54	81	64	1.18 (0.97–1.41)	0.10
Blister	100	46	69	54	1.17 (0.93–1.46)	0.19
Bruise	58	27	79	62	2.31 (1.78–2.99)	<0.01
Burn	45	21	24	19	0.91 (0.56–1.44)	0.77
Sprain	36	17	21	17	1.00 (0.58–1.67)	>0.99
Sunburn	12	6	7	6	1.00 (0.36–2.65)	>0.99
Strain	7	3	7	6	1.70 (0.55–5.29)	0.46
Other	30	14	19	15	1.08 (0.61–1.90)	0.90

[a] Respondents could report more than one injury. Numbers recalculated from raw data in article

3 Foot Blisters

Foot blisters are by far the most common load carriage-related injury [7, 21, 29, 68, 110, 176, 177, 207]. This injury may seem minor but blisters can develop into more serious problems such as cellulitis or sepsis [2, 87]. In military field environments, blisters may be exacerbated by less sanitary conditions and the fact that any limited mobility imposed by blisters may be life threatening. A survey of US soldiers in Iraq (2003–2004) found that 33 % had experienced foot blisters while on deployment and 11 % had sought medical care for treatment [30].

3.1 Mechanisms Involved in Foot Blisters

Frictional forces that oppose the movement of materials or objects across the skin appear to be the cause of most foot blisters. When the skin of the foot is in contact with the shoe or sock and external forces as a result of locomotion attempt to move the sock and boot across the skin, a frictional force (F_f) will oppose the movement. As the external force increases, friction increases and movement will occur when the external force exceeds the F_f. The magnitude of the frictional force is proportional to the normal force (F_n, force of contact or perpendicular force) according to the formula

$$F_f = \mu * F_n,$$

where μ = coefficient of friction.

Foot blisters are perhaps the only load carriage-related injury that has been produced experimentally in human subjects [1, 84, 199]. In investigations that have been performed, a variety of mechanical probes have been used to apply a constant force that repeatedly cycles over the skin in a linear or circular fashion. These techniques have demonstrated that a relatively uniform series of events are involved in blister formation and these events can be linked to specific physiological outcomes. At first there is a slight exfoliation of the stratum corneum, erythema is noted around the zone of the rubbing, and skin temperature increases. The erythemic area is the "hot spot" typically noted just before blister formation [84, 110, 176, 177]. With continued rubbing the subject may suddenly experience a stinging, burning, or painful sensation and a pale, narrow area forms around the reddened region. This pale area enlarges inward to occupy the entire zone where the rubbing is applied. The pale area becomes elevated over the underlying skin as it fills with fluid. Histological studies indicate that the pale area is a separation of cells at the level of the stratum spinosum, presumably due to mechanical fatigue. Normal, necrotic, and degenerated prickle cells are seen on both sides of the cleft, with the basal cell layer usually showing little insult and the junction between the dermis and epidermis remaining undamaged. Hydrostatic pressure gradually causes the cleft to fill with fluid. Compared with plasma, the blister fluid has a lower protein but similar electrolyte concentration. Recovery begins after about 6 h as cells in the blister base increase their uptake of amino acids and nucleosides of RNA and DNA. At 24–30 h, high mitotic activity is apparent in the base cells and at 48 h a new granular layer can be found. By 120 h, cellular proliferation declines and a new stratum corneum can be seen [2, 84, 199].

3.2 Risk Factors for Foot Blisters

The probability of blister formation appears to be influenced by a number of factors, including the magnitude of the frictional force, number of shear cycles, external loading, skin characteristics, moisture, ethnicity, foot type, and condition of the footwear. The magnitude of the frictional force and the number of times a material or object cycles over the skin appear to be the major determinates of blister formation: as frictional forces increase, fewer cycles are required for blister formation [148]. Related to this are studies showing that as external loads (weight carried) increase the likelihood of blister formation also increases [108, 176]. As loads increase there are higher maximal anteroposterior breaking and propelling forces acting on the foot [20, 106] possibly accounting for the highly blister likelihood. Runners who were prone to blisters were shown to have greater foot pressures and greater shear forces compared to individuals not prone to foot blisters [221].

Skin characteristics also influence blister formation. Blisters are readily formed in the palms of the hands, soles of the feet, and other areas with a thick stratum corneum and granulosum. This is presumably because frictional forces are more likely to result in differential movement of the upper and lower layers of the epidermis, leading to mechanical fatigue [2, 199]. Moist skin produces higher frictional forces than very dry or very wet skin. Frictional forces on dry skin result in exfoliation of cells from the stratum corneum and these sloughed cells provide a sliding lubrication analogous to graphite. Very wet skin may produce hydrodynamic lubrication. Moist skin appears to impede the movement of squamous cells by holding them to the skin through surface tension, thus increasing the frictional effect [2, 147, 149, 199].

During military road marching, blisters were less likely to develop among those of black ethnicity but more likely among those with flat feet [118, 211]. In Black skin, mechanical stress causes less tissue deformation and this may reduce mechanical fatigue in the stratum spinosum making blisters less likely. Pes planus feet may have a larger surface area exposed to frictional forces, possibly making blisters more likely.

Boots that have not been worn much (not "broken in") appear to increase the likelihood of blister formation [30, 163]. When boots are worn for longer period, the epidermis may adapt to repeated lower-intensity frictional forces by increasing the thickness of the stratum corneum and stratum spinosum [94, 132]. This may increase resistance to blister formation.

Tobacco use resulted in higher blister risk in one military study [118] but not in two others [30, 211]. More investigation is needed in this area. Tobacco use results in catecholamine release, vasoconstriction in the skin, and skin damage and these factors may reduce the ability to resist frictional forces.

3.3 Foot Blister Prevention

Interventions to reduce blister incidence have generally focused on decreasing moisture and/or frictional forces. An antiperspirant (20 % solution of aluminum chloride hexahydrate) applied to the foot has been demonstrated to decrease sweating and to reduce the incidence of blisters [117, 174], while an antiperspirant with an emollient did not reduce blisters [48, 174]. However, antiperspirants can result in a high incidence of irritant dermatitis and individuals should be monitored for this [117]. Use of a lower dosage of the active ingredient or a different application schedule may reduce symptoms.

Certain types of sock materials appear to move moisture away from the skin in the presence of metabolic heat ("wicking"). Likely because of this effect, acrylic and polyester socks have been shown to decrease blister incidence [86, 113, 211]. Tighter fitting nylon socks reduce frictional shear forces [46, 90] and may reduce blisters [4]. A sock with a blend of fabrics (50 % Merino Wool, 33 % polypropylene, 17 % polyamide) was rated by recruits in Swiss basic training as being cooler, less damp, with less friction, and more comfortable than a sock consisting of polypropylene alone. The blended sock absorbed more moisture resulting less skin hydration, likely favorable for blister prevention, but blister incidence was not specifically examined [23].

A closed-cell neoprene insole (Spenco®) reduced the incidence of foot blisters, but a cellular polyurethane substance (Poron®) did not [191, 194, 195]. The effects of skin coverings have been tested for their effects on the coefficient of friction (μ), and reducing μ may reduce blister incidence. Tested skin coverings (with μ values in parenthesis) include Bursatek® (0.57), Dr Sholl's® Moleskin Plus (0.69), Moleskin (0.94), Band-Aid® (1.01), Band-Aid® Plastic (1.03), Spenco® Second Skin® Blister Pad (1.04), New-Skin® (1.05), Nexcare® Comfort (1.08), Dr Sholl's® Blister Treatment (1.20), Blister Block® (1.37), and Tegaderm® (1.54) [168].

Drying powders have been advocated as a method for preventing blisters, since powders may absorb moisture. However, studies using talc-based powders have either shown no difference when compared with a control group, or a higher blister incidence among those using the powder [3, 172, 211]. Dry talc reduced frictional forces in exposed skin [41, 149], but when the talc was wet, frictional forces increased [41]. It is possible that when alkaline sweat combines with foot powder it creates abrasive surfaces as it clumps together.

Several studies indirectly supported the idea that blister likelihood is reduced by recent exposure of the skin to repeated frictional forces that were insufficient to cause immediate blister formation [94, 113, 132].

When loads were very heavy (>61 kg), the double pack had a lower blister incidence than the backpack [111]. This suggests that better load distribution might reduce blister incidence [20].

4 Stress Fractures

It is not unusual for military recruits involved in load carriage activities to experience stress fractures [31, 55, 70, 72, 96, 99, 109]. A study that examined all US Army basic trainees from 1997 to 2007 (n = 583, 651) reported incidences of 1.9 % for men and 8.0 % for women [109]. Besides military recruits, stress fractures have also been reported in well-trained soldiers [112, 175, 203]. For example, during the Central Burma campaign in World War II, 80 stress fracture cases were reported in one 7,000 man infantry unit (1.1 %) during a 483-km march [11, 54]. Stress fractures were initially termed "march fracture" because soldiers would often present with this injury after road marching [12, 35, 77, 208]; however, in other cases the onset was insidious and could not be linked to a single event [12, 35, 196]. Dr. Breithaupt is credited with the first description of the condition in Prussian Army soldiers (as cited in [35]). During WWII, cases and case series began to appear in the literature during both British [12, 196] and American [35, 54, 77, 92, 208] military training.

Individuals with stress fractures present with localized pain that is aggravated by physical activity and reduced by rest. Focal tenderness is present in most cases; swelling and erythema may be present. Patient history notes a recent increase in the amount or intensity of physical activity. Plain radiographs have a high initial incidence of false negatives because the injury is generally not evident on films for 2–4 weeks after the onset of pain. Bone scans can produce many false positives.

Despite the cost, magnetic resonance imaging (MRI) may be the most useful since it may detect stress fracture-related abnormalities within 1–2 days of the injury [15, 17, 63, 161, 162, 189].

4.1 Stress Fracture Mechanisms

Stress fractures are part of a continuum of changes in the bone in response to mechanical stress. The continuum spans events from normal bone remodeling to a frank fracture. A stress fracture is a partial or incomplete bone rupture associated with habitual mechanical deformation of the bone. The injury generally occurs in individuals with otherwise healthy bones, often in association with unaccustomed but repetitive physical activity [102, 146, 180, 215]. This type of fracture might be termed a "fatigue fracture" and is to be distinguished from similar ruptures that might occur due to conditions like osteoporosis, hypervitaminosis A, sarcoma, and other malignancies [32, 160, 217]. The current etiologic hypothesis is that repetitive mechanical loading results in an increase in cyclic hydrostatic pressures which are sensed by osteocytes within the bone matrix. Mechanical pressures stimulate osteocyte-directed transcription of osteoclastics which begin the bone remodeling process. If the repetitive mechanical stress is applied gradually over time (i.e., progressive overload), normal bone remodeling occurs. That is, osteoclastic processes, which resorb bone, approximately balance osteoblastic processes, that form new bone. However, if the repetitive stress is applied in a relatively short period of time, as with a prolonged load carriage march for which the soldier has not sufficiently trained, osteoclastic processes may proceed at a faster pace than osteoblastic processes. This imbalance produces a vulnerable period when the bone is weakened and susceptible to rupture [37, 107, 180].

4.2 Stress Fracture Risk Factors

Considerable research has been conducted to identify factors that increase stress fractures risk. Well documented risk factors include female gender [16, 31, 97, 99, 109, 115, 124, 133, 138, 171], white ethnicity [31, 64, 67, 105, 109, 115, 128, 138, 186], older age [31, 67, 109, 115, 128, 138, 186], taller body stature [72, 109, 115, 209], lower aerobic fitness [43, 97, 120, 173, 186], prior physical inactivity [67, 72, 186], and greater amounts of current physical training [5, 70, 100, 170]. Risk factors with limited support include cigarette smoking [6, 65], high foot arches [71, 104], older running shoes [67], genu varus [44], and irregular menses [173].

The gender difference in stress fracture risk may be related to fitness levels, bone characteristics, and/or anatomy. Women have lower average fitness levels than men and lower fitness is associated with higher overall injury rates [120, 122] as well as a higher incidence of stress fractures [14, 97, 173, 187, 209]. In the bones of the lower body, where most stress fractures occur [13, 31, 187], women have a lower bone section modulus and a lower bone strength index (ratio of section modulus to bone length), compared to men [14]. Female bones are thinner and narrower which provides less bone strength [14, 59]. Most stress fractures occur in cortical bone and female cortical bone have higher bone mineral density which may increase overall bone strength but may also make the bone stiffer and more likely to experience fractures with repetitive loading cycles [59, 202]. With regard to anatomical differences, women have wider pelves [75] that results in a varus tilt of the hip and a larger bicondylar angle [198] which places greater stresses on the hips and lateral aspects of the knee and lower leg during physical activity. Both men and women with wider pelves have higher stress fracture incidence suggesting that a wider pelvis alone (regardless of gender) increases injury risk [14].

The association between age and stress fractures may be associated with changes in bone with aging. Bone mass declines with age, primarily in the trabecular compartment. Losses in trabecular mass in the distal tibia amounted to 0.24 %/year and 0.40 %/year for men and women, respectively, in one study [178]. This loss may be due to a reduction in the number of osteoblastic stem cells or a reduction in the lifespan of the osteoblasts themselves. Most of bone remodeling occurs on surfaces and trabecular bone has a greater surface area that than cortical bone, so it is trabecular bone that is more likely to be remodeled [26, 201]. The overall loss of bone mass with age would affect bone strength, and age-related effects on osteoblasts would result in slower bone remodeling thus increasing susceptibility to activity-induced stress fractures. However, as noted above, most stress fractures manifest in the cortical bone. In this compartment, there are age-related changes in the mechanical properties of the collagen network (upon which bone minerals deposit), which reduces strength, modulus, and ability to absorb energy, while increases in bone porosity reduces bone stiffness and strength [202, 214]. These changes likely contribute to the higher age-related stress fracture risk.

The lower stress fracture risk among blacks could be partly related to the higher bone mineral density compared to other racial/ethnic groups [10, 150, 204]. This difference persists after adjustments for body composition, dietary history, sun exposure, biochemical bone markers, lifestyle characteristics, and other factors [58]. In addition to higher bone mineral density, studies of the female femur have shown that there are differences in bone geometric properties between black and white women: black women have longer and narrower femora with thicker cortical areas and smaller medullary area. These factors increase mechanical strength by contributing to lower bending stresses in the cortical areas during physical activity [150]. The higher bone mineral density and the manner in which the bone architecture is arranged may contribute to the lower stress fracture incidence in blacks.

The length of long bones in the lower body (i.e., femur, tibia, fibula) is highly related to height, with correlations in the range of 0.9 [190, 205]. Taller individuals with presumably longer bones may experience more bending and strain on their lower body long bones during physical activity. One study [14] showed that men with stress fractures had longer femora compared to those without stress fractures and the longer bones contributed to lower bone strength.

Prior physical activity may reduce stress fracture risk during current activity because of changes in bone tissue induced by the prior activity. Physical activity, especially activities involving high impact forces like running, soccer, basketball and gymnastics, increase bone mineral content, bone mineral density, bone size, bone mass, and result in activity-specific bone remodeling [19, 123, 193]. Effects of physical activity may be larger in men than in women and the maintenance of physical activity appears to be important for the maintenance of bone mass [19].

4.3 Interventions to Reduce Stress Fractures

A number of interventions have been tested in an effort to reduce the incidence of stress fractures. Successful interventions include reduced running and/or marching mileage during training [62, 98, 185] and particular types of boot insoles [192]. A multiple intervention study in Australian basic training demonstrated that reducing march speed, allowing trainees to march at their own step length (rather than marching in step), running and marching in more widely spaced formations, running on grass, and reducing running mileage was successful in reducing female pelvic stress fractures [169]. Vitamin D supplementation with calcium was shown to reduce stress fractures during basic training [127, 183]. Systematic surveillance combined with graduated, progressive-overload physical training and leadership education was found to reduce femoral neck stress fractures and other overuse injuries in US Army basic training [184].

Previous studies have suggested that when men and women carry identical loads over identical distances, women report considerable problems with load carrying equipment. Problems with shoulder straps of the rucksack, fit of pistol belts, and fit and stability of the rucksack were especially prevalent [83]. Modification of the equipment of female Israeli border police by using a shorter rifle, lighter, better fitting combat vest, and 25 % reduction in load resulted in a 56 % reduction in stress fracture incidence [42].

5 Brachial Plexus Palsy

Brachial plexus palsy is a disabling injury that has been widely reported in association with load carriage [18, 49, 51, 85, 93, 140, 154, 200, 218]. In early case studies the malady was termed “pack palsy” or “rucksack palsy” because it occurred in association with carrying heavy rucksacks [8, 25, 49, 220]. Adjusted incidence rates in US Army basic training were 0.48 cases/1,000 recruits per year [18], and in Finnish basic training, 0.54 cases/1,000 recruits per year [134, 154].

Symptoms of brachial plexus palsy include paresthesia, paralysis, and cramping with pain in the shoulder girdle, elbow flexors, and wrist extensors. The paresthesia and pain usually progress in intensity with longer load carriage time, but after the load is removed the pain is reduced or absent, although the sensory deficits and muscular weakness remain [18, 49, 134, 154]. In the longer-term, sensorimotor deficits from rucksack palsy injuries are usually temporary but, in some cases, may result in a chronic condition. In one case series of 38 Finnish recruits with diagnosed brachial plexus palsy, 79 % were asymptomatic within a median 3 months of symptoms onset (range 0 to 9 months). Others had prolonged symptoms but daily activities were not affected; a few reported some difficulty in their current profession or had to modify their free-time activities.

Brachial plexus palsy appears to be caused by rucksack shoulder straps. Heavy loads on the shoulder can cause a traction or compression injury of the C5 and C6 nerve roots of the upper brachial plexus. In minor cases, compression results in entrapment of the long thoracic nerve. Long thoracic nerve injuries usually present with “scapular winging” (i.e., the medial boarder of the scapula is more posterior) because of weakness in the serratus anterior muscle [18, 218].

Hypothetical risk factors for rucksack palsy include load weight, improper load distribution, and longer carriage distances [18, 49, 139, 176]. Muscle strength losses appear to be greater in those carrying heavier loads [154]. Age, physical characteristics (height, weight, body mass index), and physical fitness do not appear to be associated with the incidence of brachial plexus palsy [134, 154].

Use of a frame and hip belt has been demonstrated to reduce the incidence of rucksack palsy [18], presumably by reducing pressure on the shoulders [88, 126]. Use of a properly adjusted hip belt can remove 30 % of the weight from the shoulders [126].

6 Meralgia Paresthetica

Meralgia paresthetica has been reported in load carriage situations involving rucksack hipbelts [27], military pistol belts [56], parachute harnesses [188], and body armor [61]. The condition was first described in an Army officer by Martin Bernhardt in 1878; the injury has also been termed Bernhardt-Roth syndrome or lateral femoral cutaneous neuralgia [52, 165, 188]. Although the incidence of this disorder has not been reported in the military, a survey of hikers traveling ≥7 days on the Appalachian Trail found 4 % (10 of 280 surveyed hikers) reported symptoms consistent with this malady [27]. A Dutch study involving an estimated 90 % of all general practices in the Netherlands found an incidence rate of 0.43 cases/1,000 person-years [210] and a study involving 253 general practices in the United Kingdom found an age-adjusted incidence of 1.08 cases/1,000 population over an 8-year period [129].

Meralgia paresthetica is a sensory mononeuropathy in which patients report pain, paresthesia, painful burning and itching in the lateral or anterolateral thigh. Pain may be present on standing and/or walking and be relieved with sitting. The symptoms may be mild and resolve spontaneously or be more severe and limit function [27, 36, 210].

Meralgia paresthetica appears to be caused by a compression of the lateral femoral cutaneous nerve, a sensory nerve branch of the L2/L3 spinal area. The nerve passes over the anterior superior iliac spine and under the inguinal ligament and can be compressed and entrapped by these and other surrounding structures. The nerve innervates the cutaneous areas on the lateral thigh accounting for the sensory deficits reported in this area [52, 56, 61]. A report of two cases of this disorder during US Army deployments to Iraq suggested that when soldiers were wearing body armor and were seated for long periods, the lower edge of the body armor compressed the inguinal region resulting in entrapment of the lateral femoral cutaneous nerve, leading to pain and paresthesia [61].

In general clinical practice, there appears to be an inverted J-shaped relationship with age in that meralgia paresthetica cases increase after about 20 years of age, and decrease after about 60 years of age. Rates are very similar among both men and women [129, 210]. Longer duration and distance of load carriage increases the risk [27]. Higher BMI has been shown to be a risk factor, possibly because abdominal fat may compress the lateral femoral cutaneous nerve [52, 144]. When symptoms are observed the nature of the compression should be identified since symptoms generally resolve with removal of the compression [27, 56].

7 Digitalgia Paresthetica

Digitalgia paresthetica is another condition reported among those involved in load carriage. The condition has been termed "marcher's digitalgia paresthetica", presumably because of its occurrence among military recruits involved in foot marching activities [197]. The condition was first described in 1954 [216], and since then a few additional case studies have appeared in the literaure [137, 142]. A survey of hikers who spent ≥7 days on the Appalachian Trail found that 8 % (21 of 280 surveyed hikers) reported symptoms consistent with this disorder [27]. A group of 30 Israeli Army recruits were examined every 3 weeks during a 14-week basic training cycle and 14 (47 %) were diagnosed by physicians with the digitalgia paresthetica [197]. In this latter study, 9 recruits were followed up for 9 months after basic training. Seven became asymptomatic and two remained symptomatic.

Digital paresthetica is characterized by numbness, neuropraxia, pain and/or paresthesia in the toes. The condition is presumed to be due to compression of the sensory digital nerves in the foot that result in the neuropathy. In one case study, the compression was apparently due to the toes being tightly squeezed by the boots the patient was wearing [142]. In the case of military recruits, the compression was likely due to the repetitive pressure on the foot from walking with heavy loads [197].

In the study involving the Appalachian Trail hikers, those experiencing digital paresthetica had hiked for a greater number of days (144 vs. 131) and for a longer distance (1,800 vs. 1,555 miles) than those without the condition, but the differences were not statistically significant ($p < 0.05$) [27]. Military training in running shoes as opposed to the normal Israeli boot had little influence on the incidence of digitalgia paresthetica (boot incidence = 50 %, running shoe incidence = 43 %, relative risk = 1.17, 95 % CI = 0.54–2.54, $p = 0.70$) [197].

8 Conditions Associated with Anatomical Locations

Several studies have reported load-carriage related pain in specific anatomical locations but did not provide explicit diagnoses. The large majority of locations involve the lower body and lower back. Lower body/low back pain, was reported in 25 % of injury cases following a strenuous 20-km road march (Table 1) and in 22 % of cases following a 100-mile, 5 day road march (Table 2). Specific anatomical locations most often cited in load carriage studies are the foot (mostly metatarsalgia), knee, and lower back [47, 110, 156, 177, 200].

8.1 Metatarsalgia

Metatarsalgia is a descriptive term for nonspecific foot pain under the metatarsals. Based on clinical opinion, this condition has been subclassified as primary (associated with anatomical factors such as pes cavus, hypermobility, metatarsal head abnormalities), secondary (associated with other conditions such as trauma, neuromas, osteonecrosis), and iatrogenic (failed surgery) [53, 57]. During a strenuous 7-month US Army Airborne Ranger physical training program that included regular load carriage, the incidence of metatarsalgia was 20 % [200]. After a single strenuous 20-km walk during which 335 infantry soldiers carried 45 kg, 3 % experienced metatarsalgia [110] and during a 100-mile, 5-day road march 8 % of infantry 218 soldiers were diagnosed with metatarsalgia [177]. The foot accounted for 10 % of load-carriage related injuries in an Australian Army study [156].

In metatarsalgia, pain is generally localized under the metatarsals, especially the second, third, and fourth metatarsal heads [57]. During locomotion, the foot rotates anteroposteriorly around the distal ends of the metatarsal bones. While the exact mechanism for load carriage-related metatarsalgia has not been defined, it is reasonable to assume that the cyclic focal pressures and mechanical stress caused by the load can induce inflammation and trauma in the metatarsal heads [106]. The nature of the trauma is difficult to identify but may be associated with bone stress injuries (e.g., stress fractures or stress reactions) and/or strains to the foot muscles, and may be aggregated by conditions such as neuromas and arthritis [57, 101].

The size of the fat pad under the metatarsal heads was not related to the intensity or frequency of metatarsalgia [212]. Nonetheless, guided placement of polyurethane pads under the metatarsal heads reduced subjective pain scores and pressure under the metatarsal heads [103].

8.2 Knee Pain

The knee is another anatomical location in which pain is often reported in association with load carriage. A strenuous 20-km road march with a 46-kg load resulted in a 1.2 % incidence of post-march knee pain (4 of 335 soldiers) with the cases resulting in a total of 14 days of limited duty [110]. A 5-day, 161-km road march resulted in 2.5 % of infantry soldiers (7 of 281) reporting to medical care providers for knee pain but only one limited duty day was prescribed [177]. The knee was involved in 11 % of load carriage-related injuries in the Australian Army [156]. Another study [47] queried Swedish conscripts about pain in specific anatomical locations after 20–26-km road marches. Investigators found that 39 % (44 of 114 conscripts) reported knee pain. The latter study [47] involved active surveillance (querying Soldiers) as opposed to the passive surveillance (presentation to medical care providers) in the other two investigations [110, 177].

Clinical experience suggests diagnosis of knee pain should proceed from interview (history) to physical examination, to possible imaging and diagnostic tests. Interviews should characterize the nature of the pain, mechanical characteristics (locking, popping, knee giving way), mechanism of injury, and previous knee problems. Physical examination should include visual evaluation (symmetry, erythema, swelling, bruising) followed by evaluation of the patella, ligaments, and menisci with specific tests (e.g., patellofemoral tracking, drawer test, Lachman test, valgus/vargus stress test, McMurry test). Potential diagnoses are extensive but the most common overuse injuries include patellofemoral pain syndrome (chrondromalacia patellae), medial plica syndrome, bursitis (pes anserine), and tendonitis (iliotibial band and popliteus). Traumatic injuries can include ligamentous strains (anterior/posterior cruciate, medial/lateral collateral), meniscal tears, and fractures [33, 34, 91]. In a study of 1,330 Finnish military conscripts who presented with "stress-related" knee pain that disrupted their military training, 7 % (n = 88) were found to have MRI-diagnosed bone stress injuries in one or both knees (endosteal marrow edema alone, or with periosteal edema and/or muscle edema, and fracture or callus in cortical bone) and 9 % were found to have internal derangement of the knees (meniscal lesions, ligamentous injuries, patellar chondromalacia, osteochrondritis dissecans or plicae). The other 84 % of cases had no MRI-diagnosed aberrant findings [151].

Carrying heavy loads on the trunk increases the knee range of motion and increases forces experienced at the knee joint. When loads are less than 15 % of body weight, there is little change in the knee joint angle with walking [40]. As loads increase from 20 to 50 % of body weight there is an increase in the knee joint angle. After about 50 % of body weight there is apparently little additional change in the knee joint angle [9, 82, 106, 167]. The increase range of knee motion is presumably a protective mechanism to assist in absorbing impact forces [9, 106] and lowering the body center of mass so greater stability can be achieved [82]. Also, as loads increase, forces experienced at the knees (moments) increase [82, 213]. The greater range of knee joint motion combined with the greater forces at the knee joint induced by the load may be associated with knee pain experienced during load carriage.

An evidence-based conceptual model has been developed to explain the development of anterior knee pain in individuals with specific muscle weaknesses and anatomical factors [24]. Lower knee extension strength could lead to compensation at the knee joint that increases patellofemoral contact stress and the development of anterior knee pain. A greater hip internal rotation angle (possibly associated with a greater navicular drop) may lead to greater lateral displacement of the patella, increasing patellofemoral contact stress and leading to the development of anterior knee pain over time [24]. Greater knee moments experienced with load carriage [82, 213] may exacerbate anterior knee pain in individuals with these characteristics.

Prospective studies have identified greater patellar mobility as a risk factor for knee pain [125, 219]. Other risk factors for anterior knee pain include lower quadriceps flexibility; faster vastus medialis reflex time; lower vertical jump performance (lower leg power); greater hip adduction; lower isometric strength in hip adduction, knee extension and knee flexion; greater navicular drop; and a smaller knee flexion angle (unloaded) [24, 219]. There is contradictory data on taller stature and greater leg length inequality [125, 219]. Because specific weaknesses of the hip adductors, knee flexors, and knee extensors appear to be injury risk factors [24, 219], specific strengthening exercises for these muscle groups may reduce knee pain during load carriage. Likewise, increasing quadriceps flexibility and knee extension power may assist in knee pain reduction [219]. It should be noted that increasing flexibility and strength of the quadriceps has not been specifically tested for efficacy in reducing knee pain and research in this area should be conducted.

8.3 Back Pain

Back injuries can pose a significant problem during load carriage. In one study involving a strenuous 20-km road march with 46-kg load, 23 % of injuries involve back and 50 % of the soldiers who were unable to complete the march reported problems associated with their backs (Table 1). Injuries to the back were the most common in an Australian Army investigation, accounting for 23 % of all load carriage-related injuries (Table 3). In a research study involving 9 and 12-km treadmill walks with varying loads and speeds, 25 % of soldiers reported back problems (Table 4). In the active surveillance study that queried Swedish conscripts about pain in specific anatomical locations after 20–26-km road marches, 38 % (43 of 114 conscripts) reported back pain [47]. On the other hand, a 161-km road march over 5 days only resulted in a 4 % incidence of back problems with 8 days of limited duty that accounted for 12 % of all limited duty days (Table 2).

Systematic reviews and national medical societies provide evidence-based guideline for the assessment of low back pain [38, 39, 45]. These recommendations generally advise using very specific signs, symptoms and patient characteristics to rule out serious spinal pathologies (spinal cancers, cauda equine syndrome, fractures, ankylosing spondylitis, neurological defects, herniated discs, and spinal stenosis). Imaging or diagnostic tests are recommended only for very specific signs and symptoms [38, 45, 206]. Clinical estimates suggest that 82–97 % of back pain involves muscle strains, discs injuries (herniations, tears, and degeneration), facet degeneration, and spinal stenosis, with few cases involving more serious pathologies [39, 135].

When an individual walks with a backpack load there are cyclic stresses on the vertebrae, intervertebral discs, muscles, and other spinal structures [79, 153] and heavy loads do not move in synchrony with the trunk [153, 166]. Backpack load carriage results in an increase in the forward inclination of the trunk [22, 81, 136, 179] and an increase in the spinal curvature [155]. Stresses on the spine during load carriage include compressive forces, shear forces, and torques that increase as the load increases [66, 74, 155, 179]. Shearing forces are lower than compressive forces [74] but the spinal structures opposing shear forces (vertebral arch, ligaments, annulus fibrosis) appear to be weaker and may be more susceptible to large or repeated loads [66]. The magnitude of shearing forces also differs depending on the shape of the unloaded spine. Individuals with more lordotic spines tend to increase the spinal curvature more under load than individuals with smaller lordotic curves [141]. Thus, individuals with more lordotic spinal shapes likely experience greater shear stresses which have been shown to be associated with back injuries [152, 158]. In addition to stresses on the vertebra, load carriage increases trunk stiffness [89] due to both active co-contraction of the abdominal and paraspinal muscles [130, 181] and paraspinal reflexes [145]. Trunk stiffness increases as either load or speed increases [89]. The combined stresses on vertebra, discs, muscles and other spinal structures are likely associated with back pain and injuries experienced in susceptible individuals.

Backpack loads have been shown to be associated with back injuries in a dose-response manner [176, 182]. This could be because as the load increases there are greater changes in trunk angle [76, 79, 153] which may exacerbate shearing forces. A double pack (load equally distributed on the front and back of the trunk) may help reduce the incidence of back problems because it results in less forward trunk flexion and reduces prolonged bending of the back [80, 106]. However, a double pack might not be feasible in some conditions because it can induce respiratory problems and reduce the ability to dissipate heat [95]. Nonetheless, it may be possible to move some of the load to the front of the body through the use of vests to achieve a load distribution that reduces forward flexion of the trunk.

9 Summary and Conclusions

Among the most common load carriage-related injuries are foot blisters, stress fractures, various compression-related paresthesias (brachial plexus palsy, meralgia paresthetica, digitalgia paresthetica), metatarsalgia, knee problems, and back problems.

Foot blisters are cause by frictional rubbing that results in mechanical fatigue in the epidermis, separation of cell layers, and filling of the separation with fluid due to hydrostatic pressure. A stress fracture (“march fracture”) is a partial bone rupture that typically occurs when there is a sudden increase in activity and osteoclastic processes that resorb bone proceed faster than osteoblastic processes that form bone resulting in a vulnerable period when the bone is weakened and susceptible to activity-induced breakage. Brachial plexus palsy (“rucksack palsy”) occurs when rucksack shoulder straps cause a traction or compression injury of the C5 and C6 nerve roots of the upper brachial plexus or the compression results in entrapment of the long thoracic nerve. Meralgia paresthetica can arise when load carriage equipment compresses the lateral femoral cutaneous nerve where it passes over the anterior superior iliac spine. Digital paresthetica (“marcher’s digitalgia”) is due to compression of the sensory digital nerves in the foot. The mechanism of metatarsalgia is not well understood but the load carriage-induced cyclic focal pressures during locomotion may provoke inflammation and trauma in the metatarsal heads that may be associated with bone stress injuries (e.g., stress fractures or stress reactions) and/or strains of the foot muscles. Knee pain can be due to many types of injuries but may be more likely among individual with lower knee extension strength or a greater hip internal rotation angle that lead to compensations that increase patellofemoral contact stress. Back pain can also be due to a number of specific injuries but the mechanism of many of these may be associated with the fact that heavy loads do not move in synchrony with the trunk and producing cyclic stresses on the vertebrae, intervertebral discs, muscles, and other spinal structures; individuals with lordotic spines may experience more shear stress in the lower spine during load carriage.

Primary prevention techniques are available which should assist in reducing load-carriage-related injuries. For blisters, reducing moisture, reducing friction, and inducing skin adaptation will reduce the risk. Antiperspirants without emollients have been demonstrated to decrease moisture produced by sweating but can result in irritant dermatitis in some individuals. Using socks composed of acrylic, polyester and tight fitting nylon reduce friction as do some type of skin coverings like Bursatek®, and moleskin. Skin adaptations can be achieved by repeated exposure to low intensity frictional forces by systematic training with loads. Stress fractures may be reduced by proper progressive overload training that gradually increases carriage loads and/or distances over time. Vitamin D supplementation with calcium, and specific equipment modifications to better fit equipment and reduce load weights, can reduce stress fractures in women. The incidence of compressive paresthetica s (brachial plexus palsy, meralgia paresthetica, digitalgia paresthetica) can be decreased by removing or reducing the sources of the compressions. For example, the incidence of rucksack palsy can be reduced by use of a hip belt which reduces loads on the shoulder. Metatarsalgia symptoms can be relieved by use of polyurethane under the metatarsal heads. Distributing the load more symmetrically about the trunk may reduce back injuries (e.g., moving more of the backpack load to the front of the body by use of a vest).

In general, lighter loads, improving load distribution, using appropriate physical training procedures, selecting proper equipment, and choosing specific techniques directed at injury prevention will facilitate load carriage. An understanding of injury mechanisms and implementation of appropriate prevention strategies will provide service members a higher probability of mission success.

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Part III
Overuse Injuries

Overuse Injuries in Military Personnel

Jay R. Hoffman, David D. Church and Mattan W. Hoffman

Abstract The most common reason of medical evacuation for non-combat related injuries appears to be related to the musculoskeletal system. This is reported during both military deployments as well as during basic combat training. The most common cause of non-combat musculoskeletal injuries appear to occur from overuse, generally as a result of physical training. Overuse injuries are considered an outcome of the overtraining syndrome, which is considered a continuum of negative adaptations to training. Symptoms appear when the training stimulus has reached the point where the intensity and or volume of training have become too excessive, coupled with inadequate rest and recovery. These are issues that are quite common within the military during both training and deployment. During periods of deployment additional physiological stresses such as the environment (altitude, cold and heat), and nutritional and sleep deprivation may pose significant challenges on the health and performance of the soldier. This is often manifested during sustained combat operations, in which the ability to provide rest and recovery become secondary to the mission's objectives. This chapter will focus on the frequency, mechanism and risks associated with overuse injuries reported during both military training and deployment.

1 Introduction

In a recent report on medical evacuations from Operation Iraqi Freedom/Operation New Dawn the United States military reported 50,634 evacuations from the theatre of operations, 17.7 % (8944 soldiers) of the evacuations were combat related [2]. The remaining evacuations were for soldiers suffering from non-battle injuries or illness (82.3 %). The most common reason of evacuation for non-combat related

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injuries was related to the musculoskeletal system. Not only do injuries to the musculoskeletal system account for the largest portion of medical visits during deployment [7], they are also most common injury during basic combat training as well [6, 20, 36]. The most common cause of non-combat musculoskeletal injuries appear to occur from overuse, generally as a result of physical training [61].

Overuse injuries can be considered an outcome of the overtraining syndrome, which is considered a continuum of negative adaptations to training [31]. Symptoms of overtraining appear when the training stimulus has reached the point where the intensity and or volume of training have become too excessive, coupled with inadequate rest and recovery. Initial stages of overtraining are generally accompanied by subjective feelings of fatigue and staleness that may, or may not be accompanied by performance decrements. As the continuum proceeds, these subjective feelings of fatigue and staleness become associated with decreases in performance. When the training stimulus is excessive and recovery does not occur within an anticipated time, the athlete is considered to be overreaching [31, 45]. With a decrease in the training stimulus and adequate rest, complete recovery usually occurs within 1–2 weeks [45]. Recovery may also coincide with an overcompensation and improved performance. This type of overreaching is referred to as *functional overtraining* [55]. For the competitive athlete, this may be an accepted and planned part of the training program that helps the athlete reach peak performance at a desired period of time. However, for the tactical athlete (e.g., soldier), the concept of functional overtraining may or should not exist. The primary differences between the competitive athlete and the tactical athlete is that the competitive athlete is preparing to reach peak conditioning for a specific season or competition, with a known time frame. In contrast, the tactical athlete needs to remain at a high level of conditioning, but without a focused end-point. Thus, the tactical athlete is never in a peaking phase to prepare for a competition or deployment.

During periods of military training (i.e., basic training, advanced military training, etc.,) the tactical athlete is focused developing specific military skills that they may have to use during combat, possibly in extreme environments. As such, commanders often attempt to enhance the physical and mental stress during such training to enhance discipline, unit cohesiveness, and resiliency. If a balance between training and recovery is not appropriate, the soldier may progress to a serious problem of overtraining that may take months to recover, and significantly increase the risk for overuse injuries.

During periods of deployment additional physiological stresses such as the environment (altitude, cold and heat), and nutritional and sleep deprivation may pose significant challenges on the health and performance of the soldier [61]. This is often manifested during sustained combat operations, in which the ability to provide rest and recovery become secondary to the mission's objectives. As such, commanders need to understand the physiological limits of such operations and to examine potential interventions to potentially offset the physiological strain associated with combat.

2 Contributing Factors to Overuse Injuries

The examination of overuse or overtraining in military personnel should be separated into two distinct theatres; the garrisoned soldier and the deployed soldier. The main difference between these two areas of operation, in regards to overtraining the soldier, focuses on the mechanisms causing overuse. In the garrisoned soldier, the cause of overtraining and overuse is likely due to inappropriate training progression. More specifically, increases in training intensity and volume being coupled with insufficient rest and recovery. In contrast, the deployed soldier may be subject to a combat situation, that by itself will increase both physical and psychological stress [56], but during sustained combat operations a number of additional contributing factors may increase the soldier's susceptibility for injury. These factors are related to environment (altitude, heat or cold), sleep deprivation, and energy depletion. Although a single factor may be sufficient to cause overtraining, each additional factor adds to the total stress experienced by the soldier.

During periods of intense training or activity, which may occur in either garrisoned or deployed soldier, an adequate amount of carbohydrates, protein and calories need to be consumed to prevent the catabolizing of muscle for amino acids as a fuel source for exercise, and to maintain performance at a high level. Appetite may be influenced by a host of factors such as stress, sleep, and environmental conditions [83]. Strenuous work in hot and dry environments has been shown to increase energy deficit and body water loss [59], suggesting that soldiers working in the heat results in additional energy demands, which if not met, may increase the risk for overtraining. Similar effects have also been noted for the energy requirements of soldiers in hypothermic and/or hypoxic conditions [83]. In addition, special operation forces, whose mission may differ than the average infantry soldier, appear to have a greater energy requirement than the average soldier [53]. Thus, consideration of the specific demands of each soldier also need to be accounted for.

3 Overuse Injuries During Military Training

Training is an essential element for military success, as soldiers' occupational tasks require high levels of fitness and physical activity [42]. Activities common to military training include running, marching, climbing, jumping, digging, lifting and carrying heavy loads while hiking. Training is essential for new recruits to increase their level of physical fitness to appropriate standards to successfully complete basic military training. Following basic and advanced training all soldiers, but especially those in combat units, are required to maintain their level of conditioning and serve as a deployable workforce [81]. However, it is during these basic and advanced combat training courses that overuse injuries have become the leading cause of lost time, resources, and attrition of military personnel [35, 81, 82].

Considering the potential for conflicts to arise rapidly, military personnel are required to maintain combat ready levels of fitness. Thus, a balance between maintaining a high level of fitness and injury prevention becomes a critical goal for the military [42]. During basic training, many recruits experience a large increase in training volume compared to their civilian life. This sudden escalation in repetitive activity (e.g. marching, running, drilling) has been observed to increase the incidences of overuse injuries [67]. Factors such as female gender, high running mileage, low aerobic fitness, extremes in flexibility, prior injury, participation in recreational sports activities, cigarette smoking, and age of running shoes have been associated with higher risks for overuse injury in military populations [11, 27]. Although poor or inadequate levels of conditioning have been clearly identified with an increase in overuse injuries, another problem that is becoming more scrutinized is the increase of overuse injuries among highly motivated soldiers performing extreme conditioning programs [8]. Many of these well-marketed programs (e.g., CrossFit, Insanity, etc.,) are becoming popular among military personnel to enhance their level of fitness, with the belief that these programs are more specific to the needs of the soldier on the battlefield [8]. Although concern for a greater risk of overuse injuries arising from extreme conditioning programs have been raised, there is limited to no data to support these concerns. Examining a military population, Grier et al. [24] indicated that soldiers participating in advanced tactical conditioning or extreme conditioning programs experienced a 5 % increase in the incidence of injury and a 5 % increase for injuries designated as overuse, while soldiers that were not participating in these extreme programs realized a 7 % increase in injury, and a 4 % increase in injury designated as being overuse. However, the overall rate of overuse injuries in soldiers participating in extreme training programs was 37 %, while for soldiers that did not participate in the extreme training programs was 46 %. It is possible that soldiers that were more comfortable with extreme conditioning programs tended to gravitate towards those programs, and were in better overall condition to sustain their efforts in those conditioning programs. Nevertheless, others have suggested that high intensity, short rest resistance training schemes, common to extreme conditioning programs, appears to result in a greater inflammatory response and an increase in acute damage markers [28]. Thus, further examination of injury risk and extreme conditioning programs is warranted.

3.1 Musculoskeletal

The most common type of non-combat medical injury associated with overuse involves the musculoskeletal system [26, 65, 76]. Musculoskeletal injuries are a broad category of medical conditions involving muscles, tendons, nerves, ligaments, and bone tissue [90]. Musculoskeletal injuries can occur from acute physical trauma, but are more often related to the cumulative effects of repetitive micro-traumatic forces, classified as overuse injuries [26]. Investigation of an army database of all hospital admissions for active duty military personnel revealed that during a 6-year period of study, 82 % of all injuries documented (from a total of 13.861 hospital admissions) as occurring during sport or army physical training, were musculoskeletal injuries [47]. These injuries accounted for an average of 29,435 work days lost per year for military personnel.

The incidence rate of musculoskeletal injury has been reported to range from 10 to 15 per 100 male recruits per month, 15–25 per 100 female recruits per month, 6–12 per 100 infantry soldiers and 30–35 for Navy special warfare candidates [37]. A recent study reported that the injury rate of U.S. army special operation forces was 24.5 injuries per 100 soldiers in a yearlong examination of veteran operators [1]. To provide some context to this rate of injury, Kaufman et al. [37] indicated that the rate of injury for military recruits and infantry soldiers were similar to that of endurance athletes, but lower than that reported in contact sport participants. Jordaan and Schwellnus [36] published one of the first studies to examine the incidence of overuse injuries in recruits during basic training. They reported that the highest incidence of overuse injuries occurred during the first three weeks and the final week of the 9-week training period. During these periods the authors noted the majority of training time was characterized by marching (77 %) with more than 80 % of all reported overuse injuries seen in the knee, lower leg, and ankle. However, subsequent studies have reported a more even distribution of lower leg and back injuries [26, 37].

In regards to body region, the knee (22 %) and lumbar spine (20 %) are the most common site of musculoskeletal injury, followed by the ankle/foot (13 %) [26]. The high frequency of injury in these areas are likely related to marching with heavy load carriages, which is a common method of exercise within combat training. Reynolds et al. [71] examined infantry soldiers who completed a 161-km march over 5 days carrying 47 ± 5 kg of gear ($\sim$63 % of the soldiers' body mass). Results revealed that 36 % of the soldiers suffered one or more injuries, with overuse injuries accounting for 93 % of the total injuries. Most of these injuries occurred within the first two days of the march. Noteworthy, smokers had a higher injury rate than non-smokers, and young soldiers (<20 years) had a higher injury rate than older soldiers (>24 years). Although speculative, it is possible that the younger soldiers were less fit than the older soldiers. Previous studies have indicated that physical fitness levels at the onset of basic training have been shown to be a risk factor for overuse injury, and that recruits with lower levels of fitness have to work harder at a given task [6, 43, 79]. Thus, during a prolonged road march younger soldiers may not have attained the same level of conditioning as their older counterparts.

It does appear that the risk for musculoskeletal injury may be more prevalent among elite level soldiers than in infantry or other military occupational specialties [1, 37]. Whether this is due to higher motivation among those soldiers or due to greater training requirements (higher load carriages) and expectations is not well-understood. Others though, have reported a lower injury rate among Special Forces soldiers in comparison to soldiers in infantry, artillery, and combat engineering, but that the injuries among the Special Forces personnel did result in the highest limited duty days [70]. Thus, further examination of injury risk in these soldiers is warranted.

3.1.1 Stress Fractures

Stress fractures are the most common overuse injury known to the military with initial reports first raised from the Prussian Army [10]. Stress fractures occur from a rapid and unaccustomed physical stress, and develop in normal bone tissue [85]. They are a major health problem for the military, and are the most common overuse injury in recruits undergoing basic training [57]. Unlike many other overuse injuries, the military has conducted a large amount of research on stress fractures. Investigations looking at etiology, risk factors, prevention, and treatment have been conducted in an effort to reduce the frequency this injury.

Bone is a dynamic and metabolically active tissue, which responds to the stresses that are imposed on it. Mechanical stresses that occur during the application of an external load are sensed by the bone. When adequate recovery is provided the external loads experienced by bone will stimulate positive adaptation. The loading will increase osteogenesis on both the periosteal and endocortical surfaces of bone [23, 69]. Upon stimulation, osteoblasts from mesenchymal stem cells produce and mineralize matrix before differentiating into osteocytes [63]. The osteocytes orchestrate the remodeling process of bone. However, if the mechanical strain is too much the stress will induce microdamage in the matrix and potentially result in apoptosis of osteocytes weakening the matrix and potentially increasing the fragility of the bone [63]. An important point in the discussion of the pathophysiology of stress fracture is that during bone remodeling, resorption (i.e., breaking down bone) and formation are coupled, and once resorption occurs bone deposition may take up to 90 days for completion [73]. During the adaptation to stress, a temporary weakness may occur at remodeling sites of bone [52]. That potentially makes the bone more vulnerable, and if adequate recovery is not provided stress fracture may occur [57].

The understanding of the anabolic and catabolic processes of bone tissue provides a clearer picture of how and why stress fractures occur more often in military recruits during the beginning weeks of basic training. The appearance of stress fractures in new recruits is thought to occur within the first 4-weeks of basic training [21]. This period is most likely where recruits would experience new unaccustomed, repetitive, high impact mechanical stress on the body [18]. This is supported by an investigation examining 1357 United States (U.S.) Army recruits and determined that the identification of stress fractures were more likely during the early weeks of basic training [67]. These investigators also examined whether resting from running for 1-week during the second, third, and fourth weeks of basic training would reduce the occurrence of stress fracture and other overuse injuries. Interestingly, the week off of running did not provide any benefit for reducing the incidence of either stress fractures or overuse injuries. It is likely that a different recovery strategy (e.g., alternate day running schedule) would potentially provide a greater benefit [78].

The duration of marching may provide more information on the risk for stress fractures in recruits. During marching it is not uncommon for soldiers to carry loads up to 68 kg ($\sim$150 lbs) depending upon their duty position and the nature of the mission [38, 41]. The heavy weight placed in the load carriages causes biomechanical stresses to the lumbar spine of soldiers that alters kinematic behavior and subsequently may alter marching gait [74]. Not only does this cause large stress to the lumbar vertebrae, but it also may increase the stress to the lower limbs. Attwells et al. [5] reported that increases in marching loads caused a greater forward lean, which may increase the risk for overuse injuries including stress fractures. Interestingly, when the volume of marching is reduced, despite the heavy load carriages carried by recruits, the risk for stress fractures do appear to be reduced [19]. If the soldier can move through basic training unscathed by stress fracture the chances of a future incident may be reduced. Reynolds et al. [71] examining experienced 218 infantry soldiers who undertook a 161-km cross-country march carrying an average of 47 ± 5 kg load, reported only one stress fracture. It is possible that these soldiers had made positive adaptations to the stress of military training, or perhaps the more experienced soldier is more knowledgeable about appropriate foot care, footwear and load carriage packing and assembly.

3.2 *Effect of Gender on Overuse Injury in the Military*

The assimilation of females into military professions, specifically combat roles, has been controversial [17]. Despite the controversy many nations have utilized gender-integrated training, and there is an increasing involvement of women in the military worldwide [12, 86]. Considering the non-linearity of today's theatre of combat operations, the reality of women partaking in combat operations is becoming more frequent. Although the majority of female recruits embrace the opportunity to train with their male counterparts, evidence demonstrates that they are at a higher risk for overuse injury when performing the same training as men [9, 40, 49].

Women appear to suffer 3–10 times as many overuse injuries than men when training within the same unit [9, 20, 68]. Knapik et al. [40] reported that the incidence of stress fractures in men were 19.3 per 1000 recruits, but 79.9 per 1000 recruits for women. Interestingly, the gender difference observed in overuse injuries within military populations is not consistent with that seen in intercollegiate sports. The competitive female athlete appears to experience overuse injuries at a lower rate than her male counterpart [72]. These differences are likely attributed to the preparation difference between competitive and tactical athletes. Competitive athletes train in a gender specific manner within their own teams, but tactical athletes are often involved in gender integrated basic training. A lower physical capacity of women may result in a higher relative intensity during training compared to men

[43]. Interestingly, women that are training within integrated units appear to improve their performance significantly more than women who train in gender-specific units [58]. Still, if training doctrines are adjusted, to account for gender differences within mixed-gender units, evidence has been shown that women within these units can meet the physical standards necessary to complete the unit's assigned missions [17]. However, in certain close combat units there will still be an exception, primarily due to the extreme physical demands that are required of soldiers serving in those units, which are beyond the physiological capacity of the average female [17].

