A Nonlocking End Screw Can Decrease Fracture Risk Caused by Locked Plating in the Osteoporotic Diaphysis

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Background: Locking plates transmit load through fixed-angle locking screws instead of relying on plate-to-bone compression. Therefore, locking screws may induce higher stress at the screw-bone interface than that seen with conventional nonlocked plating. This study investigated whether locked plating in osteoporotic diaphyseal bone causes a greater periprosthetic fracture risk than conventional plating because of stress concentrations at the plate end. It further investigated the effect of replacing the locked end screw with a conventional screw on the strength of the fixation construct.

Methods: Three different bridge-plate constructs were applied to a validated surrogate of the osteoporotic femoral diaphysis. Constructs were tested dynamically to failure in bending, torsion, and axial loading to determine failure loads and failure modes. A locked plating construct was compared with a nonlocked conventional plating construct. Subsequently, the outermost locking screw in locked plating constructs was replaced with a conventional screw to reduce stress concentrations at the plate end.

Results: Compared with the conventional plating construct, the locked plating construct was 22% weaker in bending ($p = 0.013$), comparably strong in torsion ($p = 0.05$), and 15% stronger in axial compression ($p = 0.017$). Substituting the locked end screw with a conventional screw increased the construct strength by 40% in bending ($p = 0.001$) but had no significant effect on construct strength under torsion ($p = 0.22$) and compressive loading ($p = 0.53$) compared with the locked plating construct. Under bending, all constructs failed by periprosthetic fracture.

Conclusions: Under bending loads, the focused load transfer of locking plates through fixed-angle screws can increase the periprosthetic fracture risk in the osteoporotic diaphysis compared with conventional plates. Replacing the outermost locking screw with a conventional screw reduced the stress concentration at the plate end and significantly increased the bending strength of the plating construct compared with an all-locked construct ($p = 0.001$).

Clinical Relevance: For bridge-plating in the osteoporotic diaphysis, the addition of a conventional end screw to a locked plating construct can enhance the bending strength of the fixation construct without compromising construct strength in torsion or axial compression.

Periprosthetic fracture at the plate end is a well-recognized complication associated with plate fixation in osteoporotic bone. For compression plating, this complication has been described in classic orthopaedic texts[1,2] and in the plating literature[3-6] with an incident rate of 1% to 3%. While infrequent, these periprosthetic fractures may require revision surgery and represent a serious clinical concern[3].

Periprosthetic fractures have been attributed to the stress-riser effect at the plate end, caused by the difference in structural stiffness between the normal bone and the plated bone segment[4]. This difference in structural stiffness is especially pronounced in the presence of osteoporotic bone, in which stress concentrations at the peripheral bone-screw junction contribute to an elevated periprosthetic fracture risk. The recent

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introduction of locking plates holds the promise of better fixation in osteoporotic bone and, presumably, lower rates of fixation failure for these challenging fractures[7-9]. However, the mechanics of locking plates are fundamentally different from those of conventional plates, and their use requires a revised understanding of the bone-plate construct stability[10-12]. Conventional plates rely on compression of the plate to the bone, allowing for load transfer through the screws as well as through the plate-to-bone interface. Conversely, locked plates engage with the bone through fixed-angle screws in the absence of plate compression onto the bone. Load is transmitted from the cortex to the plate exclusively through the locking screws. Theoretically, locked plating may therefore induce higher stress concentrations caused by the focused load transfer through fixed-angle screws compared with nonlocked plating. Clinically, a recent case series on locked plating reported a 2.6% incidence of secondary fractures at the plate end[13]. Despite this theoretical and clinical concern, the potential for an elevated periprosthetic fracture risk of locked plating compared with conventional plating in osteoporotic bone has not been investigated to date.

This study comparatively analyzed the periprosthetic fracture risk between locked and conventional bridge-plating constructs in a validated osteoporotic bone surrogate. Additionally, a modified locked plating technique was explored in an attempt to minimize stress concentrations at the plate end and to maximize the load required to induce a periprosthetic fracture.

