A Review of Biomechanics and Function of Locking Plates in Orthopedic Surgeries: Advantages and Limitations

(Review Article)

Abstract

Fracture fixation using plates has evolved over time. This progress is evident both in the types of devices used—such as the development of locking plates—and in the conceptual approach of orthopedic surgeons, including greater respect for soft tissue and minimal manipulation of it. In summary, locking plates function as internal fixators, where the plate acts as a rod and the screws play the role of Schanz pins. This structure, which functions as a solid, unified unit, is less dependent on bone quality, making it especially useful in the fixation of articular surface fractures and metaphyseal fractures, particularly when using minimally invasive approaches. Numerous advancements in the evolution of these plates—such as Variable Angle Plates and Locking Compression Plates (LCP)—offer surgeons significant intraoperative flexibility. Understanding concepts such as *working length* and *screw density* greatly assists surgeons in screw placement. Although there are no absolute contraindications for the use of locking plates, their use is not recommended for simple diaphyseal fractures. Through ongoing evaluation of design, biomechanics, and surgical concept comprehension, it is anticipated that newer and more advanced locking plates will be developed in the near future through the collaboration of orthopedic surgeons and engineers. This continuous improvement reflects the dynamic nature of orthopedic technology and surgical techniques in fracture management.

Keywords: Bone plate, Biomechanics, Bone screws, Fracture fixation.

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Introduction

The field of orthopedics has witnessed significant advancements across various domains, with fracture fixation being no exception. Parren's innovation in dynamic compression plating represented a seminal advancement in orthopedic traumatology, effectuating enhanced rigid stabilization of osseous discontinuities and expediting postoperative joint kinematics[(1)]. Since the advent of fracture remediation, orthopedic methodologies have witnessed a plethora of paradigm-shifting refinements and biomechanical breakthroughs[(2)].

Although the initial paradigm of rigid fixation provided notable advantages, it frequently encumbered the physiological fracture-healing cascade by perturbing osseous perfusion, thereby precipitating end-fragment resorption and potential non-union[(3)]. During the 1990s, Swiss orthopedic innovator Davos pioneered the locked plating modality, which engendered a paradigm shift in fracture stabilization methodologies[(4)]. The integration of locked plating systems has since become an indispensable facet of contemporary orthopedic trauma management. Nevertheless, an intricate comprehension of the biomechanical principles governing locking plate constructs is paramount for their prudent and efficacious clinical deployment. Locking screws, characterized by their threaded heads, engage with the threads of the plate upon tightening, creating a fixed-angle construct that is less susceptible to loosening or toggling compared to non-locking constructs[(2,5)]. In contrast, screws in an unlocked plate are not rigidly attached to the plate, allowing for potential toggling or loosening through the bone[(6)].

First-generation locking plates necessitated the insertion of locking screws along a fixed axis[7]. The introduction of variable-angle locking plates has afforded surgeons the flexibility to insert screws at various angles, which is particularly advantageous in managing intra-articular and metaphyseal fractures[8]. Locking screws, distinguished by their precisely machined threaded heads, engage synchronously with corresponding threads in the plate, culminating in a fixed-angle construct that exhibits superior resistance against micromotion and mechanical destabilization relative to conventional non-locking constructs[2,5].

Conversely, screws in a non-locking plate lack rigid integration with the plating system, thereby predisposing them to toggle or progressive loosening within the osseous matrix[6]. The initial generation of locking plates necessitated screw insertion along a predetermined axis, whereas the advent of variable-angle locking plates has conferred an enhanced degree of intraoperative versatility, permitting angulated screw placement—an innovation of paramount significance in the stabilization of complex intra-articular and metaphyseal fracture configurations[7,8].

In this study, we will review the materials mentioned regarding the available headings and examine these tools' progress, advantages, and disadvantages.

History of Locking Plates

The transition from non-locking to locking plate constructs in orthopedic surgical practice epitomizes a fundamental conceptual metamorphosis, transcending a mere enhancement in implant engineering to embody a transformative shift in biomechanical stabilization strategies and osseous fixation principles[9]. This shift began with Carl Hansman's introduction of the monocortical fixator in 1886, considered the earliest step in the evolution of locking plate technology[10]. Through successive decades, various orthopedic fixation systems have played instrumental roles in this evolutionary trajectory, including the advent of the Litos stabilization framework in 1974[11], the pioneering Zespol osteosynthesis system in 1982, and the subsequent refinement culminating in the widespread adoption of the modernized locking plate construct in 1995[12].

