Chapter 12
Biomechanics of the Fracture Fixation

Yingze Zhang, Hongde Wang, Tianrui Wang, Wei Chen, and Yanbin Zhu

Abstract Biomechanical factors is an important aspect that affects bone healing directly. Although some biologic etiologies of fixation failure can be directly affected by the physician, there are only minimally under the surgeon's control. The surgeon should do their best to preserve soft tissue, vessel, and the zone of injury. Skillful surgical technique, tight wound closure, and appropriate antibiotic therapy could decrease the risk of infection and reduce the risk of fixation failure. If failure occurs suddenly or prior to the expected time when fracture healing would occur, a mechanical issue is usually the primary cause. Biomechanics study the role of force and energy in biological systems. The fracture fixation should follow the principle of biomechanics. Excessive stress concentration and fatigue, leading to increased pressure load and bending load, results in internal fixation failure. Understanding the biomechanical principles underlying stable fixation and failure fixation could help the surgeon determine the appropriate investigation and intervention. Appropriate biomechanical fixation technology will promote fracture healing, accelerate rehabilitation of patients, and reduce nonunion of fracture. Biomechanical study can help the design of the internal fixation and also plays an important role on the improved clinical effects; furthermore, it will help clinicians to choose reasonable diagnosis and treatment strategies.

Keywords Biomechanics · Innovation · Orthopedics trauma · Fixation principle
Fixation devices

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1 Introduction

Biologic and mechanical problems are the two underlying problems, when fracture implants fail prior to fracture union [1–3]. Patient's systemic biology as biologic factors including chronic diseases, smoking, medications, and many other causes may cause delayed union and fixation failure [4–6]. Many biologic etiologies of fixation failure can be treated by the physician, but there are still many biologic etiologies beyond the surgeon's control [7, 8]. Therefore, surgeons should make many efforts to preserve soft tissue and vascularity, and respect the zone of injury [9]. Meticulous surgical technique, wound closure, and appropriate perioperative antibiotic therapy can all reduce the risk of infection and treatment failure. Mechanical problems are usually the main cause, when failure occurs acutely or prior to the expected time that fracture healing [10–13]. To determine the appropriate investigation and intervention, surgeons should understand biomechanical principles underlying the stable fixation and fixation failure.

2 Pin and Wire Fixation

There are some biomechanical differences among pins, rods, and nails used for fracture fixation. Pins only resist alignment changes, rods resist deviations in alignment and translation, and nails resist changes in alignment, translation, and rotation. Kirschner wires and Steinmann pins are often used for both provisional and definitive fracture fixation. Due to its poor bending resistance, it should be used in conjunction with bracing or casting. They are usually inserted as final fixation with limited open reduction or percutaneously. They should be inserted slowly and stop drilling frequently. To prevent thermal damage to bone and soft tissues. To make their removal easier after fracture healing, we recommend using smooth wires.

Threaded wires can better fix the fractures compared with temporary fixation, but the fracture fragments must be fixed together when insertion wire to avoid distraction. If the cortical bone is hard, there is a risk of pin breakage. For small fragments in metaphyseal and epiphyseal regions, especially in the fracture of distal foot, forearm, and hand, such as Colles fracture, and in displaced metacarpal and phalangeal fracture after closed reduction, pin or wire fixation is usually sufficient [14, 15]. In most cases, pins are inserted under the control of an image intensifier. This can protect the soft tissues from further damage, theoretically allowing for maximum bone regeneration; however, it must be noted that nerves and tendons are not wound around the pin during insertion. Wire fixation is used alone or in combination with other implants for definitive fixation of some metaphyseal fractures, such as in the cervical spine, proximal humerus, and patella [16]. The wire should not be notched as it may reduce the fatigue life of the implant. It is rare touse wire alone to provide adequate stability for functional rehabilitation of the extremity [17].

For some intra-articular fractures, simple rigid fixation using screws or plates for a long time might result in some complications such as articular degeneration,

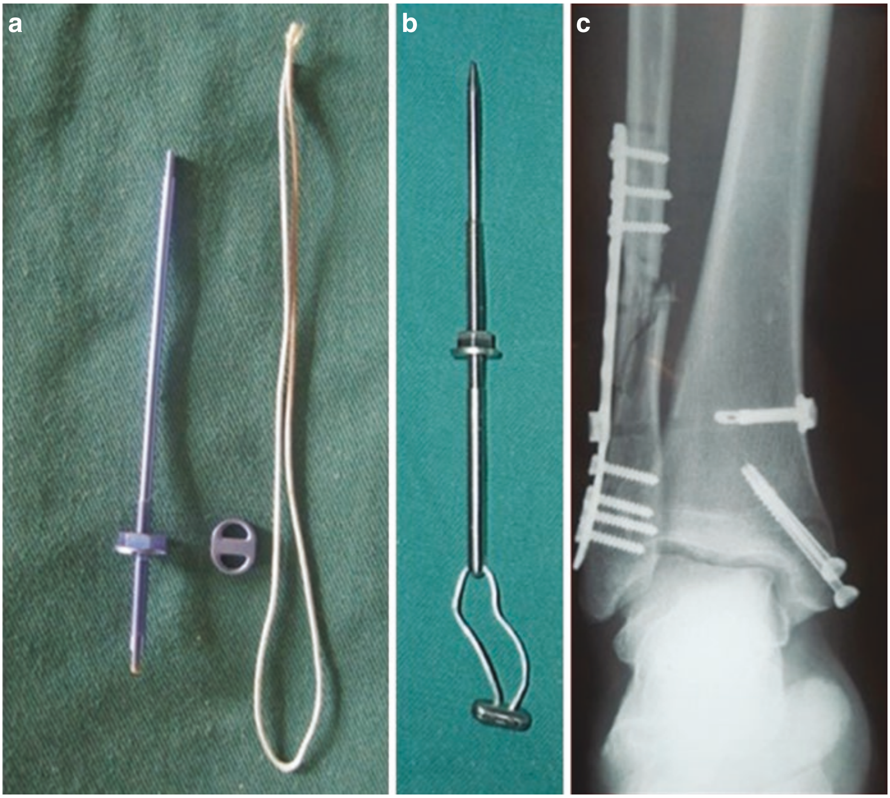

Fig. 12.1 (**a**) Components of the elastic bionic fixation; (**b**) Assembly of the fixation device; (**c**) Post-operative X ray film of the fixation device

arthritis, and the limited articular mobility. Given that, our research team used the biomechanical principle to invent an elastic fixation device for treatment of ankle fracture combined with distal tibiofibular syndesmosis (DTS). As presented, the elastic bionic fixation device included elastic cable, a bolt, and a button (Fig. 12.1), which could provide both rigid and elastic fixation, and therefore reduce the possible intra-articular issues, compared to the traditional rigid fixation. With this device, we treated 17 cases of ankle fracture combined with DTS, and at the final follow-up all of them obtained excellent and good outcomes according to the AOFAS score.

3 Screw Fixation

Screws are made up of four parts: head, shaft, thread, and tip. The head of a screw acts as an attachment for the screwdriver and can be hexagonal, cruciate, slotted, or Phillips. The head also acts as a reaction, and the pressure generated by the screw acts on the bone. The shaft or shank is the smooth portion of the screw between the head and the threaded portion. The thread is defined by its root (or core) diameter, its thread (or outside) diameter, its pitch (or distance between adjacent threads), and its lead (or distance it advances into the bone with each complete turn). The root area determines the resistance of the screw to pullout forces and relates to the area of the bone at the thread interface and the root area of the tapped thread. The cross-sectional design usually is a buttress (ASIF screws) or V-thread (usually used in machine screws). The tip of the screw is either round (require spretapping) or self-tapping (fluted or trocar). Clinically, if it is necessary to consider pulling out from the screw because of soft bone, it is more inclined to use a larger thread diameter, and if the bone is stronger and more fatigue is needed, the screw with a wider root diameter has a higher anti-fatigue failure ability. Screws also usually are divided into machine-type screws and ASIF screw.

The use of screws to convert torque forces into compressive forces through a fracture is a valuable technique. Its success requires the application of a screw in such a way that the proximal portion of the screw is allowed to slide in the near bone and the thread is formed in the opposite cortex so that the head of the screw can exert load and forces the fracture to heal. Careful selection of the angle of the screw corresponding to the fracture is necessary to prevent the fracture fragments from slipping when compressed.

Screw Breakage by Shearing During Insertion

A screw is a mechanical device that is used to convert rotary load (torque) into compression between a plate and a bone or between bone fragments. The basic components of a screw are shown in Fig. 12.2. As shown in Fig. 12.3, when the thread of the screw is released from the shaft, it is actually a slope or a slope, and the lower bone is pulled toward the fixing plate, causing compression between the bone and the bone. In order to achieve this effect, the screw head and shaft should rotate freely within the steel plate; otherwise, the compressive force generated may be limited. The locking screw passes through the plate hole; although this fixed interface is beneficial in some clinical situations, it prevents compression between the plate and the bone.