Material and Methods

Three bridge-plating constructs were applied to osteoporotic diaphyseal bone surrogates and were tested to failure in bending, torsion, and axial loading. First, a locked plating construct was compared with a nonlocked conventional plating construct to investigate whether locked constructs exhibit periprosthetic fracture at lower loads relative to nonlocked constructs in osteoporotic bone. Subsequently, the outermost locking screw in the locked plating constructs was replaced by a conventional (nonlocked) screw in an attempt to reduce the hypothesized stress concentration with use of a hybrid plating construct.

Specimens

Bridge-plating constructs were evaluated on validated surrogates representative of the osteoporotic femoral diaphysis[14]. These surrogates consisted of a cortex shell and a trabecular core. The 27-mm-diameter and 2-mm-thick cortex shell was custom-machined from third-generation composite bone cylinders (Pacific Research Laboratories, Vashon, Washington). The trabecular core was machined from 10-pcf (0.16-g/cm^3) solid rigid polyurethane foam (Pacific Research Laboratories) and was bonded to the inside of the cortex shells. Prior research has demonstrated that five structural properties of this bone surrogate (torsional rigidity and strength, bending rigidity and strength, and screw pullout strength) matched the lower 16% of the cumulative range reported for native femora[14]. Therefore, this osteoporotic bone surrogate reflects the diminished structural properties seen in osteoporotic femora.

Implants

Generic implants were dimensioned to nominally replicate standard 4.5-mm osteosynthesis plates and screws that are commercially available from several manufacturers. However, the generic implants were designed with screw-holes in the exact same position for locked and nonlocked plates, and with identical screw threads for locked and nonlocked screws (Fig. 1). These attributes are not available in most commercially available implants but were deemed necessary for direct comparison between locked and nonlocked constructs to eliminate confounding factors due to geometric differences. Generic eleven-hole 4.5-mm osteosynthesis plates in locked, nonlocked, and hybrid configurations were manufactured from Ti-6Al-4V surgical grade titanium by an orthopaedic implant manufacturer (Thortex, Portland, Oregon). The plates were 17.5 mm wide, 5.5 mm thick, and 200 mm long and had a hole-spacing of 18 mm. Generic 4.5-mm screws with a four-fluted self-tapping feature and a thread pitch of 1 mm were manufactured from the same material and by the same manufacturer in locked and nonlocked configurations. Screw heads of conventional screws had a spherical undersurface as defined by ASTM standard F543-02[15].

Fixation Constructs

Conventional plating, locked plating, and hybrid plating constructs were tested in a bridge-plating configuration with a standard 10-mm fracture gap[16-18]. On one side of the construct, the plate was applied to the osteoporotic bone surrogate with three screws placed in the first, third, and fifth plate-holes (Fig. 2, *a*)[11,19]. This construct side served for the evaluation of conventional plating, locked plating, and hybrid plating configurations in an osteoporotic diaphysis. On the opposite side, the plate was rigidly fixed with five bicortical screws to a bone surrogate that was reinforced by filling its medullary canal with bone cement. Therefore, construct failure was isolated to the osteoporotic diaphyseal side of each construct, making results applicable to both purely diaphyseal fractures and diaphyseal-metaphyseal fractures with a well-fixed metaphyseal component. For the conventional plating configuration, the three nonlocking screws were tightened with a torque wrench (Sturtevant Richmond, Franklin Park, Illinois) to 80% of their stripping torque, as determined in a pilot study on five specimens. In the locked plating configuration, the plate was placed directly on the surrogate surface and the three locking screws were tightened to 4 Nm[20]. In the hybrid plating configuration, first, a conventional (nonlocked) screw at the plate end was inserted to compress the plate onto the surrogate, and then locking screws were placed in the third and fifth plate-holes (Fig. 2, *b*).