The increased risk for overuse injury among women compared to men has been attributed to the physiological and anthropometric differences between the genders [16, 17]. Women tend to have less muscle mass, more body fat, lower red blood cell counts, lower hemoglobin levels, and smaller cardiac outputs compared to men [60]. In addition, lower bone mineral density seen in women likely reduces the ability of bone to withstand the physical stress of basic training, and may partly explain the higher frequency of stress fractures seen among women [18]. However, this is not to say that women soldiers are unable to perform the same tasks as men. Although women appear to experience musculoskeletal injuries at a greater rate than men, injury location appears to be similar between the genders [6]. Recent research has suggested that a single trait (e.g., bone density) may not be sufficient to predict risk for stress fractures, and that a broader diagnostic approach is needed to provide a more thorough understanding of the risks for stress fracture [32]. Thus, it may be other factors that contribute to the greater injury risk observed in female soldiers.

Several studies have suggested that women entering basic training are less physically fit than men [6, 17, 20, 33]. A study examining fitness level prior to basic training in a gender-integrated unit of the Israel Defense Force showed that males had a 22 % higher aerobic fitness, 28.6 % higher anaerobic fitness, and 13.1 % upper extremity endurance than females [89]. In the same group of soldiers, only 25 % of the women were able to score equivalent to the lower 5 % of men on the VO_2 max test, and the slowest male soldier ran faster than 75 % of the females. However, when physical fitness, age, and race/ethnicity are controlled, gender may not be a significant risk factor for increased injury [6]. This supports previous investigations indicating that lower aerobic capacity and slower times during a maximal effort long distance run to be associated with greater injury risk in both men [34] and women [6, 33]. A lower aerobic capacity will cause a soldier to exercise at a higher relative intensity compared to soldier who are more aerobically fit. As such, the onset of fatigue will occur sooner, leading to changes in running mechanics and place and unaccustomed muscular stress in specific area on the body [20, 43]. Therefore, the high frequency of injuries seen in women during basic training is likely a function of lower physical fitness levels, and not strictly a gender issue.

A focus on improving physical fitness prior to basic training, or extending the training period to improve physical fitness, especially among female recruits is gaining momentum. Daniels et al. [16] demonstrated a reduction from 22 to 8 % in aerobic power difference between men and women following 8-weeks of training. The reduction in performance gap between the genders may reduce the greater risk for injury among women. In support of this concept, the Israeli military extended the training period of a gender integrated unit [20]. A more gradual increase in the frequency and volume of training per week was instituted, without changing the overall total volume of training. This resulted in a significantly lower incidence of overuse injuries in the female recruits. Thus, consideration for extending the time of basic training to provide a greater opportunity for physiological adaptation resulting from a slower onset of training volume and greater recovery may serve to increase physical fitness levels of both male and female recruits, lower the risk of overuse injury and reduce the gender gap for injury.

4 Overuse Injuries During Deployment

Deployment to areas of conflict exposes the soldier to a number of physiological, environmental and psychological stresses. During the Iraq and Afghanistan conflict the United States deployed more than two million soldiers for 6–15 months, with 40 % of the service members being deployed more than once [61, 77]. Similar to what has been previously discussed in garrisoned soldiers during basic and advanced military training, musculoskeletal injuries account for the largest proportion of medical visits and evacuations among deployed soldier [61]. Of the non-combat related injuries, the most common cause for medical visits or evacuations appear to be related to overuse [26, 76]. Roy et al. [77] prospectively examined 263 American soldiers deployed to Afghanistan. Soldiers worked approximately 6-days (5.8 ± 1.8) per week, reported standing about 10.5 ± 5.3 h per day, wore their armored vest 8.2 ± 5.1 h per day, and walked 7.4 ± 7.8 km per day with an average load of 21.7 ± 13.5 kg per day. During the yearlong deployment 62 of the soldiers (23.6 %) reported at least one injury with the vast majority (71.8 %) of these injuries occurred while working. The most common site of injury was reported to be the lower back (15.4 %), shoulder (14.5 %) and the knee (12.8 %). Sprains and strains were the most prevalent diagnosis (37.6 %), which generally occurred when dismounting during patrol (21.3 %) or lifting objects (14.5 %). Injuries were also reported to occur during sport activities (e.g., resistance training and sport participation). Deployed women in this study had a nearly 3 fold greater risk for being injured (59 %) compared to men (21 %). In addition, the heavier the load carried the greater the relative risk for injury was seen. At what height the load was carried also appeared to have an impact on injury. Loads carried at the chest appeared to result in a greater the risk for injury than loads carried at waist height. Most importantly, the risk of injury appeared to increase the longer the soldiers remained deployed. In contrast to what is generally reported during basic training, the injuries seen during deployment appear to increase with prolonged stay.

Knapik et al. [44] performed an additional study on injury rate from two deployments to both Iraq (3242 soldiers) and Afghanistan (505 soldiers). Troops were examined for 180 days prior to deployment and 180 days after deployment. The rate of overuse injury in both groups of soldiers were similar prior to deployment (ranging from 6.9 to 8.3 %). However, upon return from deployment the incidence of overuse injuries among soldiers that were deployed to Iraq ranged from 24.5 to 33.1 %, while lower levels were observed in the troops returning from Afghanistan (10.9–16.6 %). The investigators suggested that the lower rate of overuse injury in the troops returning from Afghanistan may be attributed to several factors including time of return, the physical training program of the units, demographics and differences in combat environments between the two theatres of operation.

The units returning from Afghanistan came home during the holiday season and were provided a 2-week leave. This may have provided some needed recovery or simply they did not report any concerns during this time period. In addition, the specific unit that served in Afghanistan was part of a new physical readiness training program. In contrast, the units that served in Iraq were still adhering to the traditional military training doctrine. Previous reports have suggested that the new training program is more effective in reducing injury rate during military training than the previous models [38, 41]. It was suggested that the soldiers serving in Afghanistan may have returned healthier due to the continued implementation of the new training paradigm [44]. Another potential factor contributing to the lower injury rate was soldier demographics. More than 20 % (21.2 %) of the soldiers that were stationed in Iraq were 30-years of age or older, in comparison to 13.9 % of the soldiers examined and served in Afghanistan. In addition, 7.8 % (n = 254) of the troops studied from the Iraq cohort were women, while no female soldiers were among the troops examined in the Afghanistan cohort.

The combat environment was also suggested to partly account for the differences in injury rate. The troops fighting in Afghanistan were primarily involved in classic combat operations involving patrols and search-and-destroy missions in mountainous terrain. In contrast, troops within the Iraqi cohort were focused primarily in mounted counter-insurgency operations, and dealing with improvised explosive devices in an urban environment. Differences in physical and psychological stressors were suggested to differentially influence post-deployment injury rates among the troops fighting in the different combat theatres [44]. Psychological stress is known to increase the incidence of injury through a variety of mechanisms, possibly through diminished immune function and delayed healing [14, 87]. The delay in tissue healing may increase the likelihood for an overuse injury, especially during situations of repeated trauma [44]. While the stress of combat between the different theatres of operation differed is debatable, subsequent study has suggested that soldiers deployed to Iraq had a higher prevalence rate for post-traumatic stress disorder (12.9 %; 95 % confidence intervals [CI] 11.3–14.4 %), than personnel deployed to Afghanistan (7.1 %; 95 % CI 4.6–9.6 %) [29]. Whether tasks or exposure to trauma significantly differed between theatres of operation leading to greater stress is still not clear, especially with soldiers being deployed on multiple occasions to both combat arenas. Further examination of this outcome is warranted.

Another concern with deployment is the ability for soldiers to maintain their level of physical fitness. Considering that poor physical fitness is associated with an increase in musculoskeletal injury, if fitness levels are not maintained during deployment the risk of injury may rise. Sharp et al. [80] examined 110 infantry soldiers deployed to Afghanistan for approximately 9-months (range 208–318 days). Aerobic fitness, power, strength and body composition levels were obtained 1–2 months prior to deployment and as soon as possible upon the soldiers' return (18 ± 14 days). Body mass and fat-free mass significantly decreased from pre-deployment to post-deployment (−1.9 and −3.5 % respectively), whereas percent body fat and fat mass significantly increased during the same time points (10.2 and 7.9 %, respectively). Performance on the medicine ball put (upper body power) and peak VO_2 significantly decreased from pre-deployment to post-deployment (−6.9 and −4.9 %, respectively). No changes were noted in lifting strength (box lift) or in vertical jump performance (lower body body). The decrease in aerobic capacity was likely related to the decrease in aerobic training during deployment (70 % of the soldiers self-reported a decrease in the frequency of endurance training during deployment). Interestingly, access to training facilities did not appear to be an issue as 90 % of the soldiers reported having access to a training facility. Prior to deployment 80 and 82.7 % of the soldiers indicated that they performed endurance or strength training 3–7 days per week, respectively. During deployment though only 50.9 % of the soldiers reported a training frequency exceeding 3+ days per week for endurance training, while 74.9 % of the soldiers indicated that they were able to maintain a 3+ days per week training schedule for strength training. Although significant declines in aerobic fitness and body composition were noted during a 9-month deployment schedule, this may become a much larger issue during multiple deployments or deployments of longer duration.

A subsequent study examining a longer duration of deployment was conducted on soldiers serving in Iraq. Lester et al. [48] examined a cohort of soldiers comprising a combat brigade team that deployed for 13 months. Following deployment significant increases in body mass (2.9 %), lean mass (3.0 %), fat mass (8.7 %), and body fat percent (4.7 %) were noted. In addition, significant improvements in bench press (7.4 %) and squat (8.1 %) strength were observed. Change in lean body mass significantly correlated with changes in both bench press ($r = 0.63$; $p < 0.05$) and squat ($r = 0.47$; $p < 0.05$) strength, indicating that soldiers who increased their strength also gained lean mass during deployment. Upper body power (bench press throw) was significantly increased (8.7 %), with no changes in lower body power (squat jump). Similar to the previous study, aerobic performance significantly declined during deployment, as reflected by a 12.6 % increase in 2-mile (3.2 km) run time. Similar to the Sharp study (2008), a significant decline was reported by soldiers in the frequency of endurance exercise (87.7 % exercising 3+ days per week pre-deployment compared to only 28.8 % of the soldiers exercising 3+ days per week during deployment). The decline in aerobic fitness during deployment may be a contributing factor to the higher rate of injury seen during post-deployment studies [44]

4.1 Stress Associated with Sustained Operation

During deployments soldiers tasked with enemy engagement may experience highly intense combat for a sustained period of time. During specific military operations soldiers who engage with the enemy will often approach the battlefield carrying heavy loads over prolonged distances. The loads carried by soldiers have continuously elevated over the past 150 years [38, 41], and during actual combat the loads carried by soldiers often exceed recommended military doctrine [61]. Enhanced weapon systems, improved communication equipment, and individual protective body armor, carried by the soldier, potentially results in a load that exceeds 55 kg [38, 41]. Depending upon the situation (i.e., actual fighting, approach march or emergency march) the soldier may carry between 35 and 78 % of their body mass on their back [61]. Heavy loads carried in an uncontrolled and unfamiliar environment during extreme stress and accumulating fatigue challenges the physiological limits of the soldier. In addition to the physical stress, combat operations are a fluid event that changes often at a moment's notice. As such, the opportunity for recovery, nourishment and sleep are often minimized resulting in the soldiers becoming both food and sleep deprived.

The physical demands associated with the approach and engagement with an enemy force is a large challenge facing the soldier. However, this challenge is more than physical. The psychological stress associated with combat include fear, uncertainty, and information overload [56], which may further exacerbate the risk for injury. Lieberman et al. [50] investigated cognitive and physiological changes in 31 male U.S Army officers (31.6 ± 0.4 years) from an elite unit during 53 h of simulated sustained combat in a hot (19–31 °C), humid (56–86 % relative humidity) environment. During the study, soldiers were engaged in continuous military activity. They parachuted into action, traveled in small boats, performed long treks with a heavy load, and participated in simulated combat including gunfire and explosions. During the simulated combat operations soldiers were permitted to drink ad libitum, but food consumption was minimal (single standard field ration, 1250 kcal of energy) and they slept in total 3.0 ± 0.3 h in the field. There were approximately 14.4 ± 1.0 sleep intervals, with an average duration of 12.3 ± 0.8 min of sleep per interval. During the study soldiers' mood, as assessed by the profile of mood states, increased in anger, fatigue, tension, depression and confusion, and decreased in vigor. The changes in mood were similar to that reported by Nindl et al. [62] examining simulated, sustained military operations. Lieberman et al. [50] also noted decreases in both simple (e.g., vigilance, reaction time) and complex (e.g., memory, learning) cognitive function. A 20 % decrease in response time was reported, which is greater than that normally seen during severe hypoglycemia and being legally drunk [50]. These responses were also consistent with other investigations demonstrating significant degradations in cognitive function during simulated, sustained (48–72 h) military operations [25, 51].

In addition to cognitive and psychological decrements, significant decreases in testosterone and significant elevations in cortisol reflect the elevated physiological stress associated with sustained combat operations [50]. These results were consistent with other studies examining sustained high intense combat operations. Gomez-Merino et al. [22] reported significant decreases in resting testosterone, but no change in resting cortisol concentrations following 5-days of sustained operations involving limited sleep (3–4 h per day) and 3200 kcal per day from battle rations. In addition, they reported significant reductions in salivary IgA (marker of mucosal humoral immunity) and increases in plasma IL-6, a pro-inflammatory marker. These results provided further evidence of a lowered immune response and elevated inflammatory response accompanying intense, stressful combat activity, which likely decreases the recovery ability of the soldier and increases the risk of injury.

Food deprivation, sleep deprivation, and a high volume of work, common to sustained operations have been shown to compromise both psychological and physiological systems [22, 50, 51, 61]. Compromising these biological systems may have profound effects on the physical performance capability of the soldier. The magnitude of performance decrements are likely related to the soldier's ability to hydrate, nourish and rest. Knapik et al. [39] were one of the first investigations examining the effect of sustained, short term, (5-days) combat operations on physical performance abilities of infantry soldiers. During the 5-day study soldiers performed operational related tasks, carried food rations, ammunition, communication and other necessary equipment to maintain 5-days of sustained operations. The total load carriage for each soldier ranged between 25 and 29 kg, and the soldiers slept between 4–5 h per night. The soldiers were required to perform both offensive and defensive of maneuvers in a heavy wooded terrain with thick underbrush. The average daily temperature was 18 °C with a total rainfall of 10 mm. The simulated field exercises included mission relevant tasks generally performed by infantry personnel. These tasks included raid and react, establishment of rally points, react to mortar fire and evacuate, establish a patrol base, reconnaissance of specific areas, defend two hills while reacting to enemy fire, evacuate wounded, link up with other groups, secure and hold a landing strip, and supply and move to resupply points. During the 5-day operation significant decrements were observed in both upper and lower body power, muscle endurance (push-ups and sit-ups) and cardiovascular endurance performance. However, grip strength and dynamic lifting strength was actually increased following the 5-day study. In addition, no changes were noted in shooting performance. This is consistent with previous research that suggested that simple, learned tasks can be maintained during sustained operations [25]. This study demonstrated significant declines in power and endurance during the 5-days of sustained combat operations, despite adequate sleep and food intake.

A subsequent study by Nindl et al. [62] examined 3-days of sustained combat operations with both food and sleep deprivation in 14 soldiers. However, injuries to three of the soldiers and an illness to a fourth resulted in only 10 soldiers completing both pre- and post-assessments. Soldiers were permitted two periods of sleep of 1-h per day and consumed a total of 1600 kcal per day (one meal ready to eat and a smaller meal). During the three day study significant changes were noted in the profile of mood states (increases in tension, depression, anger, confusion and fatigue, with a decline in vigor), and significant declines in body mass (2.5 kg), lean mass (1.5 kg), fat mass (1.2 kg), and % body fat (18.5–17.7 %). Performance measures were conducted one day prior to the simulated mission and during days 2 and 3 of the operation. Upper and lower body power performance was assessed by a maximal effort 30 repetitions at 30 % of the soldiers' maximal strength on the bench press throw and squat jump, respectively. No significant changes were noted in upper body power performance, but significant decrements were noted in both lower body power (−9.2 %) and the total work (−14.0 %) performed on the squat jump exercise by day 3 of the operation. Performance in military specific tasks, such as repetitive box lifting, obstacle course time and wall building were significantly reduced by day 2 of the operation (−8.3, −8.6 and 25 %, respectively). No differences though were noted in marksmanship or grenade throw accuracy, again supporting the concept that simple, leaned skills are not affected, even during situations of sleep and food deprivation. Interestingly, performance on the military specific tasks improved by day 3 of the operation, and was attributed to an enhanced level of motivation as the study completed. This is likely not realistic during actual combat, in which the stress of combat (e.g., casualties) is high, and a clear end point is not known.

In a study examining a longer duration, Welsh et al. [88] studied 29 U.S. Marines during an 8-day simulated sustained operation. During the training exercise, the marines, carrying 20 kg of equipment, were expected to perform normal operational field duties including patrols, handling weapons and ammunition and demonstrate leadership capabilities. All soldiers received approximately 4-h of sleep per day and an inadequate daily ration of 1540 kcal per day (daily energy expenditure was calculated as 3834 ± 200 kcal). Similar to previous studies, significant decreases were noted in body mass (−4.1 %), fat mass (−12.7 %) and lean body mass (−2.4 %). Jump power performance was also significantly reduced over the 8-day operation. Power during a single countermovement jump was reduced by 8.9 % (1371 ± 159 W–1249 ± 165 W from pre to post assessments, respectively), while mean lower body power output during a 5-consecutive countermovement jump test declined by 9.1 % (1291 ± 89 W–1173 ± 94 W from pre to post assessments, respectively). In addition, during a 30-jump test average power output declined 8.6 % (1133 ± 204 W–1036 ± 125 W from pre to post assessments, respectively). Comparing the results of the study by Welsh et al. [88] (an 8-day sustained operation) to Nindl et al. [62] (a three day sustained operation) revealed a similar decline in jump power. However, the soldiers in the 3-day study received less sleep and provided less energy per day than the Marines in the 8-day study, suggesting that sleep and food deprivation can accelerate the onset of performance decrements.

4.2 Unique Issues Associated with Sustained Operation in Special Forces

Elite units within a military are often tasked with demanding missions, such as difficult infiltration/exfiltration circumstances, camouflage, and limited communication. During certain circumstances these operators are required to maintain stealth and provide only observations. During intelligence gathering type operations soldiers may be immobilized (i.e., lying prone 24 h/day with restricted spaced movement) for a prolonged period of time (8–12 days). These special support and reconnaissance (SSR) missions have been shown to cause physiological changes that may occur during extreme periods of inactivity [15, 84]. One study examined a real time simulated SSR mission in which five special operation soldiers from the Danish military infiltrated stealthily wooded area for an 8-day mission to observe a farmhouse for potential weapon sales [15]. Soldiers were tested prior to infiltration and immediately upon return. During this covert operation the soldiers remained in a prone position for 3-h rotations with noise discipline. They were provided 3 L of water and 3564 kcal of food rations per day. The temperature during the days ranged from 8–14 °C and between 3 and 9 °C in the evenings. A degree of threat was constant as guard dogs and helicopters were used to patrol the area. The soldiers were situated approximately 200 m from the farmhouse and had to maintain their stealthness. During the 8-day mission a 4 % loss in body mass and a 5 % loss in fat free mass was reported. In addition, vertical jump height (a measure of lower body power) was significantly reduced by 8.2 %, while maximal force was reduced by 9.2 %. However, the most profound effect was noted on the rate of force development, in which a 21.9 % decline was observed. Additional data from an expanded version of this study reported a significant increase in type IIx muscle fiber type area, and trends for an increase in the number of type IIx muscle fibers ($p = 0.09$) and an 11 % decrease in type I muscle fibers ($p = 0.06$) [84]. No change in capillary density were noted. The change in type IIx muscle fiber size may have resulted in possible changes in pennation angle, explaining the change in the rate of force development. These marked declines of muscle function; neuromuscular capability, maximal strength, functional performance and morphology could potentially compromise operational performance and overall mission success, especially during a 'hot' extraction.

4.3 Environmental Effects on Soldier Performance During Sustained Operations

As discussed earlier, the physiological stresses associated with sustained operations can be magnified when these missions are conducted under additional environmental stresses. For instance, during combat operations in Afghanistan it was reported that many missions were "aborted" due to altitude sickness and combat ineffectiveness [75]. Many of these operations in Afghanistan were conducted at altitudes of 3000 m (ranging from 2500 to 4382 m) [75]. In one operation (Operation Anaconda) conducted by the U.S. military in Afghanistan, 14.6 % of the casualties were cases of acute mountain sickness [66]. The sojourn to altitude, without the benefit of acclimatization, is by itself highly stressful. At altitudes above 2500 m the ability of soldiers to perform simple and complex military tasks become significantly reduced, including target engagement speed during tests of marksmanship [46]. The effect of altitude, coupled with the heavy load carriages, inadequate nutrition and sleep during a sustained combat operation in a difficult terrain increases the risk for both musculoskeletal injury and acute mountain sickness [75].

Significant challenges are also noted during military performance conducted in both hot and cold environments [3, 30, 50, 54]. However, the specific physiological stresses associated with these environmental challenges are beyond the scope of this chapter. It is recommended for further information on physiological challenges associated with environmental extremes readers should examine Armstrong [4], Cheung [13], or Pandolf and Moran [64].

5 Summary

The most common non-combat related injury among military personnel, during either training or deployment, is related to the musculoskeletal system. These injuries are often a function of the cumulative effects of repetitive microtrauma, and are classified as overuse injuries. These injuries generally occur when the intensity and volume of training exceed the recovery capacity of the soldier. The risk for an overuse injury in among military personnel appear higher among female soldiers, soldiers that are less fit, or have a high running volume, previous injury, or in soldiers that smoke.

During deployment a higher risk of injury is associated with the soldiers' load carriage, and the height in which this load is carried (i.e. chest vs. waist height). Fitness levels are still an important component of injury risk, and the ability to train aerobically does become a challenge when deployed. Interestingly, strength levels do appear to be maintained in deployments ranging from 9 to 13 months, and is likely related to the training facilities available on base. The risk of injury does appear to increase the longer the soldiers remained deployed. This is likely related to decreases in fitness as deployment is prolonged. During sustained operations soldiers may engage the enemy carrying large loads in difficult terrains, with added stress of both food and sleep deprivation. These stresses appear to result in significant declines in body mass, lean body mass and fat mass. Further, significant declines in mood, cognitive function, muscle endurance and lower body power are often noted. Physiological stresses are also reflected by a lowered immune and elevated inflammatory response, which may increase the soldier's susceptibility for injury. These stresses may be further exacerbated if sustained operations are conducted in an extreme environment.

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The Mechanophysiololgy of Stress Fractures in Military Recruits

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Abstract Stress fractures (SFs) are of the most common and potentially serious overuse injuries. Many athletes, naïve exercisers, and military recruits who are engaged in frequent and repetitive activity may suffer a SF; the most common site for SF is the tibia. SF is regarded as fatigue fracture—when training yields bone strains in a range where the micro-damage formation in the bone exceeds the ability of a remodeling process to repair it and ultimately this cumulative tissue damage might result with a spontaneous fracture. The registry of SFs among athletes is incomplete, but in military recruits the incidence of SFs range between 5 and 12 % (female soldiers are 2–10 times more prone to SFs compared to their male counterparts). Recovery from a SF is primarily achieved by halting any load bearing activities and on rest. This might be detrimental to athletes and military recruits, as results in loss of training days and consequently a reduction in physical capacity. The ample risk factors for SFs can be categorized as internal factors depending on the individual (e.g. gender, bone geometry) and external factors (e.g. training volume). It follows that in many cases SFs are preventable. Recruits engaged in a reasonable level of physical activity, especially impact exercise in the years prior to joining the military, and also maintain adequate nutrition, may lower their risk for SFs. Yet, several fundamental issues in regard to SFs are still left unresolved. For example, how muscle forces provide a protective effect against SFs, how many cycles (i.e. steps or strides) can an individual perform before he or she will be at a risk of suffering a SF, or is it necessary to implement prophylactic interventions in order to protect those who are identified at a greater risk? New experimental tools

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and improved computational modeling frameworks for investigating and better addressing the above questions that are reviewed in this chapter can be used to improve the knowledge on the etiology and prevention of SFs.

Keywords Fatigue fractures · Overuse · Physical activity · Biomechanical modeling

1 Introduction

Overuse injuries comprise more than one quarter of all injuries among competitive athletes [1]. For example, up to 70 % of the runners sustain overuse injuries during a one year training period [2]. Stress fractures (SFs) are of the most common and potentially serious overuse injuries. For instance, tibial stress injuries were found to be the most common injury among high school runners (41 % for girls, 34 % boys) [3]. Current methods of injury classification in sport injury epidemiology substantially underestimate the burden of overuse injuries, since in most cases there is no time loss involved [4]. Registry of SFs among soldiers is more commonly reported, however, it is highly variable as the reported incidence ranges from 1 to 62 % [5–9]. During the last decade there is a trend towards a decrease in SFs rates among soldiers with 5–12 % incidence reported [5, 10, 11]. Noteworthy, female soldiers suffer 2–10 times more SFs than their male counterparts [12–16].

For the athlete and military recruits, SFs are detrimental, as they are associated with pain symptoms that require rest, thus resulting in loss of training days [1, 17]. This loss of training days might hamper the injured soldier military readiness. Yet, the operational consequences of training days lost due to SFs could not be thoroughly assessed since the military records in this respect are lacking. In a 9-week basic training follow-up study, a total of 3.6 % of training days were lost, mainly due to bone stress injuries [18]. Among athletes, the literature regarding loss of training days is more abundant; Devas reported that the average time period before returning to full activity following the diagnosis of a tibial SF was more than 3 months [19]. More recently, it was reported that more than 80 % of the SF-injured athletes missed participation in training for more than 3 weeks [1]. In another study that examined high-grade SFs, the mean time to return to sport activity at the pre-injury level was 143 days [20]. In severe symptomatic or asymptomatic untreated SFs the outcome might result in a complete or displaced fracture, requiring surgical intervention [21–23]. The recommended time periods for reduced physical activities among soldiers range from 1 week for a low grade tibial SF, to 24 weeks for a complete femoral fracture [24].

For the active athlete and soldier populations, a more appropriate definition for a SF is 'fatigue fracture'. A fatigue fracture indicates the repeated stress loading on a bone that is lower than the failure stress associated with a single loading event that cause bone tissue failure, resulting in a partial or complete fracture [25].

2 Bone Loading

In its role of providing internal support, the skeleton is exposed to repetitive bouts of mechanical loading, which result in bone deformations and strains. The long bones of the lower extremities are loaded in compression, tension, torsion (producing shear at the transverse plane), and bending at the longitudinal bone direction (Fig. 1). Bone is typically weaker in tension than under compression. Thus, when bending causes a crack (the more frequent mode of bone failure), it will be apparent on the tensioned side of the bone [25]. In response to the loading, deformations and strains will occur on and within the bone tissue, and their distribution and magnitudes will depend on the physical, material and geometrical/structural properties of the bone. Bones also exhibit viscoelastic material properties, which means that changing the loading or deformation rate (and thus the strain rate, or the strain change divided by the time interval during which strain has been applied) will result in different stiffness properties or an elastic (storage) modulus of the bone tissue. The higher the strain rate is, the greater is the elastic (storage) modulus.

During walking, axial loads acting on the tibia range between 2 and 4 time the body weight (BW) [26–29]. Therefore, in soldiers who are marching while carrying loads that in some scenarios may equal their own BW, the resulting axial tibial loads can be as high as 8 times their BW (~ 5500 N). During running, even without any backload, the loads on and within the bones can reach about 13-times the BW [30]. Additionally, the strain rate in the loaded bones will be approximately 3-times greater than the strain rate during walking [31].

These loads normally result in bone strains which are safely below the (catastrophic) failure strength, or even the micro-damage accumulation zone (> ~ 3000 $\mu\varepsilon$) [31]. However, the nature of military activity, that is, the repetitive pattern of walking and running during training, may eventually lead to bone fatigue.

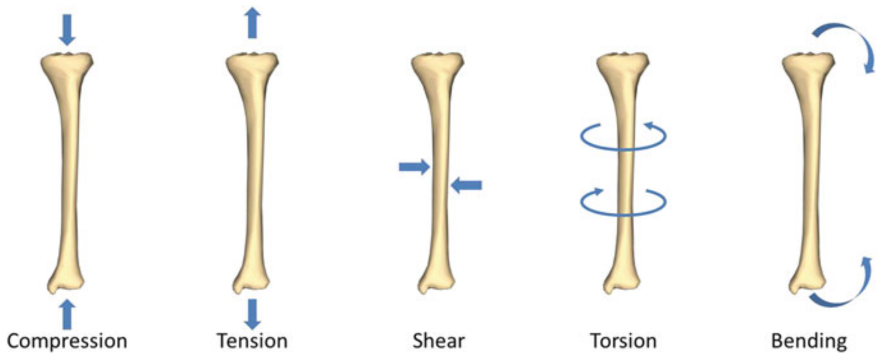

Fig. 1 The loading modes which are possible for long bones. In real-life physiological activities, the loading is compound, that is, these different loading modes act concurrently, and the femur and tibia for example are subjected to different levels and combinations of these individual modes

3 Bone Functionality

Healthy bones should have such structural characteristics (combined geometries and mechanical properties of the tissues) that provide adequate stiffness and strength in order to support and withstand the daily loads, both static and dynamic, without failure (i.e., proper functionality). To this end, bones have a unique, complex shape, which pose difficulties in calculating the loads and resultant strains using simple analytical, mathematical models. Nevertheless, for representing the whole-bone stiffness (or resistance to deformation) behavior in a simplified manner, e.g. when bones are subjected to physiological bending loads, an engineering approach adopted from the beam theory is used, as follows. Whole bone stiffness is calculated as the product of the tissue elastic modulus (E), which is a measure of the material stiffness of the bone, and the cross sectional area moment of inertia (I), which is a geometrical feature. This indicates that the bending resistance of long bones is influenced by both their material stiffness (e.g. density and quality of mineralization), and the amount of that material which exists to support/resist the load (in simple words, how thick is the bone). The elastic modulus, representing the bone material properties, depends primarily on the matrix mineralization and porosity of the tissue [32]. The cross sectional moment of inertia, representing the structural properties, depends on the mass distribution of bone material along the axis. If more bone mass is located distal to the axis, there will be a greater moment of inertia, and therefore, greater resistance against bending. When considering the bone resistance during pure axial compression loading, the cortical area (CtAr) of the bone is of most importance. A higher CtAr value will result in greater resistance against compression.

Many studies which are relevant to the topic of this chapter quantify the bone mineral density (BMD), an aerial density parameter measured by dual-energy X-ray absorptiometry (DEXA). In the context of SFs, the BMD is used as a parameter to explain gender differences in bone mechanical behaviors [11]. Although it is a useful clinical measure for detection of pathologic, massive bone loss and osteopenia, it might not be sensitive enough and capable to accurately detect subtle variations in bones of healthy young populations [33].

Poor bone geometry parameters, such as a (substantially) smaller than average cortical area for females [5], smaller area moment of inertia [34, 35] and smaller bone width [5, 36–38] were previously associated with increased SF risk.

Utilizing advanced imaging techniques, like peripheral quantitative computerized tomography (pQCT), allowed more reliable and accurate measurements of material and geometrical parameters, including for the purpose of studying the interactions between them. Utilizing this imaging technique, Jepsen et al. [39] showed in a cohort of healthy young subjects that bone robustness, which is the total bone area divided by the bone length, correlated with the calculated bending stiffness. This was further validated against actual bone bending stiffness measurements using cadaveric bone specimens. Their data demonstrated that a slender tibia is two to three times less stiff relative to body size, compared to a robust tibia.

More interestingly, a complex association was found between robustness, cortical area, and the elastic modulus of bone tissue. Specific changes in the cortical area and elastic modulus of bone appeared to serve as compensatory factors for lower bone robustness (Fig. 2) [39]. In a later study, Jepsen et al. [40] showed that there are multiple biomechanical and biological pathways leading to an increased SF risk. Specifically, the combined effect of whole bone stiffness, elastic modulus, and the moment of inertia, was associated with the occurrence of SFs in men. For men engaged in elite-forces training, 78 % of the SFs could be attributed to bone slenderness, or impaired functional adaptation. The major contribution of this functionality approach was that 22 % of the individuals who suffered a SF had average bone robustness, but reduced CtAr and BMD. These authors surmised that the individuals with the increased SF risk have experienced a failure to accumulate adequate CtAr during their growth, and also, that their matrix was either less mineralized or more porous, or both.

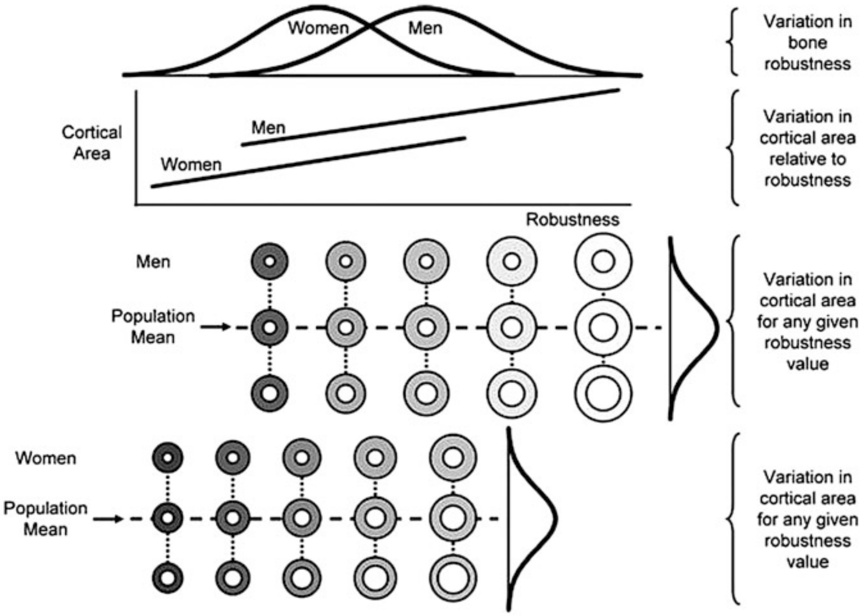

Fig. 2 A schematic diagram to illustrate how variations in cortical area are superimposed on the variation in robustness. Tibial diaphyses are represented as idealized *circular cross-sections, with grey values* representing the variation in elastic modulus of bone tissue. With the body size effects excluded, the cortical area varies with the level of robustness, with slender tibias showing lower CtAr than robust tibias. In addition, there is a variation in CtAr for any given robustness value. These relationships hold for both men and women. The natural variation in robustness is associated with specific changes in the cortical area and elastic modulus, which need to be considered when studying the effect of genomic factors and environmental factors on bone structure and function (reprinted with permission from [39])

4 Bone Remodeling and Modeling

Bone is a fatigue-prone living structure that is characterized by an ability to adapt to repeated mechanical loads by changing composition, micro-architecture, density and shape, which repairs and minimizes the fatigue damage. However, if the rate of damage accumulation exceeds the rate of tissue repair, SFs, and ultimately fractures will occur. Bone remodeling and modeling activities underlay the control of physiological bone strains, which are caused mainly by ground reaction and muscle forces (Table 1) [41].

Remodeling in bone is accomplished by synchronized activity of osteoclasts and osteoblasts, which are cells that resorb and deposit bone, respectively. These cells assemble into discrete temporary anatomic structures called basic multicellular units (BMUs). The lifespan of osteoclasts and osteoblasts is short compared to the lifespan of the BMUs, and therefore, cells must be continually replenished for maintaining the activity in the BMUs. A remodeling period in a BMU normally takes about 3 months. During this process, a temporary small gap (resorption bay) is created due to the osteoclastic action [41]. When the mechanical stimulus is low, resulting in bone strains which are chronically lower than 50–100 με (i.e. disuse), remodeling will overall cause bone resorption, leading to osteopenia. When the strains are chronically high (i.e. mechanical overload), remodeling results in the deposition of bone and hence the overall gain of bone mass [41, 42]. It has been suggested, based on a large volume of experimental data, that osteocytes (which were formerly osteoblasts that were captured in the bone matrix, and which are distributed throughout the bone tissue) are the bone's mechanosensing cells. Hence, osteocytes are also the specialized type of cells in bone which are capable to sense fatigue damage and thus activate localized remodeling, by attracting osteoclasts and osteoblasts, in order to repair the damage [42].

Modeling, as opposed to the above-described remodeling process, constitutes bone development, maturation, and shaping. This is a process that is typical mainly during the pre-natal stage of bones until adolescence and young adulthood. During

Table 1 Physiological bone strain ranges (based on Frost [41])

Range (με)	Definition	Description
50–100	Remodeling threshold	The minimum effective range (or equivalent factor) that helps to control remodeling
~1000	Modeling threshold	Strain above this range can make modeling strengthen a bone to reduce later strains
~3000	Microdamage threshold	The range above which the amount of new microdamage exceeds the ability of a remodeling process to repair it timely, so tissue damage begins to accumulate and can cause a spontaneous fracture if no rest has been given.
~25,000	Ultimate strength	Normal lamellar bone's fracture strength expressed as a strain threshold

adulthood, modeling rate is decreased, but still take part in changing outside bone diameters and is responsible for slow and intermittent bone drifts [43]. Modeling, much like remodeling, utilizes osteoblasts that slowly deposit (organic) bone tissue on larger surfaces compared to remodeling in BMU. Osteoclasts may also participate, and remove bone from other (larger) surfaces, which is distinguishable again from the localized processes of remodeling. Hence, while normal remodeling essentially maintains and repairs bone tissues, modeling slowly determines the whole bone shape (including for example its length, cross-section and outer diameter) as well as its apparent mechanical properties such as stiffness and strength. The dynamic strains that are required for modeling in bone tissue should exceed a threshold of ~1000 με in adolescents and young adults [41].

5 Pathophysiology

Other than mechanical stimulation, other conditions such as: diseases, exposure to certain drugs, or hormones and nutritional deficits may alter normal remodeling and modeling. Estrogen, for example, normally lowers the remodeling strain threshold. This would explain the association of osteopenia with amenorrhea in young females [44].

The safety factor between naturally occurring, physiological strains (which are in the range of 400–1500 με) and a single strain loading event causing failure (~10,000 με) in bone is rather large (Table 1). Nevertheless, repetitive strains below a single load–failure threshold are also harmful [45]. Typically, repetitive low strains are of little pathological consequence, as bone is capable of self-repair through targeted remodeling, as mentioned above. Under certain biomechanical conditions, however, imbalances can develop between the extent of accumulation of damage and its repair. This is because osteoblasts and osteoclasts have certain (finite) numbers, rates of recruitment to the damage sites and capacity of biological effects such as tissue resorption and deposition output per cell, which they cannot exceed. At ~3000 με the subsequent accumulation of micro-damage is believed to be the threshold of a pathological continuum that is clinically manifested as stress reactions, SF, and ultimately complete bone fractures. With increasing strains or strain rate, the number of loading cycles a bone can withstand before fatigue failure is hence reduced.

Remodeling and its given physical and biological capacities (per the individual) is the limiting factor in the cycle of bone repair. Thus, damage accumulation and consequently the development of a SF may result from cyclic over-loading, which occurs when remodeling is given insufficient time to repair damage and additional loading cycles enable damage to accumulate, as the bone tries to repair itself. Therefore, factors that increase the number of loading cycles contribute to the development of a SF [45].

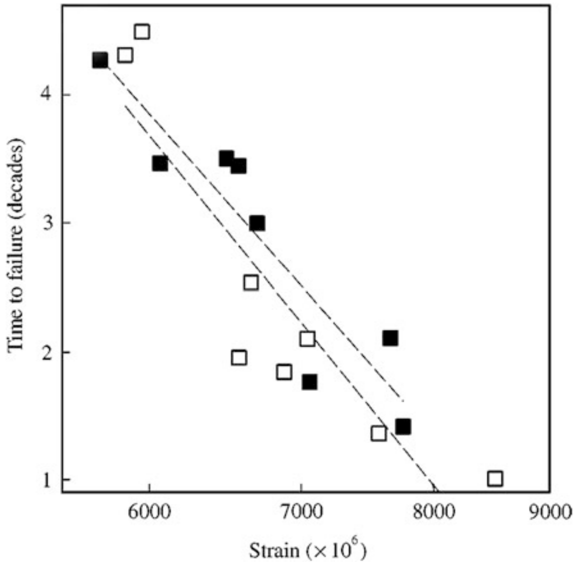

Fig. 3 Tensile fatigue behavior of human bone specimens. Time to failure as a function of strain. The exponent n in 10^n s. *Open squares* 5 Hz; *solid squares* 0.5 Hz (reprinted with permission from Ziopos et al. [46])

The fatigue life of bones is determined as the number of loading cycles that are needed to cause a fatigue fracture at a given strain amplitude and frequency. The relationship between the number of loading cycles to failure or the time to failure and the strain magnitude is logarithmic [46] (Fig. 3). Thus, when a bone is repeatedly loaded under conditions that result with 2000 με, the bone's fatigue life is ~10,000,000 cycles and for loads that result with 4000 με, the fatigue life will be 500 times lower, <20,000 cycles [41].

In regard to loading rate (i.e. the strain rate), cyclic loading at high physiological strain rates causes more damage than cyclic loading at a lower strain rate (e.g., 0.03 and 0.01 s^{-1} as measured during running and walking, respectively) due to the viscoelastic nature of bone which then produces greater bone stresses for the higher strain rates. Morphologically, these fatigue processes correspond to increased numbers of microcracks throughout the bone matrix [47]. A systematic review of the literature with regard to ground reaction forces, loading rate, and SFs found that instantaneous vertical loading rates (and not magnitudes), were significantly different between individuals with SFs and healthy subjects [48].

6 Diagnosis and Treatment

Stress fractures should be suspected in any individual who is engaged in an intense sport activity and who complains about bone or musculoskeletal pain. Pain appears with daily activities, mainly walking or running, or during physical examination, in which direct or indirect tapping on the limb elicits pain. Clinical examination

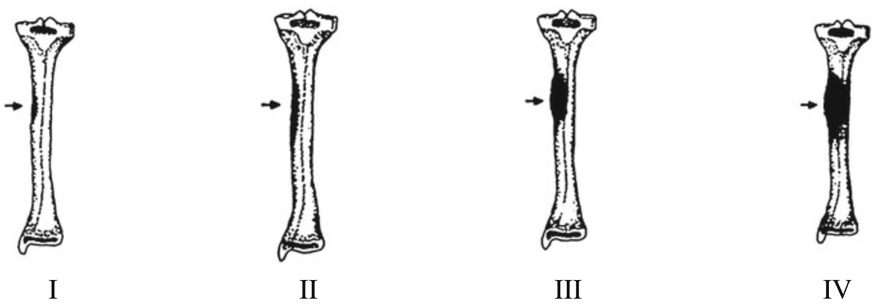

Fig. 4 Stress fractures classification (grade I–IV). A schematic presentation as seen on bone scintigraphy. Originally published by Zwas et al. [52]. © The Society of Nuclear Medicine and Molecular Imaging, Inc.

includes localizing the pain and eliciting tenderness [49]. It is noteworthy, however, that even for a well-experienced physician, not all cases can be diagnosed by means of clinical examination, and up to approximately 50 % misdiagnosis had been reported [24]. Thus, to establish the diagnosis, imaging techniques are commonly used.

Radiography can be used for the detection of SFs. Although being very specific, its sensitivity is low and depends on the time from the onset of symptoms and the affected bone. Shortly after the appearance of symptoms, radiography may be negative, and radiological evidence may be seen only after several weeks. Findings may include lucent zones, periosteal new bone formation, focal sclerosis, callus formation or later a fracture or cortical cracks [50, 51].

Bone scintigraphy is a very sensitive imaging technique, though less specific. It reflects a semi-quantitative bone remodeling activity, thus allows grading the severity of the SF starting from an early stage (Fig. 4). Grade I: Small ill-defined cortical area of mildly increased activity in the cortical region. Grade II: Larger well-defined elongated cortical area of moderately increased activity in the cortical region. Grade III: Wide-fusiform area of highly increased activity in the cortico-medullary region. Grade IV: Wide extensive lesion with intensely increased activity in the transcortico-medullary region [52]. One major drawback of this method is the large amount of ionizing radiation that the patient is being exposed to; it is estimated to be equivalent to 4200 extremity radiographs, which increases the likelihood for a related malignancy [53].

In recent years, magnetic resonance imaging (MRI) has been considered the most sensitive and specific imaging modality for diagnosing early stress injuries [54, 55]. An MRI SF grading scale, in line with the bone scintigraphy grading, was published by Fredericson et al. [49]. Noteworthy, however, MR imaging may be too sensitive in some cases, as it detects even subtle bone changes, which have no clinical implications, especially in asymptomatic patients. Nevertheless, MRI could be used as an early detection modality, given that large lesions observed in MRI scans at day of induction tend to progress during military basic training [56].

Recovery from a SF is based primarily on halting any load bearing activities and on rest. Recommended resting periods range from one week for a grade I tibial or metatarsal SF, and up to 8 weeks for a grade III or IV femoral SF. When a complete fracture is detected, the treatment is similar to that of a traumatic fracture, and can last 12 weeks for a tibial, and 24 weeks for a femoral fracture [24].

Femoral SFs are considered the most serious overuse injury. In severe cases, they might accompany with a displaced fracture of the shaft or the femoral neck, which would require surgical intervention and in some instances might even result in a nonunion [57, 58].

7 Epidemiology

The Incidence of SFs among military populations is highly variable, and ranges between 1 and 62 % [5–9]. This high variability is explained mainly by differences in study designs (e.g., prospective versus retrospective, active surveillance versus self-reported, or passive surveillance). Therefore, it is difficult to generalize or pool data from such different studies. Factors that influence the acquisition, results and interpretation of data include differences in definition of the injury and the method employed for diagnosis, with its inherent accuracy (e.g., clinical compared to radiological) [59].

The understanding that a SF is an injury that affects military readiness underlay a change in training programs which ultimately reduce the number of injured soldiers. Thus, comparing data from different eras may be misleading. One example is the experience attained by the Israel Defense Forces (IDF). During the early 1980s SFs were common in the IDF training programs and the incidence of SFs was 31 % among Israeli infantry recruits, as reported by Milgrom and colleagues in 1985. The same group of investigators reported a sharp decrease in the SF incidence within two decades, to 12 % in the same unit [10, 60]. This dramatic decrease in incidence was attributed mainly to fundamental changes in training programs that have been revisited, revised, developed and adopted with the aim of minimizing SFs. While 12 % SFs among soldiers is still a considerable rate in comparison to the approximately 5 % incidence reported by others [5, 11], it is a major step forward which exemplifies the importance of the regimen of physical activity as well as the difficulties in comparing incidence data from different armies and military units.

In the non-military population, SFs are mostly prevalent among runners and track and field competitive athletes. SF incidence was found to be 21 % in a one year follow-up prospective study, with the highest incidence observed in distance runners [61]. Likewise, SFs have been observed in 37–42 % of ballet dancers [62, 63]. High incidence of SFs was reported among elite gymnasts as well [64], and also, in figure skaters [65]. With lower incidence rates, SFs also occur in other fields of sports that include repetitive stress, such as basketball, soccer, baseball, softball etc. [66].

Although SF were reported to occur in most parts of the skeleton, 20–60 % of the SFs in the active population occur in the tibia [15, 34, 66–68]. Femoral and metatarsals SFs are also abundant among military recruits [13, 69].

8 Risk Factors

Risk factors for SFs are numerous, and can be categorized as internal factors depending on the individual and to external factors.

8.1 Internal Factors

- A female gender was the most commonly identified intrinsic factor for SFs. Women performing the same prescribed physical activities as men incur SFs at incidences that are 2–10 times greater than men [12–16]. This might be explained by gender-related hormonal differences. Estrogen deficiency during growth and ageing is one of the most important factors in the pathogenesis of bone fragility [70–73]. Estrogen deficiency increases bone remodeling and is accompanied by osteoclastogenesis, which result in a negative basic multicellular unit balance [74]. Female athletes with a history of SFs reported significantly older menarche age and were more likely to have experienced a history of menstrual disturbances compared to healthy individuals [75]. It was also reported that female distance runners with a history of irregular or absent menses and who have never used oral contraceptives may be at an increased risk for developing SFs [76].
- Sedentary lifestyle, or activity with relatively low frequency (<2 times/week) were found to significantly increase the risk for SFs. This is most prominent among young men and women military recruits who are entering basic military training with a poor history of physical activity, or poor current physical fitness level. [15, 16, 77, 78]. Factually, regular exercise protects against SFs and a longer history of exercise further decreases the relative risk of fracture [15]. Among different types of activities, basketball seem to have a protective effect [79].
- The effect of age on the incidence of SFs is currently under dispute. One military study indicated that older age may heighten the risk of SF [13]. Another study, however, showed that SF incidence was inversely proportional to age [8]. The largest study examining the relationships between age and SF incidence ($\sim$2400 recruits) found no association between them [78].
- Caucasians have been found to be at a higher risk to incur SF than Afro-Americans [13, 78, 80].
- Lower limb morphology, such as leg-length discrepancy [81, 82] and genu valgum significantly increase the risk to develop a SF [82, 83]. High-arch feet

were associated with femoral and tibial SFs, while a low-arch structure of the foot was associated with metatarsal SFs [84].

- Bone geometry—narrower tibias for males, and smaller cortical areas for females were identified as significant risk factors contributing to SF incidence [5]. Smaller tibial area moment of inertia [34, 35] and smaller tibial width were found to be associated with a greater SF incidence [36–38].
- Lower thigh muscles section area have been associated with SF, in both genders, compared to healthy control subjects [5].

8.2 Extrinsic Factors

- Active smoking and a history of smoking increase the risk for SFs [85, 86]. The association of history of cigarette smoking with exercise related injury occurrence (including SFs) was consistent for both men and women undergoing military basic training. The effects of smoking on injuries appear to persist at least several weeks after cessation of smoking [87]. Generally, smoking is more abundant among military populations compared to civilians [88], especially civilian athletes; thus, smoking and bone health among civilian populations is focused mainly on bone loss in late adulthood. Among adult men, smoking was correlated with higher long-term bone loss [89].
- Although there are reports of different SFs distributions among different sports types, few associations can be made between particular sports and fracture patterns in collegiate athletes [66]. Distance runners were more likely to suffer a long bone (i.e. tibia, femur, and fibula) or pelvic SFs, whereas foot fractures occurred more frequently in jumpers, sprinters, hurdlers, and multi-event athletes [61].
- Large training volume (e.g. cumulative long distance marches) were found to be correlated with higher SFs incidence in soldiers [90]. Mann et al. [91] have reported that exceeding 100 km per week of walking, marching or running raised the SFs incidence.
- Footwear is supposedly a risk factor, but the evidence is still debated. A review of 16 randomized trials concluded that there is insufficient evidence of preventive interventions to draw firm conclusions. However, there is limited evidence suggesting that the provision of "shock absorbing" insoles or "customized" orthotics in the boots of military recruits could reduce the overall incidence of SFs and stress reactions of bone [92, 93]. Noteworthy, there are no studies among athletes investigating the effects of the shoe type or shoe quality on the risk of SFs [15].
- Training surface as a risk factor has also not been resolved. An epidemiological study suggested that the running surface does not play a significant role in the pathogenesis of running injuries [94]. However, on the basis of lower in vivo strains and strain rates that occur during treadmill running, Milgrom and

colleagues concluded that treadmill runners are likely at a lower risk of developing tibial SFs compared to overground runners [95].