In the cortical bone, a tap is necessary so that the torque applied by the surgeon translates into compression rather than cutting the thread, thereby overcoming the friction between the screw thread and the bone (Fig. 12.3) that it is being driven into (Fig. 12.4). In some cases such as inserting screw into a dense bone or inserting a smaller diameter screw, using a separate tap and then inserting the screw, the screw can be pushed into the bone. Most modem screw designs have self-tapping screw tips that cuts the path of the thread when the screw is inserted. Screws with multiple cutting flutes at the up of the screw appear to be the easiest to insert and have a greater grip. Tapping in cancellous bone is less advantageous because tapping reduces the pullout strength of the screw. In some cases, it may beneficial to tap the cancellous bone. A clinical example is when treating a femoral neck fracture with a physiologically older patient versus a younger patient; you may need to use a tap to make a thread in the denser bone of a younger patient. The reason for using tap in the dense bone is to prevent the frictional force from causing femoral head to rotate

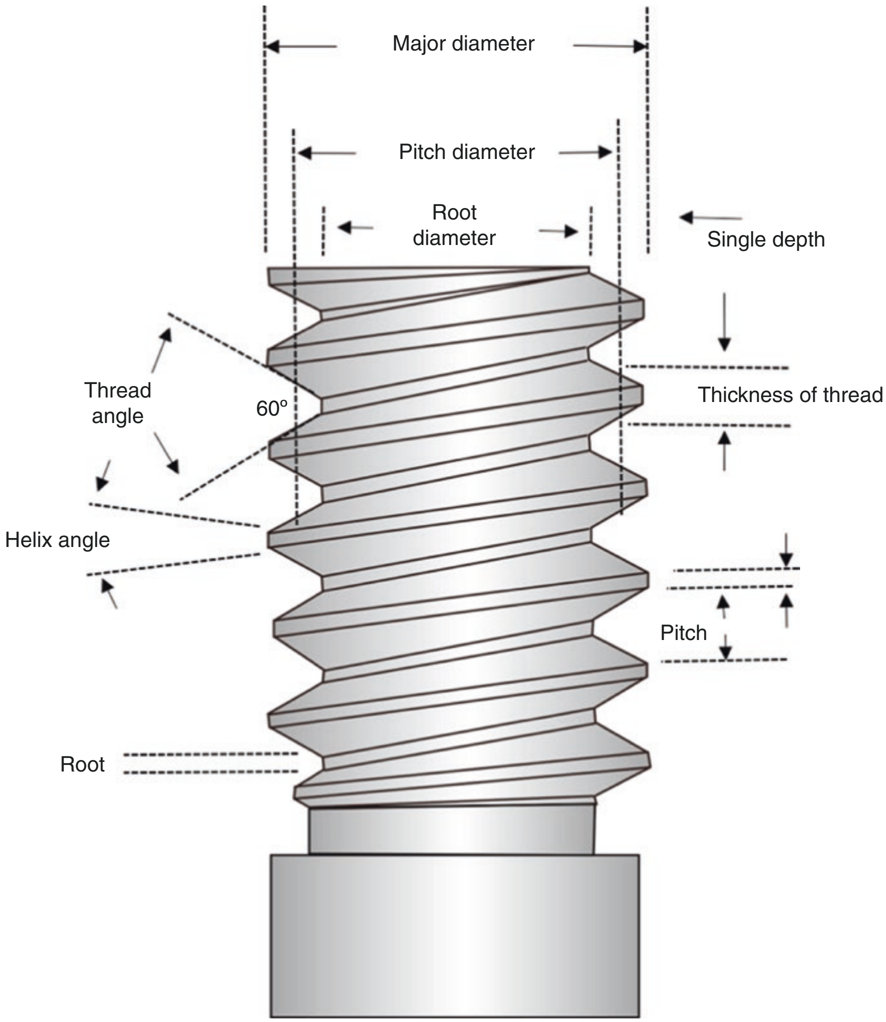

Fig. 12.2 Nomenclature of screws. The root diameter is the inner diameter of the screw and the pitch defines the distance between threads

during the insertion of screw resulting malreduction. In particularly hard bones the frictional forces become so great that it becomes difficult to advance the screw [18].

One problem with the placement of the screw is that the shear failure of the screw, the head twisting off usually, making it difficult to remove the shaft from the bone. This can occur especially if the tap is not used before insertion, or when a smaller (less than 4-mm diameter) screw is inserted into the dense bone. The stiffness and strength of the screw are related to the fourth power of its radius (the effect of moment of inertia for screws of the same material). The 6-mm diameter screw is

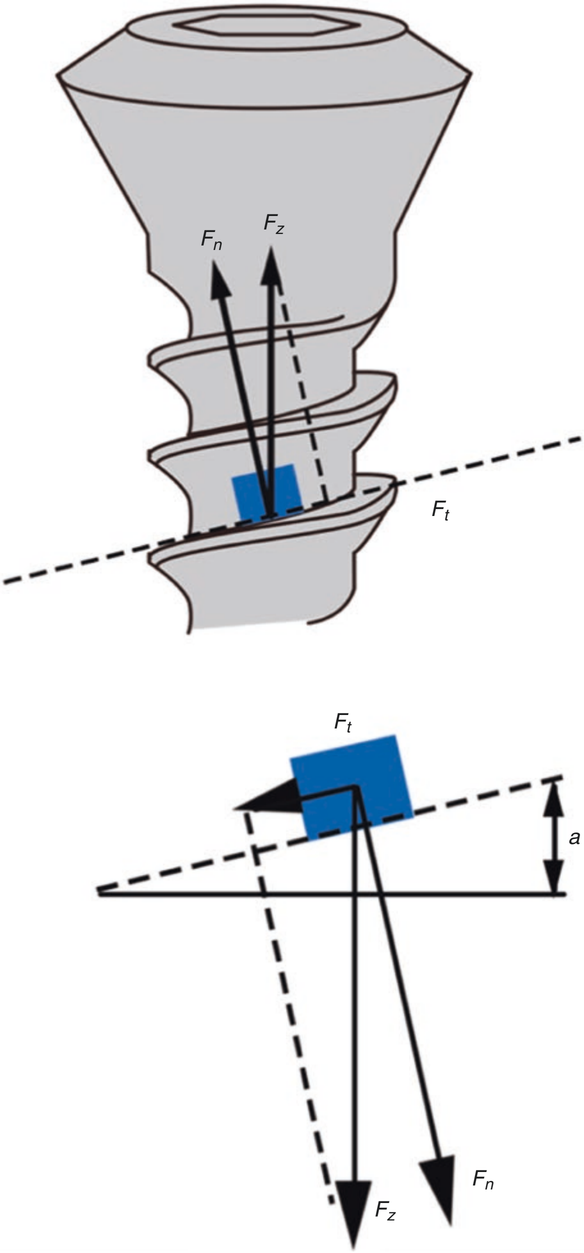

Fig. 12.3 A screw is a mechanical device that converts torque into compression between objects. The screw thread is actually an inclined plane that slowly pulls the objects it is embedded into together (F_n normal or compressive force acting against the screw head; F_t tangential or frictional force acting along the screw thread; F_z resultant of the two forces; α angle of the screw thread. The smaller the angle α [finer thread] the lower the frictional force)

about five times stiffer than a 3-mm diameter screw and 16 times more resistant to shear damage than a 3 mm diameter screw. The junction of the screw head and threaded portion of the screw is the transition point of shape and size. Therefore, it acts as a stress concentrator, usually where the screw breaks.

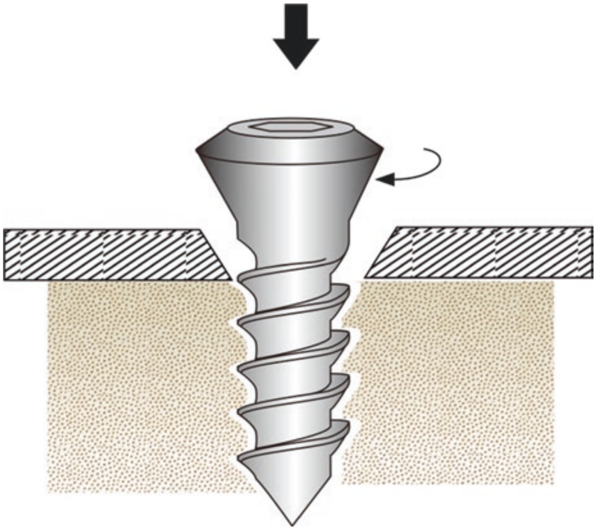

Fig. 12.4 Schematic diagram showing the approximate distribution of torque acting on a screw placed into cortical bone. With a pre-tapped hole, about 65% of the applied torque goes to produce compression and 35% to overcome the friction associated with driving the screw. When the hole is not tapped, only about 5% of the torque is used to produce compression, the rest going to overcome friction and to cut threads in bone. These observations do not apply in cancellous bone

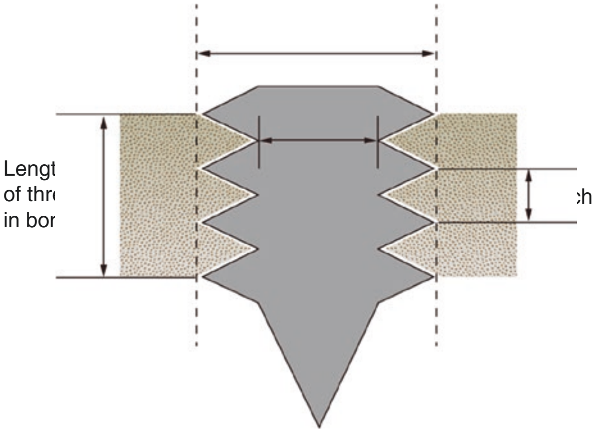

Fig. 12.5 The factors that determine the pullout strength of a screw are its outer diameter and length of engagement (this defines the dimensions of a cylinder of bone that is carried in the threads and is sheared out as the screw is pulled out of the bone) and the shear strength of bone at the screw-bone interface, which is directly related to its density. A finer pitch screw produces a small gain in purchase

Screw Pullout

Especially, in cancellous bone, the maximum force that a screw can withstand along its axis. The pullout force depends on the size of the screw and the density of the bone placed. As shown in Fig. 12.5, when the force acting on the screw exceeds its pullout strength, the screw will be pulled out or "stripped" out of the hole, and the sheared bone will be placed in thread, greatly reducing the nail holding force and fixing force. The larger the diameter of the screw, the larger the number of threads per unit length, and the longer the insertion length of the screw shaft, the greater the pulling force. And a greater density of the bone it is placed into. The diameter and length of the embedded screw can be considered to define the outer surface of a cylinder along that screw shears. Given a maximum stress that bone of a particular density can withstand, increasing the surface area of the screw cylinder increases the pullout force (because force = stress multiplied by the area over which it acts). In order to enhance screw purchase, it is conceivable to insert the screw of largest diameter into the bone of the highest density with the length of the implant as long as possible. However, placing screws of as large a diameter as possible has disadvantages. Larger screws occupy a larger volume in small fracture fragments, limiting the number of possible fixation sites and propagating adjacent fracture lines.

In cancellous bone, the extraction of the screw becomes a more important problem because the porosity of cancellous bone reduces its density and thus the shear strength. Drilling preparation, especially drilling, rather than tapping, can increase the pullout strength of cancellous bone (such as pedicle screws in the vertebral body). The reason why the cancellous bone is knocked down is that the tap is removed from the hole or placed in the hole, which can effectively increase the diameter of the hole and reduce the amount of bone material that interacts with the thread. When the bone density is reduced, the tapping is more harmful, and the pullout strength can be reduced from 8 to 27%. The pullout strength can also be related to the time after insertion. As the bone heals, it can remodel around the screw, possibly doubling its initial pullout strength [19].

