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Tangential Bicortical Locked Fixation Improves Stability in Vancouver B1 Periprosthetic Femur Fractures: A Biomechanical Study

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Abstract

Objectives—The biomechanical difficulty in fixation of a Vancouver B1 periprosthetic fracture is purchase of the proximal femoral segment in the presence of the hip stem. Several newer technologies provide the ability to place bicortical locking screws tangential to the hip stem with much longer lengths of screw purchase compared to unicortical screws. This biomechanical study compares the stability of two of these newer constructs to previous methods.

Methods—Thirty composite synthetic femurs were prepared with cemented hip stems. The distal femur segment was osteotomized, and plates were fixed proximally with either: (1) cerclage cables; (2) locked unicortical screws; (3) a composite of locked screws and cables; or tangentially directed bicortical locking screws using either (4) a stainless steel LCP system with a Locking Attachment Plate (Synthes), or (5) a titanium alloy NCB system (Zimmer). Specimens were tested to failure in either axial or torsional quasi-static loading modes ($n = 3$) after 20 moderate load pre-conditioning cycles. Stiffness, maximum force, and failure mechanism were determined.

Results—Bicortical constructs resisted higher (by an average of at least 27%) maximum forces than the other three constructs in torsional loading ($p < 0.05$). Cables constructs exhibited lower maximum force than all other constructs, in both axial and torsional loading. The bicortical titanium construct was stiffer than the bicortical stainless steel construct in axial loading.

Conclusions—Proximal fixation stability is likely improved with the use of bicortical locking screws as compared to traditional unicortical screws and cable techniques. In this study with a limited sample size, we found the addition of cerclage cables to unicortical screws may not offer much improvement in biomechanical stability of unstable B1 fractures.

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Introduction

The active aging population, coupled with the increased prevalence of total joint arthroplasty has resulted in an increasing number of periprosthetic fractures.^{1,2} Displaced fractures below a well-fixed stem (Vancouver B1) are typically treated with open reduction and internal fixation. Distal fixation is generally reliable with standard locked or non-locked bi-cortical screws, but biomechanical purchase of the proximal segment is difficult due to the presence of the hip prosthesis with or without a cement mantle. Options for proximal plate fixation include cerclage cables, locking or nonlocking unicortical screws, allograft struts, and more recently, plate designs that allow bicortical fixation by directing offset locking screws tangentially around either side of the hip stem.

A series of studies have compared the biomechanical strength of different repair approaches for periprosthetic femur fractures.^{3–19} These studies have used cadaver or synthetic bones, cyclic or quasi-static loads, and destructive or nondestructive testing. Unicortical locking screws have been reported to have superior stiffness compared to cables, in one study showing five times greater stiffness in axial loading.⁵ Studies comparing bicortical methods to other approaches are few because of the more recent development of these plates.^{9,17} Furthermore, quantification of torsional behavior in bicortical constructs compared to other approaches is lacking, an important gap addressed in the present study. The aim of the present study was to compare torsional and axial strength and stiffness of bicortical, unicortical, and cable based constructs under quasi-static conditions. We hypothesized that bicortical fixation methods would provide increased stiffness and strength compared to traditional methods including cerclage cables, unicortical locking screws, and a combination of these two.

Materials and Methods

Femoral stem implantation and simulated fracture

Thirty synthetic femurs (4th generation composite, medium, left, solid foam cancellous material, 13-mm canal; #3403 Sawbones, Vashon WA) were prepared by an orthopaedic surgical resident and an orthopaedic fellow under supervision of the senior author. Each femur was implanted with a cemented total hip arthroplasty stem component (APS left size 5, Zimmer, Warsaw, IN). A custom fixture (Fig. 1) was used to position an oblique femoral neck cut and direct initial drilling of the canal. Specimens were sequentially broached, matching the 11° anteversion of the synthetic femur. We elected to place the femoral prosthesis in a matched anteversion to create a uniform cement mantle around the prosthesis. A cement restrictor was placed 2.0 cm distal to the tip of the prosthesis. Cement (Osteobond Copolymer Bone Cement, Zimmer, Warsaw IN) was mixed under vacuum and then pressurized into the canal (third generation technique). The mounting fixture was then used to obtain uniform depth and anteversion during stem insertion and curing of the cement. Following implantation, an AO/OTA 32C3²⁸ unreduced fracture with segmental bone loss was simulated by a transverse osteotomy 2.5 cm distal to the prosthesis tip.