Patrick Sürer independently developed the Surfix system, which has remained unchanged since its inception[13,14]. In parallel, the AO Foundation catalyzed advancements in the orthopedic stabilization paradigm through successive technical innovations—exemplified by the introduction of the point contact fixator (PC-fix) in 2005 and the Less Invasive Stabilization System (LISS) in 2001[15]. Collectively, these innovations culminated in the emergence of a novel generation of locking compression plates (LCPs), devices that have been subjected to extensive iterative refinements to optimize biomechanical stability and enhance fracture fixation efficacy[16].

Introduced in 1998, the Schuli locking nut system signified a paradigm shift in orthopedic implant technology by enabling the secure integration of conventional screws within standard osteosynthesis plates. This seminal innovation catalyzed subsequent refinements in plate architecture, including the optimization of locking hole geometries, the enhancement of locking mechanisms, and the evolution of screw configurations—all of which have markedly augmented the biomechanical stability of fracture fixation constructs[17].

The evolution of locking plates has been driven by the need to balance mechanical stability with biological preservation. Early rigid fixation methods, while providing stability, often compromised the biological environment necessary for optimal fracture healing. This led to issues such as impaired blood supply, bone resorption, and fracture non-union. The introduction of locking plates addressed these concerns by allowing for stable fixation without excessively disrupting the biological processes of bone healing[18].

Locking screws, defined by their threaded head design, interlock with the plate's internal threads during insertion, thereby establishing a fixed-angle construct that exhibits diminished susceptibility to toggling or mechanical loosening relative to non-locking configurations[5,19]. Conversely, screws in non-locking plates do not achieve a rigid connection with the plate, predisposing them to micromotion and potential loosening within the osseous substrate[19].

The initial generation of locking plates required the insertion of screws along a predetermined, fixed axis. In contrast, the advent of variable-angle locking plate systems has endowed surgeons with enhanced flexibility, allowing for multi-directional screw placement—a critical advantage in the stabilization of complex intra-articular and metaphyseal fractures.

The continuous evolution of locking mechanisms and hole designs enhance the surgeon's experience, improves biomechanical stability, and ensures

superior outcomes with reduced failure rates under physiological loading conditions[20,21]. Innovations such as polyaxial locking systems, which allow for multi-directional screw placement, and hybrid plates, which combine locking and non-locking screw options, have further expanded the versatility and applicability of locking plate technology[22].
As we delve into the various topics within this field, we will encounter numerous examples of these evolutionary advancements, each contributing to the improved efficacy and reliability of orthopedic fracture fixation techniques. The ongoing research and development in this area continue to push the boundaries of what is possible, aiming to provide better patient outcomes and more efficient surgical procedures.

Biomechanical Paradigms of Locking Osteosynthesis Plates

Locking plates, engineered from biocompatible alloys such as stainless steel or titanium alloy, have become pivotal in modern internal fixation paradigms due to their optimal elastic modulus characteristics. These metallic constructs function as intrinsic orthopedic splints, effectively mitigating the elastic deformation of osseous structures under physiological loads. To augment implant flexibility and minimize stress shielding, the dimensions of these devices are meticulously reduced, thereby enhancing their conformity to native anatomical contours. Consequently, these low-profile constructs—typically fabricated from a malleable metal like titanium—are associated with superior clinical outcomes in fracture stabilization and osteosynthesis[12,23]. Although immunologically mediated hypersensitivity reactions are uncommon following implantation, they are typically associated with elevated local concentrations of galvanic, fretting, and crevice corrosion by-products within periprosthetic tissues. The biochemical corrosion of metallic implants in vivo not only precipitates an adverse inflammatory response but also compromises the implant's structural integrity by diminishing its fatigue resistance over time[24]. The locked internal fixator plate functions analogously to an external fixation system, wherein the threaded screw head emulates the biomechanical role of a Schanz pin by engaging securely within the plate's threaded aperture. This locked construct establishes a rigid, fixed-angle configuration, conferring both axial and angular stability. Under physiological loading conditions, mechanical forces are effectively redistributed across the osteosynthesis construct, facilitating load transmission between osseous segments via the locked screws, thereby enhancing fracture stabilization and mitigating micromotion at the bone-implant interface[2]. Unlike conventional plates, locking screws do not compress the plate onto the bone. Instead, they primarily endure bending loads rather than tensile forces. As a result, the core diameter of locking screws is inherently more significant than that of conventional cortical or cancellous bone screws (Figure 1), enhancing their mechanical strength and resistance to deformation. The integration of the locking plate and screws establishes a rigid monoblock construct, which exhibits reduced reliance on bone mineral density and cortical anchorage. This biomechanical independence from osseous quality renders locked fixation particularly advantageous in osteoporotic bone and anatomically constrained regions, optimizing stability and load distribution[25,26].