Recent research has focused on whether pullout strength is an appropriate measure of screw performance in cancellous bone. In nonlocking steel and screw constructions, the stability of most structures comes from the friction created by compression between the plate and the bone. When the screw is inserted into the bone, if it is able to generate high values of insertional torque, the compression of the steel plate to the bone is increased, and the stability is increased. As the maximum insertional torque is reached and then exceeded, the screw will then "strip out" and lose its supporting force in the bone. Although there is a relationship between maximum insertional torque, screw pitch, and compression forces, it has been found that the pullout strength has no correlation with either the maximum insertional torque or screw pitch. Therefore, this may be a better way to measure screw performance and optimize screw characteristics.

Screw Breakage by Cyclic Loading

Once screw is successfully inserted and the construction is completed, the screw will be subjected to cyclic bending forces as the patient begins to mobilize (Fig. 12.6). Ideally, a nonlocking screw initially tightens the plate to achieve the maximal torque, which is translated into the maximum compression between the plate and the bone (Fig. 12.4). The screw on the bone portion is in frictional contact, depending on the friction generated between the plate and the bone. The frictional force is directly dependent on the compressive force generated by the screws. If a slip occurs between the plate and the bone, the bending load will be transferred from the screw head into the plate, at which point the screw comes into contact with the plate. The bending load perpendicular to the axis of the screw, coupled with possible stress corrosion and fretting corrosion, may cause the screw to fail rapidly in fatigue. Zand et al.'s research shows that when the screw is fastened to the steel plate, it is subjected to a maximum load of less than 1000 loading cycles due to bending

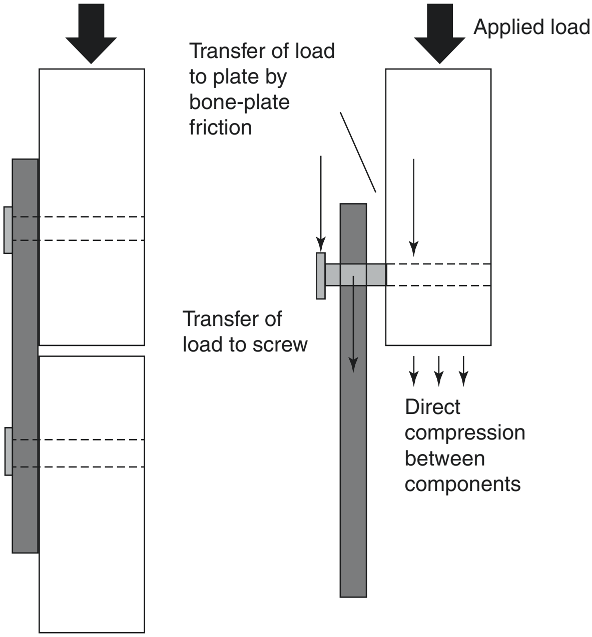

Fig. 12.6 A mechanism for rapid failure of screws in cyclic bending occurs when the screw has not been tightened sufficiently to keep the plate from sliding along the bone surface (the plate-bone gap shown here is exaggerated for clarity). The results is that bending loads are applied transverse to the long axis of the screw, which in combination with fretting corrosion caused by the screws rubbing against the plate results in early failure of the screw

fatigue, compared to a fully fastened screw capable of withstanding more than 2.5 million loading cycles. The load is less than 10–15% of the maximum load. This emphasizes the clinical importance of ensuring screw tightness during the fixation of the plate.

Locking the screws on the board can reduce the problem, since the problem is less subjective when the screw head is fully fastened to the plate hole. Small-fragment screws (3.5- to 4-mm outer diameter) can fatigue because of their small core diameters. Although the use of locking screws with a larger core diameter and shallower thread can reduce the possibility of fatigue failure, a smaller core diameter and deeper thread can increase purchase strength in the bone. Screws with smaller core diameters are more likely to fatigue than larger ones. The fatigue strength of the screw must be weighed against the purchasing power of the screw and the size of the screw associated with the size of the bone fragment. The surgeon sometimes must make a decision between a screw with a large core diameter with shallower thread, which maximizes fatigue strength, and a smaller core diameter screw with deeper threads, which maximizes purchase power.

Cannulated screws are used for fixation when the insertion of a guide wire is helpful to guide the future path of the screw. However, as the bone density increases and the diameter of the guide wire increases, the drilling accuracy of the guide wire decreases. Cannulated screws follow the same mechanical principles as solid screws, but material must be removed from the center of the screw to accommodate the passage of the guidewire. Manufacturers commonly increase the diameter of the screw at the base of the thread to fit the loss of this central material. The same size

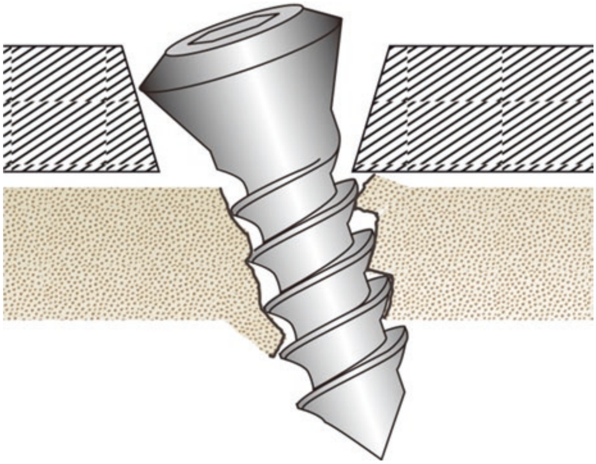

Fig. 12.7 Using a fully threaded lag screw causes the threads to engage in bone on both sides of the fracture. This inhibits the screw from compressing the bone fragments together

of cannulated screws usually has less thread depth than solid screws. The result depends on the size of screw, rather than pullout strength. For screws with a diameter of 4 mm, the tensile strength of the cannulated screws of the same outer diameter is about 16% lower. Alternatively, to maintain the same thread depth, the outer diameter of the screw may be increased. Another consideration is that the cannulated screw is much more expensive than the solid screw [20].

Fully Threaded Lag Screws
The lag screw is a very effective device which can produce amount of pressure across fracture fragments and the fracture site. The head of the screw and upper portion of the shaft must be allowed to glide in the near broken pieces to pull the far broken fragments toward it, thereby creating compression on the fracture surface. As shown in Fig. 12.7, a fully threaded lag screw can prevent the gliding action between the two fracture fragments. The fully and partly threaded lag screws were used to compare the compressive forces across the fracture site. The results showed that the average compressive force at the opposite cortex (i.e., the force in the screw itself) was about 50% greater when a partly threaded screw was used.

3.1 Machine Screws

The machine screws are threaded throughout the length and can be self-tapping or threaded before insertion. Most are self-tapping; there is a cutting flute that cuts the screw threads as the screw is inserted. Machine screws are mainly used to fix the hip compression screw on the femoral shaft. The size of the machine screw holes is critical. A hole that is too large can cause the thread to be unsafe, and a hole that is too small can cause the screw to be inserted or fractured during insertion. The drill bit selected should be slightly smaller than the screw after the screw minus the thread. For a self-tapping screw, the drill point is used to drill holes in soft bone. The size of the screw and drill tip should be checked before surgery.

3.2 Asif Screws

Screws developed by the Swiss ASIF Group for bone fixation techniques and principles are widely used. The threads are more horizontal than the machine screws, and these screws rarely seldom self-tapping; the drill must be tapped with a cutting tapper before inserting the screws. ASIF screws are available in cortical, cancellous, and malleolar designs. Mini screws for fixation of small fragments and small bones, and the standard cancellous and cortical screws, come in multiple lengths and diameters. Standard cancellous and cortical screw heads have a hexagonal recess for a special screwdriver, while the smaller screws have a Phillips head.

3.2.1 Cortical Screws

The full length thread of the cortical ASIF screw is available in a variety of diameters (4.5, 3.5, 2.7, 2.0, and 1.5 mm). If the hole in the near cortex is drilled too deep, the cortical screws can be used as a positional or lag screws for inter-fragmentary compression.

3.2.2 Cancellous Screws

These screws have larger threads that provide more support for soft cancellous bone, making them more suitable for use on the metaphyseal areas. The cancellous screws are available in 6.5- and 4.0-mm diameters with thread lengths of 16 and 32 mm, respectively. Regardless of the lengths of the screw, both lengths are threaded. The malleolar 4.5-mm screw is also included in this group, but it is unique in that it has a self-tapping thread. Choosing the right drill size and tapping the hole is the key to a safe purchase. Plastic and metal washers are commonly used with these types of screws to reattach ligament tears or to increase compression between the fragments by providing a larger cortical surface area for screw head.

3.2.3 Self-Tapping, Self-Drilling Screws

The self-tapping screws are the same sizes as cortical screws. A small portion of the ends of these screws are used to remove bone fragments. Self-tapping screws have a lower pullout strength due to their structure. These screws are preferably used for external fixation pins [21].

3.2.4 Locking Screws

The locking screw is a self-tapping screw with a locking screw at the head. These screws require accurate predrilling so that the locking plate is tightly attached to the locking plate and requires a special screw drivers for implantation.

4 Plate and Screw Fixation

Pauwels et al. firstly defined and applied the tension band principle to fix fractures and nonunion. This engineering principle applies to the transformation of the tensile force into a compressive force on the convex side of an eccentrically loaded bone. This is accomplished by placing a tension band (bone plate) across the fracture on the tension (or convex) side of the bone. The tension is counteracted by the tension band at this position and converted into a compressive force. If the plate is applied to the compression (or concave) side of the bone, it is likely to bend, fatigue, and fail [22]. Therefore, a basic principle of tension band plating is that it must be applied to the tension, and the bone itself will get a compressive force, so the tension band device does not require a heavy and rigid tension band principle and is also used for some olecranon and patellar fractures. Tension bands and axial compression are often combined when using plates and screws [23–25].