- Muscle strength (in a non-fatigue state) was found to be lower in female athletes with lower limb SFs, suggesting that muscles strength may have a protective effect [38]. However, Frost [41] suggested that bone strains are caused mainly by muscle forces and less due to ground reaction forces. Further studies are needed to determine the specific effects of muscle forces on SF development in non-fatigue conditions. Fatigue places higher demands on the postural control system by increasing the frequency of actions needed to regulate the upright stance [96] which, over time, could alter the loading patterns in the lower-limb skeleton. Although some studies suggest that fatigue is associated with a decreased impact transmission that may protect the leg, it is commonly accepted that fatigue is leading to an altered foot and lower-extremity loading [97–99]. This, for example, can explain the incidence of SFs of the metatarsals under fatiguing loading conditions. Furthermore, Mizrahi et al. [100] reported that with developing fatigue, an imbalance in the contraction of the shank muscles appears in parallel to an increase in the shank shock acceleration. Milgrom and colleagues found that tibial tension strains increased 26 % post-run and 29 % post-march (i.e. in a fatigue state) compared to these strain levels at the pre-run phase. The corresponding tension strain rates increased 13 % post-run and 11 % post-march [101]. In several different experimental methods fatigue had been attained, but no direct clinical link between fatigue and SF occurrence had been shown. The effects of fatigue on the etiology and probability of SF development deserve further studies.

9 Prevention

SF is a preventable injury. In this regard Jones et al. [15] have presented that modulation of training regimens of soldiers, which included reduction in the amounts of running and other weight-bearing activities were the most promising approach to reduce the incidence of SFs without compromising on the soldiers' physical fitness. Nevertheless, during real-world military training programs, modifying training programs in order to reduce load bearing activities and running is not always applicable, as these are basic activities that are typically needed to fulfill a soldier's mission [102]. In that context, the Cochrane collaboration review of interventions for preventing and treating SFs concluded that "There is insufficient evidence from randomized trials of preventive interventions to draw firm conclusions".

There is limited evidence from randomized trials suggesting that the provision of "shock absorbing" insoles in the boots of military recruits reduce the overall incidence of SFs and stress reactions of bone [103]. In one earlier study, the use of customized shoe orthotics could substantially lower the SF rate in military recruits who were training with military boots [93].

Calcium and vitamin D supplements were successful in reducing SF rates among female military recruits [86]. Among male military recruits, calcium supplementation alone did not have an effect on SF rates [104].

The risk to incur SF is a multifactorial but tackling the major risk factors listed above—low physical activity history and poor bone strength are among the most important ones. Accordingly, conscripts preparing themselves for military service (prior to drafting) can lower the risk to develop SF if they train adequately, which will help in reducing substantially the incidence of SF during basic training. Human studies among military recruits have shown that pre-induction ball players, and especially basketball players, have a protective effect against SFs. This was explained by the high strains and strain rates that were measured at the tibia during basketball games, which were 2–5.5 greater than those measured during walking, and 10–50 % more than during running [79]. Consistent with the above findings, animal studies have demonstrated that by enhancing the structural properties of a bone through a mechanical loading program, its fatigue resistance could be significantly improved [105]. Specifically, using a mechanical loading program to induce bone adaptation, the aforementioned study reported that small (<2-fold) changes in the structural properties of rat ulnas increased the fatigue resistance of these bones >100-fold [105]. This indicates that an adequately planned, continuous and moderate exercise program in preparation for the intense army training routine may be an effective preventative strategy to protect individuals from SFs during basic army training. Importantly, when considering the sudden increase in intensity of physical exercise on the day of recruitment to combat training, such pre-military-service exercise programs should be implemented a sufficient time before induction.

Seeman [106] previously claimed that puberty is the only opportunity for bone strengthening. However, there are data indicating that exercise strengthens bone before and after puberty as well. For example, general school-based physical activity increased bone health in elementary school children of both genders, particularly before puberty [107]. Among young adults (age of about 18), an increased level of physical activity was associated with increased cortical bone thickness, periosteal circumference, and trabecular bone mineral density. The lowest reported effective level of physical activity in that study was less than 4 h per week [108]. In a study conducted in groups of twins aged between 7 and 20 years, exercise during growth appeared to have even greater skeletal benefits than variations in protein or calcium intakes in their diets [109]. Moreover, a recent systematic review of studies investigating the interactions between nutrition, physical activity and bone mass in children and adolescents, suggested that physical activity and calcium intake play an important role in bone acquisition and development during growth, independent of the pubertal status [110]. Summarizing the above literature, it follows that in order to minimize SFs among military recruits physical preparation should not start at the day of induction. Rather, they should be engaged in a reasonable level of physical activity, especially impact exercise (involving jumping and hopping, such as basketball or other ball games), throughout their lifespan. To ensure a maximal beneficial effect on bone tissues, adequate nutrition containing sufficient calcium intake, should be maintained as well.

10 Modeling Studies of Stress Fracture Etiology and Risk Factors

Unlike epidemiological studies that rarely enable to withdraw firm conclusions regarding etiology, mechanisms of tissue damage or risk factors, in vivo studies where strain gages were surgically attached to bones have added some useful information regarding the biomechanical conditions which may lead to SFs in humans [31, 95, 111]. Nevertheless, such strain gage measurements are conducted at specific points on the bone surface (i.e. at the gage attachment sites) and thus, are likely to miss additional areas that are exposed to similar or even greater strains, which can be important in describing the damage cascade and etiology.

Computer modeling is a powerful tool that can assist with further isolating the various biomechanical risk factors, which is helpful in understanding how SFs occur and who are the most susceptible individuals. One example for such study was reported by Edwards et al. [30]. A probabilistic model of bone damage, repair, and adaptation was developed by these authors to determine the effects of stride length and running mileage on the likelihood of occurrence of a tibial SF. Peak instantaneous joint contact forces served as inputs to a finite element (FE) model, which was an important component of their modeling framework, and was used to provide the estimates of tibial strains during stance. The results of this study suggested that bone strain magnitudes play a more important role in SF development than the total number of loading cycles. Edwards and colleagues' recommendation, based on their modeling outcomes, was that runners wishing to decrease their probability for tibial SFs may benefit from a 10 % reduction in stride length. The above pioneering modeling study shed light not only on the mechanisms of SF development but also on the usefulness of computational modeling in encountering the problem and improving guidelines and training programs aimed at lowering SF risks in real-world scenarios, which is beyond the basic science understanding.

A rat experimental model of tibial SFs, which was conjugated with a computational FE model has been developed by our group [112]. This coupled animal-computational model system added, for the first time, some fundamental insights regarding the etiology. In this experimental model the loading regimen applied to the rat tibia (forces and number of cycles) was well-controlled in the animal, and determined quantitatively in terms of localized tissue strains and stresses in the FE modeling [112]. This approach has recently been extended to humans, and a model of the complete human shank (not just the isolated tibia), containing for example soft tissue components such as the ligaments and muscles of the shank had been developed (Figs. 5, 6). Our current modeling facilitates visualization and quantitative analyses of 3D distributions of strains and stresses on and within the tibial and fibular bone tissues, in several physiological and particularly military activity scenarios, including carriage of heavy weights, marching, running and resulted muscular fatigue. Specifically, variabilities in bone dimensions such as

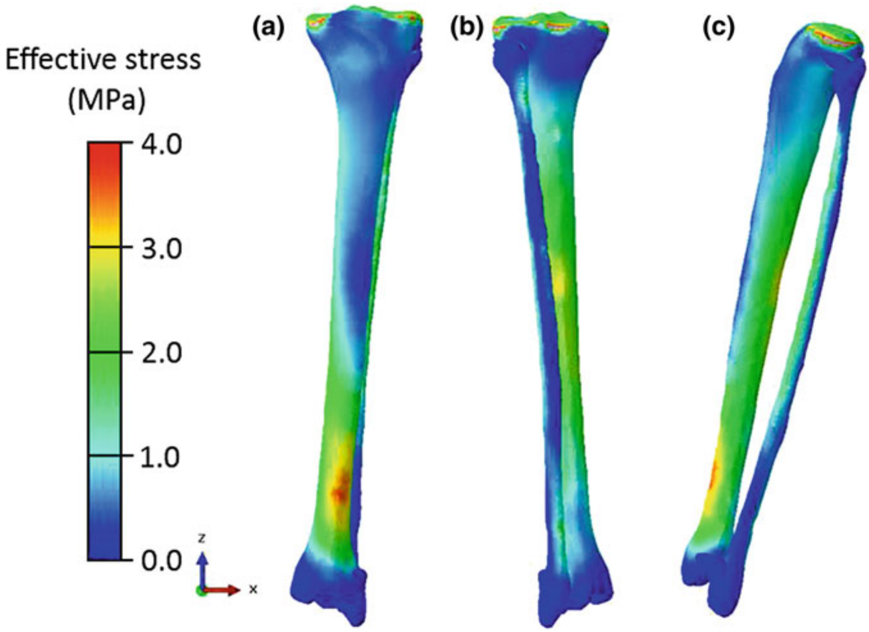

Fig. 5 Example maps of effective stress distribution over the surfaces of the tibia and fibula during walking, showing the anterior (**a**) posterior (**b**) and lateral surfaces (**c**)

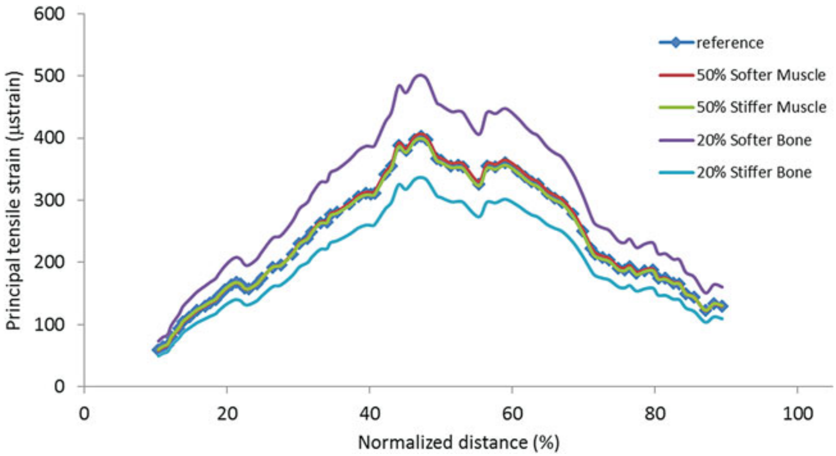

Fig. 6 Maximal principal tensile strain distribution along the posterior tibia for the reference model and tissue mechanical property variant configurations. Data are presented for 10–90 % of the length of the corresponding tibial path in order to exclude boundary effect

length and cross-sectional area, tissue stiffness properties (associated with the level of bone mineralization), muscle stiffness and muscle forces (including in fatigued states) can all be considered in our present modeling framework.

11 Conclusion

A copious number of studies and review papers have been published on the etiology and biomechanics of SFs in the last 30 years or so, which built a rich knowledge base regarding mechanisms of SF development and risk factors. The empirical and computational results and the conclusions that were drawn throughout this long period did not provide, however, a conclusive, firm, and simple-to-implement tool for the practitioners and clinicians who train or treat recruits and soldiers. Based on the current accumulated knowledge, which has been compiled in this chapter, we surmise that SFs in military recruits can be prevented indeed, or at least substantially minimized. However, several fundamental issues are still left unresolved, as follows. First, how are muscle force production patterns able to provide a protective effect that reduces or redistributes bone strains as related to SFs, and whether strengthening muscles can indeed be an effective methodological intervention in recruit populations? Second, how many cycles (e.g., steps or strides) can an individual perform before he or she will be at a risk of suffering a SF, and how does that activity limit vary across individuals, and affected by intrinsic and extrinsic characteristics including for example the bone density, muscle endurance, foot structure, footwear type etc.? Finally, is it necessary to provide prophylactic interventions in order to protect those who are identified to be at a greater risk? New experimental tools and improved computational modeling frameworks for investigating and better addressing the above questions should improve knowledge regarding the individual bone strength and endurance, and hence, understanding the SF tolerance in different recruit populations.

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The Biomechanical Basis for Increased Risk of Overuse Musculoskeletal Injuries in Female Soldiers

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Abstract An increasing number of women are serving in militaries around the world. Overuse musculoskeletal injuries (OMI) are common with military activities in both sexes but are more common in female soldiers, in part because of differences in whole body and tissue-level biomechanics. Sex-based differences in whole body biomechanics such as stride length, knee valgus, and others may help explain differences in OMI risk. Further, tissue-level sexual dimorphisms in body composition, muscle, and bone, also contribute to the higher risk of OMI in female soldiers. Understanding these biomechanical differences will help militaries tailor preventative measures towards female soldiers at high risk of OMI.

Keywords Overuse musculoskeletal injury · Biomechanics · Military · Sex differences · Gender

1 Introduction

An increasing number of women are serving in militaries around the world, and in some countries such as the United States, combat roles are opening to women who can meet physical criteria for certain operational specialties. Further, in other countries, such as Canada, Taiwan, New Zealand, and Israel, women have been serving in combat roles such as infantry positions for some time with documented success [1]. With an increase in women serving in militaries, and particularly in combat positions, it has become more important to understand the physical

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demands that military service places on female warfighters. Injuries are common with military activities in both sexes due to the physically-demanding nature of training and combat [2]. However, overuse musculoskeletal injuries (OMI) are of particular concern for female soldiers as these injuries are more common in women, in part because of differences in whole body and tissue-level biomechanics. Understanding these sexual dimorphisms will help us better understand how to prevent OMI in female soldiers and ultimately allow for effective integration of women into combat roles. Therefore, the purpose of this chapter is to review the epidemiology of OMI in male and female soldiers, review biomechanical differences in male and female soldiers, and discuss the implication of these differences in regards to countermeasures for preventing OMI in female warfighters.

2 Overuse Musculoskeletal Injuries

Overuse musculoskeletal injuries (OMI), known also as cumulative trauma disorders, may be defined as tissue damage that is a result of repetitive demand over the course of time [3]. OMI are the result of repetitive micro-trauma to the connective tissue such as tendons, ligaments, muscles, bones, and cartilage. OMI are distinct from and more subtle than traumatic injuries which are usually the result of a single, traumatic event (macro-trauma) [2, 4]. Since OMI usually result from repetitive, cyclical, weight-bearing activity, they are most common in the lower extremities and in physically active populations, such as recreational and competitive endurance athletes, and in those with physically demanding occupations such as military personnel [4, 5].

Among all injuries in US military services, OMI have been found to have the greatest impact on the health and readiness of US Army personnel than any other category of medical complaint [6]. Further, these injuries are the leading cause of lost training days in the US Army, reaching an annual cost of two billion dollars [7]. During basic combat training (BCT), about 25 % of males and about 50 % of females experience one or more training-related injury. About 80 % of these injuries are in the lower extremities, are considered overuse injuries, and include minor muscle strains, contusions, tendinopathy, fasciitis, bursitis, muscle or tendon tears or ruptures, joint sprains or complete ligament tears with joint instability, joint dislocation, bone fractures, cartilaginous disruptions, bone stress reactions and stress fractures, and other related injuries [8].

In a prospective cohort study which examined injuries and injury risk factors among 660 British Army infantry soldiers during a pre-deployment training cycle, it was reported that most injuries which occurred during training involved the lower body (71 %), especially the lower back (14 %), knee (19 %) and ankle (15 %). Activities most associated with injury included sports (22 %), physical training (30 %), and military training/work (26 %) [9]. Similar findings have been reported in the Israeli Defense Forces (IDF) where OMI accounted for 90 % of all orthopedic injuries in a study conducted on 18,651 soldiers. Low-back and lower extremity injuries accounted for the vast majority (71.5 %) of these injuries [10]. Since OMI have a direct impact on the soldier's physical performance and combat readiness and are associated with military discharge, these injuries are a primary concern for militaries.

3 Overuse Musculoskeletal Injuries in Female Warfighters

While OMI can be attributed to sport and occupational activities in both sexes, it has been noted that certain injuries are especially prevalent in female athletes and soldiers in comparison to their male counterparts. Females ultimately have higher attrition rates from the military service (Fig. 1). There are several reports that overall medical discharge of women from military service due to an injury is 2–3 times higher than men in the British Army [11], and 1.5–2 times higher in the US Army, US Marine Corps, and the Australian military [12–14].

Among the most prevalent OMI that differ between physically active men and women are knee pain and injuries, anterior cruciate ligament (ACL) ruptures, and stress fractures [5]. Patellofemoral pain syndrome (PFPS), as reflected by anterior knee pain (AKP), is one of the most common OMI among physically active individuals between the ages of 15 and 30 [15] and is diagnosed as AKP throughout

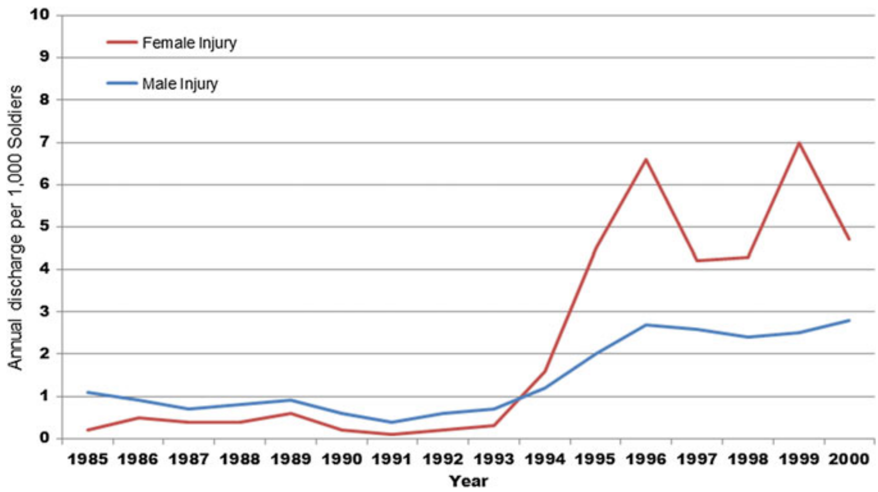

Fig. 1 Musculoskeletal injury discharge rates for males and females in the British Army, 1984–2000 [103]. Annual discharge rates due to musculoskeletal injury in the British Army were approximately two times higher in females (*red line*) than in males (*blue line*) from 1993–2000, coinciding with an increase in opportunities for females to enter military training establishments on the same terms as men

and after physical activity, during bodyweight loading of the lower extremities in walking up or down the stairs, squatting, banding, jumping, in sitting with the knees flexed, and rising from a chair after sitting [16, 17].

Military recruits undergoing initial training experience abrupt increases in physical activity and are thus considered high-risk population for developing AKP [18]. This phenomenon was reported to reach an incidence of 8.7 % among British Army recruits [19] and 5–15 % among US Army and IDF recruits, respectively [20], and is considered the primary cause of attrition from military training which has a debilitating effect on individuals, restricting or ending participation in physical activity.

As previously published in the scientific literature, females have been found to suffer from OMI 2–10 times more frequently than their male counterparts. These studies report 1.5–2.2 times higher incidence of PFPS [21, 22], 8–9 times greater incidence of ACL [23, 24], and 2–10 times greater incidence of stress fractures [25].

There are a number of proposed explanations for the higher prevalence of OMI in women [26], including speculation regarding psychosocial reasons such as occupational stress felt by females when entering a field that has traditionally been male dominant, a sense of obligation to perform in similar absolute demands [27, 28], and sex socialization which plays a role into the increased reporting of OMI by females in comparison to males, especially in integrated units [29, 30]. However, the predominant scientific explanation for increased risk of OMI in females is the numerous sex-based differences in the biomechanics of women and men. In the following sections, we review both whole body and connective tissue-level biomechanical differences between men and women that predispose women to OMI.

4 Whole Body Biomechanical Differences

There are many proposed differences in body-level biomechanics between men and women, such as the knee quadriceps angle (Q-angle), dynamic frontal plane alignment, femoral anteversion, patellar tilt angle, and peak torque knee extension [5, 17, 31, 32]. As described above, one of the most common regions of injury is the knee joint, with women experiencing higher rates of PFPS than their male counterparts. We will first discuss sex-based anatomical and biomechanical differences at and around the knee joint and then discuss sex-based differences in kinematics and kinetics during dynamic tasks such as walking, running, and landing from a jump. Whole body biomechanical differences during movement may also explain the increased incidence of PFPS in women compared with men, as well as provide a partial basis for other common overuse injuries in female service members performing these activities during military training or active duty.

4.1 Biomechanical Differences Affecting the Knee Joint

There are several well-documented sex-specific anatomical differences at the knee joint. Variation in knee valgum (knock-knee) and varum (bow leggedness) affect distribution and magnitude of patellofemoral joint stress by altering the Q-angle [33, 34]. The Q-angle is defined as the angle formed by the line in pull of the quadriceps and the patellar tendon as they intersect at the center of patella, which measures the tendency of the patella to move laterally when the quadriceps contract [35]. The Q-angle is derived from a static measurement taken while the subject is in a standing or supine position with flexed and unactuated quadriceps [36, 37].

According to the literature, a normal Q-angle is considered 14 ± 3 degrees and 17 ± 3 degrees in men and women, respectively [38]. While the literature consistently demonstrates that females have a greater Q-angle compared with their male counterparts, the reasons for this sex difference are still unknown. Although previously it was thought that the greater Q-angle in females was a result of having a wider pelvis compared with males, this has been well disputed [39, 40], and other explanations were suggested. Recently, in a study conducted on 218 (102 males and 116 females) college-aged students without any current injury in the lower extremity or any previous history that would affect the alignment of the lower extremity joints, the researchers suggested that greater Q-angle in females may be a result of females having greater structural femoral anteversion and tibiofemoral angle compared with males [41].

Regardless of the morphological basis of sex difference in Q-angles, a larger Q-angle as seen in women is associated with PFPS. For example, in a meta-analysis which analyzed 523 variables related with PFPS in 47 studies, significantly larger knee Q-angle was associated with PFPS incidence [17]. The mechanistic underpinnings of the association with larger Q-angles and PFPS are not entirely clear; however, larger Q-angles have been shown to increase lateral translation of the patella and lateral patellofemoral joint stress [33, 34]. In turn, chronic loading with high patellofemoral stress has been implicated as a cause of overloading of the subchondral bones of the knee joint and deterioration of the articular cartilage [33].

Q-angle differences are not the only sex-specific anatomical differences at the knee. Females are also more likely to have patella alta, a greater than average distance between the kneecap and the knee joint [42]. When the kneecap is in a higher position, the knee requires a greater degree of flexion before the trochlear groove and its soft-tissue tension stabilize the kneecap. Additionally, females may have greater knee volatility due to many factors that contribute to patellar instability, including trochlear dysplasia and the presence of a trochlear bump, quadriceps dysplasia (defined as lateral patellar tilt on CT scans), and differences in tibial tuberosity-trochlear groove distance (defined as a measurement of distance between the two sites on the CT scan) [43]. In support of these anatomical differences is the observation that females have more recurring patellofemoral dislocations [44]. All of these factors have been found to be more prevalent in females, and therefore likely contribute to the greater risk for patellar instability and knee injuries in women.

In comparison to males, females have increased dynamic measures of knee valgus angle, hip internal rotation angle, hip adduction moment, knee valgus moment, and decreased dynamic measures of knee flexion angle [45–49]. These sex-specific kinetic and kinematic differences are consistent across several activities such as walking, running, and landing from a jump—all common repetitive movements during military activities. The sex-based biomechanical differences during these activities are discussed below.

4.2 Biomechanical Differences During Dynamic Activities

4.2.1 Walking and Cadence Marching

Walking and cadence marching are common activities with military training [50], and sex-based differences in gait mechanics during these activities have been demonstrated in the scientific literature. During walking at a set pace, women have a higher rate of stepping due to shorter stride lengths [51]. In military training, cadence marching is a common activity, and the average male stride is often implemented by default. In this scenario, the smaller average stature and related decreased stride length in women compared to men places females at a disadvantage and at increased risk of OMI, especially in regards to stress fracture risk [52, 53]. However, sex-based differences in gait mechanics exist even at self-selected walking speeds [54]. In a study of 55 men and 36 women walking at self-selected speeds, women walked with greater pelvic obliquity and arm swing, a more stable torso and head, and had greater transverse plane pelvis and torso rotation than men [54].

4.2.2 Load Carriage

Often during military activities, loads are carried which affect gait mechanics and possibly magnify sex-based kinematic differences. In a study conducted in order to determine the effects of loads worn or carried on walking mechanics, it was concluded that males and females displayed significantly different gait patterns under no load and under four different load conditions of 9, 17, 29 and 36 kg consisting of standard military items [51]. According to this study, both males and females displayed significantly different gait patterns under all load conditions. However, females demonstrated disproportionately greater changes in gait mechanics, with a larger increase in forward inclination of the trunk during walking than men with the two greatest loads, thereby demonstrating a greater sensitivity to load magnitude. The researchers concluded that not only is it important for load requirements to be lower for females because of physiological differences but also because of biomechanical differences and the associated mechanical stresses which must be endured during locomotion [51]. Contrary to these findings, a recent study reported that when men and women walked with no added load or an additional 10, 20, or 30 % of body weight, no significant differences between men and women were detected for any measured gait parameter. The authors concluded that men and women adopt similar gait adaptations when carrying load at the same percent of body weight [55].

4.2.3 Running

Only 1.5 % of a basic trainee's time is spent running [50]. However, this means that on top of all other physical activities such as marching, lifting, climbing, walking and participating in calisthenics, a soldier will average approximately 10 h of running during eight weeks of basic training [50]. Therefore, running mechanics may play an important role in risk of OMI, and like other activities there is evidence that sex-based differences exist in mechanics of running [56]. Female recreational runners had greater peak hip adduction, hip internal rotation, and knee abduction angles compared to male recreational runners when running at the same speed. Further, female runners had greater hip frontal and transverse plane negative work compared to male runners [56]. While this study indicates sex differences in kinematic and kinetic gait patterns with running, further research is needed to determine how these differences lead to sex differences in injury rates.

4.2.4 Landing from a Jump

Knee injuries are common with physical endeavors that involve jumping, and altered landing mechanics are increasingly implicated as a contributor to ACL injuries [57]. As in other repetitive loading activities associated with increased risk of OMI, the kinematics of landing differ between sexes. For example, in a study of male and female NCAA athletes performing drop landings, the researchers reported that female athletes landed with more initial ankle plantar flexion, peak-stance ankle supination, knee abduction, and knee internal rotation than male athletes [58]. Similarly, in a study of landing biomechanics in male and females athletes, female athletes landed with greater peak knee valgus [59]. Evidence exists that these sex-based differences in landing mechanics manifest during puberty [60]. Whereas no sex differences in peak knee abduction were reported before puberty with drop vertical jumps, females increased peak knee abduction with landing between the first and second year of measurement, with the greatest sex differences in knee abduction following puberty [60]. These findings are congruent with the notion that many of the biomechanical sex differences at both the whole body and tissue level, manifest during the critical years of peak growth during puberty.

5 Tissue-Level Biomechanical Differences

Sex-based differences in whole body biomechanics are rooted in tissue-level differences in body composition and in the mechanics and function of connective tissues such as muscle, bone, ligament, and tendon. The antecedents of these sex-based differences manifest in puberty when greater rates of growth in boys compared to girls track with greater rates of body and muscle mass accrual, followed by greater deposition of bone tissue [28, 61–63]. These sex-based differences that begin in youth provide the structural and functional basis for adult sexual dimorphisms that confer greater risk of OMI in women with physically-demanding activities. In the following sections, we review sex-based differences in the composition, morphology, and function of these tissues.

5.1 Body Composition

By the time men and women reach 20 years of age, women weigh 14–18 kg less, have 30 % less lean mass, and are 13 cm shorter than their male counterparts [61, 64]. Furthermore, women have a higher body fat percentage than men (20–25 % and 13–16 %, respectively) [65], with a 10.4 % higher average body fat percent in women compared to men after controlling for body mass index [66]. These sex-based differences in body fat persist throughout the lifespan. A 20 year-old man and woman with a BMI of 23 kg/m^2 will have an average body fat percent of 13.3 and 26.0 %, respectively, while at 80 years of age, their body fat percent will be 23.9 and 32.6 %, respectively [67].

While women generally have a greater amount of body fat than men, women often have adipose tissue distributed less centrally and more subcutaneously, unlike men who have more central visceral fat which may contribute to a higher metabolic risk profile with increasing adiposity [68]. These differences in adiposity likely have their origins in youth. During puberty, percent body fat decreases in boys and increases in girls. Girls accrue 1.14 kg of fat mass per year during puberty while in boys, fat mass remains constant and fat free mass is gained at a greater rate than in girls [61].

These anthropomorphic differences between the sexes may translate to greater difficulty for the average female, compared with the average male, to safely and efficiently complete physically-demanding military tasks. For example, in a study designed to identify the effects of sex and body adiposity on physiological responses to the stress of wearing body armor, women had a greater increases in ratings of perceived exertion with addition of body armor during a treadmill test compared to men. In this study, body fat was identified as the best predictor of treadmill test completion [69]. Despite the greater body fat percent of women, female soldiers have been shown to have favorable changes in body composition with basic military training, with a decrease in body fat percent of 7.1 % in women completing 7 weeks of Army basic training [70].

5.2 Muscle

Like adiposity, there are sex-based differences in muscle mass and strength. As reviewed in the previous section, men have approximately 30 % greater lean mass than women [64]. Studies suggest greater muscle strength accompanies greater lean mass in men. In a study comparing muscle characteristics of men and women, women had 45, 41, 30, and 25 % smaller muscle cross-sectional areas for the biceps brachii, total elbow flexors, vastis lateralis, and total knee extensors, respectively [71]. The sample size of this study was relatively small (8 men compared to 8 women). Nevertheless, others have reported similar findings. In a study of 989 and 987 active duty men and women, respectively, sex-based differences in muscular strength of 57.9–103.7 % were reported for three different tasks (maximal weight lifted to 152 cm height, maximal isometric force on an upright cable pull, and maximal isometric handgrip force). The magnitude of differences in muscular strength were smaller when fat free mass was considered; however, sex-based differences in muscular strength still remained [72].

Although several studies have reported 40 % less muscle strength in both the upper and lower body of female soldiers compared to male soldiers [73, 74], other studies report disparities in sex-based strength differences between the upper and lower body. For example, Miller et al. [71] reported that women are 52 % as strong as men in the upper body and 66 % as strong as men in the lower body. Similarly, Beckett et al. [75] reported that muscle strength of the upper and lower body was 55 and 40 % lower, respectively, for woman compared to men. In this later study, strength differences remained even after adjusting for body weight (40 and 25 % differences in upper and lower body, respectively), suggesting that discrepancies in muscle strength between men and women cannot be explained solely by differences in body size. Differences in distribution of muscle strength disparities between upper and lower body regions parallel differences in muscle mass distribution between sexes. For example, women have been shown to have 39.7 and 57.7 % of their skeletal muscle mass in their upper and lower body, respectively compare to men who have 42.9 and 54.9 % of their muscle mass distributed between their upper and lower body, respectively [76].

Lower muscle mass and strength imposes a disadvantage on the ability of female soldiers to perform military activities such as carrying loads, lifting objects, and dragging injured service members or casualties. Nevertheless, studies have reported positive increases in muscle mass and strength of female soldiers following military training. Following 10 weeks of basic military training, 58 % of soldiers gained fat free mass; however, more female soldiers (88 %) gained fat free mass than male soldiers (36 %) [77]. Similarly, Knapik et al. [78] reported that both male and female recruits improved in leg extensor strength but female recruits improved more than male recruits in upper torso strength (9.3 and 4.2 %, respectively) and trunk extensor strength (15.9 and 8.1 %, respectively) following Army basic training. The greater gains in muscle mass and strength in female soldiers in these studies may reflect the lower baseline muscle mass and strength. Nevertheless, they provide evidence that the deficits in strength in women can be attenuated with physical training.

The above mentioned sex differences in body composition and muscle tissue likely place females are greater risk for OMI, especially when exposed to absolute physical demands as required in military scenarios, such as backpack carrying, lifting ammo cans and mortars, wheel assembly, and more.

5.3 Bone

Stress fractures are costly and common injuries with military training, and female recruits are more than twice as likely to suffer a stress fracture as their male counterparts [6]. As previously mentioned, sex differences in stride length, load carriage and other whole-body biomechanics partially account for increased risk in stress fracture. However, sex differences in bone morphology also likely contribute to the discrepancy in fracture risk. Women are not only at greater risk of stress fracture in young adulthood but are also at greater risk of osteoporotic fracture in later life [79]—both suggesting an inherent difference in the mechanical properties of the bones of women and men. Like sexual dimorphism in adiposity and muscle mass, the roots of intrinsic sex differences in bone morphology manifest during growth. The timing of peak bone mineral content velocity differs between boys and girls, with girls peaking at 12.5 years of age and boys peaking a year and half later at approximately 14 years of age [62, 80]. In the 2 years around this peak accrual of bone mineral content, both sexes gain approximately 25 % of adult mineral. However, during each of the 2 years prior to and after peak bone mineral velocity, boys accrue, on average, 407 g of bone mineral while girls only accrue 322 g on average [80]. A greater increase in the amount of bone mineral added to the growing skeleton in boys compared to girls is not the only advantage that creates a stronger bone in men compared to women. Rather, during growth, boys have a greater expansion in the diameter of the long bones, creating a more mechanically competent skeleton when compared to girls [63, 81].

The importance of periosteal expansion during growth is implied in the narrower tibias of fracture cases compared to controls in Israeli Army Recruits [82], and regardless of sex, greater long bone cross sectional diameter and thicker cortices are associated with decreased fracture risk in both male and female military recruits (Fig. 2). Compared to men, on average, female recruits have narrower bones with thinner cortices, and therefore, lower bone bending strength [83] (Fig. 2). These biomechanical differences result in mechanical disadvantages that help explain why female recruits have a higher risk of stress fracture with military activities.

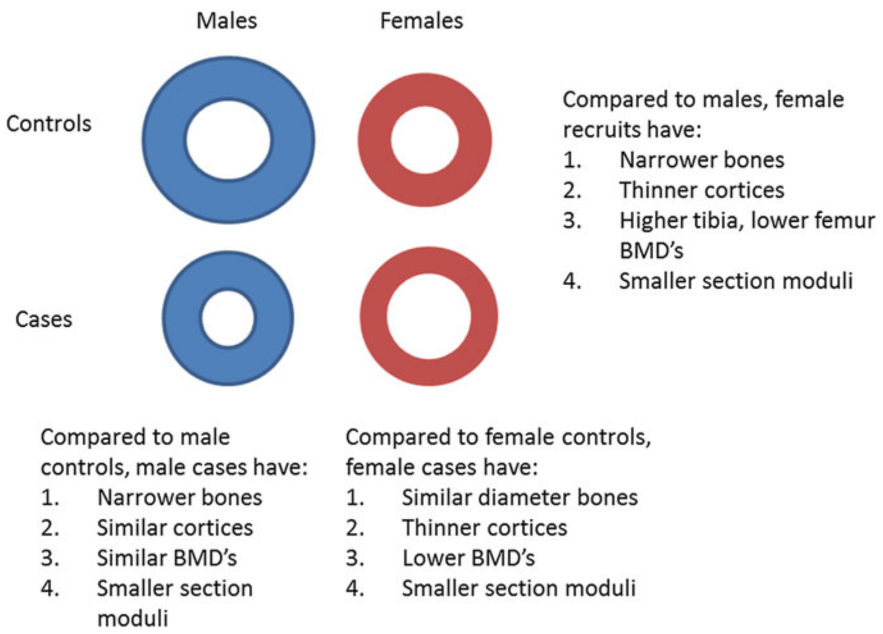

Fig. 2 Differences in cross-sectional bone geometry between stress fracture cases and controls and each gender. Recreated from [83]; Differences exaggerated for clarity

5.4 Ligament

Anterior cruciate ligament injury (ACL) is the most common ligament injury with military and athletic activity, with female athletes experiencing two to eight times greater risk of ACL rupture than male athletes [23, 84]. In the U.S. Military, male and female service members experience ACL injury rates of 2.95–3.79 cases per 1000 person-years [85]. ACL injuries occur most often with deceleration, coupled with a sudden change in direction with a planted foot [86]. As more women enter combat military positions that require increased explosive physical activities and "cutting" maneuvers, there may be a rise in incidents of ACL injuries in female soldiers reflecting the greater incidence in their athletic counterparts. However, this remains to be observed. Previously mentioned whole-body biomechanical differences such as a greater Q-angle and tissue level differences such as lower muscle strength may play a role in decreased joint stability and predisposition of females to ACL injury [87]. However, there may also be tissue-level differences in the mechanical properties of the ligament that increase risk of injury. For example, females have ACLs with an average cross-sectional area that is smaller, on average, than the ACLs of men even after adjusting for body size [88]. This smaller size may partially explain the 95 % greater peak strain of the anteromedial bundle of the ACL in the female knee compared to the male knee [89]. Similarly, the ACLs of women have been shown to have poorer material properties than the ACLs of men, with a 8.3 % lower strain to failure, 14.3 % lower stress at failure, 9.4 % lower strain energy density at failure, and 22.5 % lower modulus of elasticity in an in vitro study of ACLs tested to failure [90]. These studies suggest that the tissue-level mechanics of the ligaments, not just whole-body biomechanical differences, contribute to predisposition of ligament injury in females compared to males.

5.5 Tendon

Like ligaments, tendons are a common source of musculoskeletal injury with both athletic and military activities [6, 91]. Tendinopathy has been reported to be the second most common injury among male Army trainees, behind low back pain, and the third most common injury in female trainees behind muscle strain, stress fracture, and sprain [6]. The most common locations for tendinopathy with military training is at the achilles tendon and the patellar tendon [6], and as with other musculoskeletal tissues, there are also reports of sex-specific differences in the material properties of tendon. For example, in a study of the viscoelastic properties of the tendons, women had a 44 % lower stiffness, a 25 % lower Young's modulus, and 51 % lower hysteresis of the medial gastrocnemius tendon and aponeurosis with maximal plantar flexion [92]. Further, not only have baseline rates of collagen synthesis been shown to be 55 % lower in females compared to males, but also collagen formation following exercise has been shown to be 47 % lower in women than men [93]. The mechanisms of these sex-based differences are unclear; however, these studies suggest more favorable material properties and mechanosensitivity of tendons in men than in women.

6 Practical Solutions for Mitigating Risk of OMI in Female Warfighters

In summary, there are well-established sex-based disparities in whole body and tissue-level biomechanics that may help explain the greater rates of OMI in female military service members. In this section, we offer several practical solutions for the higher level of OMI risk conferred on females in the military. One possible solution is to identify the most physically fit women for specific military operational specialties. Another potential solution is for women at highest risk of OMI to perform physical fitness training before entering into military training. Unfortunately, the majority of whole body biomechanical characteristics such as Q-angle and stride length are intrinsic risk factors that are unmodifiable. Nonetheless, body composition, physical fitness, and bone, muscle, and tendon strength are all modifiable with physical training. We discuss the scientific evidence that supports each of these potential solutions below.

6.1 *Identifying the Most Physically Fit Individuals for Military Occupational Specialties*

One potential way to address the increased risk of OMI in female service members is to determine what the physical requirements are for a military occupational specialty and to require all candidates, regardless of sex, to meet these requirements. Such an approach may decrease risk of OMI by requiring that a warfighter be capable of performing physically-demanding tasks. In our review of the literature, we have been highlighting the *average* biomechanical difference between males and females. Naturally, some women will have greater physical advantages than some men. For example, Fig. 3 [94] shows the distribution of isometric upright pull strength among male and female soldiers. Consistent with other reports of disparities in muscle strength capabilities of men and women [95], men demonstrated greater average upright pull strength than women. Nonetheless, if a requirement to join a specific military occupational specialty was to demonstrate the ability to perform an isometric upright pull of 95 kg, then according to Fig. 3, approximately 15 % of female soldiers could pass the test, while approximately 4 % of male soldiers would fail it.

The reason OMI may be decreased in both sexes when the most physically fit warfighters are chosen for physically demanding tasks, is the strong relationship between physical fitness and OMI in the military [96, 97]. In a prospective study of 1568 women entering the Army, women who failed a 5-min step test had a 76 % higher stress fracture incidence and a 35 % higher incidence of other musculoskeletal injuries [96]. Similarly, prior to basic combat training, female trainees were categorized into quartiles by number of completed pushups in 2 min. The injury rate during training was 57 % for women who completed the least number of pushups and 38 % for women completing the greatest number of pushups [97]. In this same

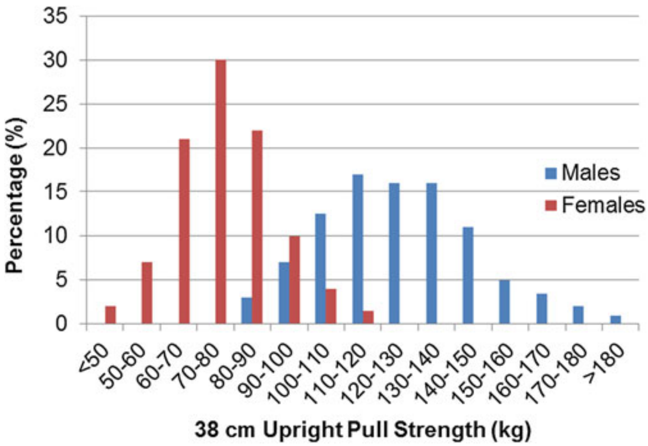

Fig. 3 Upright pull strength in female and male Soldiers in the US Army. Recreated from [94]

group, the relative risk of injury for women who could not lift more than 34 kg was 1.4 compared with women who could lift 46 kg or more [97]. There is some evidence that the gender gap in OMI risk may be overcome by physical fitness. In a prospective study of almost 400 female and male trainees followed through basic combat training, the fastest 50 % of women on a mile run test had an 18.5 % incidence of time-loss injuries compared to the slowest 50 % of men with a time-loss incidence of 29 % [12]. These findings suggest that selecting the most physically fit females for physically-demanding military duties has potential to reduce sex-based disparities in OMI. Furthermore, they suggest that increasing physical fitness prior to beginning military training will help reduce risk of OMI in both sexes.

6.2 Preconditioning Before Entering Military Service

As reviewed above, there are sex-based disparities in tissue-level biomechanical properties of muscle, bone, and tendon that contribute to increased OMI in female warfighters. Nonetheless, all of these tissues are sensitive to mechanical stimuli and have been shown to hypertrophy in response to increased physical activity [91, 98, 99]—thereby providing the theoretical basis for promoting exercise interventions in military candidates prior to military training. Several preconditioning programs prior to military training have been implemented with success in preventing OMI and improving pass rates [100–102]. In a study of over 9000 recruits undergoing basic military training, a 4–6 week preconditioning training program lowered medical attrition rates compared to both a control cohort and a cohort in which training was gradually increased to prevent attrition [101]. In a similar observational study, recruits entering basic combat training were divided into cohorts of fit, unfit with preconditioning prior to beginning basic training, and unfit with no preconditioning. The preconditioning cohort had a relative risk of injury of 1.5 for men and 1.2 for women (relative to the fit cohort) compared to a relative risk of 1.7 for men and 1.5 for women in the no preconditioning cohort. The proportion of trainees completing 9 weeks of basic training were only 59 % of men and 52 % of women in the no preconditioning cohort compared to 83 % of men and 69 % of women, respectively, in the preconditioning group and 87 % of men and 78 % of women in the fit group. These studies suggest that preconditioning can prevent injury and medical attrition for both sexes. In particular, preconditioning before basic military training has potential to benefit women at greatest risk of OMI.

7 Conclusions

Military activities require male and female warfighters to accomplish physically-demanding tasks without getting injured. However, overuse musculoskeletal injuries are common with military activities, and the average female warfighter has a greater risk of overuse musculoskeletal injury than her male counterpart. These injuries are largely rooted in sex-based differences in whole body and tissue-level biomechanics such as poorer bone geometry, larger Q-angles, lower muscle mass, greater adiposity, and lower muscle, bone, tendon, and ligament strength. Nonetheless, all tissues are responsive to physical exercise; and therefore, pre-conditioning may help bridge some of the gender gaps in OMI in militaries across the world, particularly as more combat positions are opening to female warfighters.

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Part IV
Neurological Injuries

Traumatic Brain Injury in the Military: Biomechanics and Finite Element Modelling

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Abstract Traumatic brain injury is relatively common in military and law enforcement activities, despite ongoing improvements in head protection gear and in medical aid procedures and evacuation equipment in battlefield and conflict scenarios. In this chapter, we provide the relevant anatomical and physiological background which is relevant for understanding the occurrence and consequences of a traumatic brain injury and its subcategories. Next, we review the biomechanics of traumatic brain injury, and describe biomechanical injury criteria and thresholds. Finally, we introduce the concepts of modelling brain injuries by means of finite element techniques which consider the biomechanical properties of the head and neck tissues. The possible applications of such computational modelling and simulations, particularly for developing and testing military head-protection equipment, are discussed as well.

Keywords Traumatic brain injury · Finite element modelling · Head trauma · Military ballistic helmet · Combat injuries

1 Introduction

Traumatic brain injury (TBI) is defined as damage to the brain due to external mechanical forces, and/or a fast head movement caused by rapid acceleration, deceleration or rotation [1]. Damage to the brain tissues is caused by passage of kinetic impact energy inwards into the skull, resulting from the head crashing

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against a hard surface, a bullet/fragment or a shock wave. The kinetic energy generates swift deformations of the intracranial soft tissues, which then leads to stretching, shearing, and eventual collapse of the soft tissues, resulting in their destruction—immediately or by secondary damage mechanisms [2].

In recent years, there has been a significant increase in the prevalence of head injuries among soldiers. During 2001–2005, over the course of the war in Iraq and Afghanistan, 21 % of all injuries among US soldiers were concentrated in the head, face and neck areas—a rise of nearly 20 % compared to the injury percentage of the same body parts in World War II, the Korean War, and the Vietnam War [3, 4]. Until 2005, among all the casualties of the allied forces in Iraq and Afghanistan, 35 % have died as a result of traumatic brain injuries [5], compared to 12–14 % mortality rates due to TBI 40 years earlier, during the Vietnam War [4].

Combat head injuries occur due to a variety of reasons: shrapnel and projectiles impacts, gunshot wounds, explosion shock waves, vehicle and aircraft crashes, parachute jumps and even hand-to-hand combat [5, 6]. It should be noted that even a relatively slow non-lethal rubber/sponge bullet hitting the head, can result in a TBI [7]. Although being beyond the scope of the present chapter, it is noteworthy to mention that TBI is common among the civilian population as well, instigated by car accidents, falls, and physical assaults. In the United States alone, TBIs account for almost one-third of civilian trauma related deaths each year [6]. However, the present chapter focuses on TBIs in military scenarios.

Improvement of personal protection gear, combat and battlefield medical response, and medical evacuation practices in the last few years, have reduced mortality rates following TBI [8, 9]. Nonetheless, survivors of TBIs suffer from significant short and long-term effects on their lives. In a recent study it was reported that out of 137 soldiers hospitalized with TBI, only 5.8 % have died after the injury, but nearly 60 % suffered from prolonged complications such as severe meningitis (24.8 %), systemic infection (28.5 %), venous thrombosis (15.3 %), and leakage of cerebrospinal fluid (10.2 %). Late complications that appeared up to two years after the injury included convulsion seizures (39 %) and meningitis infection (13.2 %). Death within the two years follow-up was not negligible (1.5 %) [10].

Earlier studies reported that about 6 % of TBI patients per year in the US, have suffered from a permanent disability following the injury [11] and that out of 50 patients with TBI, none regained full function one year after injury [12]. Besides the physical repercussions, TBI also has a considerable psychological impact, sometimes lasting even longer then the physical effects. It has been wildly shown that years after the injury, victims of TBI cannot be successfully reintegrated into society or return to work, typically live in isolation, and have a higher-than-average risk of committing suicide [13–16].

There are several ways to categorize brain injuries [1, 17]. In clinical settings TBI severity is commonly classified into mild, moderate, and severe categories [17]. The most commonly used system for classifying TBI severity is by grading the patient's level of consciousness, using the Glasgow Coma Scale (GCS) (Table 1). However, because the GCS grading system has limited ability to predict outcomes, other classification systems e.g. the duration of post traumatic amnesia (PTA) and the

Table 1 Clinical severity classification of traumatic brain injury

	GCS	PTA (days)	LOC (h)
Mild	15-13	<1	0–0.5
Moderate	12-9	>1 to <7	>0.5 to <24
Sever	8-3	>7	>24

duration of loss of consciousness (LOC) are also used in establishing the diagnosis (Table 1). A current model developed by the U.S. Department of Defence and Department of Veterans Affairs uses all three criteria [18].

For the present review, we will adopt the time-dependant classification into two categories; primary and secondary brain injuries.

1.1 Primary Brain Injury

Primary brain injury is defined as structural damage occurring immediately after the impact, in the scalp, skull, and brain tissues [19]. Such damage is manifested in the form of scalp lacerations, skull fractures, and injuries to brain tissues such as contusions, lacerations, shearing of white-matter tracts, diffuse axonal injury, haemorrhage, and diffusive swelling beneath the impact site in various parts of the cranium (e.g., epidural, subdural, subarachnoid, and cerebrum) [1, 19].

1.1.1 Contusions and Lacerations

The pia mater (the inner layer of the meninges) is a vascular membrane that closely adheres to the brain and provides it with nourishment, in addition to chemical and mechanical protection. A contusion refers to bruising of the pia membrane without tearing it. This injury is typical to a non-penetrating impact involving an inward depression of the skull. Tears to the pia mater are called lacerations and they are essentially an aggravated level of contusions. Both injuries are caused by brain movement inside the skull at the time of the impact, and are more likely to occur when the brain is exposed to fast acceleration/deceleration type of motion, during which the brain is pressed against or sheared by the bony prominences of the skull. The vast majority of contusions and lacerations take place at the frontal lobes, temporal lobes, and temporal poles. In many cases, the injury is not limited to a superficial wound to the brain and also includes deeper white and grey matter injuries [19].

1.1.2 Diffusion Axonal Injury

Diffusion axonal injury (DAI) is a microscopic-level brain damage, predominantly caused by shear forces acting on white brain matter, and damaging axons. DAI has been identified as one of the leading causes for coma and mortality following a head injury, particularly when the injury occurs in the brainstem [20, 21]. DAI is triggered by a fast rotation or acceleration/deceleration of the head that causes a rapid axonal stretch due to shear, along with tensile and compressive strains acting on the white matter.

Axons are considered viscoelastic and therefore react differently to slow and fast deformations. While a slow deformation could be tolerable and typically would not structurally impair the axons, fast stretching typically causes severe damage to the axonal cytoskeleton and membrane. As a result, damage to sodium channels occurs, and the axoplasmic transport abilities of the axon become impaired. Damaged sodium channels allow for a massive sodium influx which changes the intercellular voltage, thus triggering voltage-activated calcium channels which allow an excessive calcium entry into the cell. This results in local swellings along the axon. The swelling is manifested by multiple small lesions in white matter tracts, and leads to axonal permanent disconnection from the white matte, after which the axon can no longer partake in normal brain functions. High amounts of calcium inside the axon also stimulate catabolic processes and increase the intracellular concentration of free fatty acids and free radicals due to lipid peroxidases, resulting in cell necrosis and apoptosis [20, 22–25].

1.2 Secondary Brain Injury

Although triggered by a singular and rapid event, TBI evolves over time post the traumatic impact. Secondary brain damage occurs minutes to days after the initial TBI, and is caused by a cascade of biochemical events in the brain tissues, triggered by the initial impact [19, 23]. The three main and most common secondary brain injuries are haemorrhage, ischemia, and inflammation.

1.2.1 Intracranial Haemorrhage

Brain haemorrhage is a frequent impact-related injury, caused by contusions and lacerations, occurring as a result of both penetrating and non-penetrating head injuries (such as acceleration/deceleration trauma or a depressed fracture). Such trauma tears arteries or veins, leading to blood pooling inside the skull which causes elevation of intracranial pressure [26, 27]. It is regarded as an extremely dangerous condition and is considered the number-one cause of death in lucid patients after brain damage [19]. About 90 % of the TBI deaths which occur within 48 h of the injury, are due to an excessive rise in intracranial pressure associated with a brain haemorrhage [22, 23]. Brain haemorrhages can cause several additional maladies, such as rise in concentration of iron and glutamate. A rise in iron concentration due to excessive bleeding will catalyze free radicals that harm cell membranes and consequently lead to cell necrosis and death [22, 28]. Although glutamate is a common neurotransmitter, its uncontrolled release is known to be cytotoxic to neighbouring neurons and to impair brain functions in general [22, 29].

