A Biomechanical Comparison of Locked Plate Fixation With Percutaneous Insertion Capability Versus the Angled Blade Plate in a Subtrochanteric Fracture Gap Model

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Objectives: The angled blade plate has been the historical standard in fixed-angle extramedullary subtrochanteric femur fracture fixation, but it requires an extensile lateral approach to the femur. Little formal evaluation exists for specifically designed percutaneous extramedullary implants. The purpose of this study was to compare 3 locked plating constructs, all with percutaneous insertion capability, with the standard 95-degree angled blade plate to determine whether specifically designed fixed-angle extramedullary implants for subtrochanteric femur fractures were biomechanically comparable to the angled blade plate.

Methods: Forty composite adult femurs were divided into 4 equal groups. The constructs evaluated included a 95-degree angled blade plate, a broad 4.5-mm combination locking plate, and a precontoured proximal femoral locking plate (PFLP) with and without an oblique, angled strut or "kickstand" screw. A 30-degree wedge osteotomy was used to create a subtrochanteric fracture gap model. Each specimen underwent axial and torsional stiffness testing along with cyclic axial loading to failure.

Results: Axial stiffness testing revealed that the PFLP with the "kickstand" screw was the stiffest construct (92.2 ± 17.4 Nm/m), which was 211% stiffer than the blade plate, 309% stiffer than the broad plate, and 194% stiffer than the PFLP without the kickstand screw. The blade plate had the highest torsional stiffness (2.42 ± 0.08 Nm/degree), which was 151% stiffer than the broad plate, 128% stiffer than the PFLP with the kickstand, and 138% stiffer than the PFLP without the kickstand screw. The PFLP with the kickstand screw had the least irreversible deformation (6.3 mm), which was 52% less than the broad plate and 61% less than the PFLP without the kickstand screw.

Conclusions: Our data reveal that the PFLP with the "kickstand" screw provides more axial stiffness, less torsional stiffness, and equivalent irreversible deformation to cyclic axial loading when compared with the blade plate.

Key Words: subtrochanteric femur fractures, locking plates, biomechanical study

INTRODUCTION

Subtrochanteric femur fractures in young patients are high-energy injuries that require durable implants to maintain stability over potentially extended healing times. Although intramedullary fixation has a proven track record for subtrochanteric femur fractures and is probably the current gold standard,[1–12] some surgeons prefer extramedullary fixation and have used it successfully.[1,13–27] As implants and techniques continue to evolve, particularly with the use of locking plates and minimally invasive plating, extramedullary devices may have a more significant role in the treatment of subtrochanteric femur fractures.

There are advantages and disadvantages to both intramedullary and extramedullary devices. Due to the technically challenging deformities encountered in subtrochanteric femur fractures, both techniques may require an open reduction if closed reduction techniques have failed. However, open reduction can be performed without significantly devitalizing the fracture fragments. Biomechanical studies that have compared intramedullary and extramedullary devices have demonstrated that intramedullary fixation, due to its load-sharing properties, has greater axial[4,5] and bending stiffness,[4] but less torsional control[4] in simulated subtrochanteric femur fractures. Because of its relative stability, intramedullary fixation has been shown to have significantly more interfragmentary motion at the fracture site when compared with extramedullary fixation.[15] One of the greatest benefits of intramedullary fixation is that it creates an internal buttress preventing medial femoral shaft translation, which decreases the risk of fixation failure and fracture collapse.[2,5,19] Hence, there has been a lower rate of reoperations reported, due to lower rates of implant failure and nonunion, in some clinical trials that have compared intramedullary and extramedullary devices.[6,8,10,14,28,29]

Although closed intramedullary nailing techniques have been successful in treating subtrochanteric femur fractures and have allowed for nail insertion via a small access incision, there are disadvantages to this technique. Significant insertion site morbidity has been reported including trochanteric pain, abductor weakness, and heterotopic bone formation.[13,14,30,31] Furthermore, intramedullary fixation transiently ablates the endosteal blood supply,[21,32] is technically demanding with proximal fracture extension involving the nail insertion site,[3,17,21,33] can lead to iatrogenic fracture or propagation of the fracture,[31,34,35] and has a concerning risk of varus malunion, if not correctly addressed intraoperatively.[29,36,37]