We have found that almost all plate hole breaks occur at the plate hole near fracture area [26]. Therefore, the hole area seems to be the weakest part of the board, and naturally it is a place for improvement. We only widened the locking compression plate (LCP) in the hole area to make it a gourd-shaped LCP to increase the strength of the plate and reduce the plate breakage (Fig. 12.8) [27, 28]. After a series of axial loading single cycle to failure test, torsion single cycle to failure test, four-

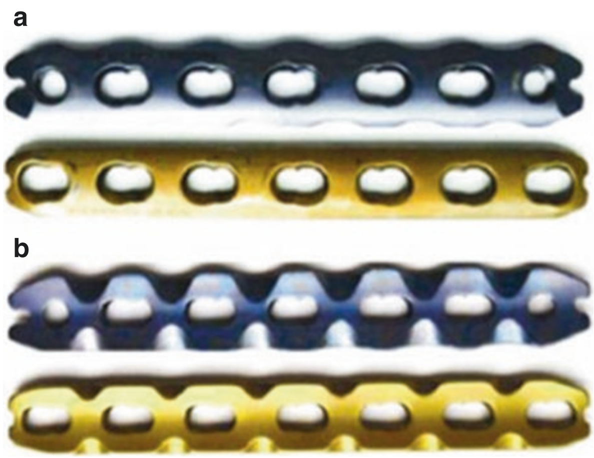

Fig. 12.8 Gourd-shaped LCP designed to enhance the plate strength in order to reduce plate breakage. (**a**) The Gourd-shaped LCP (up) and LCP (down) viewed from above; (**b**) Gourd-shaped LCP (up) and LCP (down) viewed from below

point bending single cycle to failure test, and dynamic four-point bending test, it is concluded that the gourd-shaped LCP structure has greater stiffness, strength, and longer fatigue life than LCP. This may be a more reasonable LCP that can reduce the rate of clinical breaks [29].

Our team used a new anatomical plate and compression bolt fixation technique, combined with a small incision of the posterior foot, to treat intra-articular calcaneal fractures and achieved good or excellent clinical results, and had fewer soft-tissue problems (Fig. 12.9) [30]. Compared to the conventional plate and cancellous screws technique, our fixation technique requires higher loads to cause structural failure, which may be related to the design of the implant. According to the measurement of the calcaneus specimen and the data of the three-dimensional CT image of the calcaneus, the anatomical steel plate and the compression bolt were designed. The use of conventional anatomical plates and cancellous screws to fix a calcaneal fracture is characterized by compression of the plate to the lateral wall of the calcaneus. The actual stability lies in the friction between the plate and the bone, which

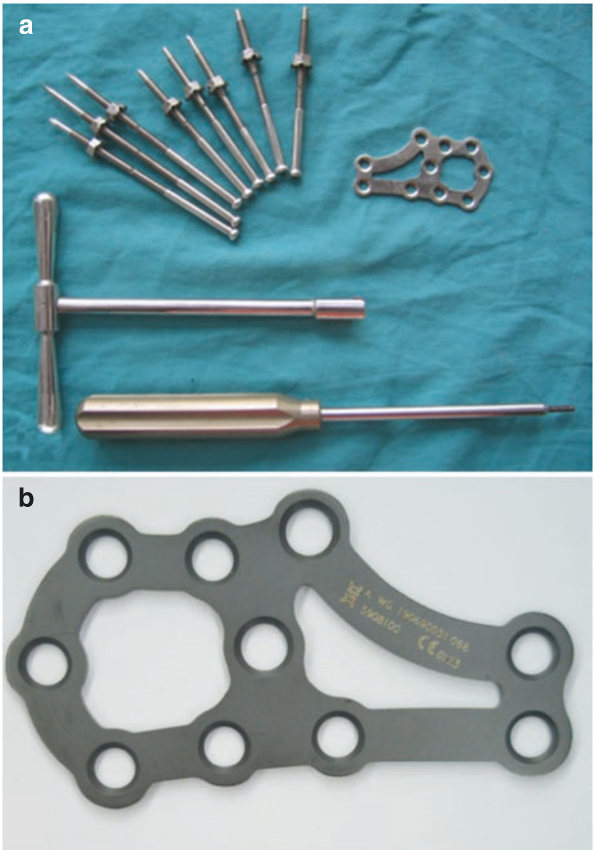

Fig. 12.9 (**a**) components of the calcaneal fracturefixation system; (**b**) the anatomical steel plate viewed from up; (**c**) post-operative lateral and posteroanterior radiograph; (**d**) post-operative CT images

Fig. 12.9 (continued)

can easily be achieved by our anatomic plate and compression bolts. Putting the compression bolts together can be seen as another piece of plate that is compressed onto the inner side wall of the bone specimen to provide higher friction and restore the width of the calcaneus.

Another example was the application of compression bolt in the treatment of Schatzker type II–VI tibial plateau fractures (Fig. 12.10), in which joint surface widening and collapse are commonly accompanied [31, 32]. The traditional metal plates and screw fixation of fractures of such types are commonly associated with a high rate of postoperative reduction loss, which likely result in the development of traumatic arthritis. Our preliminary reports of using this compression bolt presented the favorable results, both in biomechanics and clinical effectiveness.

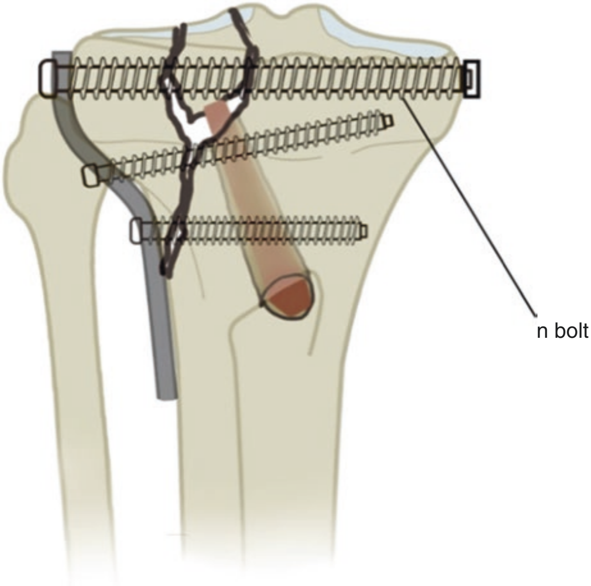

Fig. 12.10 Application of compression bolt in the treatment of Schatzker type II–VI tibial plateau fractures

4.1 Locking Plates

Locking plates are a combination of steel plate technology and percutaneous bridge plating technology using locking screws as fixed angle devices [33]. Locking steel plates provide a stronger, longer lasting fix than nonlocking steel plates [34]. They have been proven to allow for greater loads bearing than regular plates. The Less Invasive Stabilization System (LISS) (Synthes, Inc., West Chester, PA) uses unicortical locking screws that allow for greater elastic deformation than conventional plating systems. The locking plate can also be used in combination with locking and unlocking screws, mechanically similar to a pure locking structure. The locking plate works best on osteoporotic bones, where pulling out the steel plate is problematic. They also provide sufficient load-bearing strength to avoid the distal femur, proximal humerus, and medial and lateral plates of the tibial plateau. However, the locking plate structure also has an inherent that inhibits the movement of the fracture site, making it insufficient to stimulate callus formation. Therefore, the locking plate of the distal femoral fracture can result in insufficient and asymmetrical ankle formation, with minimal deposits in the proximal cortex. A recent study of locking plates for distal femoral fractures confirmed this and showed a nonhealing rate of 10–23% and a reoperation rate of open fractures of 31%.

5 Intramedullary Nail Fixation

Satisfactory stable intramedullary fixation of the fracture is possible under the following conditions:

1. When a non-comminuted fracture occurs in the narrowest part of the medullary canal, it can be considered to unlock the intramedullary nail; not only the lateral force and shear force are eliminated, but also the rotational force is controlled. If the medullary canal of one fragment is much larger than the other, it often results in poor control of the rotational force; in these cases, interlocking techniques are required. In general, the interlocking screws should be placed at least 2 cm from the fracture to provide sufficient stability for postoperative functional activity. Axially unstable fractures are best treated with static or double-locked nails.
2. The curvature of the bone must be taken into consideration when selecting the type of staple and determining the degree of reaming necessary. Biomechanically, unlocked nails achieves stability by a curvature mismatch in curvature between the bone and the nail, resulting in a longitudinal interference fit. If curvature mismatch is large, more reaming will be required. The entry portal is critical for all nails and should be in areas where the insertion force is minimized. In the femur, this is a straight nail that conforms to the medullary canal in the piriform fossa, or a nail that is curved at the proximal trochanter with a slightly lateral proximal bend. For the tibia and humerus, the offset between the entrance and the alignment of the tube create a powerful force on the posterior and medial cortex, respectively. The humeral head of the nail begins to reduce the insertion of the force into the tibia.
3. Sufficient diameter and continuity of the medullary canal are prerequisites for intramedullary nail techniques. Excessive reaming should be avoided as it significantly weakens the bone and increases the risk of thermal necrosis. We recommend reaming until the cortical "chatter" is encountered, i.e., "reaming to accommodate," but never insert a nail larger than the diameter of the tube. In general, we use nails 0.5 or 1.0 mm smaller than the largest reamer.
4. The locking intramedullary nailing technique should allow the fracture to be nailed into the joint 2–4 cm. These techniques require the use of locking screws or tightening screws. These techniques require the use of blocking screws or "Poller" screws. A new design of an intramedullary nail with an oblique distal locking screw and a screw that can be locked into the incision to form a fixed angle can increase the stability of these metaphyseal fractures.

A perfect intramedullary nail has not yet been designed. Different bone contours make this nail impossible, but improvements in the design of the intramedullary nail continue. Special nails may be designed for each bone, all types of fractures, fractures or the same bones in different areas. The intramedullary nails should meet the following requirements:

1. It should be strong enough to provide sufficient stability to maintain alignment and position, including preventing rotation; it should include the necessary interlocking screws only.
2. It should be constructed so that contact-compression forces can impact the fracture surfaces, an ideal physiological stimulus.
3. It should be easy to remove when placed; accessories are provided for easy movement.