Intracranial haemorrhage can be regarded as a secondary brain damage since it is, in most cases, a condition that evolves over time; patients can be lucid and able to talk after the initial injury but will die several hours to days after the insult [19].

1.2.2 Cerebral Ischemia, Hypoxia, and Hypotension

Cerebral ischemia along with the consequent intracranial hypotension and hypoxia are highly widespread secondary brain injury processes [22, 30]. Cerebral ischemia is very common after a traumatic brain injury; it is assumed to be present in nearly 90 % of the fatal-TBI cases and in more than a third of all TBI cases [6, 19, 22, 31]. Cerebral ischemia is defined as an insufficient blood flow to the brain which fails to meet its metabolic demands. It occurs when a cerebral blood vessel is lacerated, displaced, or compressed for a prolonged period of time [25, 32]. As the brain stores little glucose and no oxygen, ischemia quickly creates an acute shortage in nutrients and oxygen supply, causing brain hypoxia and intracranial hypotension [30, 32].

In addition, waste products are not removed and cells are unable to aerobically generate ATP. Turning to anaerobic glycolysis, an accumulation of lactic acid follows, causing cellular acidosis and death [25, 33]. Furthermore, ion homeostasis is breached as the generated ATP levels supply insufficient amounts of energy for ion regulating pumps [33, 34]. Much like with diffusion axonal injury (Sect. 1.1.2), a rapid accumulation of potassium in the brain extracellular fluid occurs, followed by brain edema. This increases the distance for diffusion, tissue osmolality, and the tissue's metabolic demand, consequently further magnifying the initial ischemia [33]. In addition, ischemia-induced fall in intracellular free calcium concentration takes place, ultimately leading to cellular breakdown [33].

1.2.3 Inflammatory Response

Inflammatory response may occur in brain tissues injured by TBI, usually after the brain tissues have been damaged by contusions and micro-haemorrhages. The inflammatory response is most pronounced within several days of the injury, although the inflammatory cascade starts minutes after the injury in the form of cytokines release, causing a breach in the blood–brain barrier and prompting cell apoptosis [1, 35]. While high levels of inflammatory response might have damaging effects lasting long after the injury, a moderate response will allow for post-injury cellular debris removal and tissue renewal [1, 35].

2 Basic Anatomy and Physiology of the Head

Before discussing head injury biomechanics, it is imperative to shortly overview the primary head and neck components: the scalp, skull, facial bones, eyes, optic nerves, brain, cerebral spinal fluid (CSF), cervical vertebrae, and the intervertebral discs.

2.1 Scalp

The scalp is a soft tissue covering the cranium and includes five layers (listed from the outside to the inside) as follows: skin, connective tissue, epicranial aponeurosis, loose areolar tissue, and pericranium, with the skin and epicranial aponeurosis joined by connective tissues [36]. During impact, the viscoelastic biomechanical properties of the scalp allow it to absorb much of the impact energy, thus providing protection for the underlying layers. As much as 35 % of the impact energy can potentially be absorbed by the scalp [26]. Therefore, even though the scalp is very vascular, causing a high level of bleeding if injured, it provides substantial shielding to the skull and increases its fracture endurance.

2.2 Skull

The skull is a bony structure that supports the facial bones and forms a protective cavity for the brain. The skull consists of two main bone clusters; eight cranial bones and fourteen facial bones. The cranial bones form the cranial cavity which houses the cerebrum, cerebellum, and brainstem. The facial bones support the vision, smell, and taste organs as well as the facial muscles [37]. An external periosteum layer covers the outer surface of the skull and acts as the pericranium layer of the scalp, connecting the skull and scalp together. An inner periosteum of the skull that envelopes the brain is the dura, which consists of a tough, fibrous tissue [26, 37].

The adult skull has a middle layer of cancellous bone, laced with vessels and fat (diploë), which is packed between two smooth, stiff, and highly dense layers of cortical bone [26, 38]. Such a structure reduces the overall weight of the skull and is highly suited for protecting the brain by absorbing external impact and crush loads. The impact energy is cushioned by the inner diploë and is absorbed by the cortical layer via bending and shearing [26].

The dome-like shape of the skull provides yet another protection component to the brain, by efficiently spreading the impact energy across the whole surface of the skull [26]. Apart from the cranium-mandible attachments, the skull bones are joined by sutures. The sutures distribute the impact energy further throughout the skull, as well as absorb much of it by themselves through localized deformations. The latter is possible due to the irregular and, therefore, increased surface area of the suture joints, and due to their high collagen contents, which has been shown to have at least 100-times better shock absorption capabilities per unit volume than the bone tissue [26, 39].

The thickness and density of the skull is greatest at the occiput and brow regions, while being the lowest at the temporal and orbits regions. This variability, along with different impact stresses (force of impact over the contact area) results in several types of skull fractures. A head impact with a large contact area, such as from a flat surface (e.g. the interior roof of a car), will cause a linear facture. An impact with a relatively small contact area, such as a shrapnel, will concentrate the impact force and cause a comminuted (or comminuted depressed) fracture [26].

2.3 Brain

The brain is composed of six main regions: (1) cerebrum, (2) cerebellum, (3) di-encephalon, (4) midbrain, (5) pons, and (6) medulla oblongata. Most of the brain's surface, including the cerebrum and cerebellum, is covered by a grey-matter layer named the cortex that has a grooved and folded structure which characterizes the human brain [37]. The cerebrum is the largest part of the brain, making 80 % of all its mass. The cerebrum is divided into right and left hemispheres, each of them composed of five sub-parts: frontal, temporal, occipital, parietal and insular lobes (the specific function of each of the brain regions is beyond the scope of the present chapter).

2.3.1 Ventricles of the Brain

The cerebral ventricles are a series of four interconnected spaces. Two ventricles, the first and the second ventricles, are lateral ventricles, located at each side of the brain hemisphere [40]. The third ventricle is connected to the first and second ventricles, forming a narrow midline space between the right and left thalamus [37]. The fourth ventricle separates the pons and cerebellum. On one side, it is linked to the third ventricle by a narrow channel (*the mesencephalic aqueduct*) and on the other it is connected to the central canal of the spinal cord [40]. The ventricles are filled with cerebrospinal fluid (CSF), which continuously circulates between them [37]. A severe traumatic brain injury (TBI) might result in ventricular enlargement due to CSF accumulation in the ventricles (a condition known as *hydrocephalus*). This can be a consequence of diffuse axonal injury (DAI) and/or due to various factors influencing CSF hydrodynamics and intracranial fluid pressure, such as traumatic subarachnoid hemorrhage, presence of a posterior fossa mass, craniotomy (partial removal of the skull), and meningitis-caused inflammation [26, 41, 42].

2.4 Meninges

The meninges, which are the membranes that envelop the brain and spinal cord, consist of three layers: the outer layer—dura mater, the middle layer—arachnoid, and the inner layer—the pia mater. Together they create a tough layer that anchors the brain and the spinal cord and protects them against movements [37, 40]. The dura is a dense, fibrous layer that envelopes both the brain (cranial dura) and the spinal cord, acting as a periosteum to the skull [26, 40]. The cranial dura is mostly two-layered while the spinal dura has only one layer—a tubular, tough, dural sheath [40]. The arachnoid is located between the dura and the pia and is a thin webbed membrane. The space between the arachnoid membrane and the pia mater (subarachnoid space) contains cerebrospinal fluid [37, 40].

2.5 The Cerebrospinal Fluid (CSF)

The CSF is produced continuously in a specialized structure—the choroid plexus fills the central canal of the spinal cord and surrounds the brain by filling the subarachnoid space. As it circulates, the CSF provides nutrients and removes cellular waste products. The cerebrospinal fluid allows for an almost natural buoyancy of the brain and spinal cord, and mechanically serves as a shock absorber, damping much of the mechanical energy that is transferred to the brain during a sudden impact [37, 40].

The entire central nervous system (CNS), including its blood circulation and CSF, is a closed physical/mechanical system requiring fine regulation of pressures. This is achieved mainly by the CSF, since it is the most significant changeable component. The CSF can be quickly redistributed so that the required amount of volume is added/removed to or from a specific intracranial location, without changing the entire volume of the system. Addition of any new volume such as blood in case of a subdural hematoma, or water in case of an edema, or volume change to one of the components such as CSF build-up in the ventricles (*hydrocephalus*) will disrupt the homeostatic pressure inside the cranium and will spiral down rapidly, sometimes ending with a comatose state or even death [26].

2.6 The Spinal Skeleton and Spinal Cord

The spinal skeleton consists of 24 vertebrae: 7 cervical vertebrae, 12 thoracic vertebrae, and 5 lumbar vertebrae, which are connected to the fused bone of the sacrum. The first cervical vertebra, named the Atlas, is connected with the skull. The vertebrae column is stabilized by ligaments and facet joints as well as intervertebral discs that are connecting the vertebrae to each other while allowing movements and flexibility to the spine, along with cushioning the bone-to-bone interaction so that spinal movements are uncompressed [26, 40]. The spine protects the spinal cord, carries the weight of the head, neck, and trunk, transfers it to the lower limbs, and maintains the upright body position while standing or sitting [37].

The spinal cord consists of a grey matter surrounded by a white matter and consists of five named regions: the cervical plexus, the brachial plexus, the lumbar plexus, the sacral plexus, and the coccygeal plexus; each of them is innervating specific regions in the body. The spinal cord extends from the brain via the foramen magnum and through the central canal of the vertebral column. The amount of grey matter is the largest at the segments that are responsible for motor and sensory control of the limbs—the cervical and lumbar enlargements, while the amount of white matter increases near the brain as nervous pathways become wider. The spinal cord is shorter than the vertebral column [37, 40]. The function of the spinal cord is beyond the scope of the chapter.

3 Brain Injury Biomechanics

Brain injuries are very common among both civilians and soldiers. In recent military conflicts, mortality following brain trauma has been reduced, mostly due to improvements in battlefield first-aid and medical evacuation practice, as well as in personal protection gear. However, the percentage of head injuries has increased and while less likely to cause immediate death, such injuries have lasting difficult physical, and mental effects upon the survivors [3–5, 8–10, 12]. Nevertheless, the great complexity of head-brain trauma biomechanics and the complex biomechanical responses of soft tissues to impact in particular, make it very difficult to quantitatively predict the outcomes of closed head injuries.

3.1 Mechanisms of Brain Injury

Brain damage occurs when a mechanical force impacts the head and its energy is transmitted through the scalp and skull, into the brain tissues. Coupled with deformations, rotations, deceleration, acceleration or a combination of acceleration and deceleration of the head, an impact causes shearing and sometimes also tearing of brain tissues, eventually leading to destruction and failure of the neural tissue [1, 2, 43].

During impact, the head/neck complex is typically affected by linear and rotational accelerations [44–46]. These two motion types are considered by several previous studies to be inter-correlated [45, 47, 48]. In reality, however, linear and rotational accelerations may not always be synchronized, as the correlation between the two motions is affected by the impact location, direction of the impact force, and the characteristics of the helmet protecting the head [49, 50]. Previous work has linked between linear acceleration and intracranial pressure differences within the cranium, due to a relative brain-skull motion, causing neurological dysfunction on a level that corresponds with the peak pressure during impact [43]. A relation between rotational acceleration and a predicted intracranial shear stress level has also been established, as the rotational motion causes tissue layers of the brain to slide over each other [44]. Most of the current research work is focused on coup-countercoup injuries, which are seen in many cases of brain injury. Coup describes the injury under the impact location, while countercoup refers to the injury at the opposite (contralateral) location [51]. Coup-countercoup injuries are often caused by the positive and negative pressures formed in the brain after the initial impact to the head, both of which are associated with linear acceleration [51]. However, the relatively poor tolerance of brain tissues to shear forces indicates that rotational acceleration is likely to be more damaging to the brain than linear acceleration [44].

The duration of the impact event affects the mechanical response of the brain. A very short non-penetrating impact by a small projectile with high kinetic energy (a ballistic impact), will cause a wave propagation of stresses, strains and pressures throughout the brain structure [52]. During the first milliseconds of the impact, positive pressures build up at the coup site, while energy is transferred from the skull to the brain. Positive pressure is assumed to be the result of the skull bending inwards and thus moving towards the stationary brain. A compressive wave is then produced in the neural tissues, causing local contusion of the brain at the regions which were instantaneously compressed. During this initial stage, negative pressures are created in the countercoup site. Subsequently, energy is transferred back from the brain to the skull, ensuing a positive pressure wave that moves throughout the cranium [46, 51, 53, 54]. Negative pressures cause cavitation, either at the countercoup site upon impact, or at the coup site when the skull deforms at the impact and then rapidly returns to its unreformed geometry, due to an elastic response of the bony structure [51]. When negative pressure at a certain location is greater than the tensile strength of the brain tissues, a vacuum cavity is created that causes the brain and/or meninges to collapse and separate from the skull. Such a collapse is destructive because it creates high localized tensile stresses in the brain tissues. However, in some cases the levels of negative pressures upon impact are not high enough to create cavitation [51, 53].

Although some basic knowledge regarding head-brain mechanics during impact does exist, the inherent complexity of the brain, along with its nonlinear viscoelastic biomechanical behavior, suggests that a simple cause-and-effect explanation that connects the impact duration and energy to the brain injury and its severity cannot be determined [51]. This is further aggravated by the fact that many of the consequences of TBI occur a significant time after the actual injury [19, 23]. Therefore, accurate injury thresholds, connecting measurable data, such as intracranial acceleration, pressure or stress changes during head impact to the resulting injury type and its severity, are yet to be determined [51]. Thus, in order to understand and quantitatively estimate the different responses of the neural tissues to impact, a suitable modelling framework of the head with a complete three-dimensional kinematics representation could be greatly beneficial, as it will allow examining in great detail the many events, scenarios and the matching biomechanical tissue responses, which are involved in head injury.

The physical and biomechanical responses of the head-brain complex and the intracranial tissue interactions under various impact forces following closed head injury can be analysed by applying the finite element (FE) method of modelling. Such modelling and analyses also enable to computationally compare the influences of different loading conditions and impact scenarios, using the resulting calculations of pressures or stresses in the brain.

3.2 Head Injury Threshold and Head Injury Criteria

An injury criterion is a numerical evaluation of the effects of engineering input variables, such as an impact force, its duration and acceleration, upon a certain body part (e.g., the brain). Input variables should uniquely characterize the head impact and should be tied together by means of a mathematical formulation with values representing various possible severities of injury to the brain. The severity of the injury will then depend on the injury thresholds of the brain tissues. Additional to the input variables, a series of output variables are calculated, defining the mechanical responses of the brain and head tissues, such as internal stresses, strains and pressures. When the levels of the output variables exceed the tolerance of tissues to the mechanical effects, an injury will occur [52]. It should be noted that while the input variables are typically feasible to measure (though could be technologically challenging to acquire), output variables are often inaccessible, as they develop inside the head and within the brain tissues. Therefore, use of biomechanical injury criteria facilitates determination of the expected influence of the specific impact scenario on the head and brain, and in particular, the potential biological damage and its likelihood to be reversible or repairable.

The most commonly used mathematical formula for predicting head injury outcomes nowadays is the Head Injury Criterion (HIC), which is used by the U.S. federal government in the National Highway Safety Administration, among others [26, 55]:

$$HIC = \max\left\{(t_2 - t_1)\left(\frac{1}{(t_2 - t_1)}\int_{t_1}^{t_2} a(t)dt\right)^{2.5}\right\} \tag{1}$$

where a(t) is the average acceleration of the three axes (X, Y, and Z), measured at the centre of gravity of the head, and t_1, t_2 are any two time marks during the acceleration pulse, creating various time intervals [55]. Two of the most commonly used time intervals are 36 and 15 ms, for which HIC is marked HIC_{36} and HIC_{15}, respectively. A good correlation was reported between skull fractures due to frontal impact in cadavers and HIC scores [52]. In particular, HIC_{36} score bellow 1000 and

HIC_{15} score bellow 700 are considered safe in the sense that the probability for a serious skull fracture (AIS ≥ 2)[1] is relatively low for the 50th percentile male [55].

In 1993, Hertz have determined a probability formula for skull fracture as a function of HIC [56]:

$$Probability(fracture) = N\left(\frac{\ln(HIC - \mu)}{\sigma}\right) \tag{2}$$

where N is the cumulative normal distribution, the expectation μ is 6.96352, and the standard deviation (σ) is 0.84664 [55, 56]. The probability of skull fracture (AIS ≥ 2) (see Footnote 1) for HIC_{36} and HIC_{15} can then be calculated using the Hertz probability formula. For example, for HIC_{36} = 1000 the probability of skull fracture occurrence is 48 % and for HIC_{15} = 700 the fracture probability is 31 %, both are calculated for the 50th percentile male [55, 56].

Despite being the most commonly used injury criterion, the HIC has several major drawbacks. The HIC is actually a skull fracture criterion, and although a skull fracture obviously involves some extent of brain trauma, a brain injury can develop without a skull fracture. Thus, an impact with no skull fracture might have a "safe" score according to the HIC, but be unsafe in reality. In addition, the HIC formula does not consider rotational accelerations, which can also be highly injurious to the brain tissues as described above (Sect. 3.1) [55]. Furthermore, HIC does not take into account internal mechanical reactions in the brain tissues (such as stresses and strains developing during and after the impact), and does not distinguish between different types of injury such as subdural hematoma or diffuse axonal injury [31]. Lastly, despite having a good correlation between skull fractures and HIC score for frontal impact in cadavers, poor correlation between the two was shown for non-frontal impact directions [57].

Finite element head models have been previously used for determining various thresholds for brain injury using different engineering output variables; e.g., effective (von Mises) stress, strain, and acceleration. Willinger and Baumgartner, for instance, used a FE head model which consisted of face, cranium, scalp, brain, and CSF, and which was validated based on cadaver impact results [58, 59]. The intracerebral pressure response, von Mises stress, and global strain energy of the CSF, were studies and compared to impact location and type of the resultant neurological lesions. The results determined that a von Mises tolerance limit for a 50 % chance for moderate to severe brain injury, would be between 18 and 38 kPa. Subdural hematoma was calculated to form for global strain energies of 4–5.5 J in the CSF. Kleiven used a FE model that included the scalp, skull, meninges, brain—separated into grey and white matter, brain ventricles, CSF, eleven pairs of the

[1] Abbreviated Injury Scale (AIS) is a standard system for classifying injuries according to type and severity following vehicle accidents. AIS score is time-independent and is given to each body area and each injury independently. AIS reflects the injury itself, and does not account for its consequences or complications. AIS scores refer to the following injury severities: 0 = non-injured, 1 = minor, 2 = moderate, 3 = serious, 4 = sever, 5 = critical, 6 = untreatable, [55].

major parasagittal bridging veins, and the neck with part of the cervical spinal cord [60]. The model was validated against cadaver impact results. According to this study, there is a 50 % probability that a concussion will occur for 26 % strain in the grey matter, 21 % strain in the corpus callosum, and a pressure of 65.8 kPa in the grey matter. According to the majority of the relevant literature, the kinetic energy of head impact forces combined with the resultant strain levels in brain tissues (as indicated by FE modelling), is the best predictor for TBI. In particular, strain levels exceeding 50 % would indicate a high likelihood of TBI [61, 62]. However, these tolerance levels and the underlying injury mechanisms are not yet well established and require additional research.

4 Modelling Brain Injury

Early FE computational studies used simplified structural models consisting of a hard spherical shell representing the skull, which was typically filled with fluid (representing the brain). As the computing power grew, and additional geometrical and tissue material properties became available, the FE head models became more sophisticated and life-like. Some of the innovative three-dimensional (3D) FE head models were published in the recent years by different groups [43, 63–66]. These models achieved more accurate description of the highly complex head-neck geometry and the inter-tissue biomechanical interactions based on MRI and CT scans [66]. Such FE models are very useful in predicting injuries after head impact, using a large range of loading conditions representing various injury scenarios, all of which would be extremely difficult to determine experimentally. The resulting biomechanical response of each tissue can be established regardless of its size or location inside the head. In addition, using various impact conditions applied to the same FE head/brain model eliminates the response deviation due to biological variability, and allows systematic comparisons across the effects of different impacts.

A set of differential equations representing the biomechanical reactions and the governing equations for each tissue is solved numerically using initial-boundary conditions, determined by the impact location, energy, duration, position, and range of motion given to the head, cranial geometry, and interface interactions between the intracranial tissues (usually set as “tided”, meaning no sliding is allowed between neighbouring tissues).

Results include, but are not limited to, spatial and temporal distributions of pressure, stress, strain, shear, displacements and acceleration that develop in each of the cranial tissues after the impact. By correlating the results with known injury thresholds and actual pathological data, simulations can be prognostic of the expected brain damage (and other tissue damage) [66].

4.1 Tissue Mechanical Properties

Biological tissues are generally recognized as inhomogeneous, anisotropic, non-linear and viscoelastic. Thus, in order to ensure that the FE computational models provide accurate information, it is essential to include in these models realistic mechanical properties for the various cranial hard and soft tissues. However, a complete material characterization of biological tissues is a challenging task and continues to be a limiting restriction in computational modelling. Difficulties arise because of the large variability in mechanical properties of biological tissues, the differences in tissue behaviours when examined in vivo versus in vitro, and due to the variance in preservation and preparation methods, as well as differences in mechanical properties between living human tissues and cadaveric tissues, or tissues from animals. Particularly important is to determine the material behaviour of the skull, CSF and the brain (with breakdown to grey and white matters), as they have the most significant influence on the overall response of the brain to impact and the consequences on survival and function [67].

Modelling the CSF as an incompressible lineal elastic layer with a low elastic modulus seems to be easiest straight-forward solution for introducing the brain-skull interaction into the FE model. A low elastic modulus is assigned to the CSF in order to mimic sliding between the brain and the skull and kinetic energy absorption by the CSF, without the additional complication of modelling an actual liquid layer with fluid-structure interactions (the brain on one side and the skull on another side). Such a valid modelling strategy has been widely used before [38, 67–70]. Characterizing the brain tissues presently seems to be a more challenging task. Constitutive laws, their related material constants, and reported values for the elastic and shear moduli of the brain, varied greatly over time and between different research groups, as brain biomechanics research evolved. The earliest FE models of the brain employed a linear elastic material behaviour [71, 72]. Other, more recent models of brain tissues, employed the large-deformation theory with linear viscoelastic constitutive laws [73, 74], or nonlinear viscoelastic materials [75, 76]. The interested reader is referred to Tables 4–6, 8 in Ref. [77] for a summary of the different elastic and shear properties of the brain tissues. Animal tissue properties have been used before [76, 78, 79], despite the known structural differences between animal and human brains [77].

Based on data retrieved from the relevant scientific literature, a collection of tissue mechanical properties is provided in Table 2. Specifically, all head components are assumed to be isotropic and elastic, except for the fat and the brain tissues, which are assumed to behave as hyperelastic, and viscoelastic materials, respectively. In some cases, for tissues that are composed of several sub-structures, such as the eyes or the lumbar disks, properties were chosen according to the largest volume component of each tissue (e.g., the vitreous body of the eye).

Table 2 Material properties for head tissues

Structure	Material model	ρ (kg/m^3) Density	E (MPa) Young's modulus	K (MPa) Bulk modulus	G (MPa) Shear modulus	τ (s)	υ Poisson's ratio	Source
Skull	Linear elastic	1412	6.5×10^3				0.22	[80]
CSF	Linear elastic	1040		2190	5×10^{-4}		0.49	[45, 68]
Skin (scalp)	Linear elastic	1000	16.7				0.42	[67]
Eyes (vitreous structure)	Linear elastic	999	42×10^{-3}				0.49	[81]
Fat (adipose tissue)	Neo-Hookean (hyperelastic isotropic)	919.6			$G_{instantaneous} = 286 \times 10^{-6}$		0.495	[82, 83]
Disks (nucleus pulposus)	Linear elastic	1020	65×10^{-3}				0.24	[84, 85]
Vertebrae	Linear elastic	1900	10^5				0.29	[86, 87]
Sinuses* (bridging veins)	Linear elastic	*1133	5.13				0.495	[45, 61]
Brains**	Viscoelastic isotropic	1040	9.21×10^{-3}		g = 0.769 (for dimensionless two term Prony series)	10^{-6}	0.4583	[45, 70, 88, 89]
Optic nerve***	Linear elastic	1040	0.03				0.49	[45, 90]

*Density of sinus tissues is taken as blood density from [45]

**A Prony series of two terms can be used to describe the relaxation modulus of the brain [76]: $G(t) = G1 + G2 \cdot \exp(-t/\tau 1)$; $G_{infinity} = G1$, $G_{instantaneous} = G(0)$. A dimensionless equation for the relaxation modulus is often used as well [70]: $g(t) = (1 - g) + g \cdot \exp(-t/\tau 1)$.

***Density of the optic nerve is taken as that of the brain.

4.2 Validation of the Modelling

Comprehensive understanding of the causes of traumatic brain injury requires careful examination of the biomechanics behind the injury that includes pressures, stresses and strains and their propagation within the brain and other soft tissues of the head. As described above, FE modelling of the head and the various impact scenarios provide a cost-effective and efficient investigative tool for uncovering the inner-workings of brain injury formation by examining the biomechanical responses of the brain to external collisions. Although MRI and CT scans and the software tools to reconstruct the head/brain shapes from them become more biomimetic with time, various simplifications to the geometry of the head/brain are usually made, in order to accommodate for the complexity of the modelling; especially since a large amount of finite elements will require substantial computation time and power. As demonstrated in Sect. 4.1 and by Yang et al. [77], tissue biomechanical properties often carry some simplifications as well, as they are based on experiments with primates and human cadavers and so the constants used in the formulation of constitutive laws might vary from properties of living human tissue [67]. Since various simplifications are inherent to FE modelling, any physical and biomechanical behaviour which is derived numerically by the FE model should be verified and validated against experimental data. Such validation process guarantees, to some degree, that any data provided by the model (e.g. strains and pressures inside the brain following an impact), would correlate with actual clinical results [69]. Validation can be performed against data from experiments with cadavers, or by using off-the-shelf or proprietary crash dummies. The two most significant advantages of using cadavers are that the viscoelastic tissues such as the brain remain viscoelastic (though possibly with different constants than a living tissue) which is difficult to replicate in a manikin, and that the mass distribution is similar to a living human [67].

One of the most cited studies regarding impact experiments with cadavers was conducted by Nahum et al. [91]. In this study, the impacts to the head were applied using rigid impactors with various speeds and weights. For each experiment, impact force and head acceleration were measured during the impact and the resultant intracranial pressure was recorded by sensors implanted in the dura. While being vastly cited and a highly respected published work, the researchers focused primarily on intracranial pressures (negative and positive), while many injuries to the brain, such as diffuse axonal injury, are caused primary by strains [69]. Thus, validation of the results obtained from a FE model should focus on strains and shear stresses as well. Additional work with cadavers was conducted by Hardy et al. [46, 92]. In these studies, intracranial pressures, shear stresses, and strains in the brain were recorded, following a struck with a padded impactor, or collision with a rigid body at a given speed, or abrupt crash onto a flat surface.

4.3 Possible Applications for Design and Testing Head-Protection Gear

Combat helmets are designed to withstand impacts with objects of high velocity and low mass, such as projectiles, fragmentation, and shrapnel. Modern combat helmets such as the Personnel Armor System for Ground Troops (PASGT), Advanced Combat Helmet (ACH), and the Modular Integrated Communications Helmet (MICH), also provide partial protection against blast waves and heat, while integrating a low-weight design and incorporating laser range finders, night vision gear, and other various sensors, making a substantial progress from the M1 steel helmet of the previous century [93, 94]. Modern ballistic helmets are woven composite structures, combining Kevlar® fibers with other materials. Composite helmets may sometimes delaminate following impacts with a high kinetic energy (i.e. react by separating the outer and the inner laminates from each other), causing an indentation of the interior helmet shell which contacts the skull and results in injuries. This is known as the “rear effect” [95].

Helmet performance under a ballistic impact is greatly influenced by the mechanical response of its fibers, as the kinetic energy of the impact is largely absorbed by straining and tearing of the fibers. The tensile strength and the stress-strain curve of the fibers are important measures for determining their performance under ballistic loading. The ballistic performance of a helmet can also be assessed by finding the failure criteria under tensile loading of the fiber-woven composite that laminates the helmet shell [93, 95, 96]. Fiber-based laminated composites under tensile loading fail mainly in an in-plane manner in reference to the fibers. In-plane tensile failure can be categorized into: (1) matrix cracking, (2) fiber-matrix shearing, and (3) fiber breakage [96]. Cracking of the matrix may occur after random micro-cracks are merged together under various mechanical and thermal loadings, often due to quasi-static or repetitive transverse tensile loading of the plies. The matrix is usually affected before the fibers as they are typically much stiffer. Fiber-matrix shearing can also weaken the composite material when the shearing forces are high enough to cause local de-bonding of fibers from the matrix, allowing them to be more elastic and promote crack progression. Fiber breakage is caused by high stress levels on the composite material [97, 98]. Chang and Chang [96] suggested a matrix cracking failure criterion for laminates with linear elastic properties, as defined by Eq. (3) [96]:

$$\left(\frac{\sigma_y}{Y_t}\right)^2 + \left(\frac{\sigma_{xy}}{S_c}\right)^2 = e_M^2 \tag{3}$$

where for each layer of the laminate, σ_y is the transverse tensile stress, σ_{xy} is the shear stress, Y_t is the transverse tensile strength, and S_c is the in situ ply shear strength measured from a cross-ply laminate [99]. If in any of the layers of the laminate, the stresses σ_y and σ_{xy} produce $e_M^2 \geq 1$, matrix cracking is predicated to occur in the said layer.

Also, according to Chang and Chang [96], both fiber-matrix shearing and fiber breakage can be predicted using an analogue failure criterion (Eq. 4). For laminates with linear elastic properties, it is stated that:

$$\left(\frac{\sigma_x}{X_t}\right)^2 + \left(\frac{\sigma_{xy}}{S_c}\right)^2 = e_M^2 \tag{4}$$

where for each layer of the laminate, σ_x is the longitudinal tensile stress and X_t is the longitudinal tensile strength. If in any of the layers of the laminate the stresses σ_x and σ_{xy} produce $e_M^2 \geq 1$, the layer will fail by fiber breakage or by fiber-matrix shearing (the complete failure criteria for non-elastic linear laminates can be found in Ref. [96]).

Another way of measuring the ballistic performance of a helmet material is by calculating its ballistic limit. The ballistic limit is classically defined as the velocity (m/s) of a projectile with mass m (kg), at which the projectile will completely penetrate the material 50 % of the times [93, 100, 101]. A summary of ballistic limits for various materials was listed recently by Kulkarni et al. [93].

Despite the many possible tests that exist nowadays, some of which were described here, there is still no efficient, low-cost and standardized procedure for assessing the effectiveness of a ballistic helmet. In fact, testing methods and procedures vary with each manufacturer, making it highly difficult to compare the ballistic protective efficacy across helmets from different manufacturers [95]. In this regard, FE modeling provides an inexpensive, cost-effective and highly flexible in silico testing framework for comparing the performances of various ballistic helmet designs, either existing or new, under the same test conditions. Alternatively, the same ballistic helmet design can be tested under different scenarios, such as various impact locations (e.g., front, top, back, and side of the helmet), or a blast wave combined with shrapnel impact, etc. All these analyses can be done even prior to manufacturing a prototype, since the only required data are the shape of the helmet and its mechanical properties.

This concept is demonstrated in Fig. 1, where two basic types of helmets (a one-layer and a two-layered helmet; b, c, respectively) are placed on a simplified head model. A “bullet” then impacts each helmet under the same conditions (same impact site, speed, and duration) and the resulting effective stresses in the brain are compared. The two-layered helmet with an outer Kevlar® layer and inner metal layer (Fig. 1c), provides far better protection over the single layered helmet with (only) metal properties (Fig. 1e, d, respectively). Using more accurate helmet models that capture the actual shape of the helmet, with advanced FE models of the head and brain, provide consistent clear distinction between desired and undesired helmet performance, though the simple model in Fig. 1 was used here for illustrating the aforementioned concepts. Relevant examples can be found in the studies by Yang and Dai [94] and by Zhang et al. [102].

By combining a biomimetic head model with detailed helmet models as presented by Tse et al. [103] and by Aare and Kleiven [95], and then comparing the

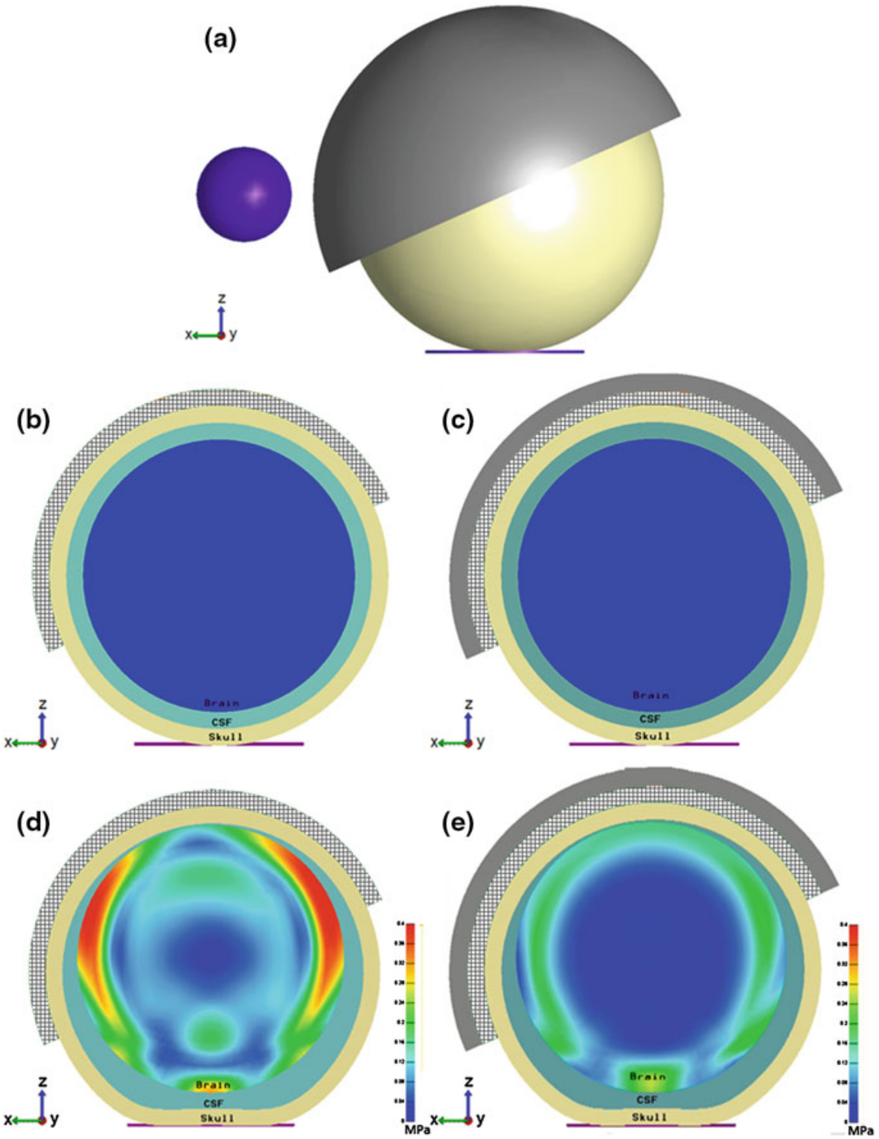

Fig. 1 **a** A finite element simplified head with a helmet. The "head" is represented by three spheres—brain, CSF and skull, each given its appropriate mechanical properties. A helmet is positioned on top of the head and a purple sphere stand for the projectile. **b** A cross section through the head model with a one-layered helmet (*reticulated* part with metal properties). The projectile is made transparent for clarity. **c** A cross-section of the head with a two-layered helmet (*reticulated* and *grey* parts, *grey* part with Kevlar® properties). The projectile is made transparent for clarity. **d**, **e** A cross-section of the heads in (**b**) and (**c**), respectively, after a non-penetrating impact of the projectile. The distribution of effective stresses (MPa) in the brain is shown using the *colour bar* specifying the stress magnitude on the right. An advantage of the two-layered helmet (**e**) over the one-layered helmet (**d**) in reducing impact-related effective stress in the brain, is evident

resultant output variables in the brain and calculating the brain injury thresholds (see Sect. 3), a simple yet elegant and highly efficient tool can be suggested for helmet performance evaluation during prototype design testing, manufacturing, and purchasing of head protection gear.

In summary, this chapter reviewed the fundamentals of military-related head/brain injuries, including the relevant anatomy and physiology aspects, research methodologies—both experimental and computational, and the state-of-science in the field. It is apparent that FE modeling plays and will continue to play a key role in applied research of head/brain injuries in armies. Its power in aiding in the process of selection, evaluation and design of helmets and other head-protecting gear, has been clearly demonstrated here.

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Biomechanics of Eye Injury in the Military

Brittany Coats and Daniel F. Shedd

Abstract Eye injury accounts for approximately 15 % of battlefield injuries worldwide. Most eye injuries are not life-threatening and are not a high priority on the battlefield. However, these injuries can result in a severe loss of function, diminished quality of life, and decreased career opportunities for military personnel following completion of service. The complexity of the eye is similar to the complexity of the brain in that trauma to the eye results in primary acute injuries with secondary and long term sequelae leading to visual dysfunction. To date, the biomechanics and resulting injury of ocular trauma has been a fairly underrepresented area of research. This has changed as a result of the recent wars and conflicts, but there is still a lot that is unknown or not well understood. This chapter will provide an overview of ocular anatomy, types and causes of ocular injury found in the military, and a review the state of knowledge in biomechanics of ocular trauma from blunt impact and blast exposure. Careful evaluation of the current literature will identify urgent areas of focus, establish guidelines for future biomechanics research, and lead to the development of better strategies for the detection, assessment, and prevention of ocular injury in military personnel.

1 Introduction

Ocular injuries are estimated to account for 8–13 % of battlefield injuries worldwide [1–3]. Most ocular injuries are not life threatening, but they result in the severe loss of functionality, diminished career opportunities, and an overall decrease in the quality of life of our military personnel. The associated medical costs are also astounding. An analysis on ocular injuries from the Vietnam War estimated that the short and long term medical expenses were approximately 4 billion USD [4].

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Incorporating historical rates of inflation for medical care, the current estimated short and long term costs for ocular trauma during military combat are 37 billion USD [5, 6].

The vast majority of ocular injury from recent military conflict is due to explosions. In one combat support hospital, 207 severe eye injuries were reported and 82 % were caused by blast or blast fragmentation [7]. Blast-related open globe injuries are typically easy to detect and immediately treated. Closed globe injuries may not be detected until symptoms are reported by the injured service member. Several studies have investigated visual impairment in soldiers with traumatic brain injury from blast exposure. They report ~75 % of soldiers with traumatic brain injury also have visual dysfunction [8–10]. One study in particular performed complete ocular exams on all soldiers with a history of traumatic brain injury from blast exposure and found retinal injuries in several military personnel that were unaware they had any ocular or visual problems [11]. Because of these studies, and many more, the US government has initiated a mandate for the evaluation of ocular trauma associated with traumatic brain injury [12].

However, government mandates such as this only help identify ocular injury if there is an associated traumatic brain injury, and blast-related closed globe injury has been shown to also occur **without** the presence of brain injury. Dougherty et al. [13] retrospectively investigated the medical records of 2254 service members with a history of blast exposure. Within 12 months after the blast exposure, 201 of the service members had developed ocular/visual disorders. More than half (54 %) of those members had no associated traumatic brain injury. This suggests that blast-related eye injury can occur with or without a traumatic brain injury, and diagnosis may be easily missed because the onset of symptoms may occur well after the event.

To develop better strategies for the detection, assessment, and prevention of ocular injury in military personnel, the biomechanics of ocular trauma from combat needs to be established. This chapter will provide an overview of ocular anatomy, types and causes of ocular injury found in the military, and review the state of knowledge in biomechanics of ocular trauma from blunt impact and blast exposure.

2 Brief Overview of Ocular Anatomy

The human eye is a complex structure that includes the cornea, pupil, lens, retina, vitreous, and optic nerve. These structures work in concert to transport visual information from the outside world to the brain. Damage to even one of these ocular components can result in substantial vision degradation or vision loss. Below is a brief overview of the major structures of the eye (Fig. 1), their role in vision, and the potential for vision loss following injury.

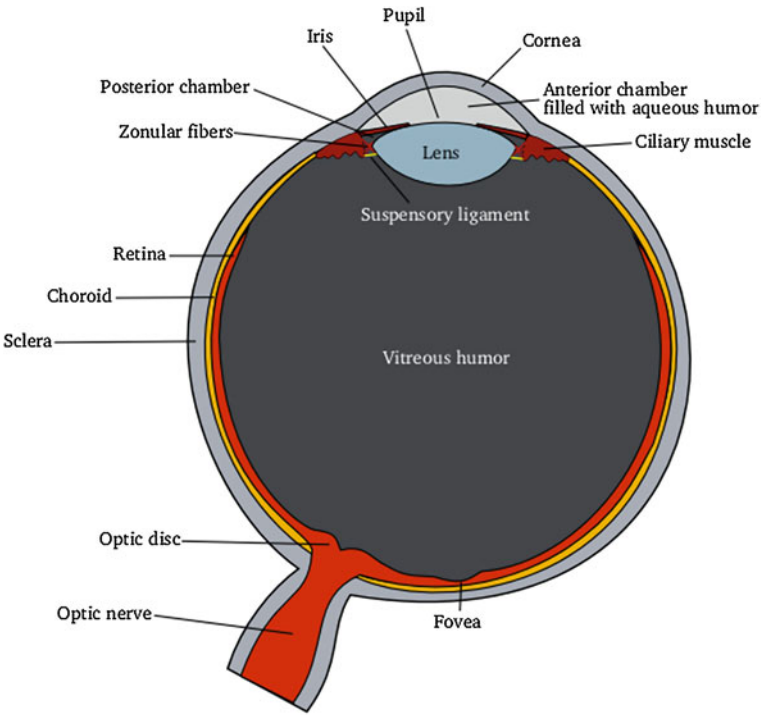

Fig. 1 Anatomical schematic of the human eye

2.1 Cornea

The cornea is the most anterior segment of the eye. Its primary functions are to protect the lens of the eye and provide optical power to focus light. To ensure optical clarity, the cornea is an avascular structure, and its nutrients are provided through the aqueous humor on the posterior side and tear fluid on the anterior side. The cornea is a five-layered structure with the epithelium as the outermost layer. The interface between corneal epithelium and tear film provides a significant portion of the optical power of the eye. Therefore, damage to the epithelium can alter the interaction and cause vision irregularities. The stroma is the thickest layer of the cornea and is made up of loosely spaced collagen fibers, arranged in parallel to create transparency. Bowman's Membrane is a thin collagen structure that protects the stroma. It is important to note that as an avascular collagen structure, this layer cannot respond to damage and will not regenerate. Descemet's Membrane and the endothelium separate the body of the cornea from the aqueous humor. These layers are permeable to allow nutrients to pass through to the stroma. As the most exposed region of the eye, the cornea is especially vulnerable blunt impact and blast exposure.

2.2 Aqueous Humor

The aqueous humor is the watery fluid that fills the space between the cornea and the lens. It adds refractive index to the eye, maintains intraocular pressure, and provides nutrients to the cornea and lens. Damage to the structures that surround the aqueous humor could disrupt outflow and result in increased intraocular pressure. A hyphema, blood in the anterior chamber, can also result from ocular trauma and decrease vision.

2.3 Lens

The lens of the eye is an elliptical, biconvex structure that changes shape and size to allow the eye to focus on objects at different distances. It works with the cornea to refract light. It is primarily protected by a smooth, transparent outer layer called the lens capsule. Long, thin transparent cells, called lens fibers, make up the bulk of the lens and are adjusted mechanically via ciliary muscles to alter the shape of the lens. Impact to the lens can lead to cataracts, which decrease the transparency of the tissue, or to lens dislocations which diminish vision quality.

2.4 Pupil/Iris

The pupil is the opening in the iris that allows light to pass to the retina of the eye. The iris can retract or expand to change the size of the pupil. The size of the pupil is varied in accordance with the amount of ambient light, to ensure that the optimal amount of light reaches the retina. The pupil is smallest in bright conditions, allowing very little light to enter the eye. Conversely, the pupil is at its maximum size in very dark conditions, allowing as much light as possible to reach the retina. Blunt impact to the eye can result in inflammation in the iris and decrease visual acuity.

2.5 Vitreous

The vitreous humor is a transparent substance that fills the body of the eye. It is considerably more gel-like in structure compared to the aqueous humor, but can degrade with age to become more liquid-like [14]. The vitreous is adhered to the retina near the vitreous base and posterior pole. Inertial head trauma may cause traction of the vitreous on the retina and lead to vitreous detachment, retinal detachment, or tearing. Vitreous hemorrhage can occur when blood from neighboring vascular tissues leak into the vitreous.

2.6 Retina

The retina is a complex light sensitive tissue that uses neural impulses to transmit images to the brain via the optic nerve. The thin membrane (160–291 μm) is composed of nine distinct layers with the nerve fibers being the most proximal to the vitreous, and the light-sensitive rod and cone cells more distal. Localized damage to the retina can cause transient or permanent blind spots in the visual field. Tears in the retina may lead to retinal detachment from the posterior layers and cause permanent vision loss.

2.7 Sclera

The sclera is the white outer surface of the eye. It is covered by conjunctiva, a clear mucous membrane that helps lubricate the eye. It is thickest near the posterior region of the eye, and is primarily responsible for maintaining the structural integrity of the eye as a unit. Impact trauma may cause hemorrhages, abrasions, lacerations, or punctures of the sclera.

2.8 Optic Nerve

The optic nerve carries information from the retina to the brain. The optic nerve passes through the sclera at the optic nerve head. The optic nerve is structurally protected by the dura, arachnoid, and pia mater layers, which are the same as those found separating the brain and skull. Inflammation and/or hemorrhage within the optic nerve or nerve sheath can occur from traumatic insult, and can lead to optic neuropathy.

2.9 Orbit, Extraocular Muscles, and Fatty Tissue

The orbit of the eye is the bony socket that contains and protects the eye. The eye is surrounded within the orbit by orbital fatty tissues and is moved within the orbit by extraocular muscles. Severe impact trauma can cause orbital bone fracture and/or extraocular muscle laceration.

3 Ocular Injuries Reported in the Military

Injuries to the eye caused by trauma are generally classified as open globe or closed globe injury. Open globe injuries occur when the outer shell of the eye (cornea and sclera) has been penetrated or torn. Closed globe injuries are less apparent and may be missed, or not present, upon first examination. Thach et al. [15] report 797 cases of severe ocular trauma from the Iraq War from 2003 to 2005. Severe ocular trauma was defined as anything that disrupted the cornea or sclera, ocular adnexa, or any trauma severe enough to cause major vision loss (e.g., vitreous hemorrhage or optic neuropathy). A little over half of the reported injuries (55 %) were open globe injuries (Table 1). Foreign body injuries were primarily to the globe (intraocular and corneal). Eyelid injuries and orbital fractures were also common and made up a majority of the remaining injuries. It should be noted that only 74 of the 797 participants wore ocular protection. In a smaller case series, Cockerham et al. [16] evaluated 46 veterans with traumatic brain injury from blast exposure. Twenty of those veterans (43 %) had closed globe injuries. The distribution of these injuries is provided in Table 2.

Table 1 Distribution of severe trauma to the eye and surrounding structures reported in 797 ocular injuries reported during the Iraq War (2003–2005) [14]

	n	(%)
Open globe injuries	**438**	**55.0**
Mean laceration size	11.2 mm	
Foreign body	**328**	**74.9**
Intraocular	116	26.5
Orbit	95	21.7
Corneal/conjunctival	86	19.6
Eyelid	31	7.1
Eyelid injury	**265**	**60.5**
Lid laceration	197	45.0
Eyelid foreign body	31	7.1
Tissue loss	12	2.7
Avulsion	11	2.5
Canalicular	9	2.1
Burn	5	1.1
Optic nerve injury	**27**	**6.2**
Avulsion	14	3.2
Optic neuropathy	13	3.0
Orbital injury	**195**	**44.5**
Fracture	100	22.8
Foreign body	95	21.7

The bold are the broader categories and the non-bold are subheadings of those categories.

Table 2 Distribution of closed globe ocular injuries seen in 46 veterans with diagnosed traumatic brain injury resulting from a blast [15]

	n	(%)
Closed globe injuries	**23**	**43**
Conjunctiva	**3**	**7**
Cornea	**10**	**22**
Scars or foreign bodies	9	20
Descemet's membrane rupture	1	2
Trabecular meshwork	**12**	**27**
Iris	**7**	**15**
Lens	**5**	**11**
Anterior cataract	3	7
Posterior cataract	2	4
Vitreous hemorrhage	**5**	**11**
Retina	**10**	**22**
Retinal tears	3	7
Retinal detachment	1	2
Peripheral retinal hemorrhages	3	7
Peripheral choroidal ruptures	**2**	**4**
Macula	**6**	**15**
Optic nerve atrophy	**2**	**4**

The bold are the broader categories and the non-bold are subheadings of those categories.

4 Mechanisms of Ocular Injury Reported in the Military

During wartime, approximately 75 % of ocular trauma in military personnel occurs during combat [7, 17]. This percentage is in stark contrast to peacetime periods where 90 % of ocular injuries are **not** combat or training related. These peacetime injuries include machinery/tool accidents (24 %), fights (18 %), motor vehicle accidents (18 %), or sports (11 %) [18]. In the same group of 797 severe ocular combat trauma patients reported above, Thach et al., tracked the source of the ocular trauma for 469 of the patients. The vast majority (73 %) of severe ocular injuries were from an explosion, which is similar to what has been reported in other studies [1, 7, 19]. Other causes of ocular injury during wartime included ballistic injury, motor vehicle accident, and blunt impacts (Table 3).

Table 3 Reported causes of severe ocular trauma in the Iraq War (2003–2005) [14]

Causes of eye injury	n	(%)
Explosion	344	73.3
Gunshot wound	51	10.9
Vehicle accident	25	5.3
Blunt trauma	17	3.6
Burn	8	1.7
Other	24	5.1

4.1 Blast as a Mechanism of Ocular Injury

Blast-related injuries are typically divided up into four categories based on mechanism of injury: primary, secondary, tertiary, and quaternary injury. Primary injury occurs when blast overpressure waves propagate through the body and cause damage. Anatomical regions at higher risk of injury from pressure waves are areas where neighboring body tissues have different densities. Pressure waves travel at speeds directly related to the density of the medium in which they travel. This results in higher density layers rupturing into lower density tissues. Air and fluid containing organs are most susceptible to this injury mechanism. The cornea-aqueous boundary and vitreoretinal interface are therefore at a greater risk than the rest of the eye. However, overpressurization of the eye may also cause damage, particularly to the optic nerve.

Secondary injury occurs when debris or fragments, accelerated from the blast, impact the body. This is the direct cause of most blast-related fatalities. Secondary blast-related ocular injuries include orbital fracture, foreign body penetration, lid laceration, and avulsion. These injuries often lead to complete destruction of the cornea or sclera and, in severe cases, are difficult or impossible to repair. Contamination from foreign body projectiles such as metal, glass, dirt, rocks, cement, plastic, and fiberglass increases the risk of local or systemic infections.