Loading

Each of the three constructs was tested under progressive dynamic loading to failure[17,20] in bending, torsion, and axial compression with a biaxial material testing system (model 8874; Instron, Norwood, Massachusetts). Five specimens of each construct type were tested to failure in each loading mode, requiring a total of forty-five specimens (Fig. 3, *a*).

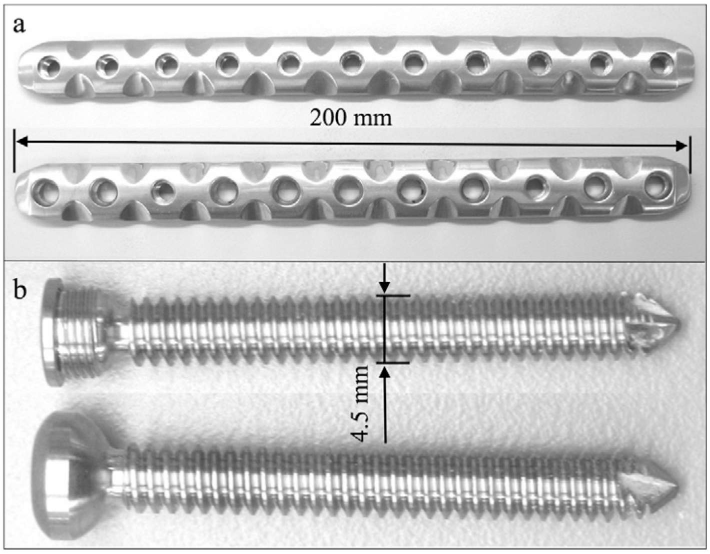

Fig. 1
Generic implants with (a) conventional (bottom) and locking (top) plates with identical screw patterns and (b) conventional (bottom) and locked (top) 4.5-mm screws with identical bone threads and a four-fluted self-tapping feature.

Bending was applied in a four-point-bending setup to generate a constant bending moment across the entire plate length. The upper and lower cylindrical supports were separated by 210 mm and 320 mm, respectively. The plate was located on the tension side to induce bending in a gap-closing mode[19]. Torsion was applied around the diaphyseal shaft axis. Axial compression was applied through a ball bearing at the osteoporotic diaphyseal side of the construct, while the reinforced surrogate portion was rigidly fixed[17,20]. After application of a static preload (L_{PRE}), sinusoidal loading with a dynamic load amplitude (L_{DYN}) was applied at 2 Hz. Every 100 loading cycles, this load amplitude was increased stepwise by L_{DYN} until construct failure occurred (Fig. 3, *b*). For bending, torsion, and axial compression, preloads of 1 Nm, 1 Nm, and 50 N, respectively, were selected. Subsequently, stepwise load amplitudes of 1 Nm, 1 Nm, and 100 N were applied in bending, torsion, and axial compression, respectively. This stepwise load increase enabled dynamic loading to failure while ensuring that failure was attained for