A wide variety of extramedullary implants have been used for the treatment of subtrochanteric femur fractures over the years and each has specific advantages and disadvantages. Currently, the most likely reason to use an extramedullary device is in subtrochanteric femur fractures that have extension into the nail entry site.[3,9,17,21,33] Fixed-angle implants have been evaluated in both clinical and biomechanical studies.[1,13–21] The angled blade plate is considered by some to be the gold standard in extramedullary fixed-angle proximal femoral plate fixation, but it requires an extensile lateral approach to the femur for insertion.[7,18,20,28,37] In an attempt to minimize soft tissue dissection and optimize the biological healing potential of plated fractures, open plating with indirect reduction techniques and submuscular plating with indirect reduction techniques for the proximal and distal femur and the femoral shaft have become increasingly popular.[7,8,18,20,21,37–39] Although minimally invasive techniques have been developed for proximal femoral fractures using traditional extramedullary implants, there has been little formal evaluation of specifically designed percutaneous stabilization devices, specifically locked plates, for the proximal femur.[8,21,39] The purpose of this study was to compare the axial and torsional stiffness and cyclic loading with failure properties of 3 locked plating constructs with percutaneous insertion capability to the standard 95-degree angled blade plate.

MATERIALS AND METHODS

A total of 40 third-generation composite adult femurs (Sawbones; Pacific Research Laboratories, Vashon, WA) were obtained and randomly divided into 4 groups, each with 10 specimens (Table 1). Third-generation composite femurs were used because of their similar biomechanical properties to human cadaveric specimens and decreased specimen variability.[40,41] The implants used in all 4 groups were made by the same manufacturer for consistency, both with plate material used and with screw design (Synthes, USA, Paoli, PA). Group 1 (control) consisted of femurs plated with the 95-degree angled blade plate. Group 2 (broad locking compression plate) consisted of femurs stabilized with a contoured broad 4.5-mm locking compression plate. Group 3 [proximal femoral locking plate (PFLP) with kickstand] femurs were stabilized with a precontoured PFLP including an oblique, locking strut or “kickstand” screw. Group 4 (PFLP without kickstand) used the same PFLP minus the “kickstand” screw. After standard plate positioning and application, a 30-degree wedge osteotomy was created at the level of the lesser trochanter to create a subtrochanteric gap model using a previously described technique.[42] This osteotomy creates an unstable fracture similar to an OTA 31A3, 32B1.1, C1.1, B2.1, C2.1, B3.1, and C3.1.[43] Although we acknowledge that fracture reduction is a key component for construct stiffness in the clinical situation, the osteotomy was created after implant application to allow for reproducible placement of the implants without radiographic assistance and to reduce the variability of fracture reduction and its effect on testing. Radiographs of each construct group after osteotomy is shown in Figure 1.

The load was applied to the head of the femur through a custom mold. The condyles of the distal end of the femur were also held in a custom mold that was secured to the materials testing machine. The specimens were supported in the testing machine by a ball bearing to avoid uncontrolled torque or bending similar to that previously described by Cordey et al.[44] Each specimen underwent axial and torsional stiffness testing along with cyclic axial loading to failure to determine irreversible deformation. All testing was performed using an Instron 5800R (Instron, Canton, MA) materials testing machine. The testing apparatus for axial stiffness and cyclic axial loading to failure protocols is shown in Figure 2. The line of action of the axial force went through the femoral head proximally and through the intercondylar notch distally to simulate physiologic loading in single-leg stance along the mechanical axis. Axial stiffness testing was performed first and consisted of loading each specimen to 500 N at a rate of 5 mm/min.