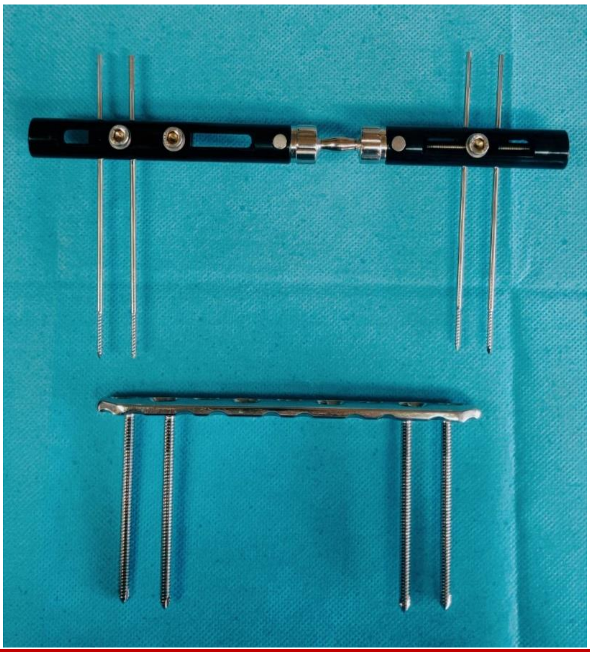

Figure 1: The locking plate acts as an external, internal fixator that has a plate-like (A), and the screws act like a Schanz pin (B).

The Locked Internal Fixator Plate (LIFP) functions as an implanted analog of an external fixator, embodying an innovative, biocompatible approach to internal osteosynthesis. This construct optimizes the physiological fracture repair process by facilitating robust callus formation via secondary bone healing mechanisms. The widely spaced locking screws, biomechanically analogous to external fixator pins, in conjunction with the plate—serving as a unifying stabilization bar—are strategically positioned in close proximity to the mechanical axis of the bone. This anatomical alignment enhances construct rigidity and load distribution, conferring superior stability compared to traditional monoplane external fixation systems[2,27]. Overall, a functional locking plate system causes minimal vascular damage compared to intramedullary nailing or conventional plating, making them a superior choice for fracture fixation (Figure 2).

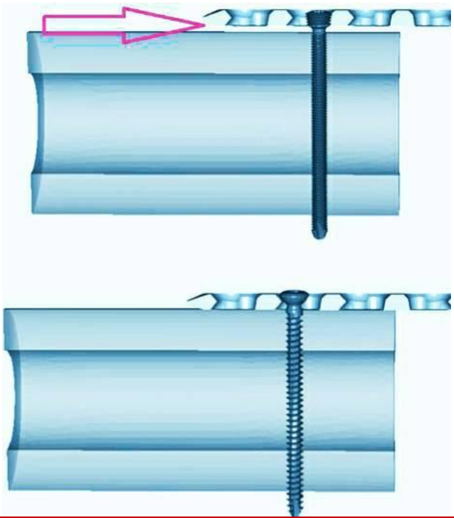

Figure 2: The distance between the locking plate and the bone and the diameter of the locking screw compared to the compression plate and screw[28], Courtesy of AO foundation

The mechanical properties of locked screw-plate assemblies in orthopedic applications reveal critical interactions between various forces during fracture fixation. A locked plate functions as a rigid fixed-angle osteosynthetic construct, precisely preserving the axial congruency and spatial orientation of the screws relative to the plate[28]. This configuration is pivotal in augmenting the stability of the screw-plate-bone assembly, establishing a stable single-beam structure[29]. The mechanism effectively transduces shear forces into compressive forces at the screw-bone interface, which is advantageous given that cortical bone exhibits superior strength under compressive loads compared to shear loads. This intrinsic characteristic of locked plates ensures enhanced angular and axial stability, significantly improving fixation[30]. The structural rigidity of external fixators is contingent upon multiple biomechanical parameters, including the material composition, length, and diameter of the Schanz pin, in addition to the dimensional attributes of the fixator bar. This intricate interplay closely parallels the mechanical relationship observed between a locked screw and its corresponding plate, wherein the screw's geometric and material properties, along with the plate's structural configuration, collectively dictate the overall stability and load-bearing capacity of the osteosynthetic construct[31]. Notably, the reduced size of screws in locked plate constructs, which are significantly shorter than those utilized in external fixators, enhances rigidity; Consequently, fracture stability is inherently dependent on the biomechanical properties of the plate and the magnitude of the applied physiological loads[32]. The intrinsic locking mechanism obviates the necessity for axial preloading of the screw, thereby maintaining the spatial relationship between the plate and the osseous substrate while ensuring robust fixation. This advanced stabilization strategy effectively negates the need for meticulous plate contouring to conform precisely to bone morphology, as the locked construct provides secure anchorage irrespective of minor anatomical discrepancies[33] (Figure 3).

