Biomechanics of Plate and Screw Constructs for Fracture Fixation

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Introduction

Bone is a unique tissue due to its mechanical properties and the ability of self-repair. Fractures result from mechanical variables, including the magnitude and direction of applied loads as well as the structural properties of the bone, which are determined by its density and physical structure [1–3]. The surgical treatment of fractures gained popularity with the introduction of the principles of fracture care by the Association for the Study of Internal Fixation (ASIF) [4]. Stable fixation of fractures has been a significant advance in fracture management allowing for bone healing while maintaining the function of the joints. In the 1960s, compression of the fracture site through absolute stability was considered the recipe for successful outcomes. Anatomical reduction and

absolute stability, however, required a more extensive surgical exposure to the fracture site, resulting in a second hit to an area where the index trauma had already compromised the vascular supply of bone fragments. In the 1990s, the emphasis changed to the internal "biological" fixation. The goals were to restore the length, alignment, and axis of the bone, utilizing indirect reduction and a bridging construct for non-articular fracture components [5, 6]. Plates and screws may be used to provide either absolute or relative stability to a fracture site, allowing respectively for primary bone healing or callus formation [7]. The same implant constructs may perform different biomechanical functions including neutralization, compression, buttressing, bridging, and tension band. In this chapter, the use of bone-plate constructs will be illustrated under the perspective of their biomechanical function and expected bone healing outcomes.

A Historical Perspective

The first plates designed for the fixation of fractures were introduced more than a century ago by Lane [8]. Those plates had poor metallurgical properties and were soon abandoned due to corrosion [9]. Robert Danis, in 1949, developed a new plate system which allowed for axial compression of the fracture [10]. This was a turning point for fixation of fractures with plates and screws. Anatomical reduction and absolute

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stability at the fracture site promoted union without callus formation. Compression of the fractures with plates became the primary goal of fracture treatment in the late 1950s. This was the mechanical era of internal fixation. Bagby and Janes proposed a plate with oval holes that would allow for axial compression of the fracture, depending on the way the screws would be inserted [11]. In 1965, the Arbeitsgemeinschaft für Osteosynthesefragen (AO) Group developed a tension device that could be coupled to one of the ends of the plate, allowing for axial compression [12]. The dynamic compression plate (DCP) was developed in 1969 [13]. This plate allowed for static axial compression of the fracture site once the screws on one end of the plate were applied eccentrically. Although compression plates proved to be beneficial for the treatment of fractures that had been anatomically reduced, there was some degree of cortical necrosis under the plate. This has been interpreted as a result of periosteal vascular compromise due to plate application [14]. Aiming to reduce the cortical necrosis under the DCP implants, newly designed plates with limited bone contact (LC-DCP) were developed [15]. It has not been proved, however, that this new generation of plates promoted less cortical necrosis.

Absolute stability with plate fixation for diaphyseal fractures created challenges. The most common issue was the lack of radiographic feedback about complete fracture healing, due to the absent callus formation, or even refractures associated with hardware removal [16]. The 1990s was the decade of the biological fixation. Callus formation was a desirable response in diaphyseal fractures, and the use of bridge plate constructs was associated with a smaller incidence of mechanical failures and infection [5, 6]. The history of plate development points out the evolution of concepts in fracture fixation. Anatomical reduction of the bone fragments is still pursued in the articular fractures, but not necessarily in the management of extra-articular ones.

The mechanical fixation of fractures with LC-DCP plates depended mainly on the torque of the screws and the friction generated between the hardware and the underlying bone. If the loading forces to the fracture site were higher than the combination of achieved torque and friction forces, the bone-implant construct would fail. In osteoporotic bones, the torque of the screws is compromised due to the thin cortices and limited thread purchase of the screws. The development of a new generation of implants was needed to overcome this challenge. The mechanical solution was to add threads to the screw heads and the plate holes. Therefore, screw-hole constructs became fixed angle units. Locking plates were designed to be more stable and biologically friendly [17–20]. Loading the fracture site once stabilized with a locking plate converts pull-out forces into compression forces to the screw-hole units. Periarticular locking plates are anatomically pre-contoured, allowing for the insertion of multiple angle stable screws into short epiphyseal segments [21, 22].