Before selecting this technique, surgeons should be aware that intramedullary fixation, like other internal fixations, may present complications. This is not a technique that can be used at will. We recommend the following considerations:

1. Adequate preoperative planning must ensure that the fracture can be fully stabilized within the working area.
2. The patient should be able to tolerate a major surgical procedure. Patients with severe pulmonary injury should be taken special consideration, because the added fat emboli from the procedure may intensify pulmonary problems.
3. The proper length and diameter of the nail must be available before the [35–37] surgery is determined.
4. Appropriate equipment, well-trained assistants, and optimal hospital conditions are necessary for successful insertion of an intramedullary nail.
5. A metal nail is not a substitute for the union and if subjected to excessive will also bend or break the strain during the recovery period.
6. Closure nailing techniques should be used whenever possible, using these techniques to improve healing rates and reduce infection; [38–41] however, surgeons must be familiar with both open and closed techniques. As more experience is gained with closed techniques, fewer and fewer fractures are needed for open reduction [42, 43]. However, limited open reduction is better than accepting a poor closed reduction. This situation most frequently occurs in high-energy subtrochanteric femoral fractures, traction does not adequately correct flexion and abduction.

5.1 Types of Intramedullay Nails

Just as plate, intramedullary nail has an anatomic and functional name. The central body nail is inserted into the bone in a straight line with the medullary canal. It interferes longitudinally with the bone through multiple points of contact [44–49]. They rely on restoring bone contact and stability to avoid axial deformation of the fracture during rotation [50, 51]. The classic Küntscher cloverleaf and Sampson nails are examples of centromedullary nails. The condylocephalic nails enter the bone of the condyles of the metaphysis and usually enter the opposite metaphyseal-epiphyseal area. They are usually inserted into groups to increase rotational stability. Ender and Hackenthall pins are examples of condylocephalic nails. Cephalomedullary nails have a centromedullary portion but it is also allowed to be fixed to the femoral head. The Küntscher Y-nail and Zickel subtrochanteric nail are examples of this type [52, 53].

Interlocking techniques further improves these classics by adding interlocking centromedullary and interlocking cephalomedullary nails [54–56]. Interlocking nails allow longer working length of the interlocking nail screw axial and rotational deformation resistance of fracture. Modney first designed the first interlocking nail. Küntscher also designed an interlocking nail (the detensor nail), which was modified by Klemm and Scheilman and later modified by Kempf et al. These pioneers developed techniques and implants that form the basis of some designs and techniques used today. Cephalomedullary interlocking nails, designed to treat complex fractures and the proximal femur, were axially and rotationally unstable, such as complex subtrochanteric fractures, pathologic fractures, and ipsilateral hip and shaft fractures. These nails can be secured with bolts, nails, and special lag screws such as Russell-Taylor reconstruction nails, Williams y nails, and Uniflex nails [57]. The current intramedullary nails for femoral fixation design reflect regional internal fixation nails. Antegrade femoral nails can be performed through the piriformis or trochanteric inlet. The retrograde femoral nail passes through the entrance between the femoral condyles [58, 59].

Interlocking fixation is defined as a dynamic, static, and double lock. Dynamic fixation controls bending and rotational deformation but allows axial load transfer of the bone. Axial stable fractures and partial nonunion can be fixed by power. Static fixation controls the rotation, bending, and axial load, so that the implant has more bearing potential and reduce the fatigue life of the equipment. It is particularly useful in crushing, nonisthmal fractures of the femur and tibia. The double-locked mode controls bending, rotational forces, and some axial deformation, but some shortening occurs due to the ability of the screw to translate axially within the nail. This type of fixation is often used for humerus fracture with delayed union and not healing.

The dynamics of the interlocking nails were originally designed to avoid fracture healing, [60] as it is theoretically believed that static interlocking will stop the repair of the fracture. This technique involves conversion of the static mode to a dynamic mode by removing the screws from the longest fragment. Dynamization increases the fatigue life of the nail by reducing the load-carrying capacity of the nail while increasing the compressive force at the break point; however, if there is insufficient cortical stability or bone regeneration before exercise, shrinkage may occur [45, 61, 62].

5.2 Reamed Versus Unreamed Intramedullary Nailing

For patients with multiple fractured long bone fractures, the need for reaming for intramedullary nailing has been controversial [63]. Physicians who support non-reamed nails emphasize the lack of physiological effects of reaming, such as fat embolism in the lungs [64, 65]. Experimental evidence suggests that reaming has an adverse effect on lung function. This adverse effect does not appear to be apparent in most clinical patients; however, some authors believe that the development of pulmonary complications may be related to the severity of the associated chest injury, rather than to the reaming of the medullary cavity. Studies supporting the reaming nail showed no statistically significant difference in the incidence of pulmonary complications in patients with and without reaming [66]. Due to various factors leading to the development of adult lung failure syndrome, it is difficult to determine which patients' lung expansion may be harmful. Whether the long bone reaming nail increases the frequency of infection is another controversial area. Current clinical data show no significant difference in infection rates between reamed and non-reamed intramedullary nail.

6 External Fixation

External fixation is accepted in trauma management, ranging from damage control to final treatment. External fixation requires more careful clinical and radiographic monitoring than internal fixation, but the general application and management principles are absolutely straightforward, [67, 68] and its versatility allows it to be used in a variety of fractures [69, 70]. However, external fixation is not suitable for all fractures; [71–73] when other forms of fixation, such as screws, plates, or nails, are more suitable, it should not be used [74].

External fixator should be used when the other methods of fracture fixation are not applicable, although the external fixation will cause inconvenience to the patient and is highly likely to cause surface needle infection, it fills the clinical vacancy [75–77]. After high energy trauma with open injuries, plates and intramedullary nails are sometimes considered an unacceptable risk of deep infection. Although the degree of comminution and the extent of involvement sometimes lead to inherent instability of the fracture morphology, the external fixator can even better mechanically control the fracture through the joint. Unlike plates and most intramedullary nails, external fixator provides an opportunity for postoperative correction. The adjustability of the external fixator has been unique until recently, partly explaining why they continue to play an important role in musculoskeletal wound care. External fixation has become the preferred method for treating some of the most challenging bone pathological diseases encountered in the clinic [78]. Although there are alternative treatments, it is still an important factor in limb salvage in early and late bone remodeling of severe limb injuries [79]. This is currently the only system that allows surgeons to control fixation flexibility during bone healing. External fixtures have undergone tremendous changes, from the most primitive combination of wood plywood design to the modern design of widely used metals and composites. The development of these devices has brought many complications and it has become a more technically demanding process. Despite these factors, many surgeons around the world continue to use external fixators to treat complex fractures, segmental defects, and congenital malformations. The work of many clinicians, researchers, and engineers around the world is responsible for the current external fixture design [80]. For example, the dispersion and compression mechanisms of modern equipment are attributed to Lambret in 1911. In 1931, Pitkin and Black field first proposed a double cortical pins to connect two external fixation clips as a bilateral frame to promote fracture healing. Anderson et al. published a series of papers from 1933 to 1945, outlining the application of half-pins and transfixation pins in various long bone fractures, arthrodesis, and limb lengthening surgery [81]. These incremental improvements have resulted in currently available designs, providing three main configurations of external fixtures [82].

Professor Gavril A. Ilizarov's contribution to the modern design of unilateral and circular external fixes should be recognized [83–85]. He also invented methods for limb salvage and bone extension by distraction osteogenesis. Distraction osteogenesis is a fixture for us to use mechanical force to stimulate the bone regeneration process clinically. This is achieved by a special form of external fixture called an annular fixture. Ilizarov found that these external frames can be used in a variety of applications, including post-traumatic and congenital limb reconstruction, treatment of osteomyelitis, regeneration of bone defects, deformity correction, and complex arthrodesis. These devices take advantage of Ilizarov's principle of stretch tissue, relying on a special type of low energy osteotomy to preserve local blood vessels. Ideally, only cortical bone fractures are made, while the medullary vessels and periosteum remain intact at the metaphysis. The initial incubation period allows the osteotomy to begin healing before the fixture is periodically adjusted to achieve a controlled gradual mechanical stretch. When the anchor is slowly extended, new bone is formed in the gap created in the osteotomy by the now familiar distraction osteogenesis process. For example, this process considers bone reconstruction through bone transport across segmental defects, using small tensioned Kirschner wires (K-wires) and circumferential ring supports. When new bone growth occurs at the metaphysis, a healthy bone gradually shifts to the defect. When the new bone grows out of the metaphysis, the normal bone gradually shifts to the defect. During the development of similar stretched tissue, the tension generated by mechanical stretch stimulates new bone formation, skin, blood vessels, peripheral nerves, and muscles. This impressive process, bone elongation and regeneration occurs at a rate of about 1 cm per month. The Ilizarov technique is important in the treatment of nonunion by mechanical stimulation and regulation of callus and can be used to reconstruct segmental defects that can be reliably filled far beyond the iliac bone graft. More importantly, this technology has produced limb salvage with superior quality of regenerative normal bone.

The circular frame contains the basic components of the rings, the tensioned wires, and the connected threaded rods. The stability of the frame depends on the configuration of the components, which will affect the local mechanical environment around the regenerated bone, and also determining the type, rate, and quality of the tissue formed. For example, the stability of the structure will change depending on the type and size of the rings (full ring, partial ring, or arches). Full ring provides the greatest stability, partially intermediate, and arches the least. The complete ring provides maximum stability, partial middle, and arch. At the very least, the diameter of the ring is also important, and the smaller ring is more stable than the larger ring of the same thickness. Stability will also depend on the distance between the rings, as well as the type and number of ring connectors, such as wires, rods, and Shantz pins. In clinical cases, different combinations of the circular frame components are used depending upon the intended application and required stability.

6.1 Biomechanical Aspects of Fracture Fixation in Specific Location

6.1.1 Fixation in the Proximal Femur

Fixation of the proximal femur fracture is particularly challenging because during normal activity, the pressure through the femoral head can reach four to eight times the body weight [86–89]. This force acts through an important forearm (the length of the femoral neck) that exerts a large bending load on the fixation hardware [90]. In addition, many of these fractures occur in the elderly, who may have trabecular bone of low density and poor mechanical quality. In addition, it is generally not possible to obtain a screw in the cortical bone of the femoral head [91–93].