Figure 3: The conventional screw structure fails when the screws lose their hold in the bone and are pulled out of the bone; the screws in this structure fail sequentially. locking structure as an integrated system, failure occurs when all the screws are rem

Locked screws establish a rigid angular-stable interface with the osteosynthesis plate, analogous to the biomechanical integrity observed in angled-blade plate constructs. The elimination of screw toggling within locked plating systems significantly mitigates the risk of fracture reduction loss, a prevalent complication associated with conventional plate constructs. Pullout failure mechanisms primarily arise from shearing forces; when pullout forces surpass pullout resistance, the screw extracts a cylindrical bone segment proportional to its diameter. Divergent screw orientations significantly bolster pullout strength, further augmented by the fixed-angle stable construct inherent in locking plate systems, which offers substantial resistance to pullout forces(12).

The fixation strength of orthopedic screws is directly dependent on cortical bone thickness, with greater cortex thickness correlating to an extended working length. Working length is determined by the total number of engaged threads, irrespective of whether the screw is anchored in a single or both cortices. The optimal working length is established when three to four threads thoroughly engage within the osseous cortical layer(34). Universally, bicortical screw placement ensures the maximal working length of the implant (Figure 4). LIFP enables uni-cortical screw application without compromising mechanical strength, provided robust anchorage is achieved within a normal-thickness cortex. Despite advantages such as simplified placement and reduced inventory requirements, uni-cortical screws exhibit a diminished holding capacity compared to bicortical screws (Less than half)(35).

Hazards associated with unicortical insertion include potential compromise of bone thread integrity if the screw tip contacts the far cortex prematurely. Measuring the length of screw post-drilling is recommended to ensure safe insertion. Bicortical fixation is essential in scenarios involving osteoporotic bones, thin cortices, anticipated high torsional forces, and specific complications during screw placement(19,36). Although locking plate constructs offer superior fixation in osteoporotic bone, they inherently restrict interfragmentary micromotion, thereby impeding callus formation. To counteract this limitation, bridging plating techniques have been developed to facilitate secondary bone healing; however, the consistency of callus formation with these constructs remains variable; In this treatment modality, it is paramount to recognize that during bridge plating across substantial gaps, screws must be positioned proximally to the fracture line(37). This proximity mitigates the stress on the plate, thereby decreasing the likelihood of plate fracture. Conversely, in cases of bridge plating over smaller gaps, screws should be strategically placed further apart, leaving two to three screw holes unoccupied. This spacing facilitates the even distribution of applied pressure along the entire plate length, lowering the risk of treatment failure(12,38).

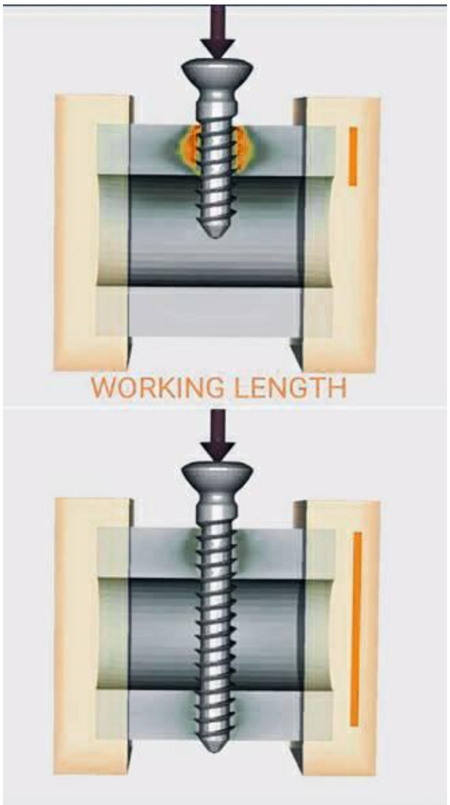

Figure 4: Unicortical& bicortical screws & working length[28], Courtesy of AO foundation