The metal alloys used to produce plates is another topic of relevance. Stainless steel has been used for decades due to its corrosion resistance, adequate strength, low cost, and intraoperative malleability allowing for easy contouring of the plates. More recently, implants made out of titanium alloys have gained popularity since their elastic modulus is closer to the bone compared to stainless steel, and they are considered to have better osseointegration properties and potentially lower infection risk. The newest trend is a generation of implants made out of carbon-fiber-reinforced-polyetheretherketone composite that has an elastic modulus even closer to the bone. Carbon-fiber plates are radiolucent and allow for easier intraoperative evaluation of fracture reduction and decreased artifact with computed tomography (CT), or magnetic resonance image (MRI) when evaluating bone healing or associated soft tissue damage. Future studies will determine if this new generation of implants proves to be beneficial in the clinical setting [23].

Plates are versatile implants which may be used for the treatment of the majority of the fractures of the skeleton. The complete understanding of the biomechanical properties of bone-plate constructs is critical for the internal fixation of

fractures. The diversity of biomechanical bone-implant constructs is the reason why many surgeons do not consider plate fracture fixation a technique, but an art.

Biomechanical Functions of a Plate

Plates may result in absolute or relative stability of a fracture site. Absolute stability requires circumferential contact of the fracture site, which is obtained by anatomical reduction. Absolute stability is the principle of fixation pursued in the management of simple fracture patterns (Fig. 12.1), articular fractures, and hypertrophic nonunions. Relative stability is based on an indirect reduction of the fracture site aiming to restore the overall length, rotation, and alignment of the bone. Relative stability is mainly applied to the management of comminuted diaphyseal and metaphyseal fractures, where the anatomical reduction of every fragment will compromise the vascular supply of the fracture site. Biomechanically, bone-plate constructs may have a variety of functions, depending on the goals of the treatment. Compression, buttress, neutralization, and bridging and tension band are the main biomechanical functions of bone-plate
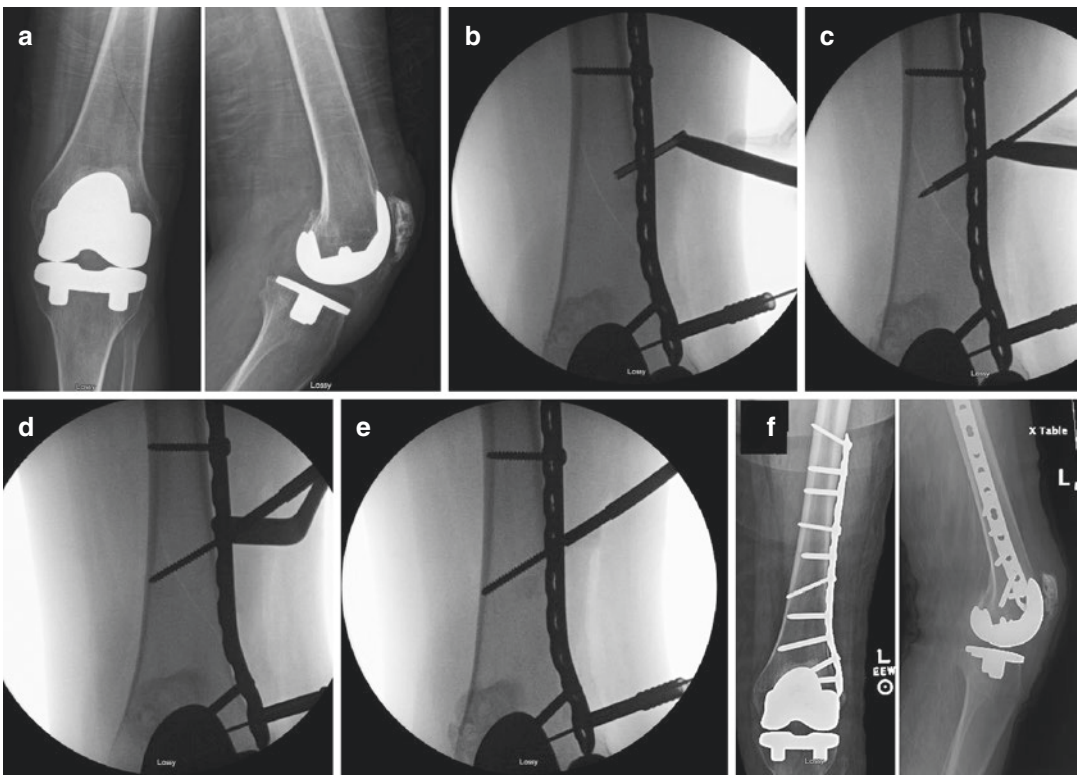