Tertiary injury results when the body is displaced causing head acceleration and/or body impact to the ground or another object. Traumatic brain injury and associated visual dysfunction is the primary concern regarding tertiary injuries. Approximately 75 % of all military traumatic brain injury patients also have visual deficits [8–10]. Therefore, closed globe injury can be classified as either a primary or tertiary injury depending on the causative factor. Common closed globe injuries from blast pressure, head rotation, or eye impact include retinal detachment, retinal tear, macular hole, injury to the trabecular meshwork, injury to the iris, hyphema, and optic nerve fiber degeneration [1, 11]. Quaternary injury results from thermal effects or noxious particulate inhalation. The primary ocular-related quaternary injuries are burns to the eye or orbit.

To understand how eye injuries can be caused from blast exposure, we must first understand the severity of the blast load applied to the eye. This load is determined by the power of the explosive (W), the height of the explosive off the ground (H), and the stand-off distance of the person from the explosive (SOD). The explosive sends off a spherical shock wave or ‘wave front’ composed of incident waves (Fig. 2). These incident waves impact the ground resulting in reflected waves that are larger in magnitude than the incident waves [20, 21]. Both the incident and reflected waves diminish exponentially as they travel away from the explosive device. However, there are locations within the shock wave where incident and reflected waves merge, called a triple point. This merger creates a new wave, called a mach stem, with a uniform front and *much greater* amplitude. Therefore, a person exposed to a blast can experience an incident, reflected, or mach stem wave, with increasingly severe injuries, respectively. The probability of exposure to each wave type depends

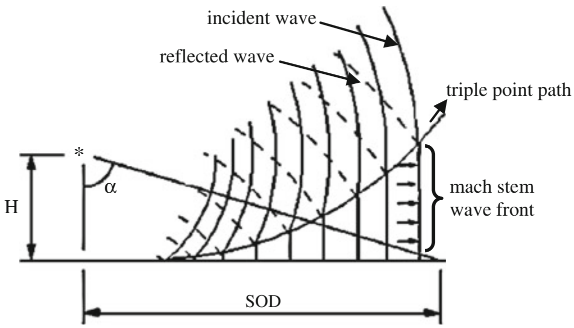

Fig. 2 Wave characteristics from an above ground explosion. Defining parameters include explosive height (H), stand-off distance (SOD), and angle of incidence (α). Image adapted from [21]

on the height of the person as well as the aforementioned blast load characteristics (H, SOD, W). Therefore, obtaining a well-documented history regarding blast exposure is critical to determining magnitude of the pressure load experienced.

4.2 *Missile and Blunt Impact*

Missile and blunt impact to the eye can occur from blast fragmentation, ammunition, motor vehicle accidents, hand combat, and other accidental or inflicted trauma. The distinction in the literature between a blunt and missile impact is with the geometry of the object impacting the eye. Missile impacts are high-rate impacts to the eye from shrapnel, blast fragments, etc. They typically lacerate, penetrate, or obliterate the eye. In the ocular trauma study by Thach et al. [15], 116 of the 438 open globe cases in military personnel required complete removal of the eye, or remaining remnants, due to missile impacts from explosions and ammunition. In contrast, blunt impacts typically involve larger objects. For this reason, there is often no ocular penetration. Instead, the eye compresses in the anterior-posterior direction, and creates distension at the equator [22]. This can result in rupture of the eye [23, 24] or damage to the anterior ocular tissues. Canavan and Archer evaluated anterior segment injuries from blunt impact in 212 eyes from 205 patients [25]. The most common injuries from blunt impact to the eye were hyphema (blood in the anterior chamber of the eye, 81.1 %) and anterior chamber angle recession (80.5 %). They also report injuries to the iris/pupil (37.3 %), lens (24.5 %), and cornea (11.8 %). Cataract opacities resulting from lens damage appeared to be localized and not associated with significant loss in vision. Corneal damage resolved quickly and did not appear to have any long-term consequences. Other anterior ocular structures are avascular for optical reasons, and tissue repair is limited. This can result in permanent damage or scar tissue, which will diminish overall vision capabilities.

4.3 Lasers

Intentional use of lasers or laser weapons to produce blindness is banned by the United Nations [26], and the reported occurrence of laser-induced ocular injury in the military is primarily limited to friendly sources [27, 28]. However, this may change in the near future with the proliferation of laser type weapons designed to shoot down missiles, drones, and disable ships and planes. Currently used laser systems include target designators, rangefinders, and the US Navy's Laser Weapon System (LaWS) [29].

Injury to the eye results when the energy of the laser is absorbed by the ocular tissue. Therefore, the location and severity of the injury depend on the wavelength of the laser, power of the laser, and the duration of laser exposure [30]. Ultraviolet and far-infrared wavelengths are absorbed by the cornea or lens. Most lasers are not powerful enough to cause damage to these regions. However, serious retinal injury can occur from visible and near-infrared wavelengths. This is because the energy of beams at these wavelengths work synergistically with the optics of the eye to magnify the energy absorbed by the retina to 100,000 times what the energy was at the cornea [30]. These wavelengths will be magnified even more when traveling through binoculars or sighting devices prior to entering the eye. When a laser interacts with ocular tissue, one of four mechanisms of injury may occur [31]. First, the laser can create a photochemical reaction that disrupts chemical bonds, resulting in opacities or haziness of the cornea or lens. Second, the energy from the laser can

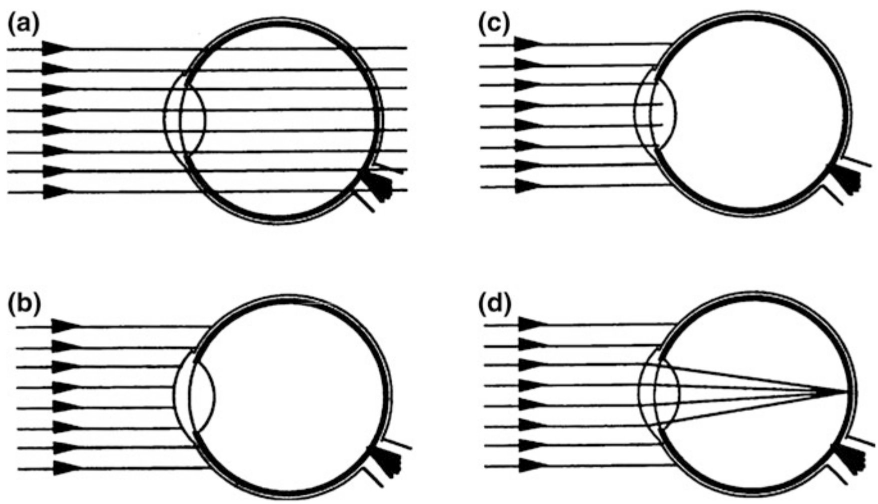

Fig. 3 Laser penetration and damage to the eye is directly related to the wavelength. **a** Microwaves and gamma rays are not absorbed by ocular tissues, while **b** far-ultraviolet and far-infrared are absorbed by the sclera and cornea. **c** Near-ultraviolet is absorbed by the sclera and lens. **d** Visible and near-infrared spectrums are the most damaging as the work synergistically with the optics of the eye to magnify the energy of the laser onto the retina. Image taken from [30]

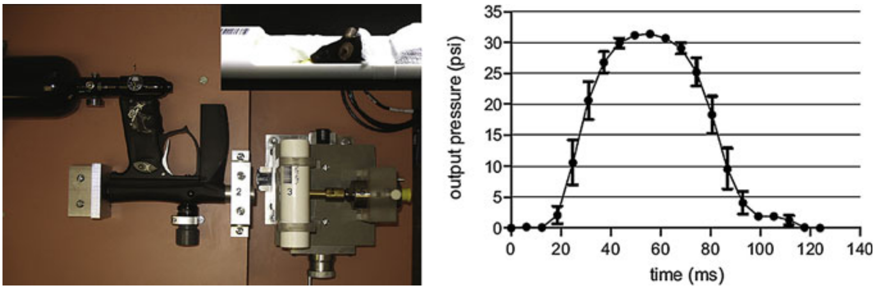

Fig. 4 Image of the gas paintball setup in [42] (*left*) and the resulting pressure-time history (*right*). Image taken with permission from [42]

increase the temperature of the tissue by 10–20 °C, causing burns or protein denaturing. More energy is absorbed by pigmented tissues, such as the retinal pigment epithelium of the retina, and therefore these tissues have a higher risk of injury. Third, the increase in temperature can cause the water in the tissue to vaporize, resulting in dehydration of the tissue. And finally, extremely high-powered lasers could cause plasma formation within the tissue (Fig. 3).

Laser injuries can be avoided by wearing protective tinted eyewear. However, these protective goggles are only useful in attenuating specific wavelengths of lasers; therefore they are likely most helpful in preventing friendly fire injuries from lasers of known wavelengths.

5 Biomechanical Research on Military-Related Ocular Trauma

With recent increased military activity and the intensified use of improvised explosive devices, there has been a heightened focus of biomechanical research on military-related ocular trauma. The majority of the research has concentrated on blast and impact trauma, as they are two of the most common ocular injury mechanisms during wartime. This section will describe the landscape of biomechanics research on military-related ocular trauma and summarize relevant findings.

5.1 Blast Exposure

5.1.1 Experimental Research

The majority of experimental studies of blast are related to traumatic brain injury [32–41]. While the visual system is clearly affected by traumatic brain injury [13], only a few research groups have evaluated the extent of visual system dysfunction and/or damage to the eyes. One of the first of these experimental studies to evaluate blast damage to the globe was by Hines-Beard et al. [42]. The researchers used gas exiting from a paintball gun barrel to simulate an explosive blast insult at 23, 26, and 30 psi in C57BL/6 mice. The duration of the 30 psi pressure wave was 125 ms (Fig. 4). Gross pathology and visual acuity were assessed for 4 weeks after blast exposure. Mortality rates were 26, 22, and 42 % for the three increasing pressure levels, respectively. Gross pathology included persistent corneal abrasions and edema that lasted up to the 2 and 4 week time points, and occasional corneal scarring. A cataract was seen at 3 days post blast in one of the animals exposed to a 30 psi blast. No significant changes in visual acuity or retinal morphology, assessed via optical coherence tomography, were identified at any of the pressures. The exception to this was the single animal that survived 4 weeks after exposure to the highest blast level. This animal had a decreased visual acuity in the blast-exposed eye compared to the contralateral eye. It should be noted that a subsequent study by the same group reports a significant decrease in visual acuity in additional mice exposed to the 26 psi insult [43]. It is unclear what the difference is in blast conditions between the two studies.

Similar to the Hines-Beard et al. studies, Shedd and Coats performed medium level blasts (~35 psi, 8 ms duration) on Long Evans Rats using an open ended 4.6 m long, 12.7 cm diameter blast tube. The blast pressure wave was generated by pressurizing a 1 m driver tank until rupture of a biaxially oriented polyethylene (BoPET) membrane with 0.01″ thickness. Preliminary studies were performed to determine the location in the tube when the pressure wave resembled a typical Friedlander wave (Fig. 5) [44]. The rats head was placed at this location which was approximately 3 m from the membrane. Rats were exposed to a lateral blast. Visual acuity was found to significantly decrease within a day following the blast, and remain decreased up to 8 weeks post-blast. A slight thickening of the corneal stroma was noted in the eye contralateral to the pressure wave 1 week after the blast, but this resolved by 2 weeks post blast. Stromal thickening was noted in the ipsilateral eye at 3 weeks post blast. This also resolved in a week, but epithelium thickening and scarring in the ipsilateral eyes was noted at 6 and 7 weeks, respectively, post blast.

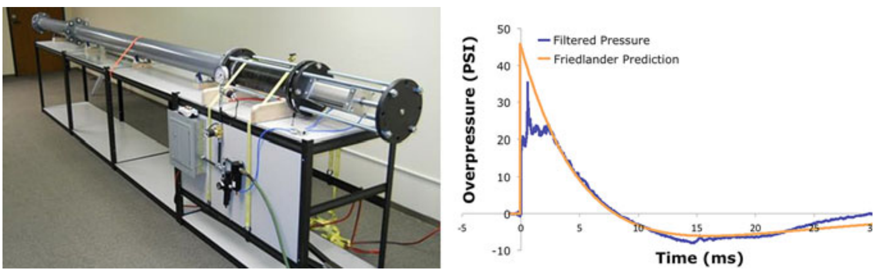

Fig. 5 Image of the 4.6 m long, 12.7 cm diameter shock tube from Shedd and Coats (*left*) and the resulting pressure-time history at the animal's head overlaid with a predicted Friedlander profile (*right*) [44]

Mohan et al. [45] studied the age effect of blast on the eye in C57BL/6 J mice. Mice were exposed to 21.7 ± 0.3 psi blast pressure levels with 10–15 ms durations [46] using a 50 cm long, 33 cm diameter enclosed tank. Part of the tank was pressurized until a mylar membrane ruptured and allowed the pressure to fill the neighboring tank. Pupil reflex was found to significantly diminish one day after the blast, but was not significantly different at 10 months after the blast. There was an immediate (<12 h) reduction in the peak amplitude of the pattern evoked electroretinography (pERG) time history, but this resolved by 24 h. This reduction in pERG amplitude appeared again at 4 months post injury and was still present at 8 months post injury. In a subsequent study by the same group, this deficit was found to appear as early as 12 weeks post injury [47]. Full-field ERG response were not affected by the blast. Age did not appear to have an effect on the results, except a significant reduction in retinal nerve fiber layer in 2 month old mice found 3 months post blast and not in 8 month old mice. It should be noted that it is unclear where the animals were placed relative to the mylar membrane. The original study reported the animals head was 30 cm from the membrane, but the graphics included with the study show the head immediately adjacent to the mylar (Fig. 6). The authors report that the rupture of the mylar membrane occurred at approximately 20.2 ± 0.2 psi (measured in studies separate than the reported blast exposure levels reported above) [45, 46]. If these two reported values are correct, then the head was likely to be close to the mylar as there would normally be a substantial loss in pressure immediately upon rupture of the 13 cm membrane into a 33 cm diameter container.

Choi et al. [35] exposed Long Evans rats to single or repeated low-level blast exposures (10.2 ± 1.0 psi, 2 ms duration) and examined the effect of repeated exposures (5 times over 5 days) on ocular injury. No pressure profile was provided. Both a single and repeated blast exposure resulted in inflammation and injury to the retina and optic nerve. The only significant difference between the repeated and single blast studies was a significant increase in caspase 3 activation in the repeated blast group, suggesting that repeated exposure to low-level blast insults may increase risk for optic neuropathy.

The challenges with experimental blast animal studies are (1) scaling peak overpressure and duration so that the tissue deformations experienced by the

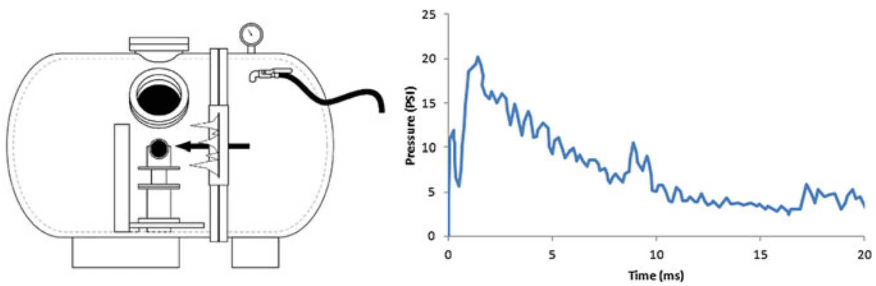

Fig. 6 Image of the enclosed pressure tank used in Mohan et al. (*left*) and the resulting pressure-time history (*right*). Image taken with permission from [45]

animals are equivalent to that experienced in humans and (2) comparing data across studies when vastly different blast injury devices and pressure profiles are applied. One of the first well known studies on scaling blast injury was performed by Bowen et al. [48]. They exposed animals as small as rats and guinea pigs, and as large as sheep and steer to blast. From these experiments, they developed scaling equations for blast positive overpressure and duration as function of animal mass based on the mortality rates for each animal. These scaling methods are generally linked to lung damage, as that was the common cause of blast fatality. There is a critical need to perform similar scaling studies for other types of blast-induced injury such as traumatic brain injury or ocular trauma.

5.1.2 Computational Research

Various computational models have been created to investigate biomechanics of traumatic brain injury from blast loading [49–58]. Only three models exist in the literature that focused on evaluating the mechanical response of the eye to blast. The first of these models was developed by Rossi et al. [59]. They simulated primary blast injury on the eye within a whole head model (Fig. 7). The model featured linear elastic properties for the majority of the eye with viscous components added to the orbital fat and vitreous (Table 4). Constants for the constitutive equations were reverse engineered by simulating experimental impact tests reported by Delori [60]. The rigid body orbit was modeled carefully to accurately represent any reflection of pressure waves that could increase damage in the eye. Close range (0.5–1 m) detonation of explosives with varying TNT masses was simulated using a Jones-Wilkins-Lee (JWL) model. Three angles of pressure wave approach to the eye (0°, 30°, and 45°) were evaluated. The authors report that all locations posterior

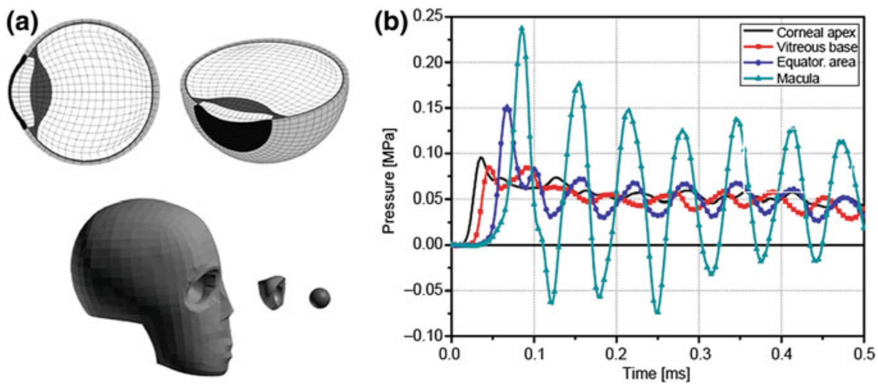

Fig. 7 Finite element model developed and used by Rossi [59] and used by Esposito [61] (*left*). Pressures experienced by the posterior retina (macula) following simulations of a 20 g TN explosion 1 m away were larger than other regions of the retina and resulted in negative pressures that could result in retinal traction (*right*)

Table 4 Material properties of eye model used by Esposito and Rossi [59, 61]

Material	Constitutive model	G_0 (MPa)	$G_{¥}$ (MPa)	Viscosity h_0 (MPa)	β	Bulk modulus K (MPa)	Density (kg/m^3)
Vitreous	Linear EOS + linear shear viscoelastic	7.60E−06	4.16E−06	5.00E−06	0.0538	2000	950
Orbital fat	Linear EOS + linear shear viscoelastic	9.00E−04	5.00E−04		50	2100	918

Material	Constitutive model	E (MPa)	G (MPa)	Poisson's modulus v	Bulk modulus K (MPa)	Density (kg/m^3)
Sclera	Linear elastic	29.5		0.454		1200
Cornea	Linear EOS + shear modulus		1.66		145	1149
Lens	Linear EOS				1550	1100
Retina	Linear EOS + shear modulus		0.035		1000	1100
Aqueous	Linear EOS				2200	1000

to the vitreous base had significantly higher pressures than the cornea. This is likely due to the wave passing from a solid (cornea) to a fluid (aqueous humor) and eventually into the vitreous. The angle of approach for the pressure wave significantly influenced the peak pressure experienced by each ocular tissue. A 0° angle (perpendicular to the cornea) produced the greatest magnitude of pressures. The largest strains were reported in the cornea, reaching values nearing 10 % from 2.5 g of TNT at a distance of 0.5 m. This correlates with the excess of corneal damage reported in the clinical and experimental literature already described in this chapter.

In a subsequent study with the same model, Esposito et al. focused on retinal deformations from primary blast exposure [61]. The highest retinal pressures were located posteriorly near the macula (Fig. 7). This finding correlates with their previous study which indicated perpendicular blast loading to a surface was more detrimental than blast loading at different angles. More interesting, however, was that the posterior retina was the only region of the retina that experienced both positive and negative pressures, suggesting that retinal traction would likely occur in the macular region. The authors attribute this to the unique cone shape of the orbit which causes substantial wave reflection, superposition with incident waves, and ultimately increased pressure amplitudes. As reflected waves hit the orbit interface with air they changes polarity and cause macular traction. One final interesting finding from the study was that the pressure in the retina was more dependent on the peak overpressure of the applied blast wave rather than the duration of the wave.

Bhardwaj et al. [62] developed two models to investigate the effect of blast on ocular injury. The first model was a full head model with rigid skin and bone components. The second model the full head model composed only of bone (no skin) with the addition of a simplified deformable eye (Fig. 8, left). The deformable eye contained a corneal scleral shell, solid inner vitreous, and solid extraocular fat. Properties for these structures are provided in Table 5. Simulations of a 2 kg TNT explosion 2.5 m from the face either straight on or emanating from the ground were first performed on the skin-skull model to determine the effect facial structures, such as the brow and nose, have on the pressure experienced by the eye. Regardless of the original source of the blast (straight-on or ground), the brow and nose clearly reflected pressure waves into the eye, amplifying their magnitude (Fig. 8, right). It should be noted, however, that the skin in this model was a rigid body. When the authors removed the skin, peak pressure forces were reduced by 36 %. In simulations with the skull and deformable eye model, the intraocular pressure was periodic due to the reflection of the pressure waves off of the bony orbit. This is similar to what was reported by Esposito et al. The highest injury risk found in the

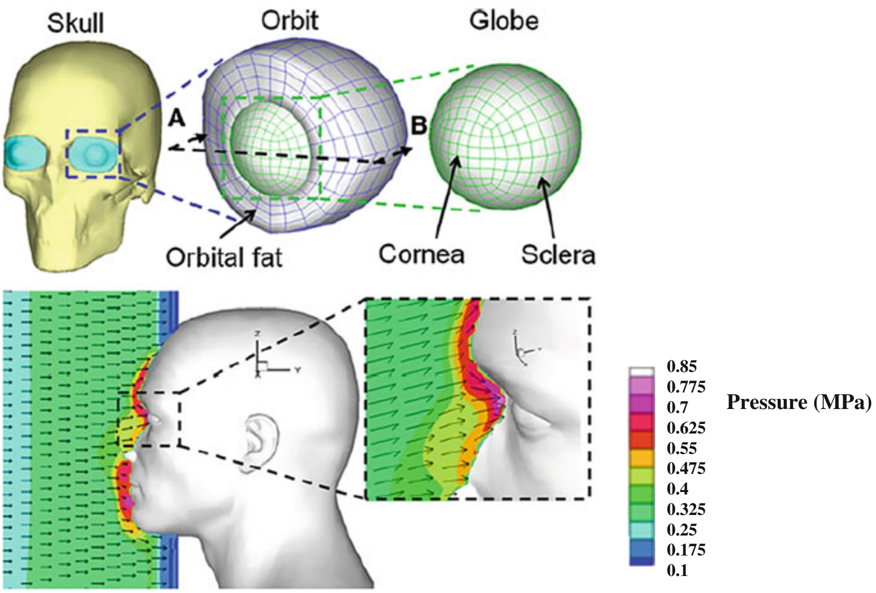

Fig. 8 Skull and simplified deformable eye developed by Bhardwaj et al. (*top*). The skin and skull full head model developed by the same group illustrates pressure wave reflection and amplification into the eye. (*Bottom*) Reproduced with permission from [62]

Table 5 Material properties of eye model used by Bhardwaj et al. [62]

Material	Density ρ (kg/m^3)	Bulk modulus K (MPa)	Shear modulus G (Pa)
Inner solid	1006	2.272×10^3	7
Corneoscleral shell	1400	3.571×10^3	1×10^6
Outer solid	1000	2.202×10^3	5×10^5

simulations was associated with corneal abrasions. All intraocular pressures were below those reported for globe rupture. Maximum IOP, however, was located in the posterior sclera which may have some implications for optic nerve injury. Due to the limited amount of data available to create the model, Bhardwaj et al. performed several sensitivity analyses in their study. Increasing protrusion of the eye from the socket reduced intraocular pressure and corneoscleral stress. Increasing or reducing the thickness of the corneoscleral shell did not substantially alter the magnitude of pressure or stress, but did create a delayed time response. Variations to the corneoscleral material properties affected intraocular pressure and corneo-scleral stress. Variations in the outer solid material properties (extraocular fat) only affected corneoscleral stress. The inner solid material (vitreous) was not varied in these sensitivity analyses.

The third model in the literature was developed by Liu et al. [63]. This model was originally design to evaluate blunt impact to the eye [64] and those simulations are described in the next section. The model was composed of cornea, sclera, lens, ciliary body, zonules aqueous, extraocular fatty tissues, and bony orbit (Fig. 9). Material and explosive parameters are provided in Table 6. Their objective was to predict globe rupture when standing 0.75, 1.0, and 1.25 m from a 1 kg TNT blast. Their results duplicate the findings of Bhardwaj et al. in that reflections of the pressure wave off of the brow and nose increased peak overpressure experienced by the cornea. Using a globe rupture threshold of 23 MPa stress in the corneo-scleral shell [65], the authors predict that globe rupture could occur when standing 0.75 m from the blast. This distance resulted in a peak overpressure of 2080 kPa at the eye and 25.5 MPa stress in the corneo-scleral shell at a single moment of time. It is recognized that a peak overpressure of 450 kPa has been found to result in a 99 % mortality risk due to lung injury [66], but the authors suggest that a smaller-scale explosion close to the face could result in peak overpressures found in this study without substantially damaging the lung.

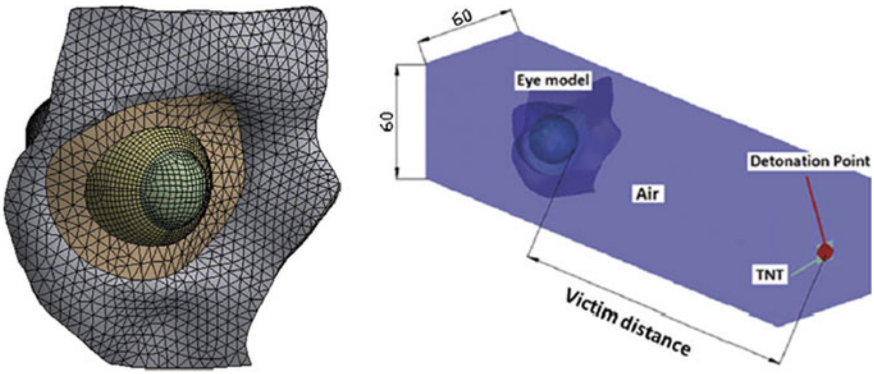

Fig. 9 Ocular blast model developed by Liu et al. to evaluate the potential for globe rupture from 0.75, 1.0, and 1.25 m from a 1 kg TNT explosion. Image taken with permission from [63]

Table 6 Summary of the material and explosive properties used by Liu et al. [63]

Structure	Element type	State	Density (kg/m^3)	Material parameters
Cornea	Lagrangian	Solid	1076	Nonlinear stress-strain
Sclera	Lagrangian	Solid	1243	Nonlinear stress-strain
Lens	Lagrangian	Solid	1078	Elastic, E = 6.88 MPa
Zonules	Lagrangian	Solid	1000	Elastic, E = 357.78 MPa
Ciliary	Lagrangian	Solid	1600	Elastic, E = 11 MPa
Retina	Lagrangian	Solid	1100	Elastic, E = 20 kPa
Fat	Lagrangian	Solid	970	Viscoelastic, G_0 = 0.9 kPa, $G_{¥}$ = 0.5 kPa, b = 50 s^{-1}, K = 2 GPa
Orbit	Lagrangian	Solid	1610	Elastic E = 14.5 GPa
Aqueous	Eulerian	Liquid	1000	Shock EOS linear, C_1 = 1530 m/s, s_1 = 2.1057
Vitreous	Lagrangian	Solid	950	Viscoelastic, G_0 = 10 Pa, $G_{¥}$ = 0.3 Pa, b = 14.3 s^{-1}, K = 2 GPa
Explosive	Eulerian	Gas	1630	Jones-Wilkins-Lee eqn.
Air	Eulerian	Gas	1.205	Ideal-gas gamma law

The simulations detailed above clearly illustrate that a straight-on blast results in significantly higher peak overpressures to the eye than a ground-up blast, and that these overpressures are substantially amplified by reflections off the brow and nose. Interpretation of injury from computational models of the eye is currently limited by the lack of data on blast ocular injury thresholds. Despite this limitation, present day models have made advancements in our understanding of pressure distribution throughout the eye during a blast exposure. Model improvements, however, such as intraocular layer interactions and optic nerve mechanics will be required before we can begin to understand the mechanics of vitreous detachment, retinal detachment, and optic neuropathy.

5.2 *Ocular Blunt Impact*

5.2.1 Experimental Research

There have been several experimental studies that have investigated injury from blunt impact [60, 67–71]. It is difficult to compare results across studies because they span species (human, monkey, pig), use different blunt objects (foam, metal rods, BBs, baseballs, golf balls, etc.), use different speeds for those impacts, and have different methods for containing the eye (bony orbit, gelatin, concrete, etc.). Regardless of this variability in the literature, unique insights into ocular trauma from blunt impact can be gleaned from these studies. Wiedenthal and Schepens [68] reported that radial expansion of the region between the cornea and equator most likely led to ocular trauma during blunt impact. This was supported when they created a stone container to prevent this expansion and noticed a substantial reduction in injury. This also suggests that the ‘container’ used to house a post-mortem eye during the experimental studies is very important to the translation of the findings in vivo. Kennedy and Duma [72] investigated the influence of extraocular muscles on globe rupture from blunt impact in five human cadavers. They report that neither the peak impact force, nor the displacement of the eye at the peak impact force were significantly altered when the extraocular muscles were transected. This suggests that the extraocular muscles contribute very little to the dynamics of eye during blunt impact leading to globe rupture and can be eliminated from computational models investigating this injury. Their influence over other ocular injuries, however, was not evaluated.

A study by Scott et al. [70], impacted adult post-mortem porcine eyes with rounded steel rods of different mass to determine whether momentum or kinetic energy were good predictors of ocular trauma. They used a gelatin-filled container to house the eye and pressurized the eyes to 18 mmHg prior to impact. Ocular injury after impact was classified as level 0 (no injury), level 1 (injury to iris, ciliary body, anterior chamber angle, or lens without dislocation), level 2 (injury to the anterior chamber and/or lens dislocation), level 3 (damage to the retina plus a lens dislocation), and level 4 (rupture of the globe). No level 4 injuries were reported. For the remaining injuries, kinetic energy had a high correlation to severity level than momentum (Fig. 10). It was noted that two of the three projectile types had the same diameter and that the cross-sectional area of the projectile likely influences risk threshold levels. A similar study on porcine eyes was performed by Sponsel et al. [73] using paintballs fired at 26–97 m/s to generate higher impact energies. The injury thresholds were higher than those reported by Scott et al. (level 1–3.5 J; level 2–4 J). A 10 J impact was required for globe rupture.

To evaluate the effect of cross-sectional area on predictions of ocular trauma, Duma et al. [74] combined eight studies of projectile blunt impact from the literature and evaluated predictive capabilities of kinetic energy, mass, velocity, and

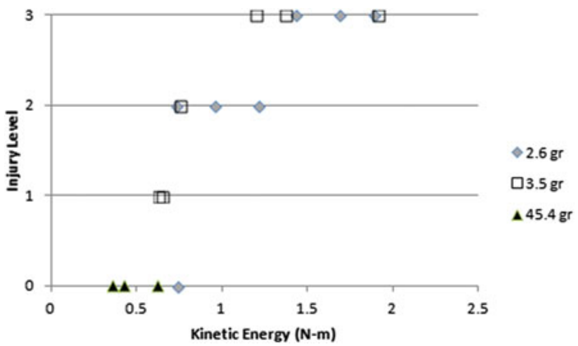

Fig. 10 Kinetic energy of steel rod projectiles plotted against ocular injury levels resulting from the impact of post mortem pig eyes. Image reproduced from [70]

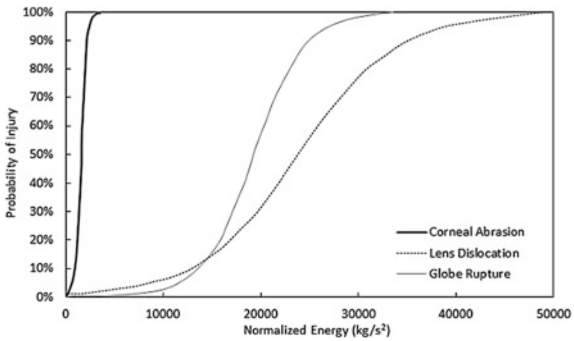

Fig. 11 Injury probability risk curves generated by combining human and animal data from blunt impact experimental studies in the literature. Normalized energy was found to be the best predictor of injury. Image reproduced from [73]

projectile diameter on risk of corneal abrasion, hyphema, lens dislocation, retinal damage and globe rupture. They also added a new predictor called normalized energy which was defined as the kinetic energy divided by the projected area of the projectile. Kinetic energy was determined to be a significant predictor for corneal abrasion, retinal damage, and globe rupture. However, normalized energy was a significant predictor for all of the injuries. Probability curves based on this prediction metric were generated and are shown in Fig. 11. The limitations of this study were the combination of findings across species (pig, monkey and human) and the small number of data available for several of the injuries (hyphema and retinal damage). This resulted in a lower normalized energy for a 50 % risk of globe rupture (23,771 kg/s^2) than retinal damage (30,351 kg/s^2). Additional data on retina injury from blunt impact needs to be gathered before injury risk curves can be generated. However, the determination that normalized kinetic energy is a better predictor of ocular trauma still stands and has been supported by a later computational study by Weaver et al. [75].

5.2.2 Computational Research

The body of literature surrounding finite element simulations of blunt trauma to the eye is greater than that of blast ocular injury. This is because blunt ocular trauma represents the largest proportion of traumatic eye injuries in the United States [76]. These studies range from finger impact [77] to high rate missiles [78] and/or blunt objects [75, 79, 80]. This review is focused on simulations of high-rate missile or blunt impact to a healthy eye because these scenarios are most likely similar to those experienced by military personnel.

The earliest ocular trauma finite element simulations was developed by Uchio et al. [78]. Their objective was to determine the dimensions and mass of a generic blunt-tipped object that results in ocular penetration. At the time, very limited cornea and sclera material property data were available, so these structures were dissected 8–24 h post-mortem from adult human cadavers (ages 64–81) and statically tested in compression. Missile impact speeds of 30 or 60 m/s were simulated with a generic object with front tip dimensions spanning 0.5–3 mm. Compressive failure strains of 18.0 and 6.8 % were used to define missile penetration thresholds for the cornea and sclera, respectively. Increasing missile size resulted in increased strain in both the sclera and cornea (Fig. 12). Most ruptures occurred at the higher velocity, and only missiles above 0.75 mm resulted in tissue strains above rupture thresholds.

Power et al. created a finite element model of an adult human eye encased in fatty tissue and ocular orbit to identify increased risks of globe rupture in helicopter pilots wearing night vision goggles when aircraft airbags are deployed [81]. The model included nonlinear elastic properties for cornea and sclera, and linear elastic properties for extraocular muscles, ciliary body, fatty tissue, vitreous, and aqueous humor. The lens, orbit, and goggles were modeled as rigid bodies. A pilot looking directly at the airbag with initial contact to the eye about halfway through its deployment resulted in the worst-case scenario and a maximum corneal von Mises stress of 11 MPa. This stress could be reduced to 4 MPa if the night vision goggles were allowed to break away upon impact. Both of these scenarios resulted in maximum stresses below estimated globe rupture thresholds. It should be noted, that contrary to their earlier experimental finding that extraocular muscles could be eliminated from computational simulations [72], this study found a significant reduction in corneal stress when the muscles were removed from the simulation.

Stitzel et al. [82] made several improvements to the constitutive material models in the Power finite element analysis, and validated the entire finite element model against impact experiments to post-mortem human eyes. Eyes were obtained from unembalmed cadavers and tested within 24 h after death. The enucleated eyes were embedded in gelatin within a synthetic orbit. All eyes were pressurized prior to testing. Foam pellets, BBs, and a baseball were launched at the eyes at speeds

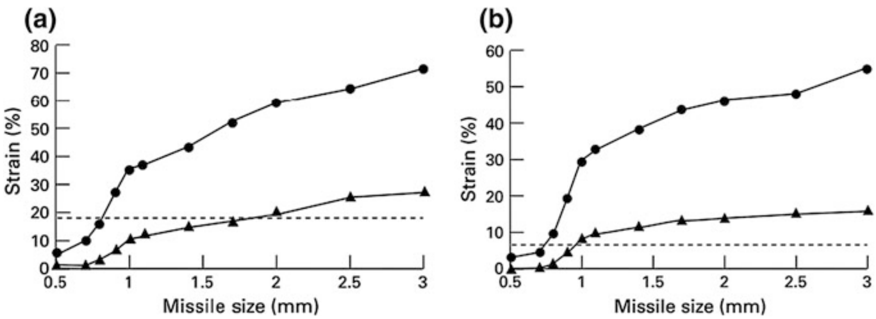

Fig. 12 Maximum percent strain in simulations of a generic blunt object with varying front tip dimensions (0.5–3 mm) impacting the cornea **a** and sclera **b** at 60 m/s (*circle*) or 30 m/s (*triangle*). The dashed line indicates the compressive failure threshold used to predict penetration. Image taken with permission from [78]

ranging from 10 to 123 m/s, and filmed using high speed video. Impact from the BBs and baseball resulted in globe rupture at speeds greater than 91.7 and 30.1 m/s, respectively. Therefore, the model was only validated using data from foam pellet impacts. In these studies, the displacement of the corneal apex was measured from the video and compared to corneal displacement from the simulation of foam impact. The lowest and highest impact speeds from the foam impact tests (10 and 30 m/s) were simulated. The simulations overestimated the peak displacement by 8.5 % at 10 m/s and underestimated the experimental studies by 6.3 % at 30 m/s. For the BBs and baseball, two impact speeds were simulated. The first was the lowest impact speed from the experimental tests. The second was the lowest impact speed that led to rupture in the experimental tests. The foam pellets never resulted in globe rupture, so the maximum measured speed was used. Only displacement from the foam impact was compared to the experimental studies. General ocular deformations and regions of high stress in the finite element simulations matched the general deformations from the experimental studies and locations of rupture (Fig. 13). Based on these simulations, Stitzel et al. predict that globe rupture will occur at stresses of 23 MPa in the corneoscleral shell. Simulations of BB impact resulted in 40 and 70 % corneoscleral strain at 56 and 92 m/s, respectively. The latter speed resulted in globe ruptures from the experimental data. Several other studies have simulated high velocity BB impact to the eye. Rossi et al. [80] investigated strain in the retina instead of the corneoscleral shell. A 62 m/s impact resulted in peak strains of 20–25 % in the vitreous base and macula, but approximately 35 % in the equator. In a later study with by the same group [83], they suggest that pressure wave moving within the eye could be a cause of retinal tears and detachment often seen in these injury scenarios. Liu et al. [84] were also interested in retinal detachment from BB blunt impact and developed a finite element model based on the model by Stitzel [82]. The retina and optic nerve were added to the model and assumed linear elastic, while the vitreous and orbital fat were modeled as viscoelastic. All contact parameters were tied, with the exception of the retina-sclera interface, which was modeled using normal and shear stress limits to simulate failure of retinal adhesion. These limits were determined experimentally by the authors. The authors showed that at speeds higher than 50 m/s, retinal detachment was predicted due to the decompression phase of impact. Lack of validation of any of the above referenced models for predicting retinal strain or retinal detachment, however, precludes any conclusions regarding retinal injury thresholds (Fig. 13).

All the models described above are variations of Lagrangian finite element analysis. Gray et al. chose a different approach to simulate blunt ocular trauma and developed a purely Eulerian model of the eye using CTH software [79]. This finite volume software was developed for high-deformation and high strain-rate events. Gray et al. [73] used the software to simulate their groups paintball impact studies described earlier in this chapter. Mie-Gruneisen equations of state with an added plasticity model were used for all components of the model. The authors noted that a more accurate approach would be the use of a coupled Eulerian Lagrangian model, with cornea and sclera modeled using Lagrangian elements. However, the simulations mimicked the general trends form the experimental studies (Fig. 14),

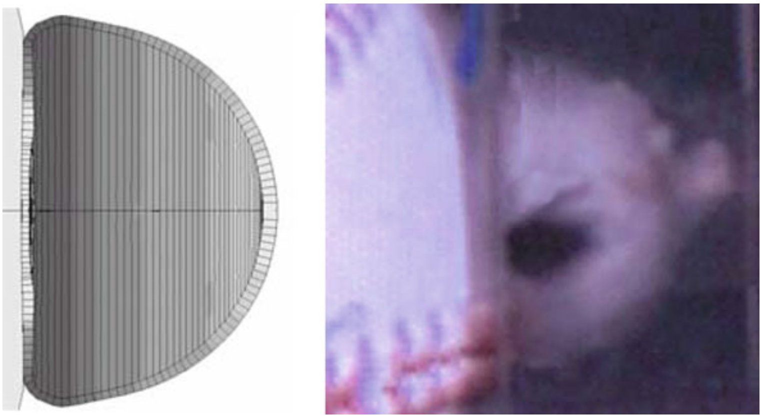

Fig. 13 *Left* deformation of the Stitzel et al. finite element eye model during simulation of blunt impact with a baseball. *Right* experimental baseball impact showing a similar deformation. Images adapted from [82]

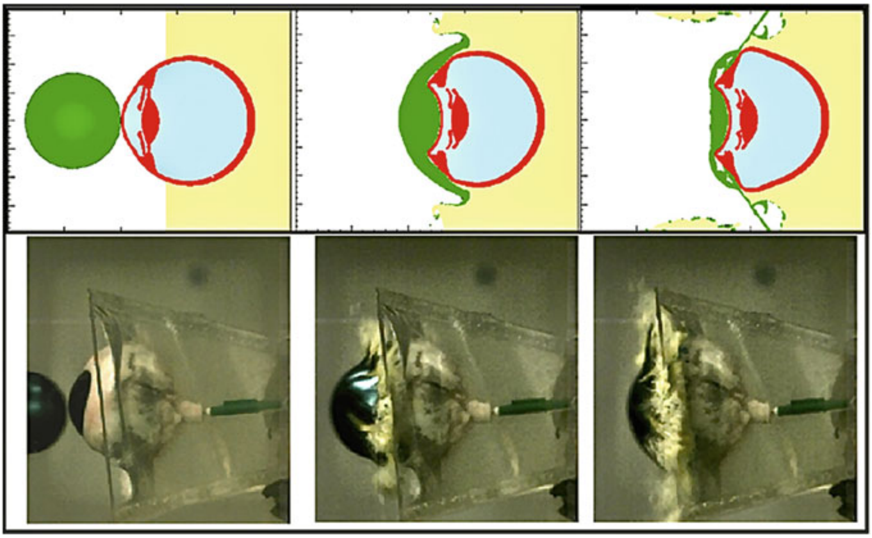

Fig. 14 3D Eulerian finite volume model of the eye developed by Gray et al. The model simulated paintball impact to the eye at rates of 26–97 m/s. Images taken with permission from [79]

and the finite volume model was able to show that the lens dislocation was not due to contact between the cornea and lens, but rather was a result of increased pressure in the anterior chamber. The authors also found that optic nerve injuries seen from the experimental studies may be a result of rotation of the globe rather than increased pressure or anterior-posterior axis motion. A negative pressure was also found to occur from reflection of pressure waves along the posterior pole, and this was hypothesized to be the mechanism of retinal detachment seen in the experimental studies.

6 Summary and Conclusions

The biomechanics of ocular trauma during military conflict and training is a substantial underrepresented area of research in the visual sciences. While ocular trauma rarely leads to death, it still results is a considerable decrease in the quality of life for our soldiers. Both the experimental and computational studies that exist have very different methodologies and use multiple animal species. This makes comparison across studies challenging. However, from the current studies, we know that prediction metrics for ocular injury are dependent on the injury. Corneal abrasions, retinal damage, and globe rupture can be adequately predicted using kinetic energy or kinetic energy normalized by the projected area of the projectile, but hyphema and lens dislocation are best predicted by normalized kinetic energy alone. Thresholds for globe rupture (2.9–10.5 J) are fairly established, but closed globe injuries such as retinal detachment, lens dislocation, and anterior chamber angle recession require additional research before thresholds and prediction metrics can be established.

The reflection of pressure waves in blast trauma is substantial concern. Not only are pressure waves reflected off of objects in the environment, but they also reflect off of facial features such as the brow or the nose. There even exists some resonance of pressure waves within the eye and orbital cavity. These reflections may play a significant role in ocular injury following blast exposure. More studies evaluating and developing prevention strategies for blast ocular trauma need to be developed, but blast ocular trauma research would substantially benefit from standardized agreement on what blast devices and pressure profiles best represent military personnel exposed to a blast in the field. Furthermore, studies need to be performed to identify scaling methods to evaluate how to represent blast pressures at the human scale in animal experimental models. The complexity of traumatic brain injury has been investigated for 70–80 years. The eye is just as complex as the brain and will require a similar amount of devotion to truly tease out mechanisms of injury and the sequelae of events that may lead to long term visual dysfunction.

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Part V
Bio-Thermodynamics and Heat Stress-Related Injuries

Modelling Human Heat Transfer and Temperature Regulation

Dusan Fiala and George Havenith

Abstract In recent years there has been a growing demand from research, military and the industry for robust, reliable models predicting human thermophysiological responses. This chapter discusses the various aspects of- and approaches to-modelling human heat transfer and thermoregulation including the passive and the active system, numerical tissue heat transfer, environmental heat exchange, and clothing. Attention is also paid to advanced modelling topics such as model *personalisation* to predict responses of individuals, and methods for coupling with other simulation models and measurement systems. Several application examples of coupled systems are illustrated including numerical and physical simulation systems and a system for non-invasive assessment of internal temperature using signals from wearable sensors. The predictive performance of the model is discussed based on validation examples covering different exposure scenarios, personal characteristics, physical activities and in conjunction with non-invasive determination of rectal temperature with measured skin temperatures as model input. It is concluded that the model is a robust predictor of human thermophysiological responses, and, the proposed numerical simulation approach to non-invasive assessment of body core temperature, a reliable method applicable to a broad range of exposure conditions, personal characteristics, exercise intensities and types of clothing.

1 Introduction

In recent years there has been a steadily growing demand for robust and reliable computer models predicting human thermophysiological and perceptual responses in industrial, civilian, and military settings. Computer simulation, in general, has

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become a widely used technology which enabled systematic analysis of complex physical, chemical, biological and engineering problems becoming thus an attractive, cost-effective alternative to expensive experimental trials.

Aspects of human heat transfer, temperature regulation and occupant comfort play an essential role in several disciplines. Expansion of human endeavour in hostile and extreme environments, tightening of industrial health and safety standards are two areas of importance. Related modelling research into human performance, tolerance limits, thermal acceptability or occupant comfort has implications for military applications, in the car and aerospace industries, meteorology, clinical, textile and clothing research, as well as medical engineering.

Diverse industrial applications require detailed analysis of the complex environmental heat transfer processes involving human occupancy in order to design safe, comfortable and energy-efficient buildings, cars, or industrial environments [7, 20]. Ongoing research efforts address e.g. the development of ‘physiologically intelligent’ thermal manikins for clothing research [55, 56], methods for assessing the impact of outdoor weather conditions on humans in meteorology [4, 5, 24], development of simulation systems to predict patients’ responses during open-heart surgery or other clinical treatments [11, 62]. Non-invasive determination of the body core temperature is a further area with implications for the development of intelligent personal protective, monitoring and heat stress warning systems for military and civilian applications [22, 23].

To adequately quantify the thermal influences to which man is exposed in different environmental settings, the specific behaviour of the human thermal and regulatory system has to be considered. Within the human body, metabolic heat is generated. This heat is distributed over body regions by blood circulation, is stored within tissues and carried by conduction to the body surface, where, insulated by clothing, heat is lost to the environment by convection, radiation, and evaporation or conduction. Humans keep their internal temperature at fairly constant levels by autonomous regulatory responses of the central nervous system including adjustments to skin blood flow, increase of metabolic heat production in muscles by shivering in the cold, or evaporation of sweating in warm and hot conditions. In addition, the perceptual responses of thermal comfort and sensation enable humans to react consciously by behavioural action which is often the far more efficient way to regulate body temperatures compared to autonomous thermoregulation.

Following general human heat balance considerations, diverse two-node models of human thermoregulation emerged some of which found widespread use [1, 25, 26, 36]. In the past decades, also more sophisticated, multi-segmental, models have been developed (e.g. [17–24, 37, 64, 65, 68]). Compared to two-node models which predict overall physiological responses, the latter simulate the human body in greater detail predicting also local temperature and regulatory responses. Heat transfer phenomena within tissues (including blood circulation) are modelled explicitly and also the specific role of extremities in human thermoregulation can be considered. Environmental heat losses are modelled in more detail taking into account typical asymmetries such as non-uniform distribution of skin temperatures, regulatory responses, non-uniform clothing and local environmental conditions.

One of the multi-segmental models of human heat transfer and thermoregulation that has gained popularity over the past decade is the *Fiala thermal Physiology and Comfort* (FPC) model. The stimulus for developing the original model [17–19] was the desire to make fundamental physiological concepts and models available to engineers and researchers working in other disciplines (for a review see e.g. [21]). The FPC model is a numerical framework of models which—linked together—predict human temperature and regulatory [18, 57], as well as overall and local perceptual (thermal sensation, thermal comfort) responses [15, 19] to steady-state and transient environment and personal conditions.

The model contains two interacting systems of human thermoregulation: the controlling *active system* and the controlled *passive system*. The passive system model simulates the physical human body and the heat and mass transfer processes which occur within the body and at its surface. The active system model simulates the human thermoregulatory system predicting responses of the central nervous system and local autonomic thermoregulation.

Diverse versions of the original 'Fiala model' are currently in use adopted by companies and institutions based on originally published material [17–19]. Model applications include the automotive sector, meteorology, architecture, textile and clothing research, research into health and safety in extreme conditions, diverse clinical and medical engineering applications (for a summary see e.g. [21]). The revised FPC-model was further developed and extended, beyond others, to (i) better represent average-population humans (ii) consider inter-individual variability to enable simulation of individuals, (iii) facilitate flexible coupling with other simulation models, systems or devices and (iv) predict global and local perceptual (thermal sensation and comfort) responses to transient and asymmetric exposure scenarios.

This chapter deals with the various aspects of modelling human heat transfer and temperature regulation based on approaches and algorithms implemented in the FPC model. Treated topics include numerical modelling of heat and mass transport within tissues, advanced modelling of human environmental heat exchange and modelling the active system. Tackled are furthermore endeavours to modelling individuals as are methods to facilitate flexible coupling with other models, sensors and devices. Examples of existing coupled systems are presented. Finally, several validation examples are discussed including the results of a study in which the rectal temperature of exercising subjects wearing protective clothing was determined non-invasively based on measured skin temperatures and metabolic rates.