Far cortex locking (FCL) screws, including innovations like Zimmer's MotionLoc and AO Foundation's dynamic locking screws, provide enhanced parallel interfragmentary movement through flexible shafts devoid of threads. FCL constructs significantly reduce stiffness by promoting micromotion at fracture sites and uniformly distributing loads, mitigating stress risers commonly observed in traditional locked plate systems. FCL screws also exhibit biphasic stiffness, initially allowing for more significant interfragmentary motion during early healing phases Plate length and working length are essential factors in optimizing fixation. Achieving relative stability is paramount for managing comminuted fractures, and

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minimally invasive plate osteosynthesis (MIPO) is the preferred method[39]. This approach requires a plate two to three times the length of the fracture. Conversely, the plate length should be maintained at 8 to 10 times the fracture length for simple transverse fractures that are short and treated with compression[16]. The biomechanical stiffness of an osteosynthetic construct is significantly reduced when a shorter fixation plate is employed instead of a more extended plate with an equivalent number of cortical anchorage points. Consequently, the utilization of elongated plating systems is advocated to enhance mechanical stability, optimize stress distribution, and mitigate the risk of implant failure, particularly in cases of comminuted or osteoporotic fractures[40].

The aggregate biomechanical rigidity and plastic deformation response of the construct is modulated by the spatial configuration and density of screw fixation, with a minimum of three screws per osseous fragment advised to optimize structural stability. Screw density is defined as the quotient of the number of fixation screws deployed and the total number of screw perforations within the osteosynthesis plate. These density indices should be maintained at 0.4–0.3 for simple fractures and 0.5–0.4 for comminuted fractures, respectively[41].

The choice of implant material, whether stainless steel or titanium, also affects stress distribution, particularly in small fracture gaps. Stainless steel implants experience more stress than titanium implants when the fracture gap is less than 1 mm. Nevertheless, no discernible disparity is observed when the inter-fragmentary gap is more pronounced[42].

The working length of an osteosynthesis plate is defined as the distance between the fixation screws positioned immediately adjacent to the fracture on either side. This length determines fracture elasticity; shorter lengths increase the likelihood of plate failure[40]. Proper screw positioning is vital for maximizing axial and torsional stiffness, ensuring adequate stabilization while mitigating the risk of mechanical failure during healing. For optimal axial stability, at least three screws should be used on each side of the fracture, with two of these screws placed in the farthest and closest holes relative to the fracture site. Screws should be used at both ends of the plate to involve the entire plate in stabilizing the fracture. The tertiary screw modulates fracture stiffness, with rigidity diminishing as the screw is positioned further from the fracture interface. For optimal torsional stability, a minimum of four screws per osseous fragment is requisite, with fixation devices deployed in the proximal and distal plate perforations relative to the fracture line[19,41](Figure 5).

Figure 5: The location of the screws relative to the location of the fracture in a radial shaft fx

Vascularization and ensuring construct stability. To optimize these factors, the plate-to-bone distance should be maintained at or below 2 mm[42].

Locking screws, a specialized variant of bone screws, are characterized by threads on the undersurface or within the countersink of the head. When tightened, these threads engage with corresponding threads in the plate aperture, thereby imparting both axial and angular stability to the fixation construct. The constituent components of a locking screw include the head, shaft, thread, and tip[12,43].

The head acts as an interface for the screwdriver, with engagement points designed to facilitate secure manipulation during surgical procedures. The undersurface may be conical or hemispherical, providing varied fixation characteristics. The presence of threads on the countersink is a crucial differentiator from conventional screws. Below the head, the runout represents the transition where the thread begins, subjecting this region to high-stress concentrations during insertion[44].

The thread encircles the core, which is vital for providing structural support. The core diameter refers to the minimal diameter across the thread base, correlating with the screw's tensile strength. In contrast, the pitch indicates the distance between adjacent threads, while the lead defines the distance the screw advances per complete turn—equivalent to the pitch for single-threaded screws(45). The primary diameter, or thread diameter, is the maximum measurement across the screw's threads, influencing its pullout strength; larger diameters correspond to more excellent resistance to pullout.

The head's design includes a recess that can be hexagonal, hexalobular, or of other geometric configurations. The hexagonal design is predominant due to its secure engagement with the screwdriver. The hexalobular socket offers improved resistance to stripping and facilitates enhanced torque transfer through its star-shaped configuration, providing greater stability during insertion(19).