Fig. 12.1 Absolute stability using a lag screw through the plate. (**a**) Anteroposterior and lateral radiographic projections of the left knee revealing a non-displaced oblique simple periprosthetic fracture. (**b–e**) Intraoperative fluoroscopic images illustrating the step-by-step application of a lag screw through the plate. (**b**) After the initial fixation of the plate proximal and distal to the fracture site, the lateral cortex is drilled with a drill bit with a diameter that matches the outer diameter of the lag screw. A sleeve is inserted in this hole for the drilling of the opposite cortex. (**c**) A drill bit with a diameter that matches the core diameter of the lag screw is inserted through the sleeve reaching the opposite cortex. (**d**) The lag screw is inserted in the drill hole, and at this point, it has not engaged yet on the opposite cortex. (**e**) Once the screw engages the opposite cortex, it will compress the fracture site, and the fracture line disappears. (**f**) Immediate postoperative radiographs depicting an anatomical reduction of the fracture and the principle of absolute stability obtained by a plate applied to the tension surface of the bone and in association with a lag screw

constructs. Those functions are accomplished according to the surgical technique, not the specific plate, adopted for the application of the hardware.

Biomechanical Properties of a Plate

Bone segments are subject to bending, torsional, and axial forces. Bending is generated when an external load is applied perpendicularly to the longitudinal axis of the bone. Bending loads will result in tension and compression stresses relative to the cortices of the bone. Torsional loads determine the twisting of the bone by the exertion of forces tending to turn one of its ends about a longitudinal axis while the other end is held fast or turned in the opposite direction. Axial loads are those that are perpendicular to the cross section of the bone (Fig. 12.2).

The biomechanical properties of a bone-plate construct are dependable on the density of the bone, the fracture geometry, the thickness of the plate, and the friction between the plate and the bone. Stiffness is affected by the plate thickness—the thicker the plate, the greater the stiffness and resistance to bending forces. The bending stiffness of a plate is proportional to the third power of its thickness [24].

When a bone-plate construct is loaded, the forces are transmitted through the interface between the hardware and the underlying cortex. The stability of the construct is dependable on frictional and mechanical interlocking forces [25].

Non-locking plates rely on the friction generated between the plate and the bone by the torque of the screws. The higher the density of the bone, the higher the torque of the screws and the frictional forces. The loading forces are transmitted to the interface between the plate and the bone and also through the screw heads (Fig. 12.3).

Locking plates have a different principle. They function as internal fixators. The threaded screw heads engage the threaded holes of the plate establishing an angle stable unit. The loads are mainly transmitted through the implant, and the mechanical interlocking forces determine the stability of the bone-implant construct [26] (Fig. 12.4).

The distribution of the screws within a plate significantly impacts the biomechanics of bone-plate constructs [27]. The working length of a bone-plate construct is the distance between the first two screws on each side of the fracture (Fig. 12.5). The closer the screws are to the fracture site, the stiffer the construct. The screws that see the most load in the bone-plate construct are
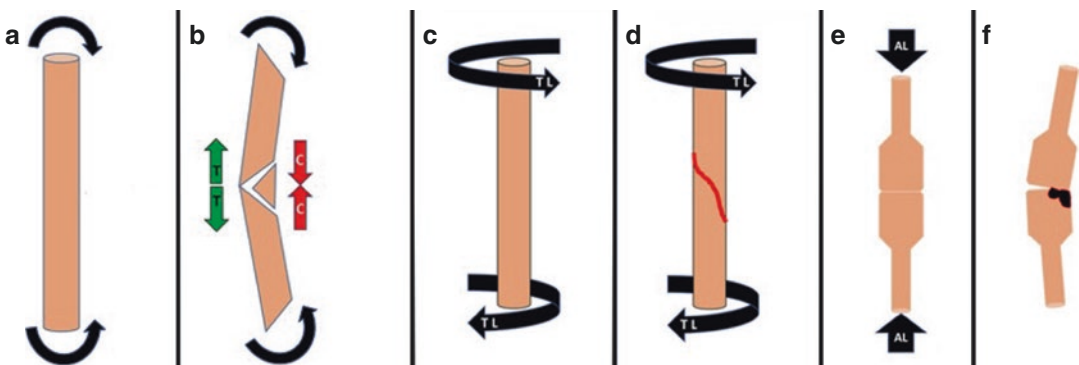