Proximal fracture fixation constructs

Constructs were created as described in Table 1 and shown in Fig. 2. The first three constructs, named Cables, Unicortical, and Cables+Unicortical, were created using a 4.5 Locking Compression Plate (LCP), stainless steel (SS) 4.5/5.0 broad curved, 14-hole (Synthes, West Chester PA). The Bicortical SS LCP construct also used this same plate but incorporated 2 Locking Attachment Plates (LAP) (Synthes, West Chester PA) that allowed the placement of four tangentially directed 3.5 mm locking screws through each plate (8 total per construct). The fifth construct, Bicortical Ti NCB, used the titanium Non-Contact Bridging (NCB) Polyaxial 9-hole Periprosthetic Proximal Femur Locking Plate (Zimmer, Warsaw, IN).

Setup for mechanical testing

Controlled mechanical loading was applied by a servohydraulic mechanical test frame (Interlaken 3300 with Flextest 40 controller, MTS, Eden Prairie MN). Vertical load was applied to the prosthetic femoral head by an attached horizontal flat plate interfaced to a 10 kN load cell. The flat plate was steel with a 6 mm layer of oil-infused polyethylene in order to allow unconstrained horizontal (x-y) motion. This setup had the goal of transmitting only vertical loads to the femoral head so that the loading direction was known and remained consistent through testing.^{4,5,7,8,18} Actuator displacement and force were measured and tests were recorded by video.

Torsional loading

Three specimens from each group were subject to torsional internal rotation (consistent with rising from a chair with the hip 90° flexed). The most-distal 12 cm of the plate below the osteotomy was fastened using three bolts to a steel fixture attached to the base of the test frame (Fig. 3). The plate was attached such that the specimen was oriented horizontally simulating neutral femoral rotation with 11° of prosthesis anteversion. A horizontal support bar, also surfaced with the oil-infused polyethylene, was positioned beneath the lesser trochanter.^{16,19} This 10-mm wide support acted as a bearing, allowing unconstrained rotational movement of the specimen but limiting bending.

The prosthetic femoral head was loaded through the flat loading plate to induce torsion. Twenty sinusoidal preconditioning cycles (100 N, 1 Hz) were first applied. Then the loading plate was translated down at 8.0 mm/min until failure occurred. Failure was defined as an abrupt drop in force of greater than 200 N, or no increase in force over 2 mm displacement. Maximum force was determined. Stiffness was computed as the best-fit slope of the linear region of the force-displacement curve up to 500 N force (using code written in Matlab software).

Axial loading

Three specimens from each group were subjected to axial loading. The distal-most 12 cm of the femoral plate was fastened to a steel fixture attached to the base of the test frame (Fig. 3). The plate was attached such that the lateral border of the femoral shaft was oriented with 13° of lateral angulation (adduction) simulating single leg stance.²⁰ The prosthetic femoral head was loaded vertically by the flat loading plate.^{4,5,7,8,18} In pilot testing with the newer

tangential bicortical locked methods, substantial plastic deformation of the fracture fixation plate at its connection to the testing apparatus was observed after exceeding about 1200 N of force. In order to address this issue we used a two-phase testing regime. In Phase I of axial loading, 20 sinusoidal preconditioning cycles (500 N, 1 Hz) were first applied. The construct was loaded at 4.0 mm/min until the measured force reached approximately 1200 N. This loading phase was used to assess stiffness of the construct and plate (slope of best-fit line up to 500 N force).