Before cyclic axial loading to failure, torsional stiffness testing (Fig. 3) was performed. The proximal end was held in a custom mold and the distal end was secured in a chuck. Precise positioning was done to ensure that the femoral axis was aligned with the axis of rotation. The custom proximal fixture was mounted to a bearing system. The Instron imparted a force to a lever attached to the bearing to allow rotation of the

TABLE 1. Pertinent Plate Design Characteristics

Group	Plate Type	Plate Length (Hole)	Proximal Fixation	Shaft Fixation
1	95-degree blade plate (control)	14	70-mm blade	36- and 4.5-mm cortical screws
2	Broad 4.5-mm LCP	14	Two 80- and 5.0-mm solid locking screws	36- and 5.0-mm solid bicortical locking screws
3	PFLP with kickstand	8	● 90-mm (hole 1) and 75 mm (hole 2) 7.3-mm cannulated proximal locking screws ● 85-mm (hole 3) 5.0-mm cannulated “kickstand” locking screw	36- and 5.0-mm solid bicortical locking screws
4	PFLP without kickstand	—	● 90-mm (hole 1) and 75-mm (hole 2) 7.3-mm cannulated proximal locking screws	36- and 5.0-mm solid bicortical locking screws

LCP, locking compression plate.

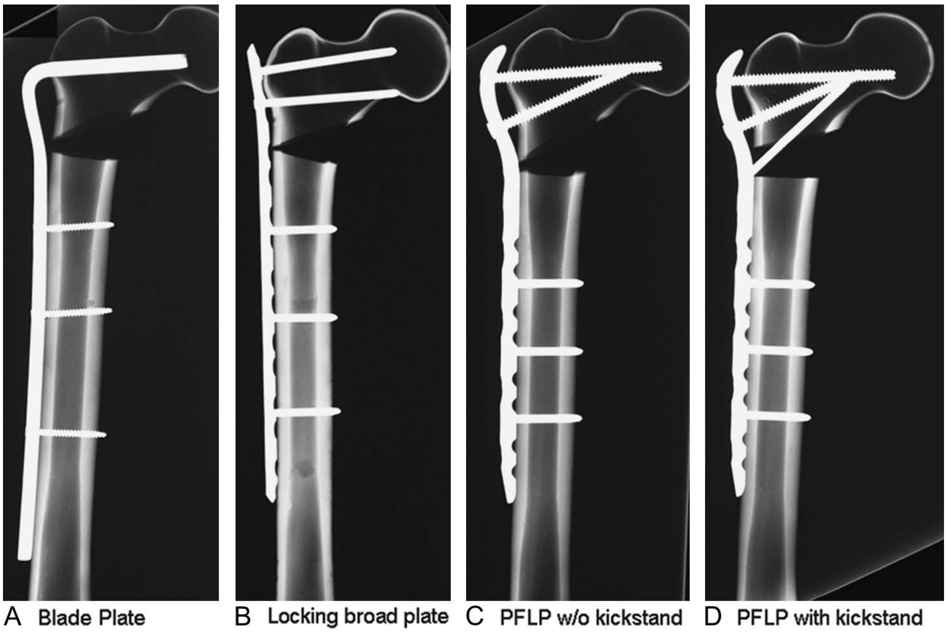

FIGURE 1. Implant construct radiographs. A, Blade plate. B, Locking broad plate. C, PFLP without kickstand. D, PFLP with kickstand.

femur about its longitudinal axis. The torsional cycle was applied in an external rotation direction based on the anatomy of the femoral model. External rotation was chosen because it is the typical rotational deforming force on the proximal femur associated with a subtrochanteric fracture. Torsional loading was performed by preloading each specimen to 5 Nm of torque and then loading the specimen to a maximum torque of 20 Nm at a rate of 25 degrees/min.

Cyclic axial loading to failure was performed last with a preload of 100 N, and each specimen undergoing 10 cycles at each peak load from 300 to 1000 N in 100-N increments at a rate of 0.75 mm/s as previously described for the distal femur.[45,46] Each load cycle was conducted as a ramp load in displacement control at a rate of 0.75 mm/s to the peak load of the cycle. The construct was considered to have failed if the implant or femur model fractured, if the implant pulled out of the femur, if the medial edges of the osteotomy contacted, or if there was irreversible deformation present on completion of the cyclic axial loading protocol.

Axial stiffness and torsional stiffness were calculated from the linear portion of the load–displacement curve and torque–angle graph, respectively. For cyclic axial loading to failure, a time–displacement curve was created (Fig. 4).[47] The irreversible deformation was determined by subtracting the amount of displacement present at the beginning of the 300 N cycle from the displacement after the final cycle.