Fig. 12.2 Typical fracture patterns in association with different loading patterns. (**a**) Bending load; (**b**) split wedge fracture as result of bending forces. Observe the side of compression (*C*) and the side of tension (*T*). (**c**) Torsional load; (**d**) helical fracture pattern as a result of torsional forces; (**e**) axial load; (**f**) compression fracture of the joint as a result of an axial load
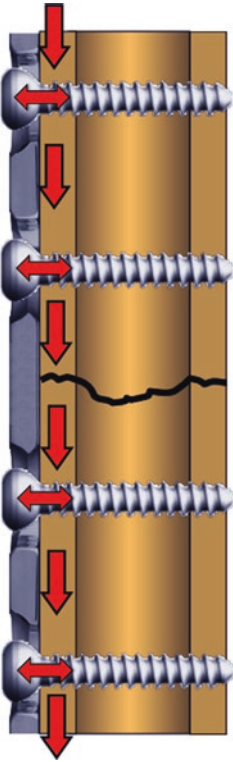
Fig. 12.3 Distribution of load through a non-locking bone-plate construct. The loads are transmitted through the fracture as well as the interface between the hardware and the underlying bone. The higher the density of the bone, the higher the torque of the screws and the friction between the hardware and the cortical bone
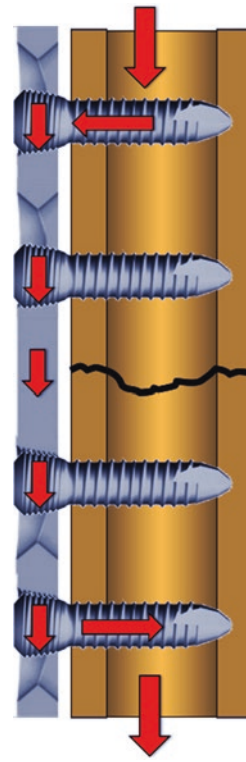
Fig. 12.4 Distribution of the load through a locking bone-plate construct. The loads are mostly transmitted through the plate and the angle stable units established between the threaded screw heads and the threaded plate holes. The density of the bone in this scenario is less relevant as high torque will be achieved at the interface between screw heads and plate holes

the screws closest and furthest from the fracture on each side. These are the screws that are subject to the higher pull-out forces.

The length of the plate and the distribution of the screws within plate holes affect the resistance of the construct to failure [28]. The longer the length of the plate on each side of the fracture and the more spread of the screws in the plate, the higher the resistance of the construct to pull-out forces (Fig. 12.6). The greater the distance between the inner and the outer screw on each side of the plate, the higher the control that the implant has over that bone segment, and the

higher the resistance against pull-out forces. Torsional rigidity is increased by adding a third screw on each side.