The major force acting in a basicervical fracture of the femoral neck, fixed with a sliding hip screw, is the joint reaction force through the femoral head, derived from body weight and the force generated by muscle movement during walking [94–96]. The joint reaction force can be divided into two parts. One (Fig. 12.11) is perpendicular to the axis of the sliding screw, causing the fracture surfaces to shear along the fracture line, which results in inferior displacement and varus angulation of the femoral head, and increases the resistance of the screw to sliding. The other parties parallel to the screw, and the surfaces are joined together by friction and

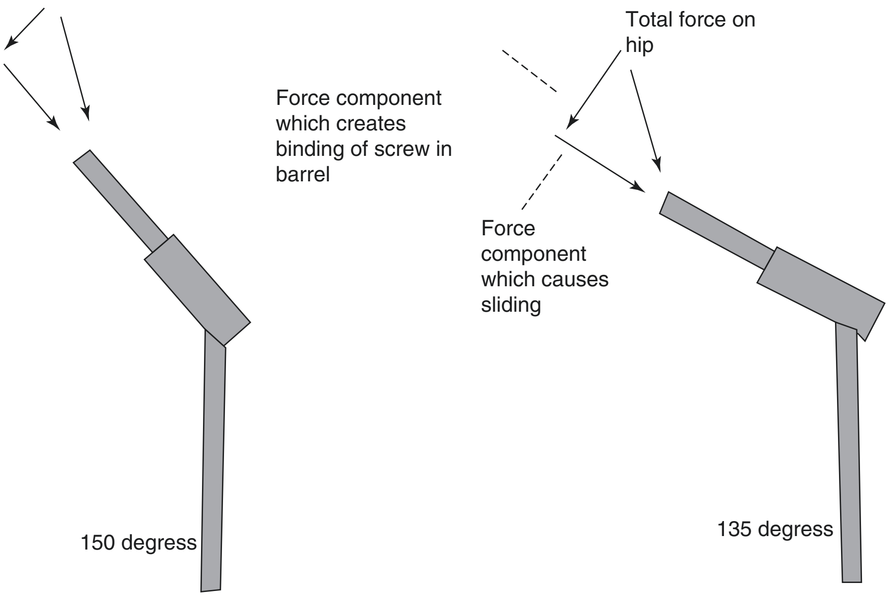

Fig. 12.11 The joint reaction force in the femoral head can be divided into two major components. The one parallel to the axis of the femoral neck produces sliding and impaction of the fracture components and the other, transverse to the femoral neck, causes the screw component of the femoral hip screw to bind, resisting sliding. The higher-angle hip screw has a screw axis more closely aligned with the joint reaction force so the force component that produces sliding is larger whereas the transverse force component resisting sliding is smaller

mechanical interlocking for improved stability. Therefore, the goal of the femoral neck fixation system is to use the component of the joint force parallel to the femoral neck to encourage the fracture surfaces to slide together. This is the basic principle for selecting high angle hip screw when possible.

The following points regarding the sliding hip screw device also apply to the nail/tension screw device. When the screw slips, since the structure is staggered by the fracture, the screw is supported by the barrel to prevent the femoral head from bending down. Adhering to two basic mechanical principles will increase the ability of the screw to slide within the side plates or nail holes. As mentioned above, higher angle hip screws are more effective in adjusting slip [97–101]. In addition, the screw should engage as deep as possible within the barrel. For the force acting on the femoral end of the screw, if the internal force of the screw in contact with the barrel is small, the remaining amount of the screw shaft in the barrel is small, and the internal force in the barrel is increased. This is because the moment (bending load) generated by the force acting transversely on the screw axis at the femoral head (Fig. 12.12) acts on the longer force arm or the vertical distance L (the force x is perpendicular to the edge of the barrel, i.e. the fulcrum)). The balance arm L_b is shorter because there are fewer screws left in the barrel. Since F_h acts on a longer arm and F acts on a shorter arm, F_b increases. When the screw is in contact with the barrel, its internal force F_b produces greater frictional resistance, which requires more friction to overcome the friction and allow slippage. Sliding hip screws with two- or four-hole side plates seems to provide an equivalent anti-physiological compression load. There are several factors that affect the fixation strength of the femoral neck using multiple screws, but the number of screws used (3 or 4) is not a significant factor.

Factors that increase this type of fixed strength include more long-axis screws with transverse fracture lines, larger femoral skull mass density in the position of the screw, and less comminuted fractures, shorter arm loads on the arm (shorter The distance from the center of the femoral head fracture line). However, the most important factor is the quality of the reduction because of the importance of cortical support in reducing fracture displacement. Under physiological load, several mechanisms of fixed failure were observed (Fig. 12.13). In some cases, the screw bends downwards, especially when it is unable to support a fracture surface below the screw due to fracture comminution. If a washer is not used to distribute the screw load to the bone, when the cortex is thin, the screw head will pass through the cortex near the greater trochanter. Finally, if the screws do not support well down through the fracture, they may rotate downwards, causing the femoral head to invert. Supporting the hypodermis with at least one screw is a mature clinical technique that may help prevent this from happening.

6.1.2 Fixation Around the Metaphyseal Region of the Knee

Both supracondylar fractures of the femur and tibial plateau fractures are challengingly stable because they usually involve the fixation of multiple small cancellous bones [102–104]. Mechanically comparable alternative methods for supraorbital

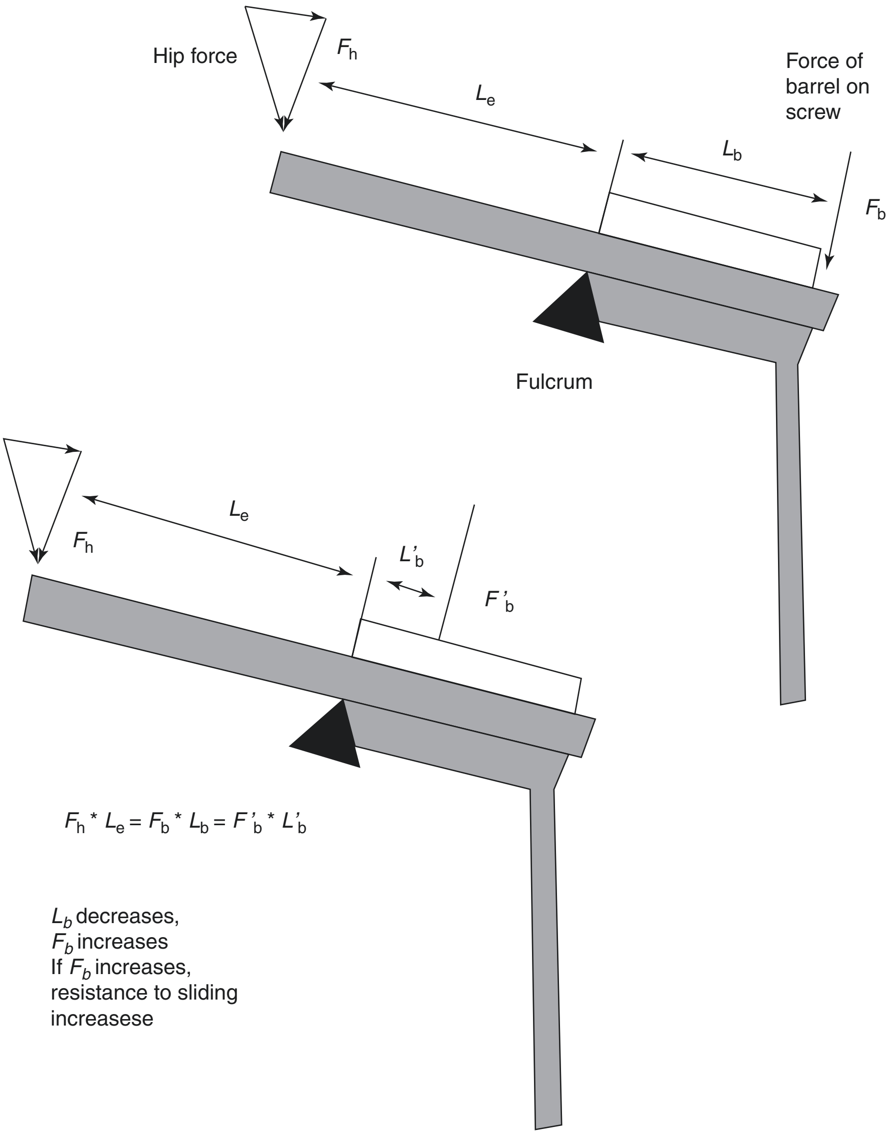

Fig. 12.12 The greater the length of the sliding screw within the barrel, the lower its resistance to sliding. In this diagram F_h is the component of the joint reaction force perpendicular to the axis of the screw. The inferior edge of the proximal end of the barrel is the location of the fulcrum in bending. An internal force, F_b from the surface of the barrel acts against the screw to counteract F_h. For equilibrium, the moments produced by $F_h(F_h \times L_e)$ and $F_b(F_b \times L_b)$ must be equal. If L_b, the distance from the point of application of internal force F_b to the fulcrum, decreases, F_b must increase to therefore the resistance to screw sliding will increase (L_e is the length of the screw beyond the barrel)