In Phase II of axial loading, a 5 × 2 cm steel tab was welded to the fracture plate 7 cm from its proximal end (5 cm from the osteotomy), and this tab was fixed to the mounting frame (Fig. 3). This testing modification prevented plastic deformation of the plate in the unsupported region, allowed higher loads to be transmitted to the proximal plate/bone interface, and in essence forced failure to occur at the proximal plate/bone interface. Loading proceeded at 4.0 mm/min until failure (defined above), or until 7000 to 7500 N (capacity of the machine) was reached.

Statistical analysis

Statistical differences between pairs of construct groups were analyzed using one-way analysis of variance (ANOVA) followed by Tukey's multiple comparison tests (SAS, SAS Institute). Normality of data distribution was assumed based on previous biomechanical studies,^{5,19} and statistical difference was considered significant at $p < 0.05$. Separate analyses were performed for stiffness and maximum force, for both torsional and axial loading. Our small sample size does present limitations in the statistical analysis (see Discussion); a small sample size was used in the study to enable testing of five different constructs destructively in two different fundamental loading modes, within a limitation of resources.

Results

Cables, Unicortical, and Unicortical+Cables constructs: The Cables constructs failed in torsion by the femur rotating and loosening within the cables (Table 1). These constructs reached significantly less maximum force compared to all other constructs, in both torsional and axial loading (Fig. 4 horizontal red lines indicate significance; $p < .0001$ for all comparisons with Cables group in torsion maximum force, $p < .05$ for all comparisons with Cables in axial maximum force; all p-values reported are after Tukey adjustment). The average maximum force reached by the Cables constructs was 27% and 41% of that reached by the Unicortical construct group in torsional and axial loading, respectively ($p < .0001$ & $p = .027$). The Cables group was also less stiff than other constructs especially in torsion (Fig. 5, $p < .001$ for all comparisons in torsion, $p = 0.019$ for Cables vs. Bicortical Ti NCB in axial loading). Unicortical and Unicortical+Cables constructs failed by catastrophic fracture of the femur with cracks generally stemming from insertion sites of the unicortical screws. Fracture displacement was somewhat mitigated by the addition of the cables, but no significant differences were detected in either maximum force or stiffness between those groups.

Bicortical SS LCP and Bicortical Ti NCB constructs

These constructs did not fail catastrophically in axial loading (Table 1); instead the force limits of our mechanical testing machine (7000 to 7500 N) were reached. In torsional loading, Bicortical SS LCP constructs did not fail catastrophically but the plates exhibited substantial plastic deformation under the higher loads. For Bicortical Ti NCB constructs, two of three specimens failed in torsion by cracking of the cement mantle and bone-cement interface surrounding the proximal femoral stem. In torsional loading both Bicortical SS LCP and Bicortical Ti NCB constructs exhibited statistically significant higher maximum forces applied to the femoral head before failure than the other three constructs, with average differences of at least 500 N (Fig. 4, Bicortical SS LCP vs. Cables $p<.0001$; vs. Unicortical $p=.0055$; vs. Cables+Unicortical $p=.046$; and Bicortical Ti NCB vs. Cables $p<.0001$; vs. Unicortical $p=.0005$; vs. Cables+Unicortical $p=.0033$). Bicortical Ti NCB constructs were stiffer than Cables+Unicortical constructs in torsion (Fig. 5, $p=.041$). Bicortical Ti NCB constructs were also stiffer than the Bicortical SS LCP constructs in axial loading ($p=.035$).

Discussion

This study demonstrates significant improvements in torsional strength associated with tangential bicortical locking screw technologies for fixation of the proximal segment compared to cables and/or unicortical screws. No significant differences were detected in maximum forces between Bicortical SS LCP and Bicortical Ti NCB plate technologies, although testing machine limits were reached in axial loading. Cable-only constructs were only half as strong, or less, than all other constructs in both torsional and axial loading.