Bones may be subjected to eccentric loading. This happens to the femur due to the eccentric position of the femoral head in relationship to the femoral shaft. In cases of bone malunions and nonunions, the convex side of the bone is the one subjected to tension, while the concave side is exposed to compression forces. Plates applied on the tension side of the bone may function as tension band devices, converting tension forces into compression ones (Figs. 12.7, and 12.8).
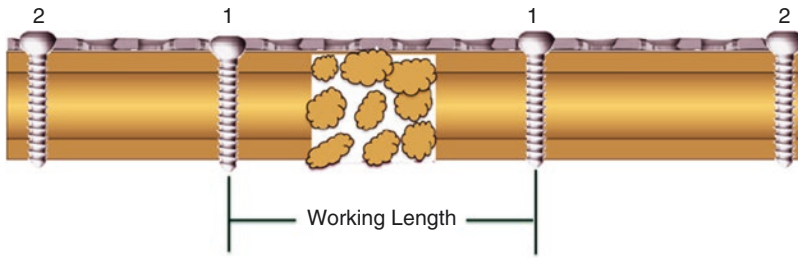
Fig. 12.5 The concept of fracture working length. A bone-plate construct is depicted. The inner screws (1) are those closest on each side of the fracture site. The outer screws (2) are the most distant ones on each side of the fracture site. The working length may be adjusted according to the fracture pattern and affects the flexibility to the fracture site
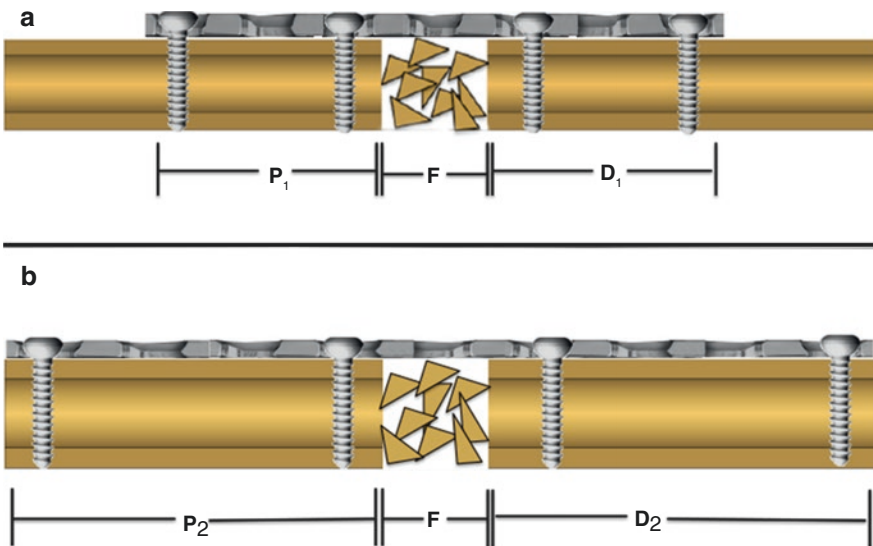