The countersink's conical design promotes superior fixation by ensuring effective force distribution between the screw head and the threaded holes in the plate. Screws with dual threads on the undersurface exhibit enhanced locking capabilities, while those with a single thread may risk slippage and unlocking. The pitch on the head aligns with that on the shaft, ensuring compatibility, while the conical design minimizes the required insertional torque(12).

The shaft includes the runout, a region critical for lag screw functionality, which should be of adequate length to promote optimal compression. This transitional area is susceptible to failure under torsional loads, especially when the screw is subjected to spiral insertion(46).

The screw thread, functioning as an inclined plane, utilizes a "V" design specific to locking screws, contrasting with the buttress thread employed in traditional cortical screws. The advancements in screw design, such as the Bone Screw Fastener (BSF), present improvements in performance metrics compared to earlier iterations(2). Core diameter, or root diameter, represents the minimal transverse dimension at the base of the threads and directly impacts the screw's vulnerability to shearing forces. The root area is critical for structural integrity, with torsional strength being proportional to the cube of the root diameter. More extensive fragment-locked screws exhibit increased core diameters compared to conventional screws(47). Pitch and lead are significant for determining screw advancement rates. Locking screws have coarser pitches than cortical screws, promoting rapid insertion. The tip design varies between self-tapping and self-drilling self-tapping configurations. Self-tapping screws are equipped with a thread-cutting flute engineered for the generation of osseous threads, whereas self-drilling self-tapping screws integrate both drilling and tapping functionalities, thereby augmenting anchorage in anatomically challenging regions(19) (Figure 6).

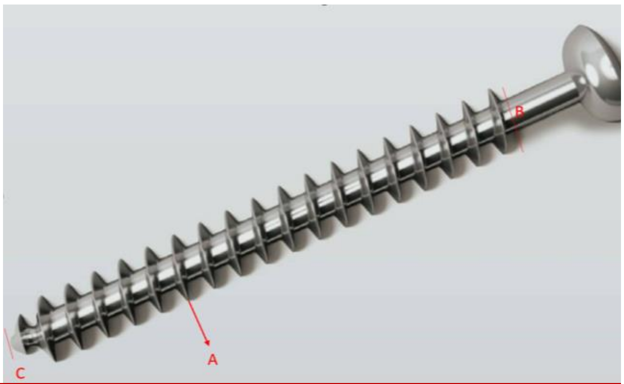

Figure 6: screw structure, A: thread, B: outer diameter C: inner diameter

Locking plates incorporate round, threaded holes designed to accommodate locking screws, establishing fixed-angle constructs that prevent screw toggling(48). Combi-holes allow for the integration of both self-compression and locking mechanisms, enhancing the versatility of the fixation system(49). Locking mechanisms can be categorized into fixed-angle and variable-angle systems, with the former requiring unidirectional screw insertion for optimal locking and the latter allowing for greater angulation flexibility(38). Variable-angle locking plates facilitate screw angulation up to 30 degrees, thereby accommodating a broad spectrum of fracture configurations. Innovations such as elastically suspended locking holes enhance stability and allow for limited axial motion, facilitating healing while maintaining fixation integrity. A novel locking plate termed the Elastically Suspended Locking Hole, has been engineered to facilitate regulated axial micromotion ranging from 1.5 to 2 mm at the fracture interface. This innovation maintains stability against bending and rotational forces, enhancing mechanical integrity and biological healing. The structural refinement of the inferior aspect of locking plates attenuates avascularity and optimizes osseous perfusion, thereby ameliorating early post-fixation osteopenia and extending the fatigue lifespan of the implant through homogeneous dissipation of biomechanical stressors.[2,19,38] (Figure 7,8).

Indications for Locking Plating

A LIFP serves as a multifaceted orthopedic implant designed to facilitate fracture stabilization by providing either absolute or relative stability, or a hybrid of both, contingent upon the specific clinical context. While the primary indications for LIFP are well-established, they can also be utilized in various other situations.

Primary Indications:

Osteoporotic Fractures: Ideal for bones with reduced density and strength. Periprosthetic Fractures: Fractures occurring around prosthetic implants (Figure 9). Periarticular Fractures with Short Metaphyseal Segments: Fractures near joints with limited metaphyseal bone.