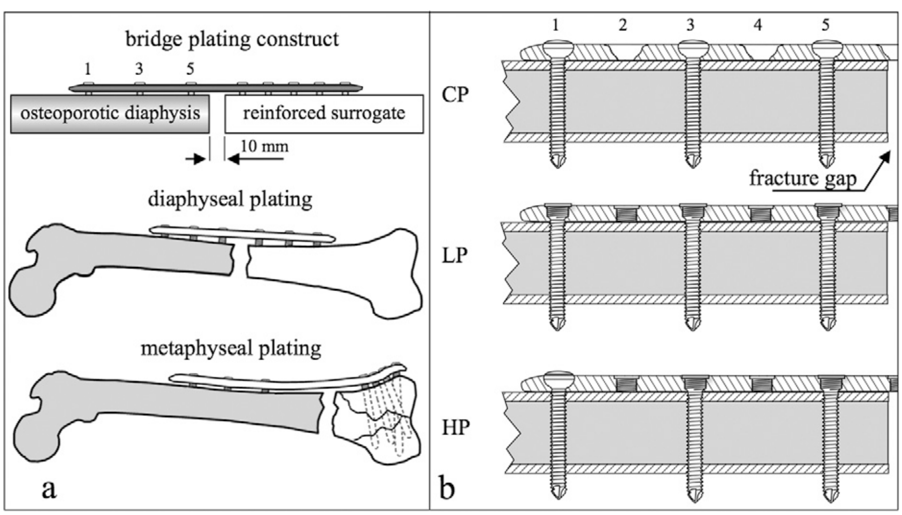

Fig. 2
a: Bridge-plating configuration with construct failure isolated to the osteoporotic diaphysis, generating results pertinent for diaphyseal and metaphyseal fracture fixation. *b:* Diaphyseal fixation configurations showing conventional (nonlocked) plating (CP), locked plating (LP), and hybrid plating (HP).

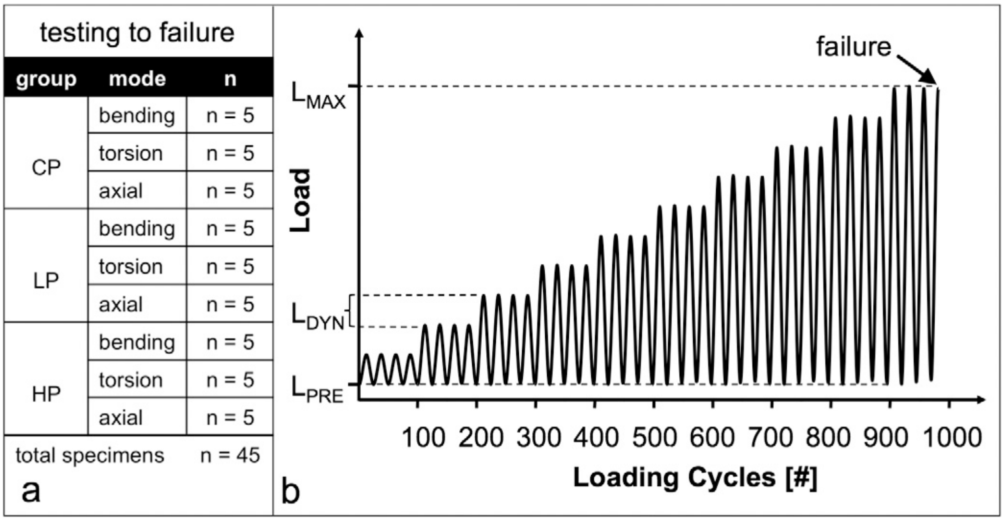

Fig. 3

a: Summary for testing of three configurations in three loading modes. *b:* Progressive dynamic loading protocol for dynamic cycling to failure in bending, torsion, and axial compression. After application of a preload (L_{PRE}), the dynamic load amplitude (L_{DYN}) was increased every 100 cycles until construct failure (L_{MAX}) occurred. CP = conventional plating, LP = locked plating, and HP = hybrid plating.

each configuration within a reasonable number of load cycles[17]. Construct failure was defined either by catastrophic fracture or by a subsidence threshold at the fracture site, whichever occurred first. Subsidence represents the nonrecoverable collapse at the fracture site after load removal and is caused by implant bending or implant loosening[20,21]. A subsidence threshold of 1 mm, 5°, and 1 mm in bending, torsion, and compression, respectively, was deemed indicative of the onset of construct failure in the absence of a catastrophic fracture. Subsidence by 5° nominally correlated to a 1-mm shear displacement between cortices at the fracture site. Subsidence was assessed with two miniature electromagnetic motion sensors (pcBIRD; Ascension Technology, Burlington, Vermont). These sensors were centered in the medullary canal at each side of the fracture gap and recorded the motion of the bone ends at the fracture site in six degrees of freedom with a resolution of 0.1 mm and 0.1° at a 100-Hz sampling rate.

Outcome Evaluation

The performance of each configuration was described by its failure load and failure mode. Failure load was quantified as the maximum load (L_{MAX}) recorded during progressive dynamic loading to failure. Failure modes were visually analyzed for the presence of a periprosthetic fracture, hardware failure, and fixation failure at the screw-bone interface.