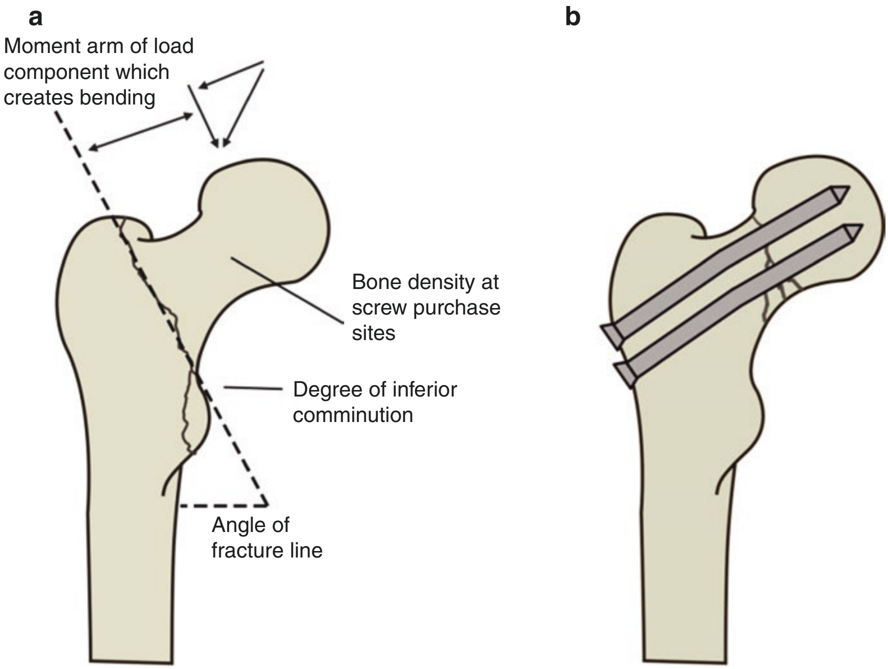

Fig. 12.13 (**a**) Some factors that decrease the strength of femoral neck fracture fixation include decreased bone density, a more vertical fracture surface (which reduces buttressing against bending), and a longer moment arm or distance of the center of the femoral head to the fracture line. (**b**) Observed mechanisms of failure of femoral neck fixation using screws include bending of the pins, displacement of the screw heads through the thin cortex of the greater trochanter, especially if washers are not used, and rotation of the screws inferiorly through the low-density cancellous bone of the Ward triangle area until they settle against the inferior cortex

fixation include condylar plates, plates and plates that use lag screws at the fracture site. All equipment tests seem to provide similar structural stiffness [105–108]. The most important factor in determining plate fixation is to maintain contact at the cortex opposite the fixture. A fixed structure without cortical contact is only 20% harder than a fixed structure with cortical support. It has been found that the use of a retrograde IM supracondylar nail results in a 14% reduction in axial compression strength and a 17% reduction in torsional strength compared to fixed-angle side panels. However, longer nails (36 cm) enhance fixation stability compared to shorter nails (20 cm). Several new fixation systems have been described as stable for supracondylar fractures of the femur. The Minimally Invasive Stabilization System (LISS) uses a low profile plate with a single cortical screw distal end, which is also locked to the plate. Compared to a conical screw or a support plate, the LISS plate produces a structure with greater elastic deformation and less sedimentation.

The tibial plateau fracture is difficult to stabilize [109–111]. Given the patient's prognosis, risk factors for reduced reduction have been shown to include patients older than 60 years, premature weight bearing, fracture comminution, and severe

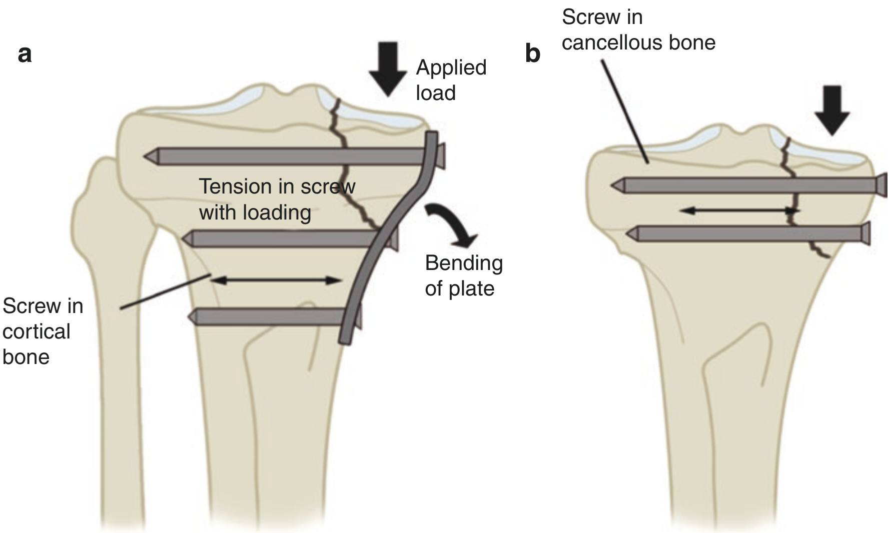

Fig. 12.14 Two alternative methods of fixation of tibial plateau fractures: (**a**) transverse screws combined with a buttress plate and (**b**) transverse screws alone. The buttress plate provides additional support in bending as the tibial fracture component is loaded in an inferior direction and allows the screws to engage the thicker, more distal cortical bone

osteoporosis [112–114]. Different fixing methods include using a wire or screw alone (Fig. 12.14), [115, 116] or placing a screw through an L-shaped or t-shaped plate to support the cortex. Wires of various shapes have been tested and the results show that the stiffness of the structure increases with the number of wires, regardless of the direction of the wires. As shown in Fig. 12.14, when the screw is used alone and the tibia fragments are pressed to the distal end through the joint, the screw needs to resist the bending force. By adding plates, not only the load is distributed to the plate, but also additional screws can be placed at the distal end of the cortical bone, which is stronger at the metaphysis of the humerus. One disadvantage of the support plate is that it requires peeling off the soft tissue during application, [117–120] which may compromise the blood supply [121–124]. Regardless of the specific configuration of the screw, the I-plate and screw fixing are most resistant to axial compression loads. Studies of different plate configurations have found that for bilateral tibial plateau fractures, bilateral (outer and medial) plates can reduce sag by about 50% under axial loading compared to single-sided locking plates. For the medial platform fracture, the medial support plate directly supports the load, and its mechanical properties are significantly better than the outer locking plate. A new option is the proximal humerus staple with multiple interlocking screws. Under the combined action of axial load, bending and rotation, the stability of the nail is equivalent to double steel plate, which is higher than the use of locking steel plate, external fixator or traditional tibial nail. The device can be used in cases without significant proximal (joint) comminution [125, 126].

6.1.3 Fixation of the Humerus

Proximal humerus fractures fixed with a locking plate provided greater stability against torsional loading, but is similar to blade plate structure when bent, as both fixation devices are loaded as tension bands in bending [127]. When comparing different types of blade plate structures, the hardest structure uses an eight-hole, low-contact dynamic compression plate that is shaped into the shape of the blade and secured with a diagonal screw that is triangular to the end of the blade. This arrangement is quite harder than other blade plates or T-plate and screw structures. A potential problem is the screw penetration of the subchondral bone in patients with osteoporosis. Due to the stiffness of the locking plate-screw structure, if there is any "settling" in the fracture site, the locking screw may penetrate the joint. The incidence of intra-articular screw penetration in the proximal humeral locking plate was significantly higher than that of conventional implants [14].

6.1.4 Fixation of Spine

For the treatment of spinal fractures, the goals are to reduce the fracture, protect the neurological function, and accelerate functional recovery [128]. The theory of 3-column model is the basis of the treatment rationale in spinal fractures [129]. Injuries that represent 3-column instability require operative stabilization even if there is no neurological deficit. The attachments of spinal fixation system consist of hooks, wires and screws, which produce different types of holding force [130, 131]. Wires could resist tension, hooks could resist driving force against the bone, while screws could resist forces from all directions except rotation. Therefore, screws are widely used for spinal fixation because of the superiority.

Posterior internal fixation system with pedicle screws has become popular for the treatment of spinal fracture. When applying lumbar spinal fixation, some principles can be considered. Screws are vulnerable to toggling when they are placed into pedicles. The screw tends to toggle about the base of the pedicle because of the cortical bone. In order to reduce toggling, the screw head should be locked to the rod of plate (Fig. 12.15).

Longer fixation could reduce forces acting on the screws because of the effect of the greater lever arm of a longer rod along with more vertebrae. Whereas it is not beneficial for a clinical perspective because the reduced spinal motion. It is also important to add a fusion cage to reduce forces in the fixation. Coupler bars could connect the fixation rods to form an *H* configuration, and prevent the rods from rotating medially or laterally, as shown in (Fig. 12.16). The coupler bars could significantly enhance the torsional and lateral bending stability of the implant.

6.1.5 Fixation for Pelvic Fractures

68.3% of pelvic fractures are unstable fractures, which are serious injuries, and the mortality rate is up to 19%. The stability of the pelvis is mainly related to the integrity of posterior pelvic ring. There are many methods available, including iliosacral

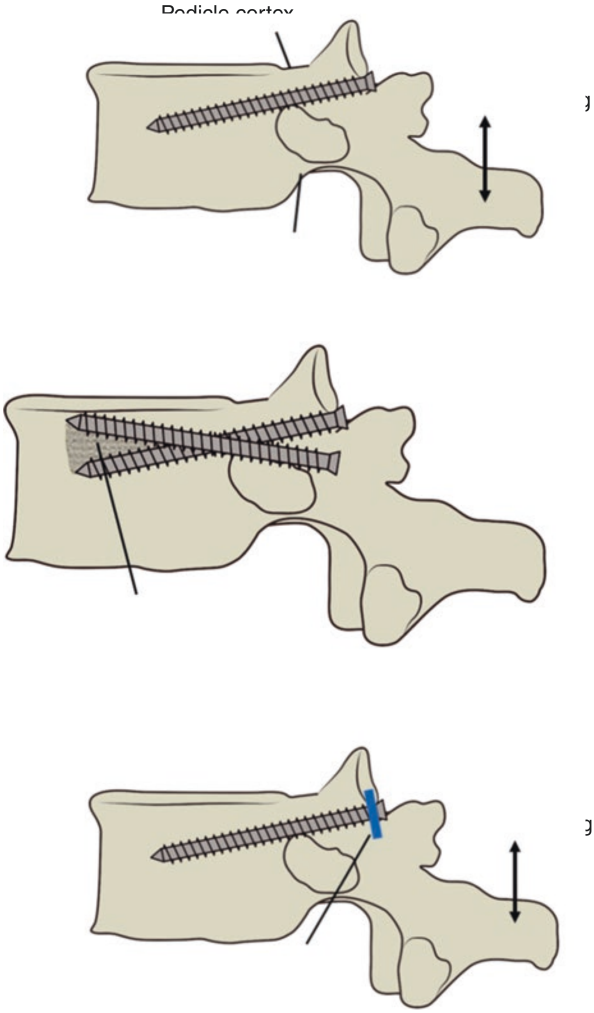

Fig. 12.15 (**a**) The mechanism of toggling of a single pedicle screw subjected to a caudocephalad loading. (**b**) The fulcrum is at the base of the pedicle, the narrowest region with little cancellous bone. The screw toggle compresses the bone within the vertebral body. (**c**) Toggling is reduced if the plate or rod to which the screw connects contacts the vertebra over a wide surface, which prevents it from rotating, whereas the screw head is locked to the plate or rod