Figure 7: Attention should be given to the angle of the variable cortical screws, their central diameters, as well as the threads present in the structure of the hole and the screw head

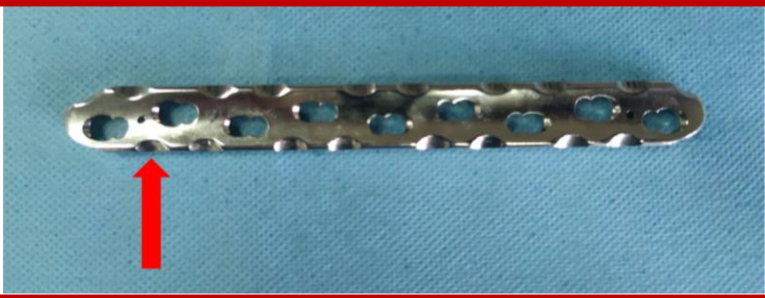

Figure 8: low contact plate

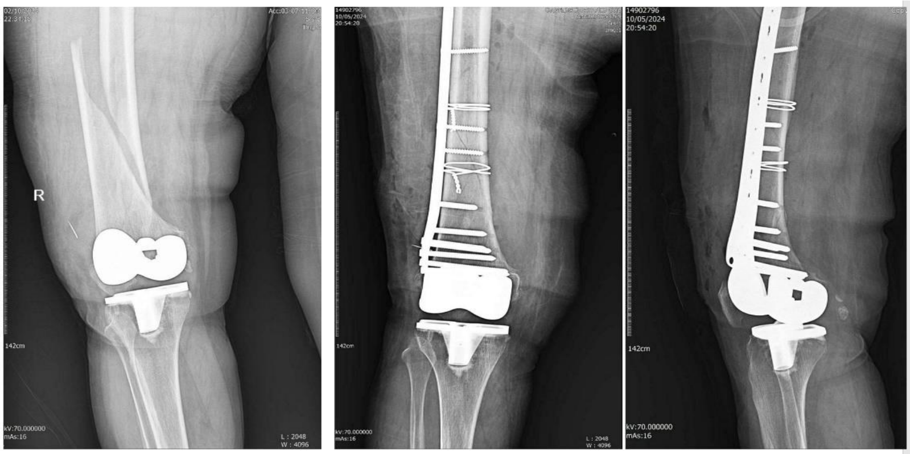

Figure 9: locking plate in a periprosthetic fx

Biological Fixation: Including MIPO for multifragmentary fractures.

Additional Clinical Indications

- Terminal Segment Fractures: Necessitate rigid structural reinforcement.
- Constructs Susceptible to Varus or Valgus Collapse: Mitigates the risk of angular deformities.
- Fractures in Osteoporotic Bone: Ensures augmented fixation stability in compromised bone architecture.
- Broken Screws from Previous Surgery: Offering a stable alternative.
- Deformities That Should Not Be Corrected: Maintaining the existing bone structure.

Utilization Modalities

The LIFP grants the surgeon the flexibility to select the most suitable stabilization approach for each clinical scenario, whether through compression mode, locked mode, or a hybrid application.

Indications for LIFP in Compression Mode

- Uncomplicated Diaphyseal and Metaphyseal Fractures: Mandates precise anatomic reduction to restore structural integrity.
- Intra-articular Fractures: Functions as a buttress plate to reinforce articular congruity.
- Delayed Union or Non-union: Facilitates osteogenic stimulation and promotes fracture consolidation.
- Closed-Wedge Osteotomies: Enables controlled bone realignment for corrective osteotomies.
- Bone Fragments with Compromised Vascularity: Maintains perfusion and mitigates ischemic complications.

Indications for LIFP in Splinting Mode

- Multifragmentary Diaphyseal and Metaphyseal Fractures: Confers relative stability while preserving biological integrity.
- Fractures in Anatomically Challenging Zones: Utilized when intramedullary nailing is contraindicated due to anatomical or pathological constraints.
- Open-Wedge Osteotomies: Particularly indicated in proximal tibial corrections to maintain structural alignment.
- Periprosthetic Fractures: Ensures stabilization around pre-existing orthopedic implants.
- Coexistence of Other Implants: Provides supplementary reinforcement when additional mechanical support is warranted.
- Secondary Fractures and Post-Nailing Instability: Serves as a salvage procedure to restore structural stability following complications from intramedullary fixation.
- Delayed Conversion from External Fixation to Internal Fixation: Facilitates the transition to definitive internal stabilization, ensuring sustained osseous integrity.
- Oncologic Reconstruction Post-Tumor Resection: Functions as a load-bearing scaffold to maintain skeletal continuity
- LIFP can be effectively utilized by combining both compression and splinting methods in the following scenarios:
- Segmental Fractures with Two Different Fracture Patterns: This includes cases where one fracture is simple and the other is multifragmentary. The combination approach allows for precise anatomic reduction of the simple fracture while providing relative stability to the multifragmentary segment.
- Intra-articular Fractures with a Multifragmented Extension into the Diaphysis: These complex fractures benefit from the dual approach, ensuring stable fixation of the intra-articular component and adequate support for the multifragmented diaphyseal extension.