To test the hypothesis that locked bridge-plating constructs can fail at lower loads compared with conventional constructs in osteoporotic bone, L_{MAX} results of the locked plating and conventional plating groups were compared. To determine if the modified locked plating technique improved the strength of the locked plating construct, L_{MAX} results of the hybrid plating group were compared with the locked plating group. Statistical analyses of L_{MAX} results were conducted individually for each loading mode at a level of significance of $\alpha = 0.05$ with use of an analysis of variance and post hoc Tukey tests.

TABLE I Summary of Construct Failure Loads and Failure Modes*

	Bending		Torsion		Axial Compression	
Configuration	L_{MAX}† *(Nm)*	Failure Mode	L_{MAX}† *(Nm)*	Failure Mode	L_{MAX}† *(kN)*	Failure Mode
Conventional plating	34.5 ± 2.6	Periprosthetic failure	25.5 ± 1.9	Periprosthetic failure (1) and fixation failure (4)	3.9 ± 0.2	Fixation failure
Locked plating	26.8 ± 3.1	Periprosthetic failure	22.5 ± 1.7	Periprosthetic failure (1) and hardware failure (4)	4.5 ± 0.2	Hardware failure and fixation failure
Hybrid plating	37.5 ± 4.7	Periprosthetic failure	24.4 ± 1.7	Periprosthetic failure (1) and hardware failure (4)	4.3 ± 0.4	Hardware failure and fixation failure

*Five specimens of each construct type were tested to failure in each loading mode, requiring a total of forty-five specimens. †The values are given as the mean and the standard deviation. L_{MAX} = maximum load recorded during progressive dynamic loading to failure.