Data were analyzed by using a 1-way analysis of variance, and Fischer post hoc least significant difference criterion was used to correct for multiple group comparisons. *P* values less than 0.05 were considered indicative of significant differences.

RESULTS

Axial stiffness testing revealed significant differences between the 4 different construct groups (Table 2). The PFLP with kickstand (92.2 ± 17.4 Nm/m) construct was axially stiffer than all of the other constructs. It was 211% stiffer than the blade plate construct ($P < 0.001$), 309% stiffer than the broad plate construct ($P < 0.001$), and 194% stiffer than the PFLP without kickstand construct ($P < 0.001$). The only other significant difference was that the PFLP without kickstand construct was 106% stiffer than the broad plate construct ($P = 0.0301$).

The blade plate group (2.42 ± 0.08 Nm/degree) had significantly more torsional stiffness than all the other constructs. It was 151% stiffer than the broad plate group ($P < 0.0001$), 128% stiffer than the PFLP with the kickstand group ($P < 0.0017$), and 138% stiffer than the PFLP without the kickstand screw group ($P = 0.002$). There were no other statistically significant differences between construct groups.

The results of cyclic axial loading to failure demonstrate a variety of differences between the groups. The time–displacement curve (Fig. 4) enabled calculation of the irreversible deformation that occurred. The PFLP with kickstand construct (6.3 ± 1.2 mm) had 52% less irreversible deformation than the broad plate construct ($P = 0.0071$) and 61% less irreversible deformation than the PFLP without kickstand construct ($P = 0.0014$). The blade plate was the only group that had significantly less irreversible deformation (59%) than the PFLP with the kickstand construct ($P = 0.0093$). No significant difference was detected between the blade plate and the PFLP with kickstand. During loading to failure, all constructs failed at the osteotomy site with irreversible bending of the implant and collapse of the medial fracture gap. Of note, none of the lateral fracture gaps had bone-to-bone contact during or at the end of testing. None of the constructs had screw or implant fracture or screw or implant pullout.

DISCUSSION

As interest in minimally invasive plating techniques increases, a percutaneous plating solution for subtrochanteric

FIGURE 2. Axial stiffness and cyclic loading to failure model.

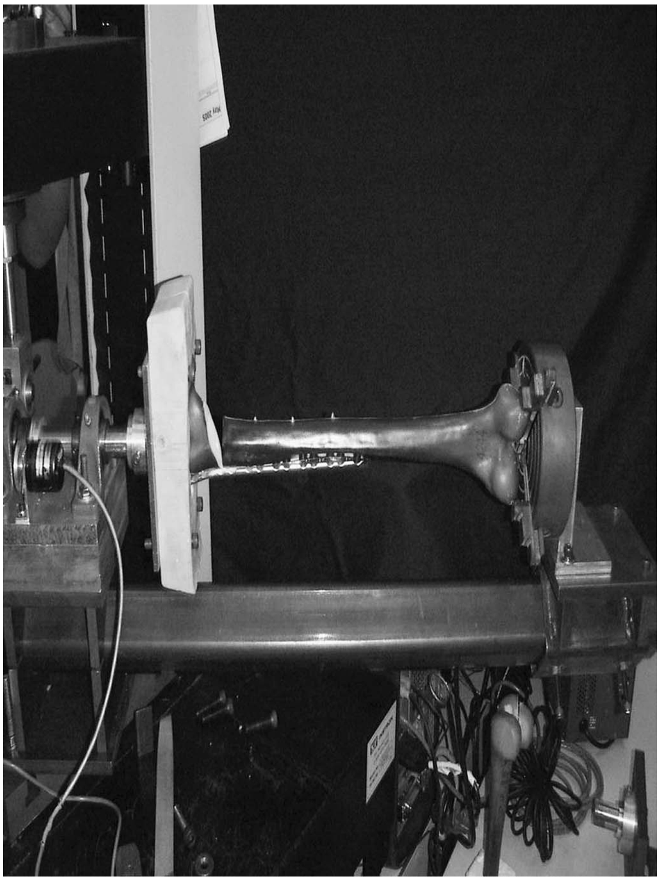

FIGURE 3. Torsional stiffness model.

femur fractures provides an attractive option. No previously published biomechanical studies have compared newer locking plate technology, which allows for percutaneous insertion, with the angled blade plate for subtrochanteric femur fractures. This study shows that locking plates can be comparable and, in some instances (PFLP with kickstand), can provide more axial stiffness than the angled blade plate.