Contraindications of Locked Plating

Although there are no absolute contraindications for the utilization of locked plates, specific clinical scenarios render their application superfluous:

- Uncomplicated Diaphyseal Fractures in Optimal Bone Quality: Conventional plating techniques are sufficient, as the intrinsic biomechanical integrity of the bone supports uneventful healing without necessitating the augmented stability conferred by locked plating.

- Pelvic and Acetabular Fractures: These fractures often require specialized fixation techniques better suited to the pelvis and acetabulum's unique anatomical and biomechanical demands.
- Partial Articular Fractures Requiring Buttress Plating: In these instances, buttress plates are more appropriate to support the articular surface and maintain joint congruity[19].

Disadvantages of LIFP

The screw length must be meticulously determined prior to insertion. Unlike non-locking screws, locked screws do not provide tactile feedback regarding their hold in the bone, which is crucial for assessing the purchase of the screw.

Challenges with Locked Screws and Plates

Reduction of Fragments: Locked screws do not facilitate the reduction of bone fragments during insertion, unlike conventional screws; This can lead to higher rates of malalignment, especially when using the MIPO technique.

First-Generation Locked Plates

Fracture Reduction Limitation: These constructs primarily maintain, rather than achieve, fracture reduction, except in specific pre-contoured designs.
Soft Tissue Irritation: Prominent hardware may induce pain and irritation in subcutaneous regions, such as the distal tibia and proximal medial tibia, particularly if the plate-screw interface lacks anatomical conformity.
Delayed Union or Non-Union: The rigidity of locked constructs may precipitate stress shielding, leading to osteoclastic resorption of fracture ends and impairing healing dynamics. Furthermore, as these implants are non-load-sharing, cyclic loading may predispose the plate to fatigue failure and subsequent loss of fixation.

Limitations of First-Generation Locked Plates

Angle Adjustment Constraint: The fixed-angle design precludes intraoperative modification of screw trajectories while maintaining screw locking, rendering certain holes unsuitable for lag screws crucial for articular reduction, complex fracture morphologies, anatomic variability, or pre-existing arthroplasty components. Contouring Challenges: Attempts to contour these plates may distort locking holes, compromising screw engagement and biomechanical stability, necessitating meticulous surgical execution.

Removal Concerns

Technical Difficulty in Extraction: Explanation of locked plates can be arduous, particularly in cases of screw osseointegration or excessive torque application, leading to cold welding. Employing torque-limiting screwdrivers can mitigate such complications and facilitate hardware retrieval[50].

Challenges and prospects

With the widespread adoption of locking plate technology, emerging literature suggests that stainless steel constructs are associated with elevated Non-union rates, particularly in distal femoral fractures. These locking plates, designed to foster secondary bone healing over primary bone healing, were often too rigid to facilitate the necessary motion for callus formation. Innovative methods for dynamizing stainless steel locking plates have surfaced, aiming to provide essential interfragmentary motion. Biomechanical research about far cortical locking screws shows that this device can reduce construct stiffness by employing a smaller-diameter screw shaft and over-drilling the near cortex. This allows controlled motion within the near cortical hole, dynamizing the fracture while maintaining structural integrity. Additionally, active plates have been introduced, incorporating silicone-suspended locking holes as a dynamization strategy. This paper does not directly compare newer locking strategies linked with second-generation locking plates, necessitating further research to evaluate long-term outcomes. Preliminary findings indicate the rationale behind these implants' technological advancements may be flawed. Experienced surgeons can still utilize first-generation locking plates in scenarios suitable for second—or third-generation plates, with no significant difference in outcomes. However, a learning curve may exist for surgeons unfamiliar with these plates, though this point might be moot due to the potential unavailability of this kind of plate.
Future research should also consider the cost-effectiveness of newer implant designs, assessing whether the clinical benefits experienced by patients justify the added costs.

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