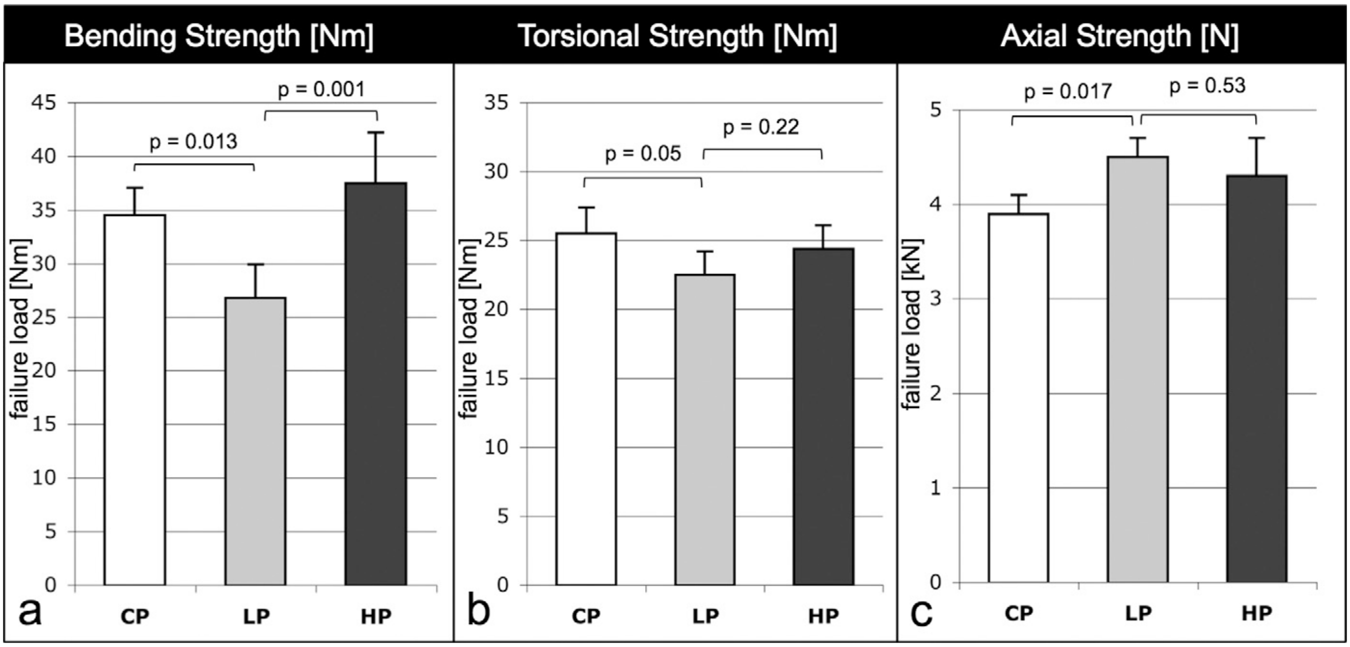

Fig. 4

Summary of failure load results, representing the mean strength of the constructs in bending (a), torsion (b), and axial compression (c). Each reported failure load is based on the testing of five specimens. CP = conventional plating, LP = locked plating, and HP = hybrid plating.

Source of Funding

Funding was provided by the Legacy Research Foundation and by the Orthopaedic Research and Education Foundation (OREF).

Results

Compared with conventional plating constructs, locked plating constructs had a 22% lower bending strength ($p = 0.013$), an equivalent strength in torsion ($p = 0.05$), and a 15% higher strength in axial compression ($p = 0.017$) (Table I, Fig. 4). In bending, all conventional plating and locked plating constructs failed by transverse periprosthetic fracture through the screw-hole at the plate end (Fig. 5). No screw bending, plate bending, or elongation of the screw-holes was noted. Inspection of the screw-holes showed the threads to be intact, indicating that no screw pullout had occurred. In torsion, one conventional plating construct and one locked plating con-

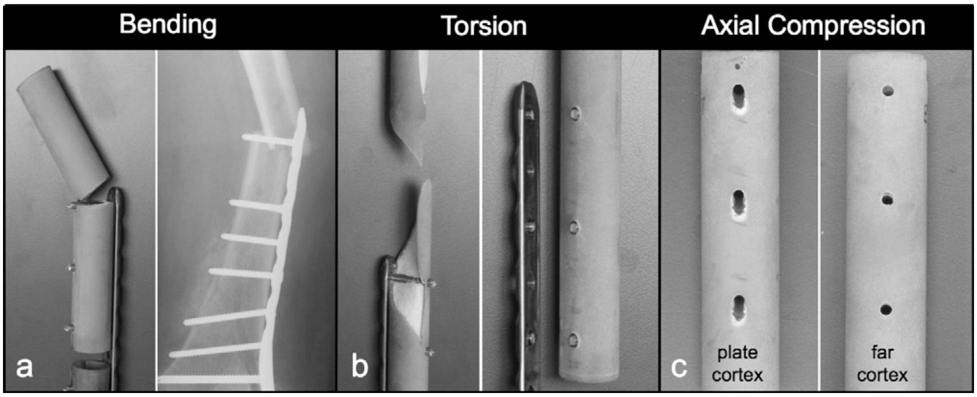

Fig. 5

Failure modes specific for loading conditions. *a:* In bending, all constructs failed by transverse periprosthetic fracture at the end screw. This failure mode was associated with periprosthetic fractures seen in the osteoporotic femoral diaphysis. *b:* In torsion, one specimen of each configuration failed by spiral periprosthetic fractures at the end screw. The remaining specimens failed by screw breakage in locked plating constructs (at a load before periprosthetic fracture occurs), by screw loosening in conventional plating constructs, or by a combination in hybrid plating constructs. *c:* In axial compression, all specimens failed by subsidence because of migration of the screws in the cortex underlying the plate, leading to loosening of nonlocked screws and bending of locked screws.

struct failed by spiral periprosthetic fracture through the screw-hole at the plate end. The remaining conventional plating constructs failed by subsidence because of screw loosening. The remaining locked plating constructs failed by screw breakage at the screw-bone interface. In axial compression, all conventional plating and locked plating constructs failed by subsidence because of screw migration in the near cortex, resulting in elongation of the near cortex holes and progressive loss of fixation. In conventional plating constructs, migration caused screw rotation about a pivot point in the far cortex of the bone. In locked plating constructs, migration caused screw bending.