2 Modelling the Passive System

2.1 Body Construction

The passive system is modelled as a composite of cylindrical and spherical (head) body elements built of concentric tissue layers (section A-A″ in Fig. 1) with skin represented as inner cutaneous layer (incorporating the cutaneous plexus) and outer

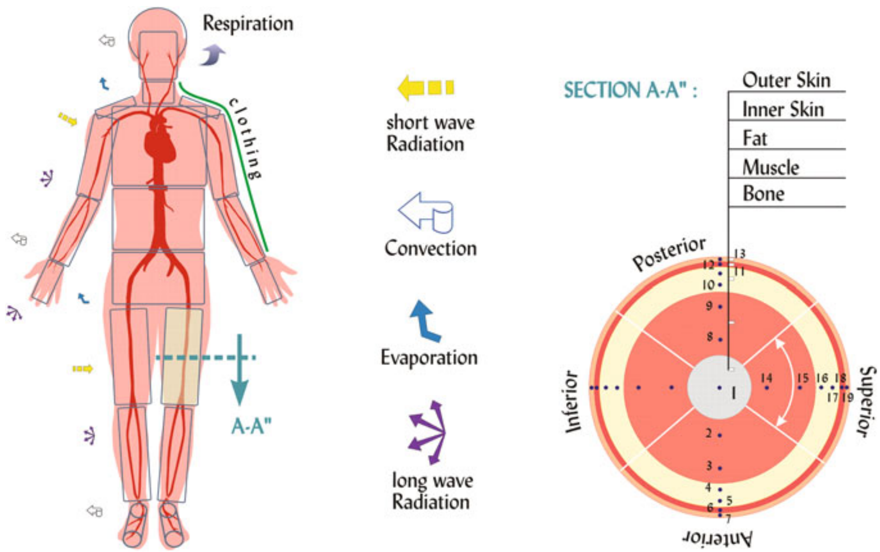

Fig. 1 Schematic diagram of the passive system model including body subdivisions, components of the environmental heat exchange, and a crosssection through the upper leg (*right*)

skin. The latter contains sweat glands but no thermally significant blood vessels [17, 24]. As indicated in Fig. 1, body elements are also subdivided into sectors to enable adequate simulation of asymmetric exposure scenarios.

The reference FPC passive system model resembles an average-population person based on analysis of anthropometric field surveys [22] which included older [50, 51] as well as recent, large-scale, studies [29, 53]. Thereby, the original (left-right symmetry) model consisting of 12 compartments [17] was refined to simulate the human body in greater detail consisting of 20 body elements. This so called *'reference'* human anthropometry model is an average, i.e. '50-percentile' (35 years old, unisex) 169.7 cm tall person weighting 71.4 kg (24.8 kg/m^2 body mass index). It features a skin surface area of 1.83 m^2, body fat content of 22.6 %, and an average body density of 1.05 g/cm^3 [22]. The unisex anthropometry represents the average of a 50-percentile male and female person according to the above surveys which agree well also with other published data [52]. It was defined to (i) obtain a common basis for simulating individuals (see further below) and (ii) provide a well-defined reference model for simulating average-population responses in cases when no subject details are available or of interest.

The resultant anthropometric characteristics of the reference person are compared—in terms of relative body element lengths—with the corresponding field survey data and data employed in a biomechanical model of Daanen and Heerlen [8], in Table 1. The results agree with field survey data within 0.7 % deviation for main body parts. Larger discrepancies result for hands and feet as the incorporated thermal models required additional factors to be considered in order to simulate these body

Table 1 Comparison of relative body part lengths of the reference anthropometry model with the published data

Body part	NASA [50, 51]	Ansur I [29]	Daanen [8]	Ansur II [53]	Reference model
	%	%	%	%	%
Stature	100.0	100.0	100.0	100.0	100.0
Head + neck	14.5	13.5	17.4	13.7	13.9
Head to trunk	53.1	52.5	50.3	52.6	53.1
Trunk	38.7	39.0	28.0	38.9	39.2
Upper arm	18.4	19.5	17.4	18.9	19.0
Lower arm	15.9	15.2	15.7	15.2	15.6
Upper leg	20.2	20.5	24.0	20.5	20.4
Lower leg	22.9	23.2	25.1	23.2	22.7
Ankle (height)	3.7				3.7

parts adequately in thermophysiological terms [17]. Larger discrepancies result also for data used by Daanen and Heerlen due to the inherent differences in the definition of individual body parts in the thermal (FPC) and biomechanical models.

2.2 Scalable Anthropometry Model

The scalable anthropometry model was designed to be an easy-to-use model which, despite the complexity of the subject area, would only require the basic four individual parameters as model input, i.e. body height, weight, age, and gender. The scaling procedure employs the above *reference human anthropometry model* as a basis upon which the personal anthropometric characteristics of the person to be simulated are modelled. The individual parameters are then used to perform calculations of the overall and local body characteristics as indicated in Fig. 2.

To simulate an individual, body elements of the *Reference Model* are 'scaled' based on the four overall input parameters characterizing a person.

Results of anthropometric surveys indicate that the dimensions of individual body parts do not change uniformly as the stature varies. Tall subjects, for example, tend to feature longer extremities in proportion to the body height than smaller subjects. In the model, the length of arms and legs is derived from the gender-specific length of the *tibia, femur, humerus* and *ulna* bones calculated as [51]:

$$L = a_1 H + a_0 \tag{1}$$

where L is the length of the bone in cm, H body height in cm, and a_1 and a_0 are the corresponding regression coefficients [50]. The remaining (cylindrical) body parts are scaled proportionally to changes in the body height while the trunk is sized to retain the body height of the simulated person.

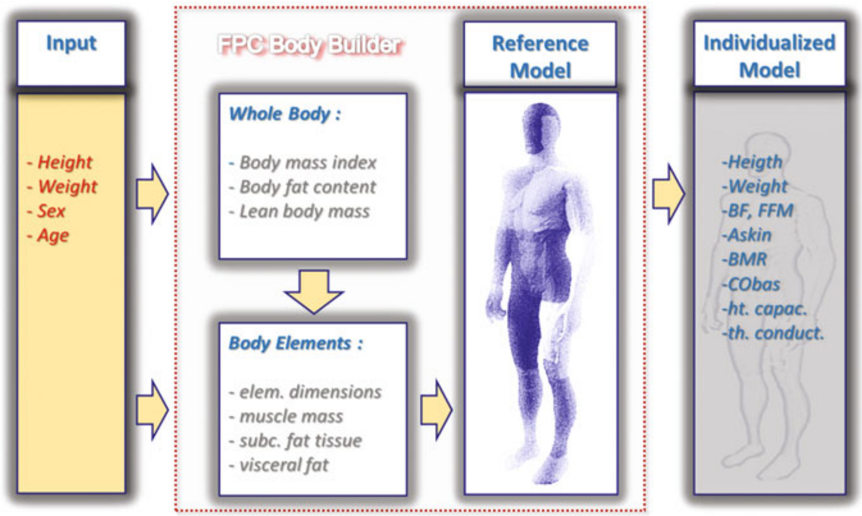

Fig. 2 Schematic diagram of the calculation process constituting the scalable FPC human anthropometry model

The predicted length of the main body sections are compared with measurements obtained from the CEASAR anthropometric survey [9, 59] for male subjects grouped in ten height categories ranging from 155 to 202 cm average height in Fig. 3. The predicted dimensions reproduce measured data within an overall average relative error of 1.8 %. The largest average relative error of 3.6 % resulted for upper legs. For lower legs, the crotch height, and the trunk the error was 1.1, 1.8 and 1.4 %, respectively. A similar level of accuracy was obtained also for upper extremities with 1.5 and 2.2 % average relative error for lower and upper arms, respectively [22].

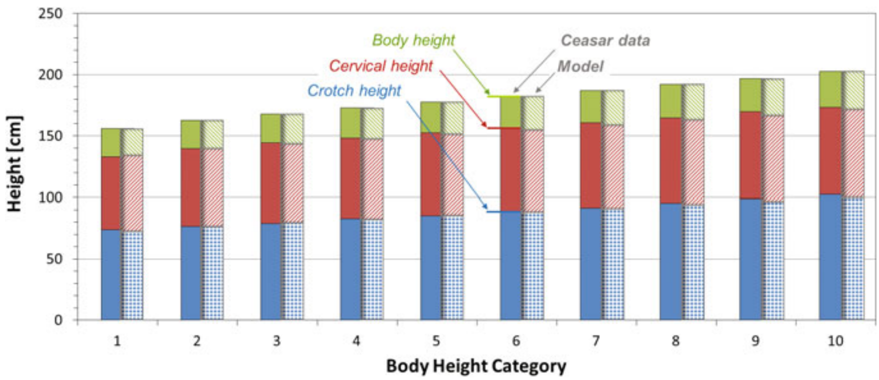

Fig. 3 Comparison of predicted body part lengths forming the stature with data obtained for male subjects from the CEASAR Project [9, 59] for 10 body height categories [22]

2.3 Body Composition

Body composition is another factor to consider in order to adequately represent a human in thermal simulations. The reference person was defined to represent an 'average' human with respect to the body dimensions, fat content, average tissue density, skin surface area, basal metabolic rate, cardiac output, as well as the overall body weight and body weight distribution. The relative weights of different body compartments, provided as percentages of the total body weight, are compared with the corresponding measured data in Fig. 4.

To simulate a person, not only his/her body dimensions but also the overall and local body properties have to be 'individualized' to fit the personal characteristics [22]. Thereby, the overall body fat content is either direct model input or is calculated according to Han and Lean [30] using the body mass index and the age of the simulated person:

$$\mathrm{BF} = c_{bf,b} \cdot \mathrm{BMI} + c_{bf,a} \cdot \mathrm{age} + c_{bf,0} \tag{2}$$

where BF is the body fat content in %, BMI body mass index in kg/m^2; and age in years. The coefficients $c_{bf,b}$, $c_{bf,a}$, and $c_{bf,0}$ with 1.330, 0.236 and −20.20 for males and 1.210, 0.262 and −6.70 for females, respectively, indicate significant differences in the body fat content among sexes [30].

Especially in males a notable portion of body fat may be retained as abdominal subcutaneous adipose tissue (ASAT) and visceral adipose tissue (VAT). VAT is determined based on Kuk et al. [44] who investigated the influence of the personal factors age and sex on VAT and ASAT in 483 young and older male and female subjects covering a wide range of body compositions. Taking into account the dependency of the waist circumference on the overall body fat content and age according to Han and Lean [30] the amount of the visceral adipose tissue is obtained as:

$$\mathrm{VAT} = \mathrm{BF}\left(c_{vat,bf0} + c_{vat,bfa}\,\mathrm{age}\right) + c_{vat,aa}\,\mathrm{age}^2 + c_{vat,a}\;\mathrm{age} + c_{vat,0} \tag{3}$$

where VAT is visceral fat tissue in kg and BF body fat content in % body weight. The coefficients $c_{vat,bf0}$, $c_{vat,bfa}$, $c_{vat,aa}$, $c_{vat,a}$, and $c_{vat,0}$ are gender specific with

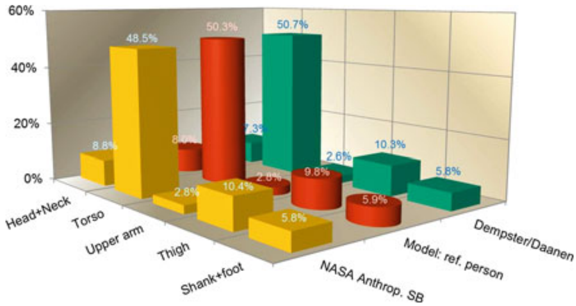

Fig. 4 Comparison of relative body part weights: model (reference person) versus measured data [8, 50]

0.459, 0.003, 0.0003, −0.071 and 12.892 for males and −0.005, 0.003, 0.0008, −0.076 and 0.529 for females, respectively.

The abdominal subcutaneous fat content is calculated similarly based on experimental results of Kuk et al. [44] and Han and Lean [30]:

$$\mathrm{ASAT} = c_{as,b}\,\mathrm{BF} + c_{as,a}\,\mathrm{age} + c_{as,0} \tag{4}$$

where ASAT is the amount of abdominal subcutaneous fat in kg. The coefficients $c_{as,b}$, $c_{as,a}$ and $c_{as,0}$ are 0.194, 0.020, and −1.400 for males and 0.251, 0.055 and −3.387 for females, respectively.

An iterative procedure distributes the overall quantities to obtain local adipose and fat-free mass portions by scaling the thickness of local subcutaneous fat tissue layers relative to the corresponding proportions of the reference model. This approach recognizes that while the largest portion of the body fat is contained in central body parts, the remainder is distributed with decreasing share over proximal limbs towards the outer extremities and the head [13].

The individualized properties are 'mapped' onto individual body elements of the reference model resulting in an updated numerical representation of the human body in the model. As illustrated in Fig. 5, the model's resultant total body fat content after iterative scaling and integration reproduces experimentally based values [30] with and average error of 0.56 % for a wide range of body height and fat content combinations.

It should be noted that the scaling processes not only changes the anthropometric and morphological body characteristics but indirectly changes also other important body properties including e.g. the basal metabolic rate and skin surface area. According to WHO [67], the basal metabolic rate of males and females varies in

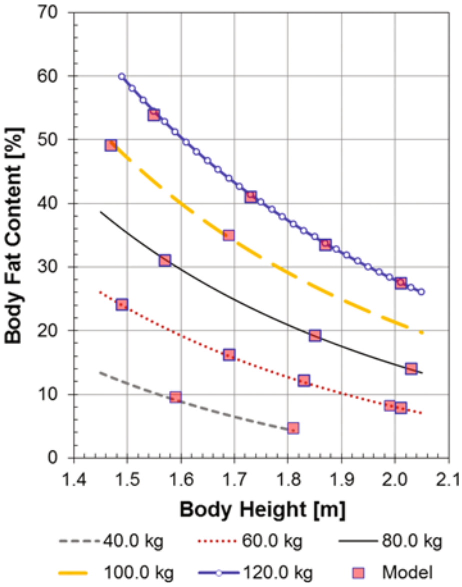

Fig. 5 Comparison of model's body fat content with experimentally derived data as a function of body height and body weight obtained for male (35 years old) subjects [22, 30]

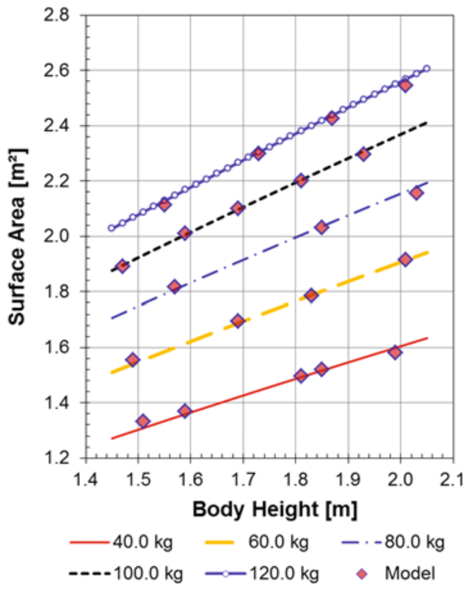

Fig. 6 Model's total skin surface area as a function of body height and weight obtained for male (35 years old) subjects compared with experimentally based data [12, 22]

proportion to body weight though with significant differences among sexes. The scalable model reproduces the WHO formula model within 5 % relative error for both sexes. The overall skin surface area affects the heat exchange with the environment and thus has important implications also for temperature regulation. The total skin surface area which results from the above scaling processes is compared with the Dubois body surface area [12] for different combinations of body height and weight in Fig. 6. It reproduces the Dubois formula with an average error of $0.010 \pm 0.008\ m^2$.

2.4 *Tissue Heat Transfer*

Pennes formulated the so called bioheat equation to describe the dynamic heat transfer processes that occur within living tissues [54]. Extended for heat dissipation in spheres in the model (head), the differential equation can be written as:

$$\rho c \frac{\partial T}{\partial t} = k\left(\frac{\partial^2 T}{\partial r^2} + \frac{\omega}{r}\frac{\partial T}{\partial r}\right) + q_m + c_{bl}\ \rho_{bl}\, w_{bl}(T_{bla} - T) \qquad (5)$$

where ρ, c, and k are tissue density (kg m^{-3}), heat capacitance (J kg^{-1} K^{-1}) and conductivity (W m^{-1} K^{-1}), respectively. T is the tissue temperature (°C), t time (s), r radius (m); ω a geometry factor (ω = 1: polar, ω = 2: spherical co-ordinates), q_m metabolic heat generation (W m^{-3}), ρ_{bl} blood density (kg m^{-3}), w_{bl} blood perfusion rate (m^3 s^{-1} m^{-3}), c_{bl} heat capacitance of blood (J kg^{-1} K^{-1}), and T_{bla} (°C) arterial blood temperature.

In the numerical model, tissue layers are discretized as nodes using a numerical form of the bioheat equation that employs a finite-difference (*Crank-Nicholson*) scheme [17]. Applied to a tissue node *n* (with *n* − *1* and *n* + *1* being the preceding and succeeding adjacent nodes, respectively) and separating the (unknown) 'future' temperature terms (*t* + *1*) the numerical form of the general bioheat equation yields:

$$
\begin{aligned}
&[\gamma_n - 1]\, T_{n-1}^{(t+1)} + \left[\frac{\zeta_n}{\Delta t} + 2 + \delta_n\, \beta_n^{(t+1)}\right] T_n^{(t+1)} - [1 + \gamma_n]\, T_{n+1}^{(t+1)} - \delta_n\, \beta_n^{(t+1)}\, T_{bla}^{(t+1)} \\
&= [1 - \gamma_n]\, T_{n-1}^{(t)} + \left[\frac{\zeta_n}{\Delta t} - 2 - \delta_n\, \beta_n^{(t)}\right] T_n^{(t)} \\
&\quad + [1 + \gamma_n]\, T_{n+1}^{(t)} + \delta_n \left[q_{m,n}^{(t+1)} + q_{m,n}^{(t)}\right] + \delta_n\, \beta_n^{(t)}\, T_{bla}^{(t)}
\end{aligned}
\tag{6}
$$

where

$$\gamma_{n(cyl.)} = \frac{\Delta r_n}{2r_n};\ \gamma_{n\,(sph.)} = \frac{\Delta r_n}{r_n};\ \zeta_n = 2\Delta r_n^2 \frac{\rho_n\, c_n}{k_n};\ \delta_n = \frac{\Delta r_n^2}{k_n}. \tag{7}$$

The time step Δt (s) approximates the differential *dt*, and β_n ($\mathrm{Wm^{-3}K^{-1}}$) represents a time-dependent calorimetric equivalent of the nodal blood flow rate:

$$\beta_n = \rho_{bl}\, c_{bl}\, w_{bl,\,n} \tag{8}$$

Applying Eq. 6 to each tissue node of the numerical model with the corresponding material properties, nodal heat generation and blood perfusion rates, constitutes a system of coupled linear equations which is to be solved for each time and iteration step of a simulation [17]. Solving the whole body matrix for the (undressed) reference person exposed to thermo-neutral, steady state (still air) environmental conditions of 30 °C, 50 % RH results in a mean skin temperature of 34.3 °C and body core temperatures of 37.0 °C in the head core (hypothalamus) and 36.9 °C in the abdomen core (rectum). The resultant overall physiological data replicate a reclining subject with an overall basal body metabolism of 75.5 W, basal evaporation (i.e. moisture diffusion) rate from the skin of 19 W, and basal cardiac output of 4.9 L $\mathrm{min^{-1}}$ [22].

The q_m-term in Eq. 6 is a sum of the local tissue's thermo-neutral basal metabolic rate, $q_{m,0}$ (W $\mathrm{m^{-3}}$) and any additional heat gain, Δq_m (W $\mathrm{m^{-3}}$):

$$q_m = q_{m,0} + \Delta\, q_m \tag{9}$$

Δq_m includes variations in basal metabolism due to changes in tissue temperature from the local setpoint, T_0, which refers to the above conditions of thermal neutrality. In muscles, additional heat may be generated by exercise, $q_{m,w}$, or by regulatory shivering, $q_{m,sh}$, as local portions of the respective overall quantities:

$$\Delta q_m = q_{m,0}\left(2^{\frac{T-T_0}{10}} - 1\right) + q_{m,w} + q_{m,sh} \tag{10}$$

Similarly to q_m, local tissue blood perfusion rates, w_{bl}, are defined as a sum of the thermo-neutral basal rate $\beta_{m,bas,0}$ ($Wm^{-3}K^{-1}$) and variations $\Delta\beta_{bl}$ ($Wm^{-3}K^{-1}$):

$$\beta_{bl} = \beta_{bl,0} + \Delta\beta_{bl} \tag{11}$$

the latter being proportional to changes in the local metabolic rate: $\Delta\beta_{bl} = 0.932\Delta q_m$ [64]. The largely variable blood flows within the cutaneous plexus are subject to central nervous system regulation as described further below.

Blood circulation plays a dominant role in the human heat transfer. In the passive system model, each body element is supplied with arterial blood from the central pool (heart). Before perfusing local tissues, blood is ‘conditioned’ by counter-current blood streams of adjacent veins. Arterial blood at local arterial temperatures perfuses then tissues exchanging heat in the capillary beds where, according to Eq. 5, it reaches equilibrium with local tissues. Depleted blood is then collected in veins being re-warmed by counter-current heat exchange with adjacent arteries as it flows back to the central pool. Finally, venous blood from the whole body is mixed in the central blood pool perfusing the lung to constitute a new central blood pool temperature.

The blood perfusion term in Eq. 5 only accounts for heat exchange with tissues in the capillary bed. In the model also counter-current heat exchange between pairs of adjacent arteries and veins is considered [17]. The net heat exchange between adjacent vessels, Q_x (W) is expressed as [28]:

$$Q_x = h_x(T_{bla} - T_{blv}) \tag{12}$$

where h_x (WK^{-1}) is the counter-current heat exchange coefficient [24]. These coefficients are equal to zero in central body parts and are >0 in extremities. In the individualized model they vary with the length of individual extremities.

Considering the counter-current heat exchange between adjacent vessels, the heat loss from an artery equals the heat gain of the adjacent vein. Assuming mass-continuity in blood vessels, the decrease in arterial blood temperature, $T_{blp} - T_{bla}$ is thus equal to the rise of venous blood temperature, $T_{blvx} - T_{blv}$, after passing the counter-current heat exchanger [17]:

$$\sum_i^{nodes} \beta_i\, V_i\left(T_{blp} - T_{bla}\right) = \sum_i^{nodes} \beta_i\, V_i\left(T_{blvx} - T_{blv}\right) \tag{13}$$

where T_{blp} (°C) is the central blood pool temperature, V_i (m^3) tissue nodal volume, and T_{blv} (°C) and T_{blvx} (°C) is the body element’s venous temperature before and after passing the counter-current heat exchanger, respectively. Since the bioheat Eq. 5 assumes capillary blood to reach equilibrium with the surrounding tissues, T_{blv} yields:

$$T_{blv} = \frac{\sum_i^{nodes} T_i \, \beta_i \, V_i}{\sum_i^{nodes} \beta_i \, V} \tag{14}$$

With the above equations the local arterial blood temperature, T_{bla}, (°C) of a body element can be calculated as:

$$T_{bla} = \frac{T_{blp} \sum_i^{nodes} \beta_i \, V_i}{h_x + \sum_i^{nodes} \beta_i \, V_i} + \frac{\frac{h_x}{\rho_{bl} \, c_{bl}} \sum_i^{nodes} T_i \, \beta_i \, V_i}{\sum_i^{nodes} \beta_i \, V_i \left(h_x + \sum_i^{nodes} \beta_i \, V_i \right)} \tag{15}$$

The blood pool temperature, T_{blp}, is a function of local tissue temperatures from all body parts:

$$T_{blp} = \frac{\sum_k^{b.elem.} \left(\frac{\sum_i^{nodes} \beta_{k,i} \, V_{k,i}}{h_{x,k} + \sum_i^{nodes} \beta_{k,i} \, V_{k,i} \sum} \times \sum_i^{nodes} T_{k,i} \, \beta_{k,i} \, V_{k,i} \right)}{\sum_k^{b.elem.} \left[\frac{\left(\sum_i^{nodes} \beta_{k,i} \, V_{k,i} \right)^2}{h_{x,k} + \sum_i^{nodes} \beta_{k,i} \, V_{k,i}} \right]} \tag{16}$$

3 Modelling the Active System

3.1 Concept and Model Definition

The active system is a cybernetic model of human thermoregulation that simulates the four essential responses of the central nervous system: production of sweat moisture, *Sw*, increase of metabolic heat generation in muscles due to shivering, *Sh*, and changes in cutaneous blood flows due to peripheral vasodilatation, *Dl*, and vasoconstriction, *Cs*. The implemented concept adopts principles of a set-point temperature based control system that has been formulated and implemented previously in other models (e.g. [25, 26, 64]). Set-point based systems define regulatory stimuli generating afferent signals as 'error' signals, i.e. the difference between a variable of the actual thermal state, x (e.g. temperature) and the corresponding setpoint that refers to conditions of thermal neutrality (see Sect. 2.4), x_0:

$$\Delta x = x - x_0 \tag{17}$$

Rather than using postulative methods, meta-regression analysis was employed to define a statistically founded reference active system [18] for simulating responses of an 'average' person. Thereby, published experimental studies were simulated and measured regulatory responses were correlated with predicted afferent signals to (i) investigate their involvement and responsibility in each response based on their statistical significance and to (ii) formulate the governing

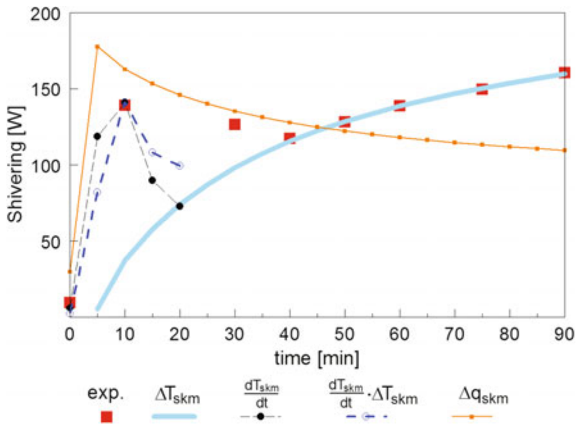

Fig. 7 Example of linear regression lines as functions of different afferent signals to predict the shivering response observed in semi-nude reclining subjects (exp.) suddenly exposed to a cold environment of 5 °C. Adopted from [18] with permission

regulatory equations. Considered were established signals associated with skin temperatures (ΔT_{skm}), head core (hypothalamus) temperatures (ΔT_{hy}), rates of change of skin temperature (dT_{skm}/dt) as well as theoretical signals associated e.g. with muscle temperatures or skin heat fluxes (Δq_{skm},), Fig. 7.

Overall, the regression studies confirmed the body core (hypothalamus) temperature, T_{hy}, and (mean) skin temperature, T_{skm}, to be the main driving impulses for human thermoregulatory action. A further signal, i.e. rate of change of the skin temperature, dT_{skm}/dt, weighted by the error signal associated with the skin temperature, was identified as the driving impulse that governs the dynamics of regulatory responses during rapid skin cooling such as the typical shivering 'overshoot' (e.g. Fig. 7). A schematic diagram of the reference active system model is provided in Fig. 8.

The analysed conditions included 26 different experiments covering exposures to steady state and transient environments ranging from cold, over moderate to heat stress conditions and physical activities from reclining to intense exercise [18]. The results obtained for all exposures were then subjected to meta-regression analysis to study the consistency and functional dependencies of the linear regression coefficients on the subjects' thermal state. The coefficients for individual responses are plotted together with the respective fitting non-linear functions in Fig. 9.

With the dynamic term for skin cooling, i.e. negative rates of change of skin temperature, dT^{-}_{skm}/dt, and (the statistically significant) temperature error signal from the head core included, the control equation for shivering, Sh (W), yields:

$$\begin{aligned} Sh = {} & 10[\tanh(0.48\Delta T_{skm} + 3.62) - 1]\Delta T_{skm} - 27.9\Delta T_{hy} \\ & + 1.7\Delta T_{skm} \times \frac{dT^{-}_{skm}}{dt} - 28.6 \end{aligned} \tag{18}$$

The Sh–response is limited to a maximum of 350 W in the model [18]. The response is compensated by any extra metabolism due to exercise, $q_{m,w}$, and is distributed over body regions using distributions coefficients, c_{sh} [18].

The overall sweat excretion rate, Sw (g min^{-1}), is predicted as:

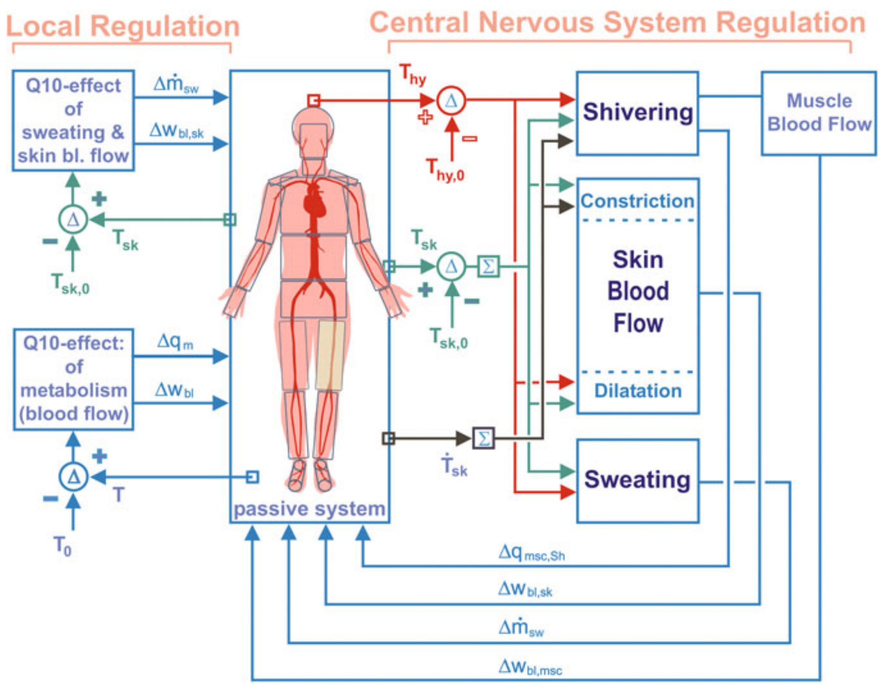

Fig. 8 Schematic diagram of the reference active system model. *Left* responses of local autonomic control; *right* responses of the central nervous system regulation accounting for overall changes in muscle metabolism by Shivering, *Sh*, (and the corresponding changes in muscle blood flow), skin moisture excretion by sweating, *Sw*, and skin blood flow by peripheral vasodilatation, *Dl*, and constriction, *Cs*. The model uses temperatures of the skin (T_{sk}) and of head core (T_{hy}) as well as the rate of change of skin temperature as input signals into the regulatory centre. Setpoint temperatures $T_{sk,0}$, T_0 and $T_{hy,0}$ refer to the nude reclining body's thermo-neutral state at 30 °C operative temperature

$$Sw = [0.8\tanh(0.59\,\Delta T_{skm} - 0.19) + 1.2]\,\Delta T_{skm} + \left[5.7\tanh\left(1.98\Delta T_{hy} - 1.03\right) + 6.3\right]\Delta T_{hy} \tag{19}$$

An upper limit for *Sw* of 30 g min^{-1} applies as a typical maximum rate for an average person [18] in the reference active system model. The resultant, local sweat rates, dm_{sw}/dt (g/min), are obtained as portions of *Sw* distributed over body parts using coefficients, c_{sw} [24], and being weighted by local influences of the Q10-effect due to local skin temperatures deviations from setpoints, $T_{sk,0}$ [18, 48] and local skin wettedness, f_{wsk} [6, 49]:

$$\frac{dm_{sw}}{dt} = c_{sw} f_{wsk}\, Sw \times 2^{\frac{T_{sk} - T_{sk,0}}{10}} \tag{20}$$

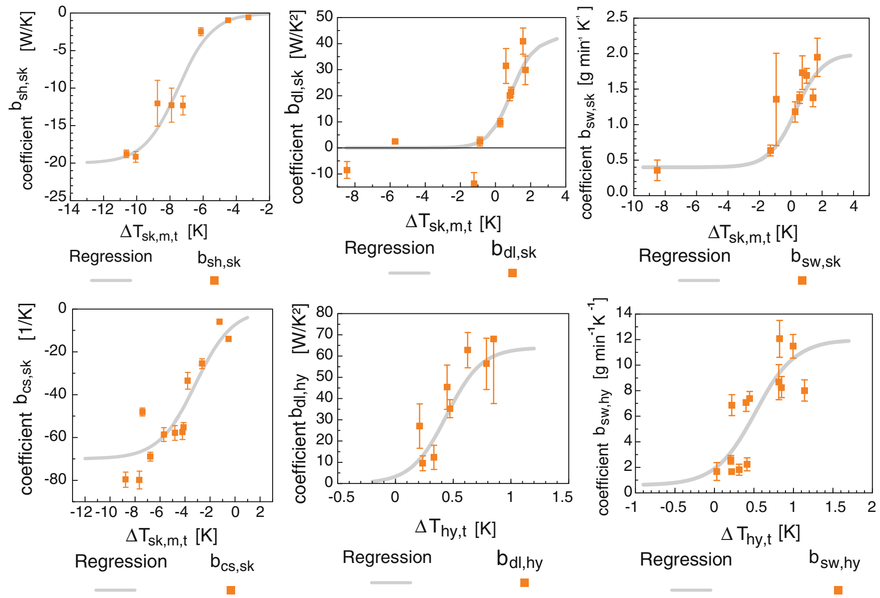

Fig. 9 Dependency of linear regression coefficients b obtained for different exposures on skin and head core temperature levels, i.e. error signals ΔT_{sk} and ΔT_{hy} obtained for the *Sh-*, *Sw-*, *Dl-*, and *Cs*-response, and the corresponding fitting functions (adopted from [18] with permission)

No statistically significant effect of the body core temperature was found in the response of peripheral vasoconstriction, Cs [−]. The control equation thus only includes static and dynamic afferent signals associated with the skin temperature:

$$Cs = 35[\tan h(0.34\Delta T_{skm} + 1.07) - 1]\Delta T_{skm} + 6.8\Delta T_{skm}\frac{dT^{-}_{skm}}{dt} \tag{21}$$

In contrast to constriction, vasodilatation, Dl (W K^{-1}), is regulated by statistically significant afferent signals associated with both the skin and body core temperature in the model:

$$\begin{aligned} Dl = {} & 21[\tan h(0.79\Delta T_{skm} - 0.70) + 1]\Delta T_{skm} \\ & + 32\left[\tan h\left(3.29\Delta T_{hy} - 1.46\right) + 1\right]\Delta T_{hy} \end{aligned} \tag{22}$$

The resultant local blood perfusion rates, w_{bl}, within the skin are obtained as local portions c_{dl} and c_{cs} [24] of the overall vasomotor responses also being modulated by local skin temperatures:

$$\beta_{sk} = \frac{\beta_{sk,0} + c_{dl}\,Dl}{1 + c_{cs}Cs \cdot e^{-Dl/80}} \times 2^{\frac{T_{sk} - T_{sk,0}}{10}} \tag{23}$$

Literature indicates a maximum skin blood flow that varies with the type and level of exercise [61]. In the model, the maximum overall skin blood flow, $B_{sk,\max}$ (W K^{-1}), decreases in proportion with any overall increase in muscle blood flow, ΔB_{msc} (W K^{-1}) [18]:

$$B_{sk,\max} = 386.2 - 0.32 \times \Delta B_{msc} \tag{24}$$

If the sum of all local blood flows in the skin as prescribed by Eq. 23 exceeds $B_{sk,\max}$ at any exercise condition then the local rates are trimmed in proportion to $B_{sk,\max}$ in the model.

3.2 Personalized Thermoregulation

Human thermoregulatory responses are known to be affected by personal characteristics such as gender, age, physical fitness, anthropometry and body composition, acclimatisation, etc. Of the various personal characteristics there are four main factors which affect the human individual heat stress response [33]: aerobic fitness, acclimatization, and anthropometric and morphological properties of the body. Other factors including age and gender lose their influence when 'correcting' observed responses for the effect of the maximum aerobic power, $VO_{2,\max}$, and the body fat content [33].

Of the four main personal factors the individualized passive system described above implicitly accounts for the effect of a person's anthropometry and morphology including related factors involved in the environmental heat exchange, e.g. skin surface area, bodily thermal insulation (fat content), heat capacity (body mass), and the body surface-to-volume ratio.

The individual heat stress response model of Havenith [34] was adapted for use with the FPC-model to account for the effect of the remaining two personal factors, i.e. the aerobic fitness and acclimatization status. Personal parameters that directly and indirectly affect responses of the active system model are depicted in Fig. 10.

The approach is based on the assumption that individual variations in thermoregulatory responses are associated with a shift of the setpoint of the body core temperature, i.e. hypothalamus temperature in the FPC-model, $\Delta T_{hy,set}$ (°C):

$$T_{hy,set} = T_{hy,set,0} - \Delta T_{hy,set} \tag{25}$$

where $T_{hy,set}$ (°C) is the adapted setpoint temperature, and $T_{hy,set,0}$ (°C) the setpoint temperature of the reference active system simulating an 'average' person. According to Havenith [34] the shift is a function of the physical fitness, *fit*, and the acclimatization status, F_{ac}:

$$\Delta T_{hy,set} = 0.1 \times \frac{fit}{10} + 0.25 \times F_{ac} \tag{26}$$

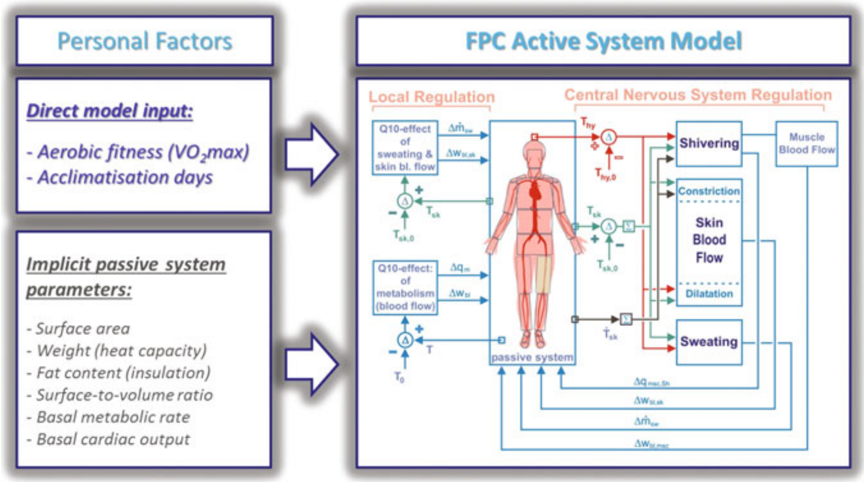

Fig. 10 Personal factors used as 'direct' and 'indirect' (passive system variations) input into the active system model to simulate responses of individuals

The personal fitness, *fit*, is dealt with in terms of the individual maximum aerobic power, $VO_{2,\max}$ (ml kg^{-1}min^{-1}) as direct model input. The difference between the individual and the average maximum aerobic power of an average-fit person, i.e. 40 ml kg^{-1}min^{-1}, is used as a measure of the individual fitness, *fit*:

$$fit = VO_{2,\max} - 40 \tag{27}$$

where $VO_{2,\max}$ may vary between $20 \leq VO_{2,\max} \leq 60$ ml kg^{-1}min^{-1} for unfit and trained individuals, respectively [34]. Quantities exceeding these limits are trimmed accordingly in the model.

The acclimatization status, F_{ac}, is a function of the number of acclimatization days, n_d:

$$F_{ac} = 1 - \exp(-0.3\, n_d) \tag{28}$$

Thereby the number of acclimatization days, which is another direct input parameter, varies between $0 \leq n_d \leq 14$ days in the model [34].

The central effector outputs for sweating, *Sw*, and vasodilatation, *Dl*, are computed as in the reference model by Eqs. 19 and 22, respectively, but using modified afferent signals from the head core, i.e. $\Delta T_{hy} = T_{hy} - T_{hy,set}$ with T_{hy} being the actual head core temperature and $T_{hy,set}$ the corresponding modified setpoint temperature, Eq. 25. If both *fit* and F_{ac} are zero also $\Delta T_{hy,set}$ is zero and the original setpoint temperature of 37.0 °C applies. However, any shift $\Delta T_{hy,set} > 0$ causes a shift of the onset of sweating and skin blood flow towards lower body core temperatures and vice versa.

In addition to the shift of the body core temperature setpoint due to acclimatization and physical fitness, the gain factor, g_{sw}, influences the sweating response in the individualized model:

$$g_{sw} = \left(1 + 0.35 \times \frac{fit}{20}\right) \times (1{+}0.15 \times F_{ac}) \tag{29}$$

g_{sw} modulates the sweating response, Sw, predicted by Eq. 19 to obtain the individualized response, Sw_{ind} (g min^{-1}):

$$Sw_{ind} = Sw \times g_{sw} \tag{30}$$

The reference upper limit for sweating, $Sw_{\max}$, of 30 g min^{-1} is also subject to personal variations which is accomplished using the factor $f_{sw,\max}$:

$$f_{sw,\max} = 1 + 0.25 \times \frac{fit}{20} + 0.25 \times F_{ac} \tag{31}$$

with $f_{sw,\max}$ the individual maximum sweat rate, $Sw_{\max,ind}$ (g min^{-1}) yields:

$$Sw_{\max,\mathrm{ind}} = Sw_{\max} \times f_{sw,\max} \tag{32}$$

In the model the maximum overall skin blood flow, $B_{sk,\max}$, is not constant but varies with the intensity of exercise. The individualized maximum is derived from $B_{sk,\max}$ using the factor $f_{Bsk,\max}$ which is a function of the aerobic fitness and acclimatization status:

$$f_{Bsk,\max} = 1 + 0.25 \times \frac{fit}{60} + 0.25 \times F_{ac} \tag{33}$$

For any activity level, the individualized overall maximum skin blood flow, $B_{sk,\max,ind}$ (W K^{-1}) then yields:

$$B_{sk,\max,\mathrm{ind}} = B_{sk,\max} \times f_{Bsk,\max} \tag{34}$$

4 Human Environmental Heat Exchange

4.1 Heat Exchange Components

Humans exchange heat with the environment by surface convection, evaporation of sweating, thermal radiation with enclosing surfaces, short wave irradiation from high intensity sources, or conduction with surfaces in direct contact with the body. Part of bodily heat is also lost to the ambient air by respiration from within the lungs and nasal cavity. The surface heat losses may vary considerably over body regions due to inhomogeneous thermoregulatory responses, non-uniform clothing or asymmetric environmental conditions. Multi-segmental models are capable of accounting for such typical human inhomogeneities by establishing local heat and mass balances at surface sectors of individual body elements. Thereby, the net heat exchange between a body sector and the environment is a sum of the individual heat exchange components:

$$q_{bs} = q_c + q_e + q_r + q_{rs} \tag{35}$$

where q_{bs} (Wm^{-2}) is the total heat loss from a body sector, q_c (Wm^{-2}) heat loss by convection, q_e (Wm^{-2}) sweat moisture evaporation and skin moisture diffusion, q_r (Wm^{-2}) long wave radiation, q_{rs} (Wm^{-2}) short wave irradiation. If the body surface is in direct contact with another surface, then q_{bs} equals to the rate of heat that is transferred by conduction from the interface to the material in contact with the human body, q_{cn} (Wm^{-2}).

Convective heat losses, q_c (W m^{-2}), are calculated using local surface temperatures of body sectors, T_{bs} (°C), the corresponding local air temperatures, T_a (°C), and convective heat transfer coefficients, h_c (W m^{-2}K^{-1}):

$$q_c = h_c(T_{bs} - T_a) \tag{36}$$

In the FPC-model, h_c are obtained for individual body sectors considering both free and forced convection. The coefficients are thus a function of the temperature difference between the sector surface and air, as well as the local air speed, v_a (m/s):

$$h_c = \left[a\,(T_{bs} - T_a)^{1/2} + bv_a + c\right]^{1/2} \tag{37}$$

where a, b and c are regression coefficients fitting experimental observations. Basis for h_c are measurements of local heat losses from heated full-scale manikins with realistic skin temperature distributions [24]. For special-purpose applications such as face cooling effects in windy outdoor climate conditions or water immersions dedicated local convection coefficients are computed for exposed or immersed body parts within the model [3, 40, 66].

Sweat moisture may or may not evaporate from the skin depending on the evaporative potential between the skin and the air, $p_{sk} - p_a$ (Pa), and the local evaporative resistance of the clothing, R_{ecl} (W m^{-2}K^{-1}). Therefore, evaporation of sweating is not considered a direct response of the human thermoregulatory system in the FPC-model; rather the regional portion of regulatory sweat moisture secretion dm_{sw}/dt (kg s^{-1} m^{-2}) is used to constitute the latent heat balance for each skin sector:

$$\frac{p_{sk} - p_a}{R_{ecl}} = \frac{p_{ss} - p_{sk}}{R_{esk}} + \lambda_{H_2O}\frac{dm_{sw}}{dt} \tag{38}$$

where p_{sk}, p_a and p_{ss} (Pa) are partial water vapour pressures at the skin surface, of the air, and the saturated value within the skin (location of sweat glands), respectively [17]; λ_{H_2O} (J kg^{-1}) is the (temperature dependent) heat of vaporisation of water [60], and $R_{sk,e}$ (kg s^{-1} m^{-2}) the moisture permeability of the outer skin [17]. The maximum possible skin evaporation rate is reached when p_{sk} rises to the saturation level. Excessive sweating, dm_{ac}/dt (kg s^{-1} m^{-2}), then cannot evaporate may accumulate at the skin:

$$\frac{dm_{ac}}{dt} = \frac{dm_{sw}}{dt} - \frac{p_{sk} - p_a}{R_{ecl}\,\lambda_{H_2O}}. \tag{39}$$

The accumulation capacity, however, is not unlimited. According to Jones et al. [39], quantities exceeding 35 g m^{-2} drip off.

The local long wave radiative heat loss from body sector surfaces can be expressed, in analogy to surface convection, as:

$$q_r = h_r(T_{bs} - T_{env}) \tag{40}$$

where h_r (W m^{-2} K^{-1}) is the radiative heat transfer coefficient:

$$h_r = \sigma \varepsilon_{bs}\, \varepsilon_{en}\, \phi_{bs-en} \left(T_{bs}^2 + T_{en}^2\right) \left(T_{bs} + T_{en}\right) \tag{41}$$

with $\sigma = 5.67 \times 10^{-8}$ W m^{-2} K^{-4} being the Stefan-Boltzmann constant, ε_{bs} and ε_{en} emissivity of the sector surface and the environment, respectively, $\varphi_{\ bs\text{-}en}$ view factor between the body sector and the enclosure 'seen' by that sector. T_{bs} and T_{en} (K) are surface temperatures of the body sector and of the corresponding radiant enclosure, respectively. The view factors $\varphi_{\ bs-en}$ of individual body sectors [22] with respect to the entire enclosure 'seen' are based on numerical radiation simulations of projected area factors, f_p, calculated using detailed 3D human geometry models [41].

Simulation of the human radiative heat exchange in asymmetric environments (e.g. exposures to proximal cold surfaces, Fig. 11, right) uses f_p-factor-based calculations of view factors, φ_{bs-en}, between individual body sectors and surfaces of the envelope, A_w. This approach employs the concept of directional mean surface temperatures of the enclosure, T_{en}, to enable fast and accurate simulations. The geometry configuration parameters involved in the calculation of view factors between a body sector and a surface of the envelope, is depicted in Fig. 11, left.

The short wave radiation q_{rs} (W m^{-2}) absorbed by a body sector is given by:

$$q_{rs} = \alpha_{bs}\, f_p\, I_s \tag{42}$$

where α_{bs} is the short-wave absorptivity of the body surface which depends on the colour of skin or the material covering the sector, I_s (Wm^{-2}) is the incident short-wave (e.g. direct or diffuse solar) radiation, and f_p the sector's corresponding projected area factor. The local human f_p -factors were modelled for both direct and diffuse radiation using detailed 3D human geometry and radiation models [42, 43] as a function of the azimuth and altitude angles (direct radiation, Fig. 12) and the ground albedo (diffuse radiation).

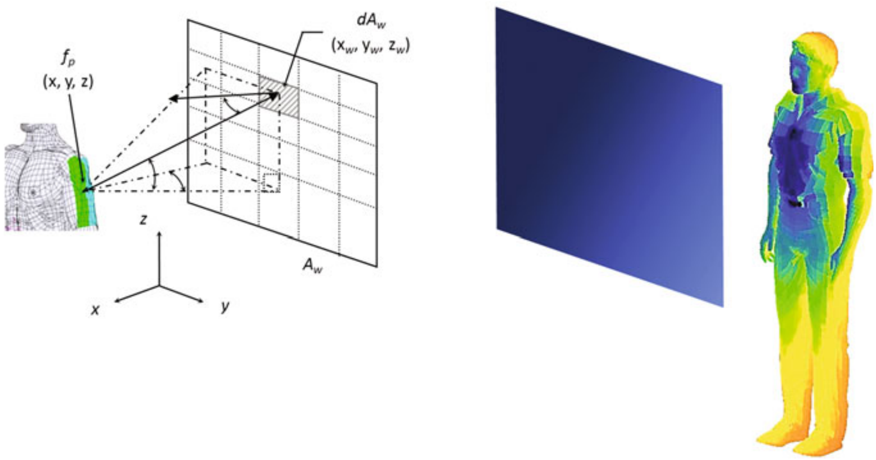

Fig. 11 Geometry-related parameters used to calculate view factors between body sectors and a wall surface, A_w (*left*). Regional body cooling due to asymmetric thermal radiation exchange with a proximal cold surface (*right*). Adopted from [41] with permission

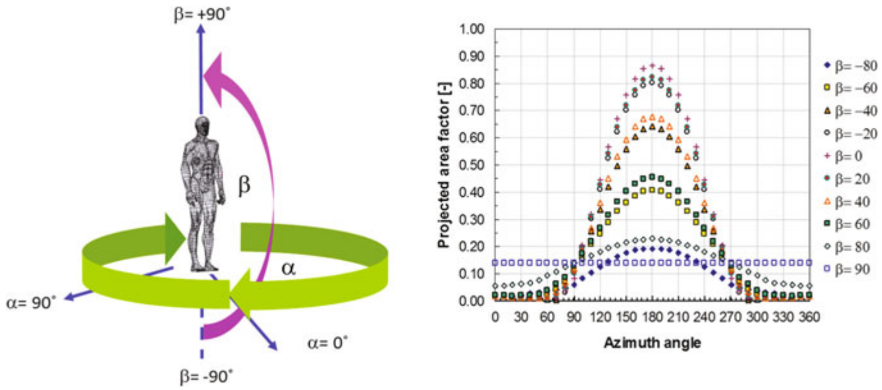

Fig. 12 Variation range of altitude (β) and azimuth (α) angles studied in the numerical radiation simulations (*left*); example of resultant local projected area factor curves: posterior thorax (*right*). Adopted from [42] with permission

Respiration heat losses typically only represent a small fraction of the heat loss humans release from their body surface to the environment. According to Fanger [16], the latent part of the respiratory heat loss, E_{rsp} (W), is a function of the pulmonary ventilation rate which depends on the whole body metabolism, Q_m (W), the latent heat of vaporisation of water, and the difference between the humidity ratio of the exhaled and inhaled air which depends on air temperature, T_a (°C), and the partial vapour pressure of the ambient air, p_a (Pa). An explicite inclusion of T_a in the formula and re-arranging yields:

$$E_{rsp} = 3.233 \times 10^{-3} Q_m (27.7 - 0.065\, T_a - 0.00491\, p_a) \tag{43}$$

Since the enthalpy of the exhaled air depends to a certain degree upon the condition of the inhaled air, E_{rsp} appears as a function of both T_a and $p_{a.}$ The inclusion of T_a allows E_{rsp} to be calculated for a wide range of ambient air conditions.

The dry respiratory heat loss by convection due to the difference between the temperature of the exhaled and inhaled air is a function of the pulmonary ventilation rate, and the temperature and vapour pressure of the ambient air, T_a and p_a, respectively [16]:

$$C_{rsp} = 1.44 \times 10^{-3} Q_m \left(32.6 - 0.934\, T_a + 1.96\, \times 10^{-4} p_a\right) \tag{44}$$

The total respiratory heat losses $E_{rsp} + C_{rsp}$ is distributed over compartments of the pulmonary tract [17].