This study has clinical implications for the treatment of Vancouver B1 periprosthetic proximal femur fractures, especially regarding torsional behavior of bicortical constructs compared to other approaches, an important gap in knowledge in the literature. First, the study supports the biomechanical inferiority of cerclage wires as a fixation technique. Second, it shows that, in unstable Vancouver B1 fractures, the addition of cerclage cables to a unicortical screw construct may not improve mechanical performance in either torsion or axial loading compared to unicortical screws alone. This was not the case in a previous study of stable (reduced) B1 fractures¹¹, where the addition of cables was reported to increase stability, a difference possibly attributable to difference in type of fracture modeled. Third, it demonstrates that tangentially directed bicortical screw fixation likely provides stronger proximal fixation compared to unicortical and cable methods. A consideration of bicortical plate systems is that they require placement of screws tangentially around the prosthesis, and it may not always be possible to achieve bicortical fixation with multiple screws in the presence of a larger femoral component.

The small sample size in the present study limited statistical power, making it more likely that some clinically significant differences were not detected. Statistically insignificant results should be viewed with caution because of the increased probability of a Type II error. Small sample size also makes it difficult to assess data qualities such as normality of distribution which are assumptions in an ANOVA analysis. Normality was assumed based on the fact that previous studies of proximal periprosthetic femur fixation using quasi-static

loading in synthetic bones also assumed data normality (albeit with larger sample sizes).^{5,19} Statistical simulation studies have shown that the *t*-test (a parametric test, like ANOVA) can be an acceptable approach for $n=3$ in the context of the aforementioned limitations.²⁷

Many previous biomechanical studies of periprosthetic proximal femur fracture have mechanically loaded both the proximal and distal bone segment together. Many failures observed during these tests have actually occurred on the distal side^{8,21,19}, a less frequent clinical finding unrelated to the proximal purchase problem. Distal failure has sometimes led to difficulty in discerning among proximal constructs. To address the above issue, in the present study the attachment of the plate to the proximal fragment was tested in isolation, an approach also used recently by Lenz et al.¹² The present model is most appropriate for the worst case clinical scenario in which a bone gap exists at the fracture site preventing fragment-to-fragment load transmission.

Dennis et al. reported an early study of periprosthetic fixation techniques.⁵ They loaded femoral heads with a flat plate in axial compression, torsion, and lateral bending. The authors reported larger stiffness for unicortical screw constructs over cable-only constructs and allograft strut constructs. Fulkerson et al.⁷ compared fixation by proximal unicortical locking screws to fixation with cables in embalmed cadavers. Axial and torsional stiffnesses were larger for the unicortical locked plating, both before and after 10,000 cycles of axial cyclic loading. No significant differences were detected in failure loads in torsion, and specimens were not loaded to failure axially. Zdero et al.¹⁹ reported testing of twenty third-generation synthetic femurs with flat fixation plates. Plates were fixed proximally with either four locked unicortical screws, two locked unicortical screws with cables, two nonlocked unicortical screws with cables, or two nonlocked unicortical screws with cables plus strut allograft. Axial loading was applied to the femoral head using a conforming cup, whereas torsional and bending loads were applied using a flat plate. No differences between the constructs were detected in axial load-to-failure. The construct with the allograft strut showed larger stiffness than the other constructs, but no differences in stiffness were detected among the other constructs. Additional biomechanical studies prior to 2011 are compared in a review from Moazen et al.¹⁴ Clinical studies have demonstrated treatment of these fractures by internal locked plate fixation alone, without cortical allograft strut and cerclage wires or cables.^{22–24}