In hybrid plating constructs, replacing the locked end screw with a conventional screw significantly increased the bending strength by 40% (p = 0.001) compared with locked plating constructs (Table I, Fig. 4). No significant difference between hybrid plating and locked plating constructs was detected with regard to torsional strength (p = 0.22) or axial strength (p = 0.53). In bending, all hybrid plating constructs failed by transverse periprosthetic fracture through the screw-hole at the plate end (Fig. 5). In torsion, one hybrid plating construct failed by spiral periprosthetic fracture through the screw-hole at the plate end. The remaining hybrid plating constructs failed by breakage of the locked screws and subsequent fixation failure of the nonlocked end screw. In axial compression, hybrid plating constructs failed by subsidence because of screw migration in the near cortex and subsequent screw bending, resulting in elongation of the near cortex holes and progressive loss of fixation.

Discussion

We analyzed for the first time the strength and failure modes of locked plating constructs in a diaphyseal osteoporotic model to assess the periprosthetic fracture risk. The results of this study demonstrate that, in bending, locked plating constructs may cause a greater periprosthetic fracture risk in osteoporotic bone than do conventional plates, and that this decreased bending strength can be mitigated with use of a conventional end screw. For conventional plating, periprosthetic fracture in osteoporotic bone has been recognized as a clinical concern for over two decades[1,2,5,6]. Computational models have shown that most of the load transfer occurs through the outermost screws[22,23]. The Association for the Study of Internal Fixation[2] and Harkess and Ramsey[1] recommended unicortical fixation of the screws farthest from the fracture site to reduce the periprosthetic fracture risk in conventional plates. Their recommendation was based on the assumption that a unicortical end screw would yield a more gradual transition in the normally abrupt change in stiffness at the end of the plate. Conversely, two independent biomechanical studies demonstrated that unicortical end screws increase the stress-riser effect and cause fracture at a lower load compared with bicortical screws under bending (gap-closure) and torsion[3,4]. As a consequence, the concept of unicortical end screws has been abandoned today. Alternative recommendations to improve the fixation strength of plate constructs in osteoporotic bone include the use of longer plates, more screws, and oblique screws at the plate end[19,24,25].

Locked plates are commonly perceived as providing stronger fixation on the basis of the superior pullout resistance of fixed-angle screws, especially in metaphyseal fixation and in osteoporotic bone[26]. However, the strength of a fixation construct is not necessarily determined by pullout resistance, but by the failure mode that occurs first, depending on the loading mode and bone quality. A recent one-year follow-up assessment of 151 fractures treated with locked plating had a 2.6% incidence of secondary fractures at the plate end[13]. While the timing of secondary fractures has not been reported, it can reasonably be assumed that the risk of periprosthetic fracture may persist until removal of the locking plate.