(IS) screws, sacral bars, tension band plate (TBP), triangular osteosynthesis, and so on. IS screw fixation is a well-recognized technique for treating the posterior pelvic ring disruption. It is implanted in the supine or prone position and has such merits as short operative time, slight trauma, and minimal invasion. However, it remains a technically demanding procedure, and both doctors and patients are exposed to large amounts of radiation as continuous fluoroscopic or computerized tomography (CT) guidance for appropriate screw insertion. In addition, higher rates of iatrogenic injury is one of the disadvantages, seriously affecting the clinical use of this technology. To avoid these limitations, our team developed a novel minimally invasive adjustable plate (MIAP) (Fig. 12.17). This MIAP is designed according to the anatomy of the pelvic ring and simulated the sacroiliac complex structure of "bridge." It can be better attached the posterior aspect of the sacroiliac joint without bending and adjusted the length of the connecting rod to pressure or separation of

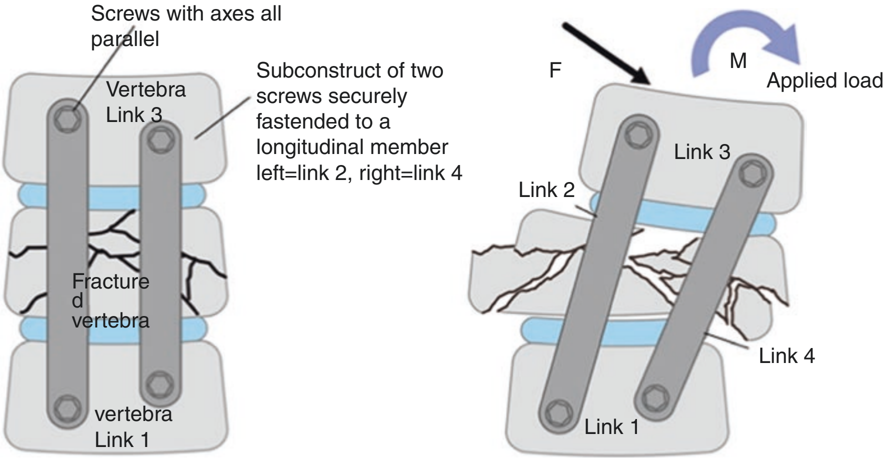

Fig. 12.16 Without a coupler bar between two longitudinal rods (left), they can rotate when a lateral moment or axial torsion is applied (right). A coupler connecting the rods to form an *H* configuration reduces this effect

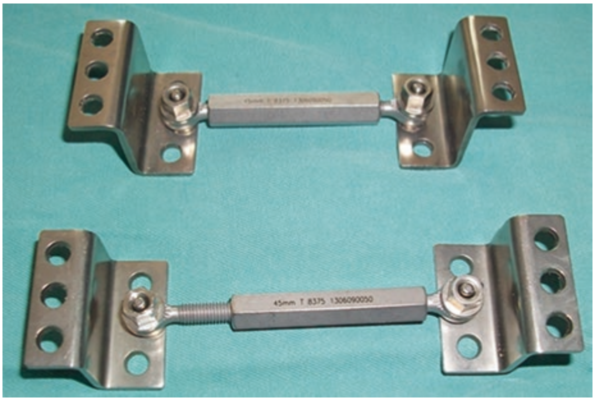

Fig. 12.17 Structure of MIAP for posterior pelvic ring injury

fracture end. Moreover, during the operation, two small incision were made for placing the MIAP, which can effectively reduce the blood loss and shortened the operation time.

6.1.6 Fixation for Tibiofibular Syndesmosis Injuries

Operative fixation and anatomic reconstruction of the distal tibiofibular syndesmosis is important to achieving an optimal outcome. An ideal implant to stabilize the tibiofibular syndesmosis should allow early mobilization for weight-bearing and be strong enough to maintain reduction in the syndesmosis. The screw fixation has been considered the standard management which can provide rigidity of the distal tibiofibular syndesmosis and easily be performed. However, this rigid fixation may reduce the physiologic motion of the syndesmosis and the screw breakage may occur. In recent years, the suture button as a flexible fixation has been applied. The suture button allows physiologic motion in the tibiofibular joint and maintains the reduction of the ankle. However, the suture between buttons can gradually release under daily motion. To avoid these drawbacks, our team developed a novel technique called "bionic fixation" (Fig. 12.18). The screw segment may afford an improved rigidity and stability. The high strength non-absorbable suture located between the

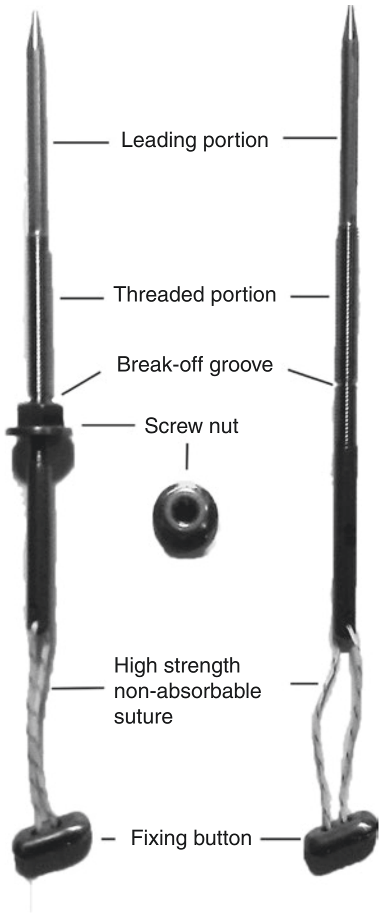

Fig. 12.18 (**1**): Schematic diagrams showing the bionic fixation construct (**2**): Schematic diagrams showing the three techniques for fixation of the tibiofibular syndesmosis. (*a*) A hole was drilled with a 2.8-mm drill bit from posterolateral fibula to anteromedial tibia. (*b*, *c*) A 3.5-mm cortical screw was then inserted through the hole from the fibular side. (*d*) After the cortical screw was removed, the hole was over-drilled with a 4.0-mm drill bit. (*e*) A 3.5-mm main screw was passed through the hole from the fibular side and the fixing button of the screw-tail was tightly attached to the fibula, then the screw nut was installed and adjusted on the tibial side to make the construct tightened properly. (*f*) The exposed leading portion of the screw was broken off. (*g*) The bionic fixation construct was removed. (*h*) The non-absorbable suture of the fixing construct was pulled from the fibular side to the tibial side. (*i*) The suture was threaded into the tibial button, looped, traversed through and securely tied over the fibular button. This process was repeated until there were three independent groups of sutures in the channel

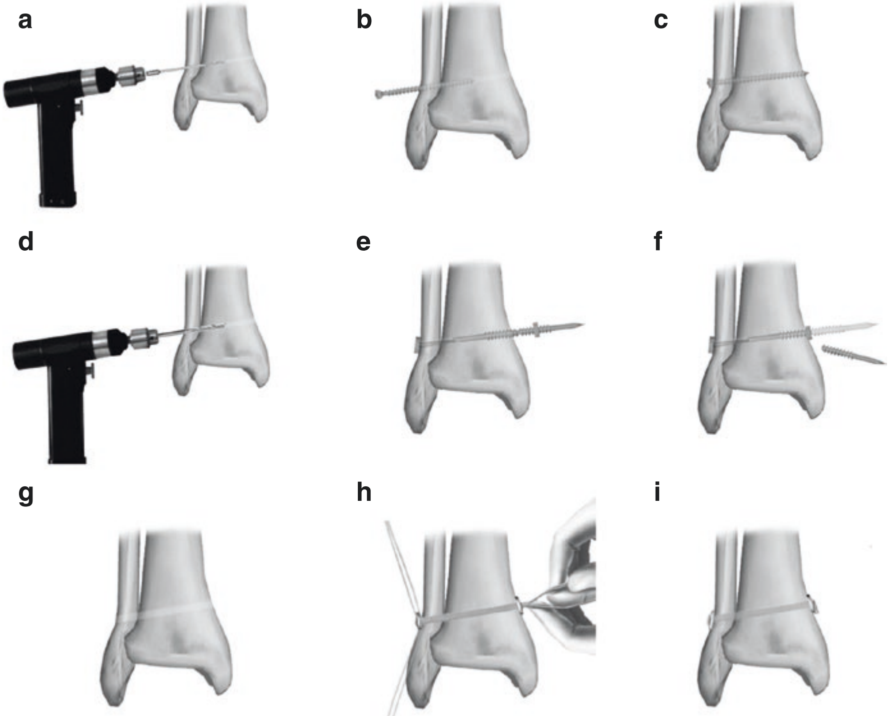

Fig. 12.18 (continued)

tibia and fibula may retain the motion of the syndesmosis to the maximum degree. Comparing with the Endo button fixation, the bionic fixation can provide more stable fixation force and retaining the motion function of syndesmosis. Besides, this technique has a low cost and is easy to perform.

6.1.7 Fixation for Posterior Column Acetabular Fractures

Operative reduction and internal fixation is the standard treatment for unstable posterior column acetabular fractures to allow early mobilization and decrease the risk of posttraumatic arthritis. The conventional methods of fixation involve lag screws and reconstruction plates, or both in combination. Conventional fixation depends on the structure of the acetabulum and the surgical technique because of the specific anatomy of posterior column of acetabulum. The conventional reconstruction plates need to be bended based on the size of the size of the acetabulum. Using screws and two reconstruction plates to obtain better fixation is a potentially serious traumatic complication. Our team designed a W-shaped acetabular angular plate (WAAP) for posterior columns of the acetabulum fractures (Fig. 12.19). This novel fixation includes a W-shaped locking plate and the guide apparatus. Comparing with other

Fig. 12.19 The W-shaped acetabular angular plate. *R* right, *L* left

reconstruction plates, the WAAP provides some advantages. First, the WAAP is anatomically pre-contoured and could match the surface of the posterior acetabulum column properly. Second, the extended fixation range spans from the greater sciatic notch to the rim of the posterior acetabulum. Third, the WAAP has locking holes which can achieve angular stability.

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