Fig. 12.6 The impact of plate length and screw distribution on bone-plate constructs. (**a**) Comminuted shaft fracture stabilized by a short plate. Observe the relationship between the length of the fracture site (F) and the length of the bone fixed by the plate on the proximal (P1) and on the distal (D1) bone segments. The smaller the ratio between the length of the plate on each side of the fracture and the length of the fracture, the higher the likelihood of a mechanical failure. (**b**) A comminuted fracture fixed by a long spanning plate. The length of each fixed bone segment (P2 and D2) is much higher than the length of the fracture. This allows for better control of each bone segment and increased stability to bending and torsional loads
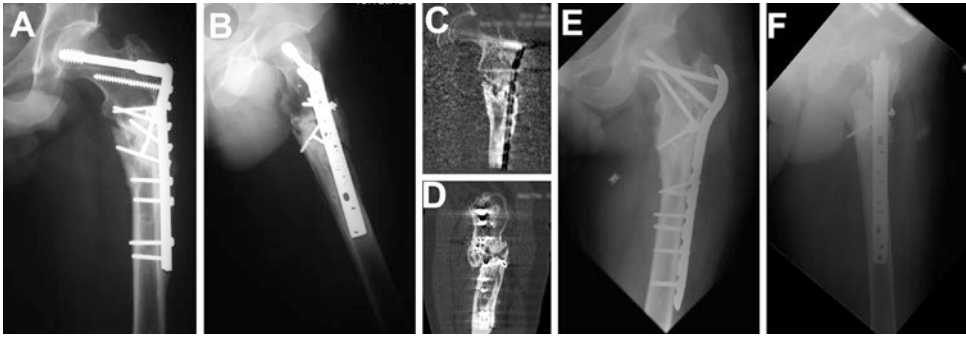
Fig. 12.7 Example of absolute stability with the use of a plate on the tension surface of the bone. (**a**, **b**) Anteroposterior and lateral radiographic projections of the proximal femur revealing a subtrochanteric nonunion, after multiple attempts of surgical treatment. Observe a broken lag screw at the fracture site and the varus angulation. (**c**, **d**) Computed tomography scan confirming the presence of a nonunion at the subtrochanteric level. (**e**, **f**) Final radiographs after surgical treatment of the nonunion and complete bone healing. The strategy was to perform a subtrochanteric closing wedge osteotomy at the level of the nonunion to resect the fibrous tissue associated with an atrophic nonunion. The osteotomy aimed to correct the varus deformity and was fixed with a plate applied to the tension surface of the femur. An articulating tension device was applied to the distal aspect of the plate to promote extra compression, before inserting the distal screws of the plate. A lag screw was applied outside of the plate to reinforce the compression. The sequence of the fixation was osteotomy, reduction, a plate fixed proximally, articulating tension device applied distally, eccentric screws applied distally, a lag screw applied from anterior to posterior, perpendicular to the fracture site. This is an example of multiple strategies to achieve absolute stability at the fracture site
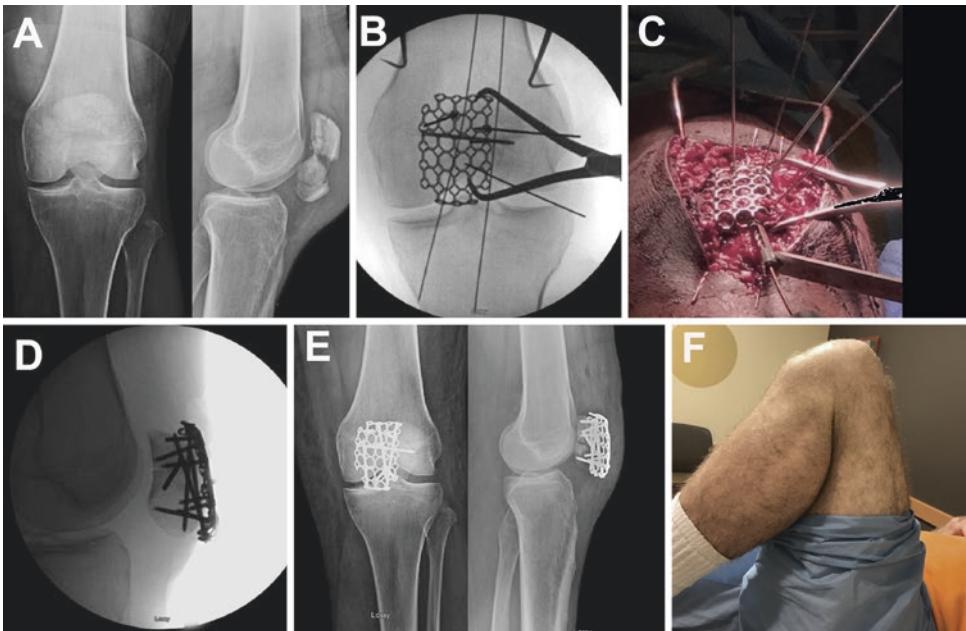

Fig. 12.8 Example of a dynamic tension band plate. (**a**) Radiographs illustrating a comminuted patellar fracture. (**b**) Intraoperative fluoroscopic control of the application of a low-profile locking plate to the anterior surface of the patella. (**c**) Intraoperative image illustrating the clinical application of a plate on the tension surface of the bone. (**d**) Fluoroscopic control after completion of the fixation revealing a satisfactory reduction. (**e**) Immediate postoperative radiographs. The plate is applied to the anterior surface of the patella, and it will convert tension forces into compression forces once the patient mobilizes the knee from extension to flexion. (**f**) Clinical outcomes after 6 months of the fracture fixation. The patient is asymptomatic and has a full range of motion

Conclusions

Plates and screws are essential tools in orthopedic surgery. They may be used with a broad spectrum of biomechanical functions allowing for either absolute stability or relative stability. Bone fixation with plates requires precise preoperative planning and meticulous execution to accomplish with the biomechanical goals of the fixation. The length and thickness of the plate, the distribution of the screws on the plate, the density of the bone, the friction generated between hardware and the underlying bone, the mechanical interlocking of the screws, and the characteristics of the screw heads and the plate holes are all determinants of the biomechanical properties of the bone-plate construct. Although many developments have been achieved in the area of hardware design and technology, the principles of fracture care remain the same, and the outcomes of treatment are directly related to the proper indication and application of the hardware. Subsequent chapters in this section will address the individual functions and biomechanical properties of both locking and non-locking plates.

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