The reported failure modes in the present study included periprosthetic fracture, hardware failure by screw bending or screw breakage, and fixation failure by screw loosening and migration. In bending, all configurations failed by periprosthetic fracture through the end screw. The locked plating configuration fractured at significantly lower bending loads than did the conventional plating and hybrid plating configurations, which suggests that a locked end screw induces a higher stress concentration than a conventional end screw. In torsion, locking screws failed primarily by screw breakage between the plate and the bone since the load transfer with angle-stable screws is focused on the screw-bone interface. These results are in agreement with a recent study by Stoffel et al. that compared conventional plating with unicortical locked plating. They reported that, in torsion, all locked plate constructs failed by screw breakage[20]. Gardner et al. tested hybrid plate fixation of osteoporotic fractures in torsion[27]. They simulated weak screw purchase in osteoporotic bone by overdrilling of screw-holes in third-generation Sawbones humeri. In contrast to the present study, their hybrid construct substituted the locked screws closest to the fracture site with conventional screws. In line with the present study, they did not find a significant difference between hybrid and locked constructs. However, they did not test constructs in bending or axial compression. In axial compression, the locked plating configuration was significantly stronger than the conventional plating configuration since the fixed-angle screws could better resist fixation failure by screw loosening than could conventional screws. The hybrid plating configuration had strength in axial compression that was comparable with that of the locked plating configuration and had strength in bending and torsion that was comparable with that of the conventional plating configuration. Therefore, the hybrid plating configuration combined the structural benefits of locked plating and conventional plating by delivering fixed-angle stability under axial loading, stress-riser reduction under bending, and comparable strength under torsion. Conversely, the need for plate compression in the hybrid plating configuration prevents the potential benefits of more biological fixation by plate elevation[28]. However, while plate elevation can preserve periosteal perfusion, controversy remains regarding its benefits for fracture-healing and cortical perfusion[29]. Contemporary combination plates are routinely compressed onto the bone with use of conventional screws to facilitate reduction, and they are supplemented by locking screws to improve pullout resistance. Finally, the use of fixed-angle locking screws at the end of the plate presents a clinical challenge, as the plate often diverges from the midline of the bone, resulting in single cortex fixation. Utilizing a conventional screw in the outermost screwhole allows angulation of the screw toward the diaphyseal midline, thus avoiding this potential problem.

The results of this study are limited to load-bearing bridge-plating of diaphyseal and metaphyseal fractures in osteoporotic bone. These so-called problem fractures constitute the primary indication of locked plating[30]. Only a standard configuration with the screws in the first, third, and fifth plateholes was evaluated. This configuration represents the minimal number of screws recommended for adequate plate fixation[31]. The results are further limited to titanium-alloy plates and screws. Since stainless steel has almost twice the stiffness of titanium, stainless steel plates may induce greater stress concentrations compared with titanium plates. Furthermore, the results in the present study are limited to the generic implant design that was used. However, the relative differences observed between plate configurations are likely to apply to geometrically similar, commercially available plating systems that accommodate locked and conventional fixation options.

A validated synthetic bone model was used to extract relative differences between constructs under highly reproducible test conditions. Given the large deviation in the structural properties of cadaver specimens and the considerable number of experimental variables under investigation, this comprehensive evaluation benefited from the use of reproducible surrogates. Nevertheless, the absolute results obtained with the bone surrogate do not allow a direct comparison with the clinical setting. However, the relative relationship in outcome parameters between the plating constructs should retain its clinical relevance. In vivo, absolute failure loads and failure modes are likely to depend on the bone quality. In the case of healthy bone, implant bending or breakage may occur before a periprosthetic fracture occurs. Furthermore, fixation constructs are loaded in vivo under combined loading modes, making the clinical failure mode more complex than that described in this study. However, to gain a comprehensive understanding of mechanisms leading to construct failure, it was vital to isolate the three principal loading modes. It should also be noted that the synthetic bone model has been specifically validated for loading in bending and torsion but not in axial compression. Finally, progressive dynamic loading yielded construct failure within 1800 to 4400 loading cycles, with the highest load level being applied for a maximum of 100 cycles. In the absence of a traumatic event, fixation constructs in the clinical setting endure a far greater number of loading cycles before fracture-healing can occur, which may lead to construct failure at lower loads than those reported in this study.

In conclusion, the focused load transfer of locking plates through fixed-angle screws can increase the periprosthetic fracture risk in osteoporotic bone under bending compared with conventional plates. Replacing the outermost locking screw with a conventional screw reduces the stress concentration at the plate end and can significantly increase the bending strength of fixation constructs compared with all-locked constructs. Given the severity of periprosthetic fractures, the benefit of an increased bending strength achieved with a conventional end screw may outweigh the theoretical benefits of plate elevation on fracture-healing of an all-locked plating construct. ■

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