There are few published biomechanical studies comparing the angled blade plate with nonlocked extramedullary devices in a subtrochanteric fracture model.[48] Most of the biomechanical studies available have compared nonlocked extramedullary devices with intramedullary devices.[2,4,5,15] Few have actually included the angled blade plate. Although there are biomechanical data for subtrochanteric femur fracture models, no other published study has compared the devices used in this study, this specific unstable fracture model with extramedullary devices, and the specific testing protocols used. Therefore, comparing our results with previously published biomechanical studies would not be valid.

The limitations of this study are those that are similar to all biomechanical studies. Although synthetic femurs may not ideally recreate the in vivo environment, the third-generation synthetic composite Sawbones have been shown to be similar to human cortical bone in axial and torsional stiffness and also feature decreased variability, which enhances statistical resolving power.[40,41] Similarly in the clinical situation, fracture reduction is a key component for construct stiffness. By creating the osteotomy after plate fixation, we attempted to remove fracture reduction as a variable that could affect the biomechanical testing. Although this did not recreate the clinical situation, it did create uniformity in the constructs and allowed us to evaluate the biomechanical properties of the different implants.

The PFLP with the kickstand screw provided comparatively greater axial stiffness, less torsional stiffness, and equivalent irreversible deformation in cyclic loading when compared with the blade plate. The kickstand screw adds important axial stability but has no significant role in

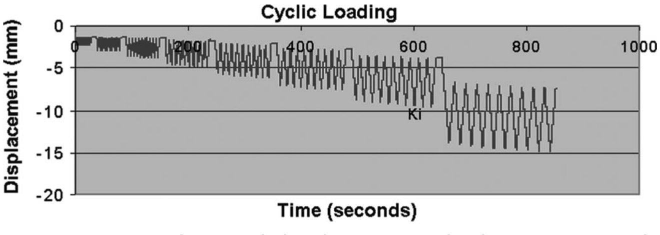

FIGURE 4. Cyclic axial loading time–displacement graph showing irreversible deformation for the PFLP with kickstand screw.

TABLE 2. Axial Stiffness, Torsional Stiffness, and Irreversible Deformation Due to Cyclic Loading Versus Plate Design

	Axial Stiffness (N/mm)	Torsional Stiffness (Nm/degree)	Cyclic Loading Irreversible Deformation (mm)
Blade plates (control)	43.6 ± 3.7	2.42 ± 0.08*	9.5 ± 2.7
Broad plates	29.8 ± 8.9†	1.60 ± 0.11	13.2 ± 2.7‡
PFLP kickstand	92.2 ± 17.4§	1.89 ± 0.39	6.3 ± 1.2
PFLP no kickstand	47.6 ± 12.6	1.76 ± 0.12	16.1 ± 4.5¶

Values are expressed as mean ± SD.

*Significantly different from broad plates ($P < 0.0001$) and PFLP with ($P = 0.0017$) and without ($P = 0.0002$) the kickstand.

†Significantly different from PFLP with ($P < 0.0001$) and without ($P = 0.03$) the kickstand.

‡Significantly different from PFLP with the kickstand ($P = 0.0071$).

§Significantly different from control (blade plates, $P < 0.0001$), broad plates ($P < 0.0001$), and PFLP without kickstand ($P < 0.0001$).

¶Significantly different from control (blade plates, $P = 0.0093$) and PFLP with the kickstand ($P = 0.0014$).

contributing to torsional stiffness. The PFLP with the kickstand screw is biomechanically equivalent to the angled blade plate, but it allows for percutaneous insertion and avoids the potential morbidity accompanying an extensile lateral approach to the femur. Further studies are required to determine if these biomechanical data are reflective of clinical outcomes.

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