4.2 Clothing

Current clothing standards define 'global' thermal and evaporative resistances of garments, I_{cl} (m^2 K W^{-1}) and $R_{e,cl}$ (m^2 Pa W^{-1}), for the human body as a whole [14]. As this global insulation approach assumes all body parts (incl. head, face, hands, etc.) being covered by a (fictitious) uniform layer of clothing it appears inappropriate for use with multi-segmental models. To adequately model the local heat and mass transport processes the non-uniformity of clothing has to be considered using the actual, i.e. local, insulation properties of clothing at individual body parts.

The FPC-model is used in conjunction with a 'local' clothing model that predicts the required local resistances of garments from measured resistances of textile fabrics [46, 47]. The model simulates the *local* (steady state) heat and mass transfer processes within cloth composites considering also radiation effects and the insulation of any air trapped between clothing layers based on an approach by McCullough et al. [47]. The local clothing model employs an iteration algorithm to fine-tune predicted resultant local resistances of garments and clothing ensembles to ensure agreement with the overall resistances as measured in thermal manikin experiments.

Within the physiological model the resultant total local sensible thermal resistance R_{lT} (m^2 K W^{-1}) of a multi-layered composite covering a body sector is calculated as:

$$R_{lT} = \sum_{i=1} R_{lcl,i} + \frac{1}{f_{lcl}(h_c + h_r)} \tag{45}$$

where $R_{lcl,i}$ (m^2 K W^{-1}) is the *local* thermal resistance of the *i*-th clothing item covering that body part, h_c and h_r (W m^{-2} K^{-1}) are local surface coefficients for convection and long wave radiation heat exchange as defined above, and f_{lcl} [−] is the local clothing area factor defined as a quotient of radii of the outmost clothing layer, r_{cl} (m), and the skin, r_{sk} (m):

$$f_{lcl} = \frac{r_{cl}}{r_{sk}} \tag{46}$$

The corresponding total local evaporative resistance, $R_{e,lT}$ (m^2 Pa W^{-1}), is obtained from local evaporative resistances, $R_{e,lcl,i}$ (m^2 Pa W^{-1}), of individual clothing layers applied to the body sector as:

$$R_{e,lT} = \sum_{i=1}^{m} R_{e,lcl,i} + \frac{1}{f_{lcl}\, h_e} \tag{47}$$

where

$$h_e = L_a \times h_c \tag{48}$$

with h_e (W m^{-2} Pa^{-1}) being the local evaporative heat transfer coefficient and L_a the Lewis constant for air, L_a = 0.0165 K Pa^{-1} [27].

The incorporated clothing model offers options for considering additional phenomena and effects such as wind penetration, the impact of walking and the clothing habits of the population (Fig. 13) as discussed elsewhere [35].

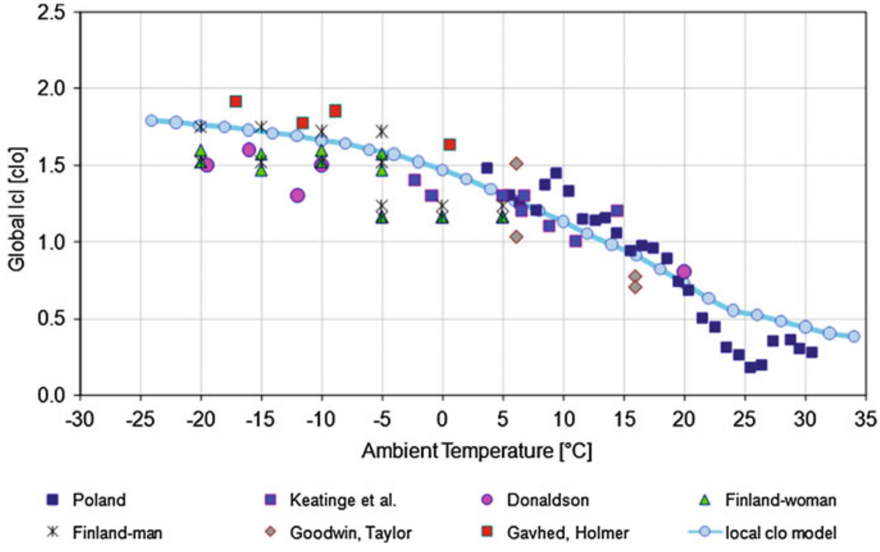

Fig. 13 Resultant global thermal clothing insulation values as a function of the ambient temperature reflecting the clothing habits of different European countries. Reproduced from [35] with permission

5 Coupled Simulation and Measurement Systems

Mathematical models of human thermoregulation typically incorporate simplified calculations of the human environmental heat exchange as an integral part of the simulation process. In reality, however, these processes as well as the environmental conditions themselves may be very complex so they either have to be measured directly or simulated using sophisticated numerical models of conjugate heat transfer.

From the mathematical point of view, the heat exchange with the environment represents a type of boundary conditions at the body surface that principally may be replaced by alternative formulations using e.g. measured or predicted surface heat fluxes or surface temperatures as model input. These quantitates too contain all the information that is required and is sufficient to fully describe the boundary conditions of an exposure for thermophysiological simulations.

As illustrated in Fig. 14, the FPC-model accepts the traditional definition of boundary conditions (Standard Model Input), i.e. environmental conditions with implicit calculations of environmental heat losses, as well as alternative definitions (Extended Model Input) using e.g. real-time measured skin temperatures, predicted heat fluxes, etc. as model input [22]. Thereby the extended model enables the type

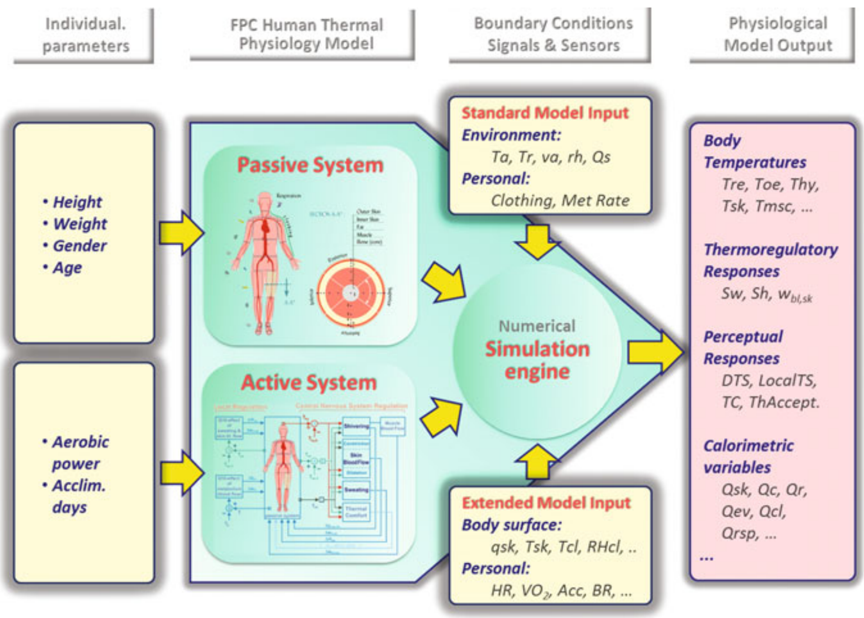

Fig. 14 Schematic diagram of the 'individualized' FPC-model extended for use with different types of boundary conditions at the body/skin surface (e.g. skin temperatures, Tsk, clothing temperatures, Tcl, or skin heat fluxes, qsk) and personal states (e.g. heard rate, HR, oxygen consumption, VO_2, breathing rate, BR, etc.) for easy and flexible coupling with other simulation tools or measurement systems

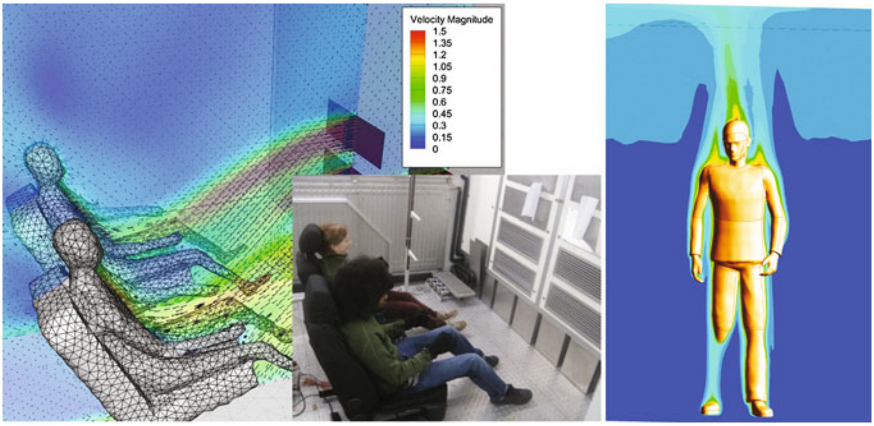

Fig. 15 Examples of 3D geometry models used for detailed human environmental heat transfer analysis: air flow and temperature distribution around a thermoregulated human body using coupled CFD-FPC-model simulation. Reproduced from [45] with permission

of boundary conditions to be defined flexibly for different body regions, body elements or individual sectors of the passive system.

Various industrial applications require detailed analysis of the complex environmental heat transfer processes involving human occupancy. Numerical simulation systems have thus been developed which couple the model e.g. with dynamic thermal simulation of vehicles [20, 21], numerical radiation scenario models [69], behavioural clothing models [35], temperature regulation during cardiac surgery [62], or sophisticated Computational Fluid Dynamics (CFD) simulation to aid the design of comfortable and energy-efficient cars and buildings [7, 45]; for further examples see e.g. [21].

With the extended definition of boundary conditions the FPC-model enables an easy and flexible coupling and integration with other simulation tools. In the examples illustrated in Fig. 15, detailed 3D human geometry models were used to facilitate the bi-directional data exchange between CFD and physiological models. In the coupling process CFD provided the environmental boundary conditions at surfaces of the simulated person needed for thermophysiological simulation while the physiological model delivered the required boundary conditions for CFD simulation. In the current FPC-model version users can set-up the coupling with CFD (or other tools) flexibly without the need to modify the source code of the models.

Coupled systems may also involve physical models such as thermal manikins. Accurate numerical simulations generally require detailed knowledge of scene-specific factors and parameters of the human environmental heat exchange. This information may be readily available for well-controlled laboratory conditions but difficult to obtain for human exposures in the field. Combined hardware-software solutions may thus be superior when predicting human responses to the complex conditions of uncontrolled real-world exposure scenarios.

As an example, the system illustrated in Fig. 16 uses the FPC-model to provide a heated sweating cylinder *Torso* with ‘thermophysiological intelligence’ [55, 56]. This

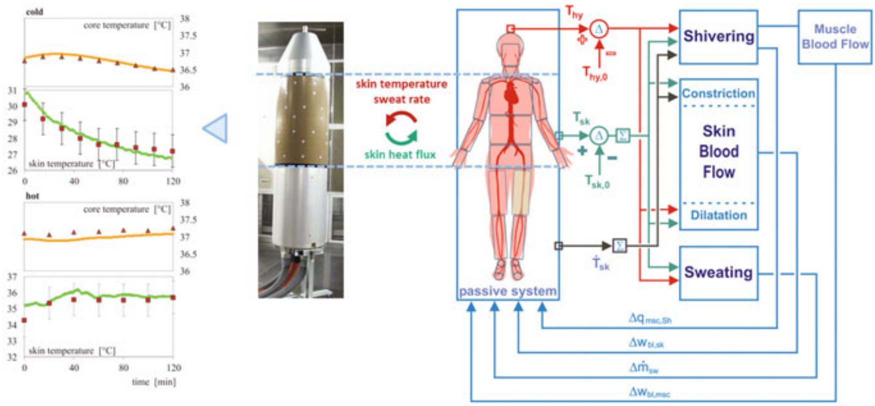

Fig. 16 Scheme of data exchange in the single-sector thermophysiological 'human simulator' [55], reproduced with permission, graphs on the left reproduced by permission of IOP Publishing.

'human simulator' device measures the net surface heat fluxes which result from the complex heat and mass transfer processes taking place within clothing systems thus enabling measurement of clothing properties under physiologically realistic conditions.

Other examples include 'intelligent' measurement systems which deploy numerical simulation of human heat transfer to postprocess and interpret measured data. An example of such a system is the development of a non-invasive monitoring and warning system (MWS) that uses wearable sensors to determine internal temperature and heat stress levels of people working in hot industrial environments [22, 23]. MWS can measure skin/clothing temperatures, Fig. 16, providing the required boundary conditions at the body surface of the wearer. This information is used by the FPC-model to assess the body core temperature of people exposed to different environmental and personal conditions. Body core temperatures are determined non-invasively by means of numerical simulation—but without the need for detailed knowledge of scene-specific factors and parameters. Validation examples of predicted rectal temperatures using the personalized model are provided further below (Fig. 17).

6 Predicting Human Thermophysiological Responses

In the past, the standard model was subject to various general as well as application-specific validation studies. The tests included climate chamber experiments with exposures to wide-ranging steady state and transient environmental and personal conditions [18], diverse indoor climate conditions with/without asymmetric radiation in cars and buildings [19, 20, 42, 43], exposures to heat stress conditions of subjects wearing protective clothing [22, 23, 58], etc. A large-scale, international

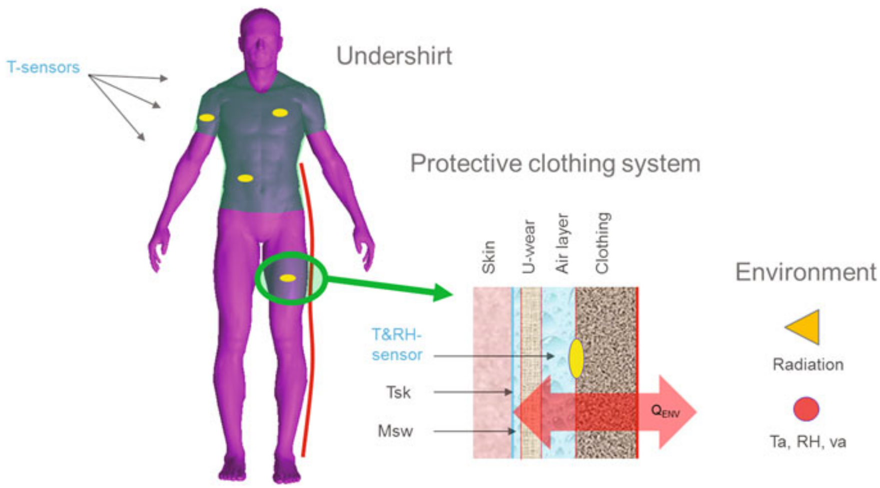

Fig. 17 Peripheral sensor set-up of a monitoring and warning system for non-invasive body core temperature assessment in individuals exposed to different environmental and personal conditions and wearing different types of (protective) clothing systems

study also addressed field surveys and exposures to uncontrolled outdoor climates including weather extremes [57].

Fig. 18 compares measured and predicted responses of semi-nude subjects exposed for one hour to different steady-state ambient temperatures ranging from 5 and 48 °C

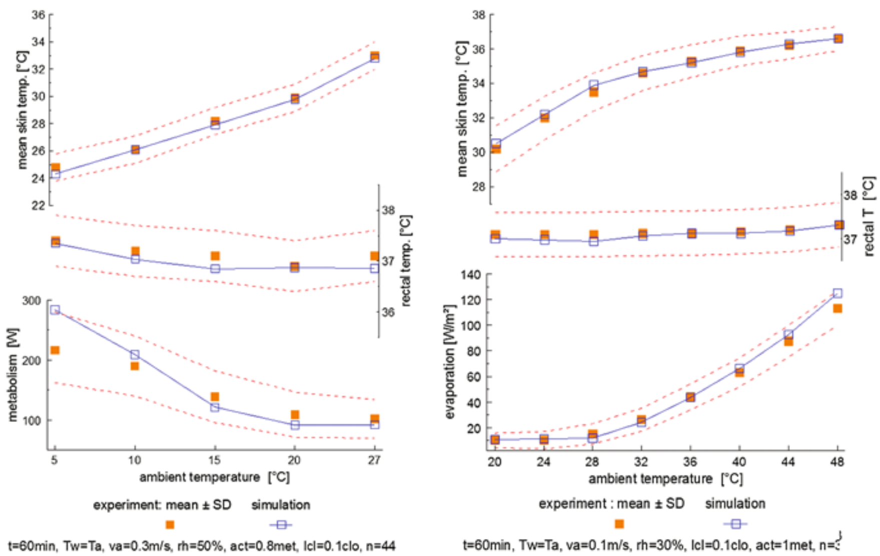

Fig. 18 Mean skin temperature, body core temperature and thermoregulatory responses as measured and predicted over wide-ranging steady state environmental conditions. Reproduced from [18] with permission

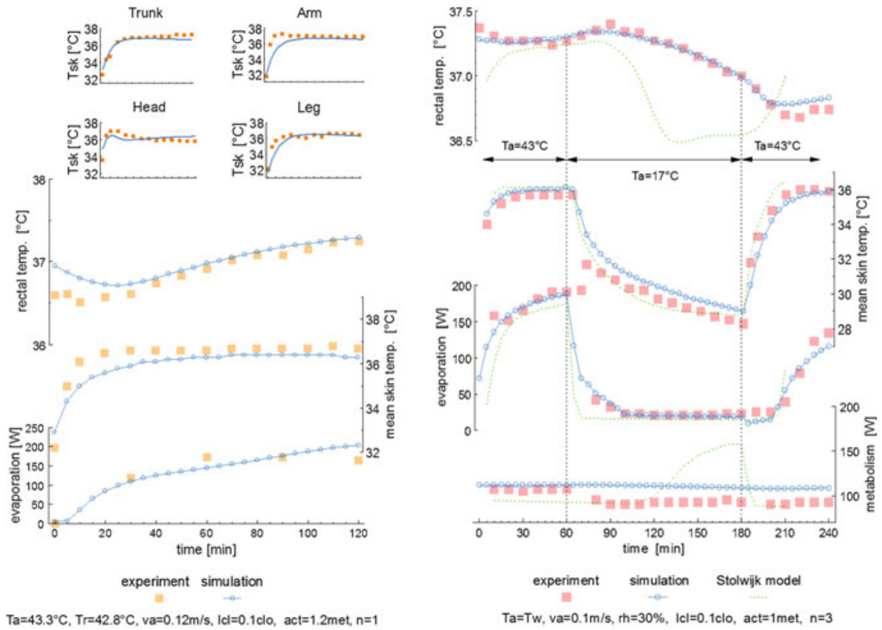

Fig. 19 Body temperature and thermoregulatory responses as measured and predicted for step changes in room temperature into a hot environment of Ta = 43 °C (*left*) and from a hot environment of 43 °C to a cold environment of 17 °C and back again (*right*). Reproduced from [18] with permission

[18]. The experimental results collected from several climate chamber experiments are provided as averages with the corresponding standard deviations (SD).

The dynamic behaviour of the model was examined extensively for different types of transient exposure conditions. Comparisons of predicted body temperature and regulatory responses with experimental data obtained using the standard model for sudden changes in environmental temperatures from neutral to hot and from hot to cold and back again are illustrated in Fig. 19.

Also the individualized FPC-model was subjected to several validation tests [22, 23]. In one test, unacclimatized semi-nude male subjects featuring different personal characteristics and fitness levels [38, 57] were exposed to steady indoor climate conditions of 28 °C ambient temperature ($T_a = T_{wall}$), va = 3.8 m/s air speed, 50 % relative humidity while exercising on a treadmill at levels up to 13 met (90 % $VO_{2,max}$). The following validation tests focused on the performance of the individualized model to predict body core (rectal) temperature responses [22]. In the simulations, information on personal characteristics including stature, body weight, gender and age, as well as information on fitness and acclimatization levels were used as input into the model along with environmental conditions and exercise intensity.

In Fig. 20 predicted rectal temperatures are compared with measured data for average-fit, semi-nude male subjects exercising at an intensity level of 8.9 met. To simulate group-average responses the model was run with the following data

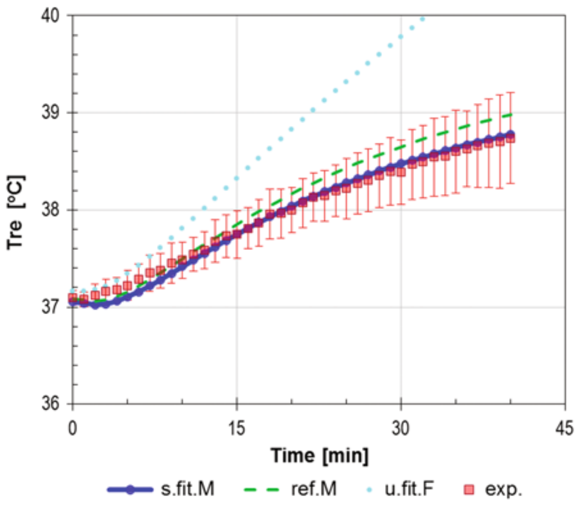

Fig. 20 Average rectal temperature responses of semi-nude male standard-fit subjects (n = 8) exercising at 8.9 met with **s.fit.M**: standard-fit male ($VO_{2,max}$ =40 ml $kg^{-1}min^{-1}$), **ref.M**: reference model (40 ml $kg^{-1}min^{-1}$), **u.fit.F**: unfit female (20 ml $kg^{-1}min^{-1}$), **exp.**: experiment

representing averages of eight subjects: stature = 1.81 m, weight = 74.1 kg (BMI = 22.6 kg/m^2), age = 33.4 years, unacclimatized, 'standard' fitness (s.fit.M). Rectal temperature predicted using this configuration agreed with measured responses within a *root mean square deviation* [2], *rmsd*, of <0.1 °C. For comparison also rectal temperatures predicted using the Reference Model (ref.M) and an unfit female (u.fit.F) are plotted in Fig. 20. The role of aerobic fitness and body composition (sex) on predicted rectal temperatures is apparent from the results obtained for a standard-fit male and unfit female both featuring the above group-average body height and weight. Substantial discrepancies from measured rectal temperatures resulted for the simulated female indicating the significance of the physical fitness and body composition under conditions of intense exercise.

The above example studied average-group responses. Rectal temperatures predicted for two individual athletes are compared with measured temperatures in Fig. 21. The left-hand diagram refers to the smallest and lightest subject of the athlete group who was 19 years old, 1.72 m tall, and weighed 62.7 kg (BMI = 21.2 kg/m^2). The right-hand diagram of Fig. 21 depicts the response of the tallest and heaviest athlete in the trials with a body height of 1.84 m, and weight of 79.7 kg (BMI = 23.5 kg/m^2), aged 20 years. The simulations (fit.M) reproduced measured rectal temperatures (exe.) by rms-deviations of <0.2 and <0.1 °C, respectively.

For comparison also rectal temperature predicted for a standard-fit male (s.fit.M) is plotted in Fig. 21, left. The predicted rectal temperature reached 41 °C at the end of the exposure, again underlining the need to consider physical fitness in conditions of very high-intensity exercise, i.e. 13.2 met in this case.

In Fig. 22, two different types of boundary conditions (BC) at the skin were used to predict body core temperatures of semi-nude sedentary subjects exposed to step changes in air temperatures of 29-38-28 °C [22, 32]. In the 'classic' BC-type mode

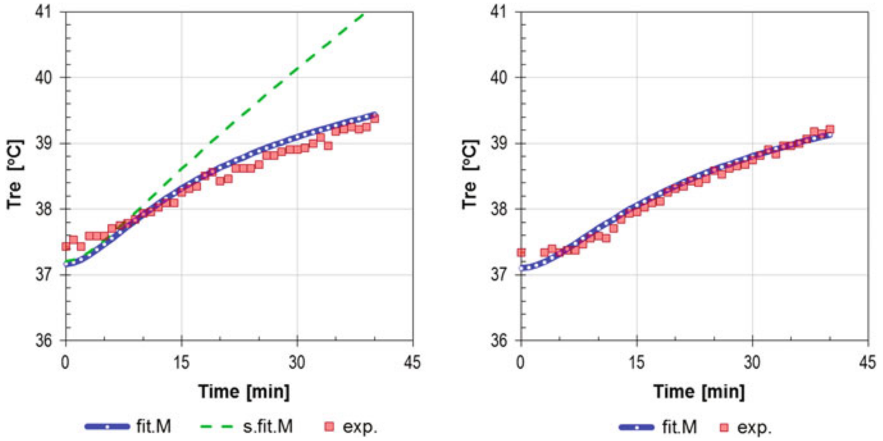

Fig. 21 Comparison of predicted and measured rectal temperatures of single male athletes: (i) response of the smallest and lightest athlete in the group exercising at 13.2 met (*left*) and (ii) group tallest and heaviest athlete exercising at 11.5 met (*right*) with **fit.M**: fit male ($VO_{2,max}$ = 60 ml $kg^{-1}min^{-1}$), **s.fit.M**: standard-fit male (40 ml $kg^{-1}min^{-1}$), **exp.**: experiment

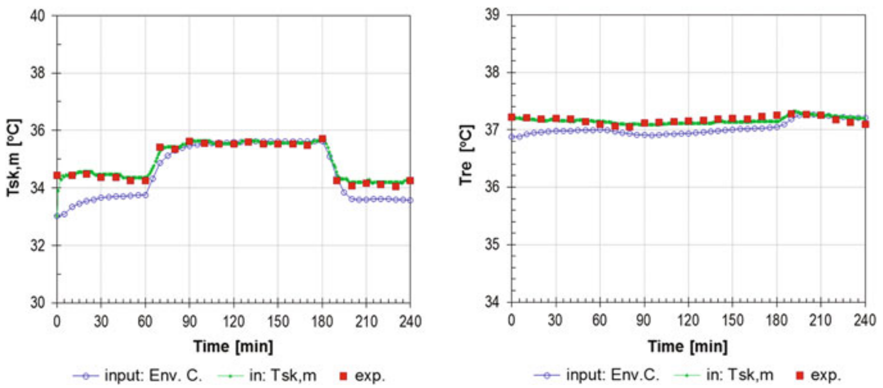

Fig. 22 Exposure of sedentary (act = 1.0 met) semi-nude (Icl = 0.1 clo) subjects (n = 3) to sudden changes in environmental temperature (T_{wall} = Ta), from neutral (29 °C) to warm (38 °C) and back to neutral (28.0 °C), v_a = 0.1m/s, and RH = 40-33-41 % [32]. Rectal temperature was predicted using two definitions of boundary condition at the skin: classic (input: Evn.C.) and alternative (in: Tsk,m) method using environmental conditions and mean skin temperature, respectively, as model input (see text for more detail)

('input: Env.C.') rectal temperature was predicted by the FPC-model simulating a standard person from measured climatic conditions (and metabolic rates) as model input calculating the subjects' dry and latent heat losses to the environment. In the alternative BC-type mode ('in: Tsk,m' in Fig. 22) observed mean skin temperatures and metabolic rates were used as model input avoiding so calculations of environmental heat losses.

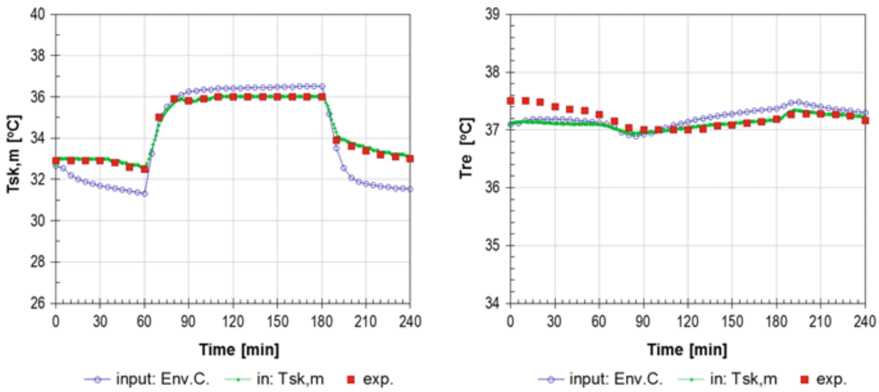

Fig. 23 Exposure of sedentary (act = 1.0 met) semi-nude (Icl = 0.1 clo) subjects (n = 3) to sudden changes in environmental temperature ($T_{wall} = T_a$), from cool (22 °C) to hot (44 °C) and back to cool (23 °C), v_a = 0.1 m/s, and RH = 40 % [63]. Rectal temperature was predicted using two definitions of boundary condition at the skin: classic (input: Evn.C.) and alternative (in: Tsk,m) method using environmental conditions and mean skin temperature, respectively, as model input (see text for more detail)

In both cases rectal temperature was predicted with an error of less than 0.2 °C rmsd; though the alternative method better reproduced measured data.

The results obtained for a 22-43-22 °C scenario are plotted in Fig. 23 [22, 63]. Except for the initial period, both the level and the dynamics of the measured response were reproduced well by the alternative method using measured skin temperature as model input (note: in the classic Env.C.-method skin temperature, Fig. 23 left, was predicted as a result of calculated environmental heat losses). In both cases rectal temperature was predicted within 0.2 °C rmsd.

The performance of the individualized model predicting body core temperature from measured skin temperatures was tested for subjects wearing permeable and impermeable protective clothing [23]. In the climate chamber trials subjects were undergoing periods of rest and exercise under different combinations of environmental and clothing conditions [10]. Although the multi-segmental FPC-model accepts individual local skin temperatures, mean skin temperatures were used as model input in this study. In the base-case simulation series (Tsk,m), the mean skin temperature was calculated as a weighted average of measured local skin temperatures using a nine-point relationship [31]. In the second simulation series (Tsk,c) local skin temperatures solely from the central body parts (five-point average) were used. The third simulation series (Ehx) was conducted for comparison calculating environmental heat losses from measured climatic conditions and thermophysical clothing properties. In all simulations, experimental metabolic rates were used as an additional model input as were personal body characteristics aerobic fitness and the acclimatisation status of the subjects.

A comparison of predicted and measured responses for subjects wearing impermeable coveralls exposed to moderate environmental conditions of 25.7 °C

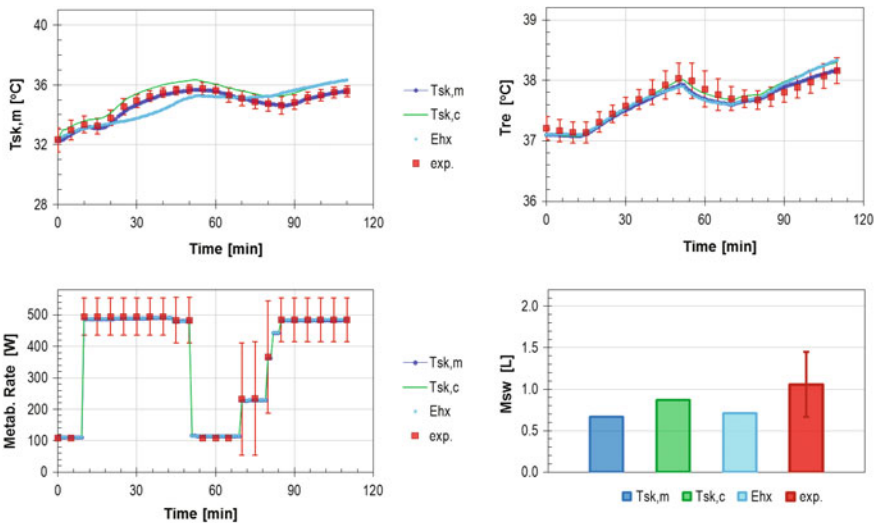

Fig. 24 Responses of subjects (n = 13) wearing an impermeable suit exposed for 110 min to a moderate environment of 25.7 °C air temperature (50 % RH, 0.3 m/s air speed). The predicted group-average rectal temperature (Tre) and total body weight loss (Msw) were predicted from different configuration of skin temperature sensors (Tsk,m and Tsk,c) and environmnetal conditions (Env) as input boundary conditions (see text)

air temperature is provided in Fig. 24. Using the Tsk,m-method rectal temperature was predicted within the range of experimental standard deviation throughout the exposure. The Tsk,c-method overestimated the (nine-point) mean skin temperature by about 1 °C. As a result, rectal temperature was overpredicted slightly at the end of the exposure although the method still delivered a comparable accuracy, rmsd <0.1 °C. No substantial differences from experimental observations resulted also for rectal temperatures predicted using the Ehx-method with rmsd <0.2 °C.

Predicted and measured group-average responses obtained for a hot environment of 40.4 °C with subjects wearing impermeable clothing are plotted in Fig. 25. The inability of the subjects to evaporate sweating caused a notable increase of the heat stress level associated with a rise of the mean skin temperature, rectal temperature, and total body moisture losses by 0.8, 0.3 °C and 0.1 L, respectively. Under such uncompensable heat stress conditions only the Tsk,m- and Tsk,c-mode methods predicted rectal temperature within the range of the experimental standard deviation with rmsd <0.15 °C. The Ehx-mode method obviously overestimated the cooling effect of sweat evaporation causing rectal temperature to be under-predicted by up to −0.6 °C with a rmsd >0.4 °C.

The impact of an additional thermal load of 600 W/m^2 due to (simulated) direct solar radiation incident on the back of the subjects wearing permeable clothing is apparent from Fig. 26. Also under these asymmetric radiation conditions the Tsk,m- and Tsk,c-methods reproduced experimental observations of the rectal temperature

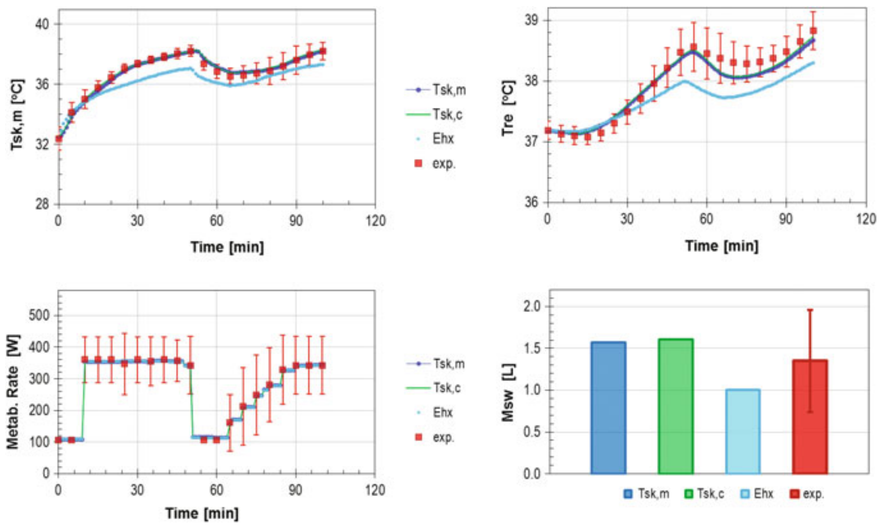

Fig. 25 Responses of subjects (n = 14) wearing an impermeable suit exposed for 100 min to a hot environment of 40.4 °C air temperature (23.4 % RH, 0.4 m/s). The predicted group-average rectal temperature (Tre) and total body weight loss (Msw) were predicted from different configuration of skin temperature sensors (Tsk,m and Tsk,c) and environmnetal conditions (Env) as input boundary conditions

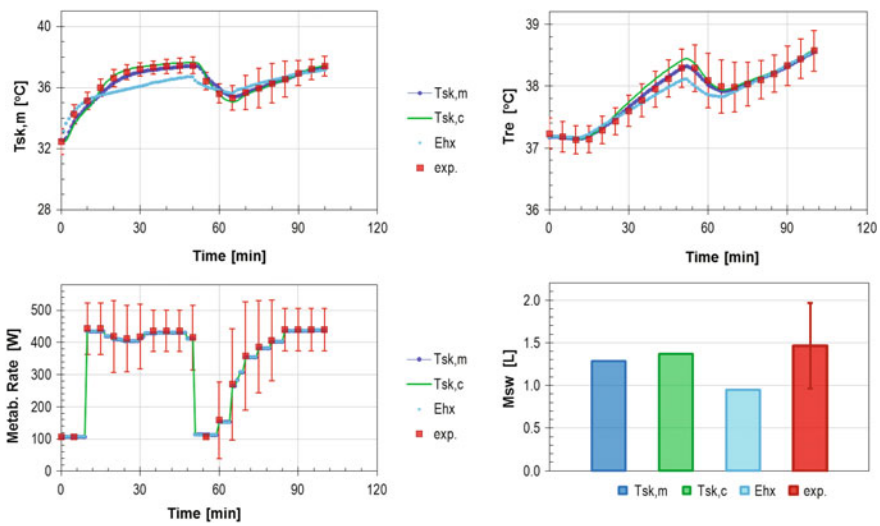

Fig. 26 Responses of subjects (n = 17) wearing a permeable suit exposed for 110 min to hot environmental conditions of 40.8 °C air temperature and 600 W/m^2 direct short wave radiation (22.8 % RH, 0.3 m/s). The predicted group-average rectal temperature (Tre) and total body weight loss (Msw) were predicted from different configuration of skin temperature sensors (Tsk,m and Tsk,c) and environmental conditions (Env) as input boundary conditions

within 0.1 °C rmsd. The Ehx-method required preliminary simulations of the 3D-radiation scene and the amount of radiation incident on irradiated body parts. With this information the Ehx-method predicted the rectal temperature with rmsd <0.2 °C.

7 Summary and Conclusions

This chapter dealt with aspects of modelling the human heat transfer and thermoregulatory system. Approaches to simulate the passive system included modelling the human body based on results of anthropometric field surveys, and heat and mass transfer processes that occur inside the body and at its surface. The statistically founded active system was formulated by means of meta-regression analysis of thermoregulatory responses observed in published experimental studies collected to cover a wide range of exposure conditions. The set-point based feedback active system uses afferent signals associated with temperatures of the skin and head core, and the rate of change of skin temperature to predict the responses of sweating, shivering and peripheral vasomotion to static and dynamic thermal stimuli.

The passive and active system models were extended to simulate individuals. The personalized models account for variations in anthropometric and morphological body properties, aerobic fitness, acclimatization status and their effect on human thermophysiological responses using readily-available personal data as model input. The individualized FPC-model accepts different types of boundary conditions at the surface of the simulated person. This way an easy and flexible coupling with other simulation models or measurement systems is possible. The model can so be 'connected' e.g. to different wearable sensors to assess, non-invasively, internal temperatures of people in real-life exposure settings.

The model was validated against experimental data in diverse general as well as application-specific validation studies. Validation tests discussed in this chapter included responses to wide-ranging steady state and transient environmental conditions, and responses of sedentary and exercising subjects wearing different types of clothing and featuring different personal characteristics.

The tests also included non-invasive assessments of the rectal temperature using measured skin temperatures and metabolic rates as model input. The numerical simulation approach proved to be a robust predictor of the body core temperature under a broad range of exposure scenarios. Rectal temperatures were predicted with a rms-deviation of <0.2 °C for individuals and <0.15 °C for group-average responses irrespective of the test environmental conditions, exercise intensities, type of (permeable/impermeable) clothing, and presence or absence of high intensity radiation sources. It is concluded that—unlike empirical data-driven algorithms—this first-principles numerical approach is applicable beyond the range of experimental data used to develop it including different exposure circumstances, personal characteristics, exercise levels or types of clothing.

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Military Clothing and Protective Material: Protection at the Limits of Physiological Regulation

Nigel A.S. Taylor and Mark J. Patterson

Abstract Contemporary military environments almost invariably require the use of personal protective clothing and equipment, and the burden accompanying its use can sometimes challenge the integrated regulation of critical physiological variables, pushing some individuals to the limits of regulation. Indeed, it is not uncommon for work to be prematurely terminated due to cardiovascular insufficiency. In such states, operational capability is reduced. In this Chapter, four topics will be addressed, including the impact of battle-dress uniforms, ballistic protection, undergarment moisture management, and chemical and biological protection. The principal emphasis is upon thermal and cardiovascular regulation in the person-clothing-environment system. For battle-dress uniforms, body heat storage is modelled using thermodynamics algorithms, with a three-dimensional summary presented to identify combinations of work rates and thermal exposures that yield positive heat storage. When ballistic protection is considered, one must evaluate both the impact of the added mass and the impediment it presents for dry and evaporative heat exchanges. Various moisture management practices are being marketed to address these matters. However, evidence will be presented that these do not offer measurable thermoregulatory or perceptual benefits when used beneath battle dress and ballistic protection in operational simulations. Finally, the most stressful scenario relates to protecting individuals from chemical, biological and radiological challenges. Indeed, working in such encapsulating ensembles can only be tolerated for short durations without supplementary cooling.

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1 Introduction

Contemporary military environments almost invariably require the use of personal protective clothing and equipment. These items can, however, behave as double-edged swords, since the burden associated with wearing protective equipment can sometimes challenge physiological regulation and reduce operational capability. Whilst the encumbrance of individual items is quite variable, and sometimes of limited consequence, protective clothing and equipment are typically configured into task-specific, and often integrated ensembles. Moreover, these ensembles are frequently worn in hot and thermally demanding environments, and in states where metabolic heat production is considerable. In this Chapter, the collective impact of such ensembles and these stressful states will be explored using examples that relate to the ambulatory activities of infantry personnel in warm to hot environments. To set the physiological context for this discussion, concepts of physiological regulation and compensability will first be summarised, along with the metabolic impact of protective loads. Four protective themes will then be addressed in which the impact of battle-dress uniforms, ballistic protection, undergarment moisture management, and chemical and biological protection will be isolated and examined. However, the principles described herein relate to personnel across all services and the protective ensembles they use, as there are many scenarios in which protection in the workplace can push individuals to the limits of physiological regulation.

2 Concepts of Physiological Regulation and Compensability

Humans are physiological regulators that constantly strive to maintain internal conditions that are most conductive to good health, optimal function and survival [90]. This was first recognised by Bernard [6], and more fully described as homeostasis some 50 years later by Cannon [15]. Thus, when a disturbance of the internal environment occurs, various organs and structures (physiological effectors) are activated to restore the *status quo*, and thereby defend homeostasis (accommodation). For instance, when blood pressure falls, baroreceptors instantaneously inform dedicated cells within the *medulla oblongata* of this change, and they modulate cardiac and vasomotor function to restore mean arterial pressure (Fig. 1). If these acute responses are ineffectual, longer-term outcomes are triggered, in the form of renal and sudomotor compensation. When protracted heavy work elevates mean body temperature, and when progressive dehydration elicits both a blood volume reduction and its haemoconcentration, the thermoreceptors, baroreceptors and osmoreceptors are triggered, and they provide feedback to the hypothalamus and *medulla oblongata* (Fig. 1). However, only one of six controlled organs so activated (posterior pituitary gland) was not involved in the correction of mean arterial pressure, and all seven controlled variables were again modified. Therefore, these homeostatic processes are absolutely integrated, and an underlying theme of

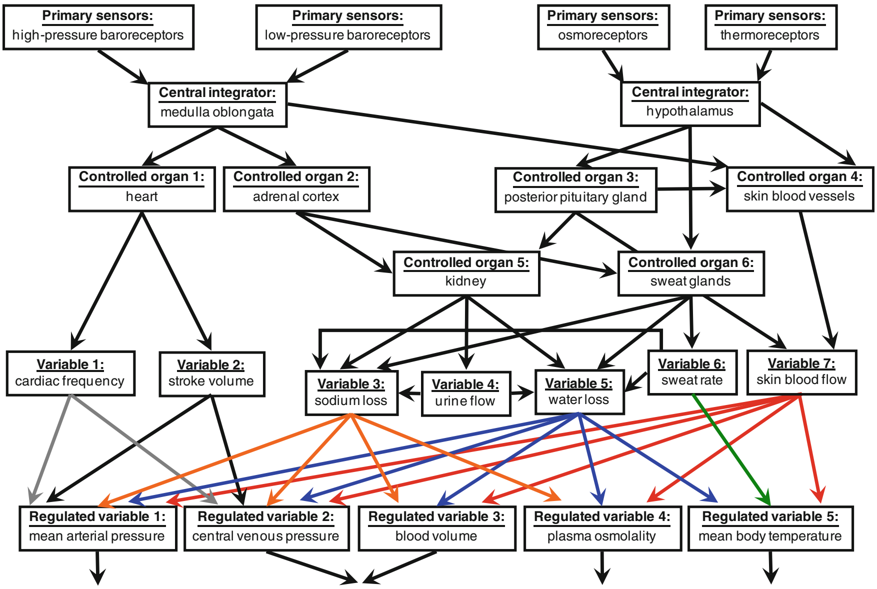

Fig. 1 Homeostatic mechanisms and responses most frequently associated with hot thermal challenges encountered when working and wearing protective clothing and equipment (from Taylor [78] and used with permission). This summary of these integrated physiological responses contains five levels of control: (1) the regulated variables (*bottom row*) that are kept within narrow ranges; (2) dedicated sensors (*top row*) that continually monitor and provide feedback concerning, the status of the regulated variables (the *arrows* link to those sensors); (3) the central nervous system structures that integrate and evaluate this feedback, and modulate (directly or indirectly) the function of the controlled or effector organs (level 4); and (5) the controlled variables through which one monitors physiological strain

this Chapter is the broad physiological impact that personal protective clothing and equipment can have upon the physiology of military personnel.

Through these regulatory mechanisms, the autonomic nervous system compensates for, and thereby accommodates, various homeostatic disturbances. However, it would be unusual for these changes to occur in isolation. Indeed, for military personnel, one would expect all five regulated variables to eventually be perturbed. Accordingly, it is inappropriate and overly simplistic to consider homeostatic defence in these circumstances through just one of these regulatory paths, as such a singular view could lead one to ignore other essential physiological interactions. Therefore, physiological compensation relies upon a well-orchestrated and tightly integrated sequence of controlled responses (Fig. 1). Since two variables dominate when wearing protective clothing and equipment, our attention is directed to those mechanisms; mean body temperature and arterial pressure regulation.

From a thermal perspective, one's ability to accommodate work-related stress can be approximated using the Heat Strain Index [4, 30]; a rational heat-strain scale based on the *First Law of Thermodynamics*. This index provides an evaluation of heat production and heat exchanges, with thermal compensability determined by comparing the extent that the required heat loss is likely to occur. Two computations are performed. The first estimates the required evaporative heat loss (E_{req}: physiological strain; Eq. 1) necessary to prevent an excessive rise in body temperature. The second is an estimate of the maximal evaporative cooling that each environmental and clothing configuration will support (E_{max}: environmental stress; Eq. 2). It is from the dimensionless ratio of these computations that one can determine thermal compensability, with quotients greater than 100 defining uncompensable states. Examples of such states, induced by the wearing of encapsulated clothing ensembles, are provided by Goldman [32] and McLellan et al. [59].

$$E_{req} = H - E_{vent} + 6.45 * A_D * (T_{sk} - T_a) / I_{TOT}[W] \tag{1}$$

where:

E_{req}	required evaporative cooling [W]
H	metabolic energy transformation: sum of resting and exercising metabolic and external work rates [W]
E_{vent}	evaporative cooling accompanying ventilation [W]
6.45	constant [dimensionless]
A_D	body surface area (DuBois equation) [m^2]
T_{sk}	mean skin temperature [^oC]
T_a	air temperature [^oC]
I_{TOT}	total insulation (trapped air, clothing insulation) [m^2 K W^{-1}]

$$E_{max} = 6.45 * A_D * i_m/I_{TOT} * 2.2 * (P_{sk} - (RH_a * P_a))[W] \tag{2}$$

where:

E_{max}	maximal attainable evaporative cooling [W]
6.45 and 2.2	constants [dimensionless]
i_m	clothing moisture permeability index [dimensionless]
P_{sk}	water vapour pressure at the skin surface [kPa]
RH_a	relative humidity of the air [%]
P_a	ambient water vapour pressure [kPa]

The regulation of mean arterial pressure is also complex, even though blood might be considered as if it were simply trapped within a closed circulatory system, since the systemic capacity exceeds the total blood volume, and both the blood volume and its viscosity are variable. These changes after encountered by clothed individuals engaged in hard work. Thus, whilst water pressure within closed and rigid pipes is a function of the pump output (cardiac output), flow resistance within the pipes (total peripheral resistance) and the downstream pressure (central venous pressure), the mean arterial pressure of living organisms is further complicated by autonomically driven changes in vaso- and venomotor tone (blood flow redistribution), and fluid (plasma) losses through urine and sweat flows that affect central venous pressure and blood viscosity. As is evident from Fig. 1, four of the five regulated variables are therefore influenced, in addition to mean body temperature, by the protracted activation of control mechanisms involved in accommodating thermal loading. For complete discussions of these interactions, readers are directed to recent thematic reviews on blood pressure regulation [39, 44, 46].

The sections that follow are based of human experimentation (person-clothing-environment system), although there are several very different stages in the evaluation of clothing ensembles [52]. The initial stage takes the form of a physical analysis of each fabric used within the garments that make up the total ensemble (clothing system). Critical characteristics include, but are not limited to, the thermal insulation, moisture permeability and the wicking capability of the fabric. This information can be obtained via several avenues, although insulation and permeability should be determined using guarded, dry and sweating hot-plates (respectively). These data can be used in the first-principles equations provided above to predict the likely physiological strain. Indeed, such estimations can, and should, lead to the elimination of some fabrics during garment development. The second testing stage involves evaluating assembled garments and clothing systems using heated, sweating and moving manikins. Output from such tests generally enable one to predict the physiological consequences that will obtain under controlled laboratory trials (stage three evaluations), and these are the focus of this Chapter. Occasionally, this stage is followed field trials, although such experiments are often hard to administer and interpret.

3 Principles of Load Distribution

Before discussing the possible adverse impact of protective ensembles on physiological regulation, it is instructive to consider how the different equipment components might influence whole-body metabolism, and therefore heat production, vasodilatation of the cutaneous and skeletal muscle beds, and mean arterial pressure regulation. This analysis is important, since the impact of equipment is location dependent [22, 34, 50, 62, 70, 80, 95], with even the protective clothing adding to this burden, particularly when it contains multiple layers [23, 64, 80, 83].

Results from the most recent of those studies are summarised in Fig. 2 [80]. When walking on a horizontal, smooth surface at 4.8 km h^{-1} with minimal clothing (control), the whole-body oxygen consumption was 0.89 L min^{-1} (standard error of the mean [±] 0.04, $N = 20$), and this increased to 1.31 L min^{-1} (±0.04) when the full complement of protective clothing and equipment worn by firefighters was donned ($\sim$20 kg). This ensemble is worn by some military personnel, and the clothing component falls within the spectrum of military ensembles, while the loads carried are not too dissimilar; helmet (1.40 kg), self-contained breathing apparatus (11.30 kg), clothing (station wear and thermal protection: 4.72 kg) and boots

(2.44 kg). The walking speed matched that used by Soule and Goldman [76], and was chosen to represent typical military route marching speeds.

Seated rest, standing and walking, all without loading, accounted for 69 % of the total oxygen cost of loaded walking whilst wearing the complete protective ensemble (Fig. 2), and 74 % of the heart rate response. It therefore follows that approximately 30 % of the physiological burden of this activity could be assigned to items of personal protection, and is subject to manipulation. In that investigation [80], both men and women were studied, with both groups chosen to provide a wide range of body sizes (body mass: 53.7–91.7 kg; height: 1.55–1.88 m). Thus, those general outcomes translate to a broad range of individuals. Nevertheless, for any one individual, the metabolic impact of a fixed load-carriage task is greater for smaller people, since its relative impact is a function of its relationship to the body mass of that person [79].

The fractional contribution of each protective component was then derived, revealing that the torso load (breathing apparatus), which was more than four times the mass of the boots and over twice the clothing mass, imposed a significantly smaller absolute metabolic burden than the boots (Fig. 2; [80]). This outcome confirmed earlier observations [76]. Indeed, when the additional, mass-specific oxygen cost of each item was derived and compared with the unloaded and

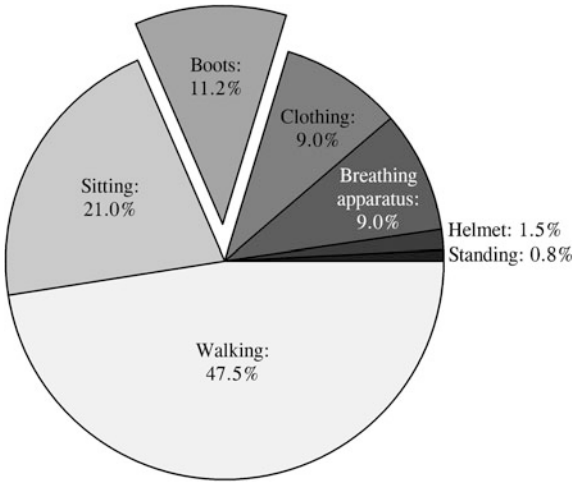

Fig. 2 The relative and location-dependent contributions to the overall, steady-state oxygen consumption (percentage) of ambulatory load carriage (walking at 4.8 km h^{-1}, 0 % gradient; $N = 20$) when wearing the complete ensemble of firefighter protective clothing and equipment. The separate contributions were determined over eight trials, first moving from unloaded sitting, to standing and then walking. Trials were then performed in which participants wore individual components of the personal protective clothing and equipment worn by firefighters: helmet (1.40 kg); self-contained breathing apparatus (11.30 kg); clothing (station wear and thermal protection: 4.72 kg); boots (2.44 kg). Relative to walking in the complete protective ensemble, unloaded sitting, for example, accounted for 21 % of the total oxygen cost. Data summarised from Taylor et al. [80]

Table 1 The mean, mass-specific oxygen cost of protective clothing and equipment (firefighters) during steady-state walking (4.8 km h^{-1}, 0 % gradient; N = 20)

Component	Oxygen cost (mL kg^{-1} min^{-1})	Relative impact (unloaded)	Relative impact (torso)
Control: minimal clothing	12.94		
Boots: 2.44 kg	88.75	6.9	8.7
Clothing: 4.72 kg	32.63	2.5	3.2
Helmet: 1.40 kg	13.67	1.1	1.3
Breathing apparatus: 11.30 kg	10.21	0.8	1.0
All components: 19.86 kg	22.02	1.7	2.2

The relative impact of each component was derived as the ratio of its oxygen cost to that of both the control (unloaded) state and the heaviest individual load, which was worn on the torso (breathing apparatus). Modified from Taylor et al. [80] with permission

minimally clothed state (control; Table 1), agreement between these contemporary [80] and past data [76] was observed during walking at an identical speed; foot loading: 88.75 (recent study) versus 73.60 mL kg^{-1} min^{-1} and head loading: 13.67 versus 13.40 mL kg^{-1} min^{-1}. Even the whole-body load carriage data were very similar (12.94 versus 12.80 mL kg^{-1} min^{-1}), and these verifications, despite considerable difference in the loads carried, reinforce these site-dependent phenomena.

Furthermore, when the impact of each item was compared to that of the heaviest load carried on the torso (breathing apparatus), it was found that head loads had a metabolic impact 1.3 times that of an equivalent torso loading (Table 1). For the clothing, this was 3.2-fold larger, and for the feet, it was 8.7 times greater. Therefore, it is foot loading that exerts the most dramatic influence, as observed by others across various experimental conditions [51, 62, 76], and changing the mass of each boot by just 100 g is metabolically equivalent to altering the torso mass by 1.74 kg. Such mass considerations are extremely efficient (or inefficient if ignored) ways to manipulate physiological strain and presumably operational capability.

These outcomes are not necessarily intuitive, although they are easily explained. When loads are added to the torso and evenly arranged around the midline, they behave in the same manner as an elevated central-body mass [7, 20, 34, 62, 76]. This relationship exists for loads up to the maximal load one can carry.

When the feet are loaded, there is not a smooth translation of the additional foot mass in the direction of movement, as that mass must move in an arc about the hip joint with each step. This occurs independently of the torso. Indeed, this foot mass moves ahead of the centre of gravity during the swing phase, remains stationary during the support phase and ends the stride cycle behind the centre of gravity. This discontinuous movement is associated with inertial work at the initiation and termination of each swing phase [16] as well as muscular work to lift the mass above the floor and move it forwards with each stride. Thus, the further limb loads are positioned away from the centre of rotation (hip joint), the greater is the arc of rotation and the work performed during each movement phase. This holds true for both the upper and lower limbs.

The layered clothing (station-wear and thermal protective garment) also exerted a significant influence, having a three-fold greater impact than an equivalent torso load. In this case, all clothed segments gained mass. However, each limb was loaded, albeit minimally, and this increased their relative metabolic contribution [62]. Since full limb covering will potentially modify joint stiffness, then this too elevates its metabolic burden, as does friction within and between clothing layers [23, 64, 83, 91]. The consideration here is to reduce clothing layers and impediments to joint movement.

4 Battle-Dress Uniforms

It is clear from Eqs. 1 and 2 that heat loss to the environment is a function of two gradients that exist between the skin and the adjacent layers of air; one of a thermal nature and the other related to water vapour pressure. These air layers are known as boundary layers (Bazett and McGlone [3]), and their thickness is largely determined by the existence of convective currents, which are of a forced (e.g. wind), natural (e.g. the rising of warmer, less dense air) or mixed form. Thus, for unclothed individuals in cold-dry climates, favourable conditions exist for dry heat loss (conduction, convection, radiation), and also for the phase change of water (evaporation) and the corresponding reduction in skin temperature. These conditions should remain thermally compensable across a wide range of exercise intensities (metabolic heat productions), with heat transported from the deep-body (core) tissues within the circulatory system (convective delivery) and dissipated at the skin surface. However, it is also clear that three changes can independently modify this outcome, all of which are primary *foci* of this Chapter: elevated air temperature, increased ambient water vapour pressure and the use of clothing [33]. These variables can transpose an exercising person from a state of thermal compensability into one that is uncompensable, and with that change and the associated elevation in body temperature, the cardiovascular system must now serve the elevated needs of at least two demanding regulatory systems (Fig. 1; [46, 72, 92]).

When more completely clothed, as is the case for most military clothing configurations, the boundary layer of air becomes trapped and it increases in thickness, forming an insulating microclimate above the skin. Two changes in the characteristics of that air now occur. Firstly, its temperature increases somewhat independently of the external environment, and the resulting warmer microclimate impedes dry heat loss, whilst it simultaneously reduces environmental influences on the skin [5, 36]. This also occurs when pilots are siting in a cockpit exposed to solar radiation [29, 66]. Consequently, the covered skin becomes warmer and more uniform in temperature [59, 69] as the skin blood vessels become engorged [27]. Secondly, since transepidermal water loss (insensible perspiration) occurs constantly, with water vapour diffusing down the vapour pressure gradient within the *stratum corneum* [74] at a whole-body rate of 26–43 g h^{-1} [81], then the trapped boundary layer also becomes progressively more humid. This obstructs, and eventually prevents, evaporative heat loss. As a consequence of these changes, both the dry and evaporative heat losses become dependent upon the thermal and moisture permeability characteristics of the clothing [35], which, in the military context, are less than ideal [33, 36]. Thus, these two forms of heat loss must now rely upon clothing ventilation [19, 41, 55, 89].

The ability of any fabric to allow (or resist) the passage of thermal energy (dry heat exchange) is known as its thermal insulation, which is primarily dictated by the volume of trapped air [45, 68]. However, since even the uncovered skin has an adjacent insulating boundary layer, then it too must be considered when deriving total insulation (Eq. 1). For any garment, insulation is determined by both its thickness and the number of layers [53], as these attributes dictate the volume of air that is trapped. From these considerations, it becomes evident that several variables contribute to the final insulation of any garment, and these include goodness of fit, posture and both the body and external air movements [41, 73]. Nevertheless, fabric insulation is typically determined using hot-plate devices [52], whilst clothing insulation can be obtained either through thermal manikin testing [48, 67] or approximated using first-principles calculations [53]. For typical battle-dress uniforms (cotton and polyester), the last of these methods would yield a clothing insulation of 0.29 m^2 K W^{-1} [86], while manikin testing might put the value at 0.21 m^2 K W^{-1} [35]. These data are combined with the skin surface layer insulation, which will vary from 0.24 m^2 K W^{-1} (stationary) to 0.07 m^2 K W^{-1} (walking at 5 km h^{-1}), and entered into Eqs. 1 and 2.

In addition to this thermal impediment, limited sweat evaporation occurs from the covered skin surfaces, so unevaporated sweat must instead be transmitted through the clothing to a more favourable site for evaporation, or it will fail to cool either the skin or the clothing [94]. This transmission is a function of the moisture permeability of the fabric [93] and how it is used to construct garments [35]. The permeability index is a non-dimensional value (range: 0–1) that describes the evaporative efficiency of fabrics, and in the case of chemical and biological protective garments, it will approximate zero, depending upon the characteristics of the protective layers. Full (ideal) permeability should approach unity, but this value obtains only for uncovered skin under conditions of high wind, as even a still boundary layer of air impedes evaporation [35]. For battle-dress uniforms, moisture permeability is about 0.41 [35].

When clothed, evaporation does not occur at the skin surfaces to a significant extent. Instead, sweat droplets are sorbed by the fabric and desorbed onto other clothing surfaces [26, 54]. This occurs in hydrophilic fibres such as cotton and wool. Sweat may also diffuse through the clothing without sorption [42]. This occurs in fabrics made from synthetic and hydrophobic fibres (nylon, polyethylene, elastane). In both instances, evaporation occurs on the surface of the fabric, and this provides less effective cooling than does equivalent evaporation at the skin surface. Moreover, the cooling efficiency of evaporation from fabric surfaces changes inversely with the distance between the skin surface and the site of evaporation

[40]. Finally, each clothing layer inhibits this moisture transfer [8, 55, 89]. Thus, while some fabrics may have superior wicking characteristics, their utility within a multi-layered ensemble is questionable, and this point will revisited in a subsequent section.

From this discussion, it is apparent that both the thermal insulation and moisture permeability of fabrics and garments are individually very important. However, it is their ratio (Eq. 2) that dictates the performance of a garment within any thermal environment: moisture permeability index/clothing insulation (Eq. 2; [32]). Within the context of all clothing systems, but particularly the complex systems used by military personnel, both of these attributes need to be considered for individual fabrics (dry and sweating hot-plate testing), the separate garments in which they are used (manikin testing) and the multi-layered components that constitute the entire ensemble (manikin testing).

Before leaving this topic, it is useful to gauge how the battle-dress uniform may influence body heat storage. This can be modelled using thermodynamics algorithms [12, 31, 77], and the corresponding heat transfer for a 75–80-kg individual working at different intensities and exposed to a wide range of ambient temperatures is illustrated in Fig. 3. Combinations of work rate and thermal exposures producing heat transfers above the zero heat-transfer plane will result in positive heat storage. Thus, only the foreground conditions (external work rates <110 W and

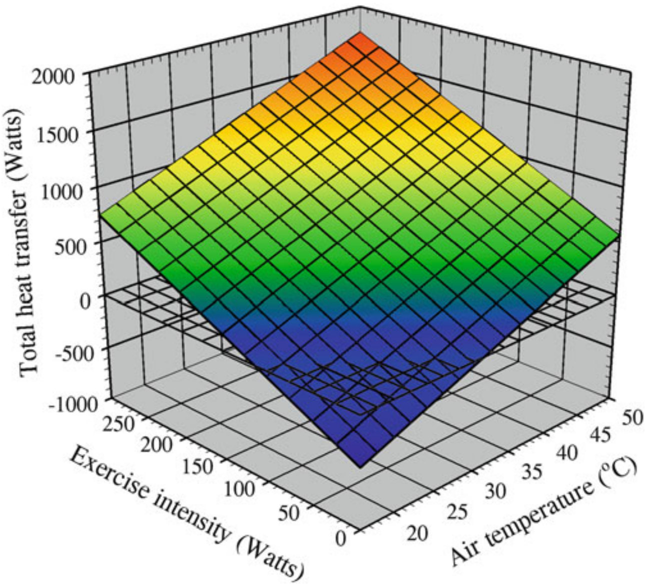

Fig. 3 A three-dimensional surface for predicting heat transfer for a 75–80-kg male wearing battle dress (insulation 0.29 m^2 KW^{-1}, moisture permeability 0.4), working across a wide range of external work rates (rest to 300 W) and exposed to air temperatures between 15–50 °C (60 % relative humidity). Combinations above the zero heat-transfer plane are associated with heat storage. Modified from Caldwell et al. [10] and used with permission

air temperatures <33 °C) yield zero or negative heat storage. The upper work rate within this region is somewhat light, albeit it similar to that which may be expected during an urban patrol. However, the apparent sensitivity reflects more on the fact that these values are estimates than it does on the precision of the model, and so it should not be construed to reflect actual physiological observations; by definition, models only approximate reality. Nevertheless, they do provide a means through which to estimate those responses. It must also be noted that heat storage is not necessarily a problem, as slow storage rates are well tolerated. On the other hand, rapid heat storage is problematic, and when exposed to hotter climates, clothed personnel are close to the cusp of positive heat storage, particularly during heightened operational states or when wearing body armour. In the following sections, the results of human trials are presented for people working under these conditions during simulated urban patrols. These circumstances are frequently uncompensable, and are illustrated by points on the heat-transfer surface that move towards the red end of the colour spectrum (Fig. 3).

5 Ballistic Protection

Many contemporary operational scenarios require the use of ballistic protection (torso armour and helmet), and this is particularly common within urban patrols. In addition to the added mass carried (*e.g.* torso: 3–11 kg; head: 1.0–1.5 kg), this equipment further impedes heat dissipation and moisture permeability. Accordingly, it alters the balance between the required and maximal evaporation (Eqs. 1 and 2), and thereby displaces personnel into less compensable conditions (Fig. 3). Indeed, when individuals are sufficiently stressed, the added burden of ballistic protection can push them beyond the limits of physiological regulation.

During urban patrols, the work intensity is generally slow and deliberate, unless an engagement occurs. Therefore, whilst various combinations of external thermal and equipment loading frequently result in uncompensable operational conditions (positive heat storage), it is unlikely that physiological strain will be excessive. This possibility was evaluated by Caldwell et al. [9, 10], with participants performing two levels of low-intensity exercise (walking: 2 km h^{-1} [90 min] and 4 km h^{-1} [60 min]) in hot-humid conditions (36 °C, 60 % relative humidity), with radiant heating (750 W m^{-2}) and a large-diameter fan used to match walking speeds. Trials (2.5 h) were completed in each of three levels of armoured protection: control (battle dress [insulation 0.29 m^2 K W^{-1}] with cloth hat (total mass 2.05 kg); torso armour (6.07 kg) with battle dress and cloth hat, and full armour (battle dress, torso armour, helmet [1.29 kg]). As predicted from Fig. 3, physiological strain increased inexorably as the simulations continued (Fig. 4), attaining progressively higher terminal levels with increments in armoured protection. Indeed, deep-body temperatures deviated significantly among trials, rising at 0.37 °C°h^{-1} (control), 0.41 °C h^{-1} (torso armour) and 0.51 °C h^{-1} (full armour); others have also described this impact of body armour [21, 56]. Notwithstanding, these changes were unlikely to lead to either impaired

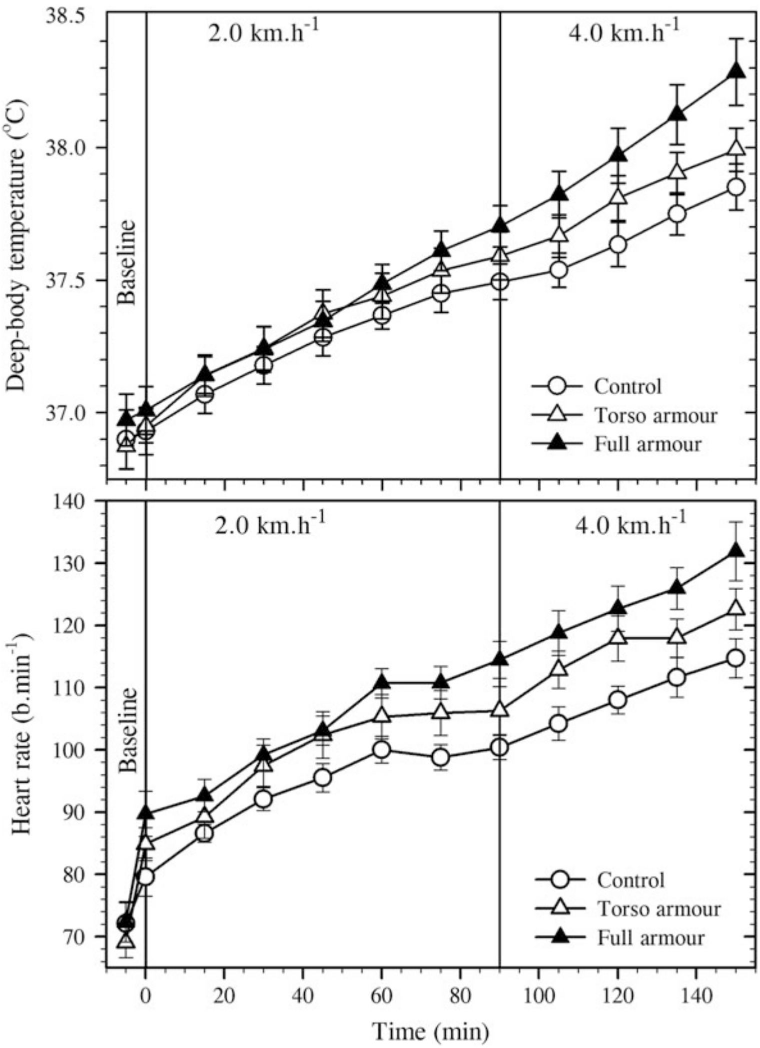

Fig. 4 Deep-body (auditory canal) temperature and heart rate responses during steady-state walking at two speeds in a hot-humid environment with participants ($N = 9$) wearing each of three battle dress and combat armour configurations. From Caldwell et al. [10] and used with permission

thermal or blood pressure homeostasis (exertional heat illness; [38, 63] during uneventful urban patrols in hot-humid conditions.

Whilst the physiological strain accompanying the use of body armour in those trials was reasonably well tolerated [9, 10], it remained uncertain whether or not soldiers could endure the higher workloads encountered during enemy engagements or the strain associated with hot-dry conditions, such as the Middle East (summer). Moreover, the interaction of different levels (tiers) of armoured protection with changes in these independent variables was also unknown. Accordingly, van den Heuvel et al. [87] investigated five tiers of armoured protection. The objective was to obtain information upon which defence personnel could make informed decisions when balancing ballistic protection against the threat of reduced capability and soldier collapse. Five, 2-h trials were completed, each with fully-hydrated subject wearing battle dress (insulation 0.29 m^2 K W^{-1}) and combat helmet: control (vest and webbing but no armour) and armoured tiers one to four (Fig. 5). The body armour varied in its mass (3.4–11.0 kg) and the surface area it covered (0.24–0.52 m^2). The experimental conditions were hot and dry (45 °C, 20 % relative humidity), with additional heating provided using infrared lamps (750 W m^{-2}) and a

Fig. 5 Ballistic torso protection ranging from no torso armour through to maximal protection. Only tiers zero and three are illustrated. Tier 0 (control): vest and webbing only; Tier one: lightest armour mass (3.4 kg) with smallest surface area coverage (0.24 m^2); Tier two: intermediate armour mass (6.8 kg) and surface area coverage (0.30 m^2); Tier three: intermediate armour mass (7.8 kg) and surface area coverage (0.44 m^2); and Tier four: heaviest armour mass (11.0 kg) with the greatest area coverage (0.52 m^2). Pouches were filled with solid, light-weight foam to ensure they behaved in the manner expected in the field, but without mass affecting the thermal properties. Participant carried a simulated weapon to limit arm motion and thereby produce realistic uniform ventilation. Modified from van den Heuvel et al. [87]

constant wind velocity (4 km h^{-1}) was applied throughout. Participants first walked at 4 km h^{-1} (treadmill: 1 % gradient) for 90 min with 2-min rests at 28, 58 and 88 min. This was slightly faster than a tactical patrol, and this speed was adopted to replicate the anticipated total energy expenditure of patrolling since assault loads were not carried. The walking speed (6 km h^{-1}) and gradient (4 %) were then increased, and subjects attempted 30 min of further exercise; this simulated the energy expenditure of a section attack, but again without using the assault loads.

The most direct indicator of the overall impact of these body-armour variations was the work tolerance time (Fig. 6). Whilst one would normally expect all subjects to complete the least loaded condition, this was not realised since even that state placed a significant physiological burden on participants (Fig. 3); two individuals failed to complete Tier 0. Of course, this probably reflects day-to-day tolerance variations when operating close to the limits of regulation. Everybody completed the Tier 1 trial, with work tolerance times then changing reciprocally with increments in armoured protection (Fig. 6). When expressed in the form of successful trial completions, or the ability to undertaken the full 120 min of work, Tiers 0, 1 and 2 had 75, 100 and 75 % trial completion rates (respectively), whilst Tiers 3 and 4 had only 12.5 and 25 % successful completions (respectively). Consequently, tolerance times between Tiers 1 and 4 differed significantly ($P < 0.05$), with most physiological responses then following this dichotomous pattern [87].

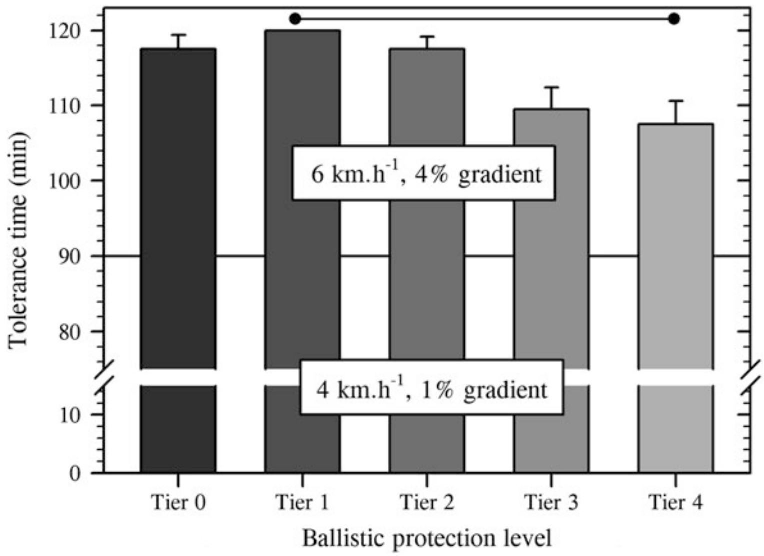

Fig. 6 Work tolerance times during combined heat (45 °C, 20 % relative humidity) and exercise stress (walking at two speeds) whilst wearing battle dress, combat helmet and each of five levels of armoured protection (Fig. 5). Subjects all completed 90 min (4 km h^{-1}) before attempting 30 min of faster walking (6 km h^{-1}) in a simulated section attack intensity with an estimated total metabolic rate of approximately 700 W. Data are means with standard errors of the means (extracted from van den Heuvel et al. [88])

For example, beyond 40 min of exercise, deep-body temperatures started to deviate, yielding significant time-by-tier interactions: Tier 0 was less stressful than Tier 4 ($P < 0.05$), Tiers 1 and 2 were both lower than Tier 3 ($P < 0.05$), and Tiers 1 and 2 were also both less stressful than Tier 4 ($P < 0.05$). Since thermal differences did not exist among Tiers 0–2, or between Tiers 3 and 4 ($P > 0.05$), then these configurations could be pooled into two isothermal categories: Tiers 0–2, and Tiers 3 and 4 [87]. Very similar between-category differences were evident for mean skin temperatures, although there now existed a non-physiological interference that one sees within every clothing ensemble; the trapping of thermal energy. In the thermoneutral baseline stage of each trial, before donning any clothing, the mean skin temperature averaged 35.0 °C, but with clothing in position, it increased to 35.7 °C across all tiers. Within the first 10 min of exercise, mean skin temperature rose to 36.8 °C and then to 37.0 °C at 20 min, after which it remained relatively stable until the work rate was further elevated (90 min; [88]. Therefore, the combined battle dress and ballistic ensembles were impeding dry heat loss from the torso by keeping thermal energy trapped within the boundary and clothing layers, and thereby minimising the thermal gradient between the skin and its microclimate.

The effect of this trapping was apparent within the heart rate responses, with between-tier separation commencing immediately (0 min), followed by a pronounced divergence, and resulting in heart rates at 90 min of: 116 (Tier 0), 113 (Tier 1), 117 (Tier 2), 127 (Tier 3) and 138 beats.min^{-1} (Tier 4). However, whilst those differences were not significant ($P > 0.05$), comparisons between Tiers 1 and 4, and between Tiers 2 and 4 were significant for the time-by-tier interactions ($P < 0.05$; [88]. Thus, as thermal strain increased, the cardiovascular system was called upon to more heavily support heat dissipation. In clothed situations, skin blood flow rises to maximal levels much earlier than during an equivalent thermal loading encountered in unclothed states [27]. In addition, due to a gradual loss of sweat and the accompanying blood (plasma) volume decrease (Fig. 1), heart rates will also reflect a cardiovascular drift [18]. Indeed, military clothing, and in particular when worn with body armour, will exacerbate that state [9, 10]. Therefore, it is not uncommon for work and exercise to be terminated due to the exhaustion associated with cardiovascular insufficiency [1, 2, 63, 71], rather than to frank hyperthermia, and brought about by changes that precipitate hypotension and syncope [39, 47]. Of the physiological control mechanisms activated during exposure to these combined stresses, it is a failure to adequately regulate blood volume that may now be the most significant (Fig. 1).

6 Garments Worn Under Protective Clothing

In the above examples, the impact of military clothing and protective material on heat trapping was observed. Equally important is moisture, primarily sweat, and water vapour accumulation within the boundary and trapped air layers, as this will impede evaporative cooling. For military personnel engaged in fire suppression (e.g. shipboard firefighters), it will also interact with radiant heat flux. For example, when the external radiant heat flux is high, moisture within these air layers facilitates heat penetration, with vapour tending to condense on the skin surface, whereas in situations of low heat flux, internal moisture retards heat penetration and vaporisation occurs, now with water vapour able to move into the clothing and its air spaces [49, 61]. For these reasons, various active and passive moisture management practices have emerged that may help restrict heat storage to physiologically compensable levels. However, for most military personnel, the only viable option is a reliance upon the capacity of some fabrics to remove sweat from the skin surface so that evaporation may occur elsewhere.

When working, an elevation in mean body temperature stimulates sweat secretion [90], the production of which is widely variable across the body surface [81]. In unclothed states, sweat will evaporate at the skin surface, cooling the skin and dissipating heat delivered from the deep-body tissues via the circulatory system. However, as already described, clothing interferes with this process. Nevertheless, in some working environments, clothing is not an optional extra; it is mandatory. In such circumstances, the wearing of impermeable clothing is illogical, unless protection is sought from chemical or biological agents, and so it follows that it is better to wear a fabric that facilitates water movement away from the skin [57, 94], as these fabrics represent the lesser of two evils, and they are generally more comfortable to wear [37]. Moreover, wicking moisture through a fabric can be beneficial when the sweat rate is not high and the skin is partially wetted, since this increases the evaporative surface area [40].

These truisms hold for single-layer ensembles, but do they remain valid for multi-layer ensembles? The first principles of thermodynamics and clothing science dictate that, with each added clothing layer, the characteristics of the most favourable fabric within the ensemble move towards those of the least favourable fabric, since successive layers inhibit both heat and moisture transfer [8, 55, 89].

The validity of this prediction was tested by van den Heuvel et al. [86], and the merits of varying the composition of torso undergarment fabrics, and the impact of this on thermal strain, moisture transfer, and the thermal and clothing comfort of fully clothed, armoured individuals was evaluated. Eight fully-hydrated males participated in five exercising heat exposures wearing a different torso undergarment on each occasion. Trials were performed in hot-dry conditions (41.2 °C, 29.8 % relative humidity, 4 km h^{-1} wind speed [large fan]). While the ultimate objective was to evaluate the suitability of such garments for the Middle East, these conditions optimised the skin-air water vapour pressure gradient, and were thereby most conducive to the use of moisture management garments. Participants first walked for 120 min (4 km h^{-1}), then alternated running (2 min at 10 km h^{-1}) and walking (2 min at 4 km h^{-1}) for a further 20 min. Whilst this may be regarded as a difficult challenge, it must be remembered that some garments are purported to alleviate heat strain through their superior wicking capabilities and their suitability for use under armour; this experiment included garments marketed under this claim.

Subjects wore battle-dress uniforms (clothing insulation 0.29 m^2 K W^{-1}), ballistic protection armour (5.91 kg) and a combat helmet (1.29 kg). In one trial (control), no torso undergarment was worn (Ensemble A). Three different undergarments were tested: two made from high sorbing and good wicking fabrics (Ensemble B: 100 % cotton; Ensemble C: 100 % merino wool), and the third had low sorbing but good wicking characteristics (Ensemble D: 66 % nylon, 19 % polyethylene and 15 % elastane). Finally, a hybrid combat shirt was evaluated without an undergarment (Ensemble E). Its torso segment was made from 100 % merino wool, with the collar and sleeves made from cotton (75 %) and polyester (25 %). Thus, whilst Ensembles B–D added an undergarment layer, Ensemble A permitted an assessment of whether or not such undergarments had any thermal utility, and Ensemble E enabled examination of the possibility that such undergarments might be useful if worn without adding a clothing layer.

The results of this experiment were striking. Under the conditions tested, neither the undergarment textile composition nor the presence (or absence) of an undergarment offered any significant thermal, central cardiac or comfort advantages. Data for deep-body and skin temperatures are presented in Fig. 7; three skin sites beneath the torso undergarments (chest, scapula, upper arm) were averaged to derive the latter measurement. At no time were any of the inter-ensemble differences significant. Indeed, the most striking feature of these data was their reproducibility across trials, as this duplication precision more closely resembled repeated trials on the same ensemble than it did trials performed using different ensembles. Furthermore, this reproducibility was evident for heart rate, sweat production, evaporation, and both thermal and clothing discomfort.

Clothing water vapour pressures were derived for the skin and trapped air spaces above the upper chest and back (Ensembles B–D). The vapour pressure gradients at those sites facilitated vapour transfer. Between the undergarment and the second air space, these gradients were negligible at the chest, but negative at the back, and a negative gradient inhibits vapour transmission. However, neither the inter-ensemble water vapour pressures nor the vapour pressure gradients were significant ($P > 0.05$). Therefore, no evidence existed to indicate that any of the fabrics tested created a significantly drier microclimate between the skin and the first clothing layer. Moreover, the evaporation of sweat from the clothing (evaporative heat loss) was not improved by any of the experimental fabrics or undergarment configurations beyond that observed without an undergarment.

It was therefore concluded that none of the experimental undergarments offered measurable thermoregulatory or perceptual benefits when used beneath battle dress and ballistic protection when subjects were simulating operational work rates. In fact, every ensemble yielded responses equivalent to those observed in the control state. Fogarty et al. [28] reported similar results for high-intensity work under hot-humid conditions. It might be argued that such outcomes were a function of the combined influences of metabolic heat production (forcing function) and the personal protective clothing and equipment worn. This is correct. Nevertheless, it is not sufficient to only demonstrate the utility of thermal or moisture management fabrics or practices on the benchtop; this must also be established under operationally relevant scenarios and forcing functions. In these situations, increasing clothing

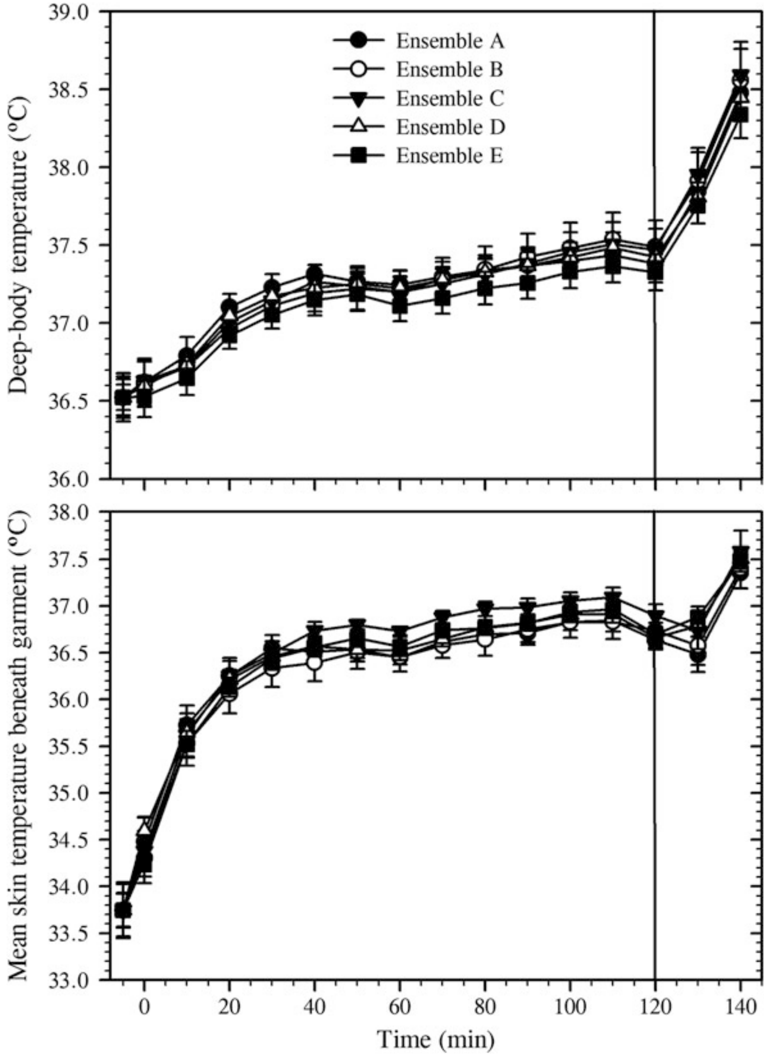

Fig. 7 Deep-body (auditory canal) temperature and mean (clothed) skin temperatures beneath the torso undergarment (chest, scapula, upper arm). These data were gathered during exercise in hot-dry conditions (41.2 °C and 29 8 % relative humidity) whilst wearing battle dress and ballistic protection armour. Five torso undergarments (Ensembles) were evaluated: A (no undergarment; control), B (100 % cotton), C (100 % woollen), D (66 % nylon, 19 % polyethylene, 15 % elastane) and E (hybrid shirt). Data are means with standard errors of the means. The vertical line at 120 min signifies the start of an elevated exercise intensity. From van den Heuvel et al. [86] and used with permission

ventilation would produce superior cooling [19, 55], as would auxiliary cooling [11, 17, 24, 65, 75], although some practices are ill-suited for situations where personnel must operate independently.

7 Chemical and Biological Protection

Our final operational scenario is potentially the most stressful, as it relates to protecting individuals from chemical, biological and radiological challenges (encapsulating clothing systems), and readers are directed to reviews of this topic for detailed discussions of encapsulating clothing [32, 59]. Herein, the focus is on hot environments of relevance to military personnel, and observations from two projects will be summarised, one involving the testing of experimental ensembles [13], and the second comparing the performance of a novel ensemble made from a low-burden and higher air-permeability fabric with that of two traditional lower air-permeability fabrics [85].

In these encapsulated situations, the person-clothing system approximates a closed thermodynamic system with an outer, impermeable membrane (diathermal wall) that is permissive to energetic, but not to material, exchanges with the surrounding environment. Thus, whilst conductive and radiative heat exchanges may occur, there is minimal or no water vapour flux across the protective membrane. This limits evaporative cooling to the point at which the water vapour pressure of the trapped air approximates that of the skin. Furthermore, when the external temperature exceeds the average temperature of the body, even dry-heat dissipation ceases. In these scenarios, the limits of physiological regulation are very rapidly approached with operational capability compromised soon after [25, 43, 58, 60], although not all encapsulating ensembles are equally stressful [25, 60].

Three novel, prototype variants of a flame-retardant chemical and biological protective textile (Defence Science and Technology Organisation, Australia) were compared with the traditional flame-retardant, chemical and biological protective textiles used by the Australian Defence Force. These fabrics were assembled into prototype ensembles that varied in their heat and moisture transmission ratings, with the aim of facilitating heat loss whilst providing appropriate chemical and biological protection. The resulting ensembles were evaluated in four experimental trials ($N = 7$) using fully-hydrated participants. This was a double-blind study with trials administered in a balanced order and with subjects acting as their own controls. While each ensemble had an approximately equivalent mass (5.7–6.0 kg), independently derived, sweating hot-plate data demonstrated the average thermal resistance of the experimental system to be 66 % of that obtained for the existing protective system (0.036 versus 0.054 $m^2\ K\ W^{-1}$: Defence Science and Technology Organisation). In addition, the total insulation of the existing system, which was worn over battle dress, was $\sim$0.396 $m^2\ K\ W^{-1}$, while that of the experimental systems averaged 0.342 $m^2\ K\ W^{-1}$, as they were designed to be worn without battle dress. Each exposure was in a hot-dry environment (40 °C, 30 % relative humidity), and the objective was to simulate metabolic rates and postures observed in military vehicles, such as aircraft or armoured vehicles. Therefore, low-intensity exercise (cycling) was performed with subjects a semi-recumbent position (external work rate 25 W [60 min] and 50 W [60 min]; [13].

The existing chemical and biological protective ensemble had a significantly more adverse physiological impact upon the wearer. That was most evident for deep-body temperature (Fig. 8), mean body temperature and heart rate (Fig. 8), each of which revealed significant time-by-ensemble interactions ($P < 0.05$) such that

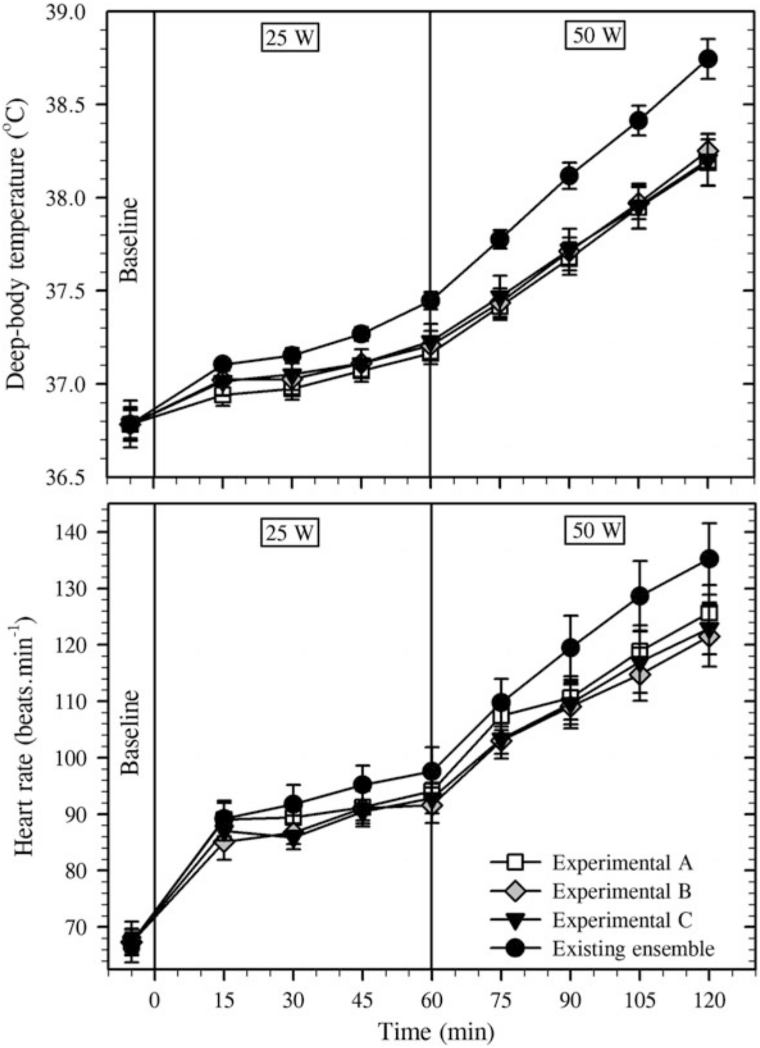

Fig. 8 Deep-body (auditory canal) temperature and heart rate responses during low-intensity exercise (cycling) in hot-dry conditions (40 °C and 30 % relative humidity) with subjects wearing each of four chemical and biological protective ensembles: experimental ensembles A–C and the existing ensemble used by the Australian Defence Force. Data are means, normalised to common, thermoneutral baselines, with standard errors of the means. The vertical line at 60 min signifies a work-intensity change. From Caldwell et al. [14]

strain when wearing the current ensemble gradually deviated from that experienced within the experimental systems. This was presumably due to its greater thermal insulation. Indeed, the experimental ensembles came close to supporting thermal and cardiovascular steady states during very-light work, although significant differences were not apparent amongst those ensembles ($P > 0.05$). Conversely, uncompensable physiological strain was experienced in all trials at the heavier workload, but less so in the experimental clothing [13].

Considerable differences were observed among the skin temperatures beneath each clothing system (Fig. 9). Whilst inter-site variability existed, when these data were averaged across eight sites, experimental ensemble C was found to be clearly and significantly superior ($P < 0.05$), and this was due to lower forehead and chest temperatures, and to a lesser extent, scapula, forearm and hand temperatures. Skin temperatures were not different at the calf or thigh ($P > 0.05$). Therefore, even though equivalent deep-body temperature and heart rate responses were obtained for this clothing system, it appeared to have superior heat loss or moisture transmission characteristics that may provide wearers with a slight thermal advantage. Nevertheless, these differences had no measurable impact on psychophysical assessments, including the perception of either thermal or clothing discomfort [14].

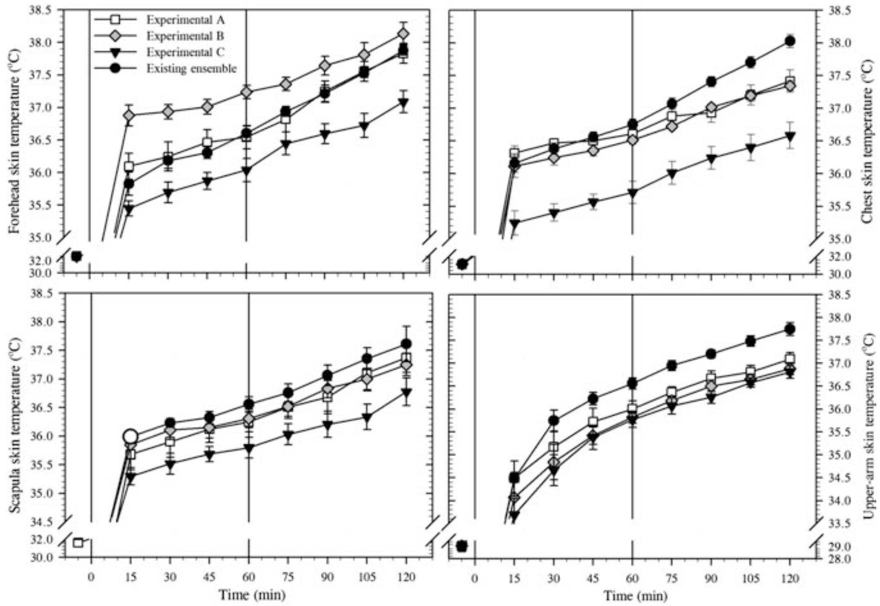

Fig. 9 Forehead, chest, scapula and upper-arm skin temperatures observed during low-intensity exercise (cycling) in hot-dry conditions (40 °C and 30 % relative humidity) with subjects wearing each of four chemical and biological protective ensembles: experimental ensembles A–C and the existing ensemble used by the Australian Defence Force. Data are means, normalised to common, thermoneutral baselines, with standard errors of the means. The vertical line at 60 min signifies a work-intensity change from 25 W to 50 W of external work. From Caldwell et al. [14]

Developments in chemical and biological clothing have enabled some air and moisture penetration into the air spaces, and such improvements can reduce the strain below that which might otherwise be experienced [25]. It is also possible these modifications could make uncompensable conditions more tolerable, and thereby elevate the operational capability of personnel. This possibility was evaluated by comparing a low-burden, chemical and biological protective system with two civilian protective systems [85].

The three protective systems (Fig. 10: Ensemble A: Australian low-burden ensemble [CBCS, Melba Industries, Australia]; Ensemble B: French ensemble [SWAT, Paul Boye, France]; and Ensemble C: British ensemble [CR1, Remploy Frontline, U.K.]) were tested in temperate conditions (29.3 °C, 56.0 % relative humidity), using fully-hydrated subjects. Ensembles B and C incorporated flame-

Fig. 10 Chemical, biological and radiological protective ensembles: *Ensemble A* (Australian low-burden ensemble; CBCS, Melba Industries, Australia), *Ensemble B* (SWAT, Paul Boye, France) and *Ensemble C* (CR1, Remploy Frontline, U.K.). Each protective system was evaluated using a standard respirator mask, gloves, over boots, torso body-armour vest and standard military webbing

retardant textiles. The trial sequence was balanced, and subjects wore the same t-shirt and shorts under ensembles A and B. However, thermal undergarments (long-sleeved shirt, long pants) were worn under ensemble C, as was specified. In addition, the same respirator mask (removed only for drinking), gloves, over boots, torso body-armour vest (without plates) and standard military webbing were worn in every trial. Participants walked at two speeds: very-light intensity (60 min, 3 km h^{-1}, 0 % gradient), seated rest (10 min; full protection retained [MOPP4]), and brisk walking on a slight gradient (30 min, 6 km h^{-1}, 3 % gradient). A large fan directed air onto each subject to match the walking speed.

All subjects completed these exposures when wearing ensemble A (Fig. 11), while the work tolerance time for ensemble C was significantly shorter than for either of the other ensembles. For this experiment, the deep-body temperature index was rectal temperature, as preferred by the Defence Science and Technology Organisation. Whilst this index is not always preferred due to its slow response time [82], it is acceptable when the thermal load is relatively small and constant. Beyond 60 min, and particularly when the work rate increased, ensemble C was associated with a significantly faster elevation in deep-body temperature ($P < 0.05$). This was reflected in cardiovascular strain, with subjects terminating exercise with average heart rates of 168 b min^{-1} (ensemble A), 173 b min^{-1} (ensemble B) and 182 b min^{-1} (ensemble C), with the time-by-ensemble interactions establishing lower strain for ensemble A than each of the other clothing systems ($P < 0.05$; [84]).

Mean skin temperatures reflect the thermal energy content of the skin tissues, as influenced by local metabolic heat production and thermal exchanges between these tissues and the surroundings [82]. In the first instance, more heat is delivered to the skin when the cutaneous vascular beds dilate (convective heat flow). In addition, heat trapped between the skin and clothing will modify the transfer of thermal energy to the surrounding boundary layer and ambient air. Therefore, these temperatures provide important information concerning both physiological strain and thermal stress, and they influence thermal discomfort. In these trials, mean skin temperatures were uniformly and significantly elevated relative to both the other ensembles ($P < 0.05$), and these data are assumed to have resulted from the combined influences of a higher skin blood flow and greater heat trapping within that ensemble ($P < 0.05$). Finally, subjects lost significantly more sweat, relative to the low-burden ensemble A (0.84 L), when wearing each of the other ensembles: ensemble B: 0.95 L; ensemble C: 1.05 L ($P < 0.05$; [85]).

From these observations, it was concluded that ensemble C placed the wearer under the greatest physiological strain. Because of its extra clothing layer (undergarment), that ensemble had a higher clothing insulation. This layer also impeded water vapour transmission, and in warm-humid conditions, this would be a distinct disadvantage. One must therefore consider whether or not this extra clothing layer was actually necessary, unless it was integral to the chemical and biological protection. Conversely, ensemble A had the least physiological impact upon these individuals. Nevertheless, working in such encapsulating ensembles can only be tolerated for short durations without supplementary cooling.

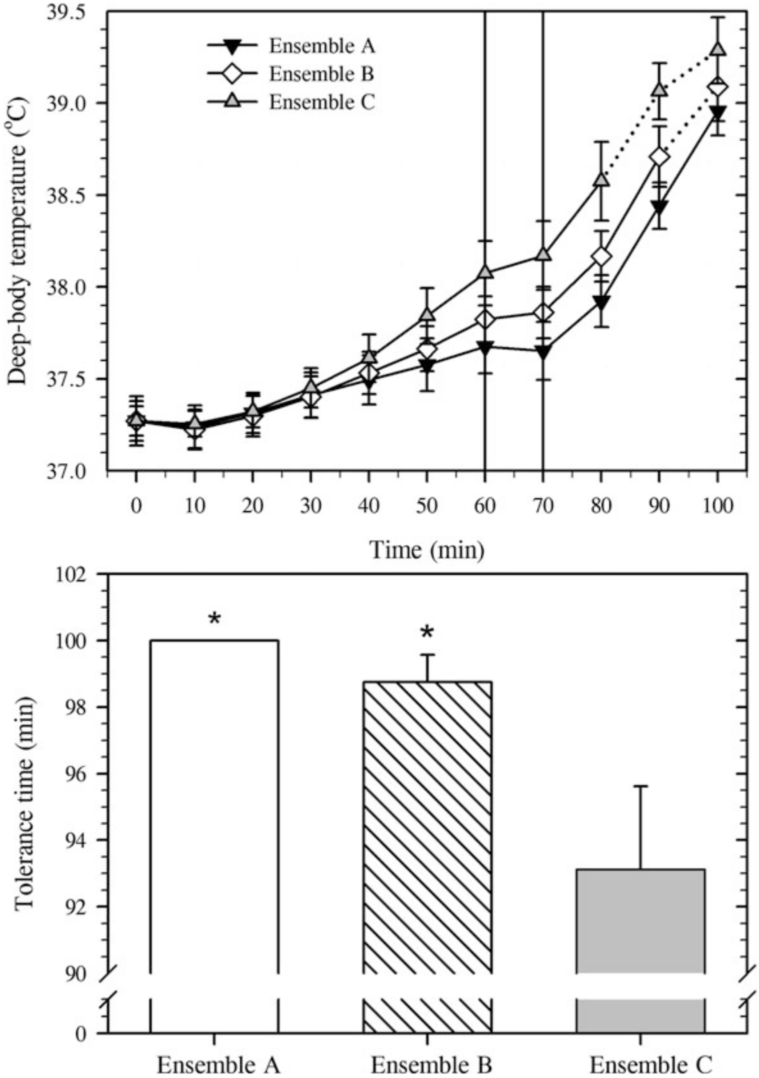

Fig. 11 Deep-body (rectal) temperatures and exercise tolerance times during two steady-state work rates (walking) in a temperate environment (29.3 °C, 56.0 % relative humidity). Subjects wore each of three chemical, biological and radiological protective ensembles identified in Fig. 10. Broken lines indicate times where individuals prematurely terminated a trial, due either to excessive thermal strain or volitional fatigue. Data are means with standard errors of the means, with vertical lines signifying work-intensity changes (3 km h^{-1} to rest to 6 km h^{-1}). From van den Heuvel et al. [85]

8 Conclusion

From this communication, it is apparent that the physiological burden of personal protective clothing and equipment can challenge the integrated regulation of critical physiological variables. Some individuals may be pushed to the limits of regulation, and it is not uncommon for work to be prematurely terminated due to cardiovascular insufficiency. In such states, operational capability is reduced. Whilst prophylactic strategies exist, passive systems that incorporate the moisture transmission characteristics of certain fabrics to promote evaporative cooling do not offer any benefit when worn under battle dress and ballistic protection. Instead, active cooling should be used, but these are power hungry and will add to the load that is already carried. Thus, such systems are really only viable for those working in and around support vehicles. For those wearing encapsulating ensembles, perhaps the only viable option is rest and the rotation of personnel.

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Conflict of Interest
There are no conflicts of interest.

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