There have been several recent studies of bicortical proximal fixation, although data on torsional behavior is lacking which is addressed by the present study. Konstantinidis et al.⁸ used formalin-fixed fractured femurs fixed proximally with either unicortical or bicortical screws. Under axial loading no significant differences in fracture gap displacement or failure load were detected. Wähnert et al.¹⁷ compared a titanium LCP+LAP to the titanium NCB system. Cadaveric femurs were loaded cyclically in axial compression until failure. Axial stiffness and number of cycles to failure were significantly higher for the NCB system, possibly due to its stiffer larger plate. Lenz et al. published recent studies that used axial cyclic loading of cadaver specimens and a fracture model with closed osteotomy gap.^{9,11} The LCP+LAP construct demonstrated longer survival under cyclic loading compared to unicortical and cerclages only. The authors also performed a comparison of the LAP to unicortical plus cables in a study of eight synthetic bones subject to cyclic loading.¹²

Torsional loading was not included in these studies. Biomechanical testing of alternative fixation such as the cement-in-cement technique have also been reported.³

The femur is subject to a complex loading environment in vivo. Large forces are transmitted through the hip joint dynamically, and these forces vary across activities and persons. In the present study two loading approaches were tested to failure: axial and torsional. Loadings were configured based on prior biomechanical studies,^{5,19} and on findings from clinical studies²⁰ in which hip implants were instrumented with wireless sensors. Those in vivo studies reported that the mean resultant hip force during walking was oriented 13° away from the femoral axis. For axial testing this angle was used to orient the femur relative to the loading actuator, resulting in a small component of bending in addition to the axial compression.

The choice of how loading is applied to the femoral head may affect stiffness and failure loads.²⁵ Loading was applied using a flat loading plate with the goal of transmitting pure vertical forces.^{4,5,7,8,18} This approach would have similar results as if loads were transmitted via an acetabular cup that was unconstrained in the horizontal plane. Although many previous studies have also used a flat plate to apply loading, it does allow horizontal translation of the head during load-to-failure testing, which increases bending moments applied to the construct. Some studies have instead constrained the head to move purely vertically, which eliminates the aforementioned horizontal movements, but results in unknown side loads building up during loading and changes the resultant force direction applied to the head.

The present study has several limitations in addition to small sample size. Some complexities in femoral loading were not considered. Loads not acting through the hip joint from muscles and other soft tissues were ignored.²⁶ The distal segment of the femur was not considered in the model, precluding failures of this segment in the testing. The proximal plate was constrained to the testing machine during axial load-to-failure tests, which forced failure to occur at the plate-bone interface as intended; however bone cracking at failure occurred at or proximal to the plate constraint location, and it is unclear whether the manner in which the plate was constrained could affect failure strength. Specimens were dynamically preconditioned, but cyclic fatigue loads were not applied. Talbot et al.¹⁶ showed in synthetic femurs fixed with combinations of locking plates and cables that 100,000 cycles of loading generally did not affect axial, torsional, and bending stiffnesses. The present study is also limited by the use of synthetic composite bones instead of human specimens. Synthetic bones were used because of the desire to compare five different constructs in load-to-failure in both axial and torsional loading, and the increased variability and cost associated with cadaver specimens. Human bone in these patients may have compromised bone mineral density, and future study should test the effects of bone quality on the differences between the fracture fixation constructs. Surgical implementation of these constructs, especially bicortical constructs, in the operating room is more technically challenging than represented by our benchtop setup. A cemented femoral component was used in this in vitro experiment in order to create a mechanically stable interface between the femoral stem and the synthetic femur bone. In a clinical setting, this long term stability could be created either via the ingrowth of a porous coated prosthesis (not possible in our

experiment) or using bone cement. In our cemented constructs the screws had purchase with, or traversed the cement mantle. While the presence or absence of cement may affect result magnitudes, we hypothesize that the key findings of our paper in cemented constructs also apply to cementless constructs.

The two bicortical constructs used in this study consisted of two different materials, and direct comparisons between these two constructs should be interpreted with this in mind, although clinically both constructs are used for the same types of fractures. Interestingly the Ti NCB plate consists of titanium alloy TiAlV which has a lower elastic modulus than the stainless steel in the SS LCP construct, but this effect on overall construct stiffness is offset in the Ti NCB plate by the larger cross-sectional area and area moment of inertia proximally.

In conclusion, this study supports the hypothesis that fixation stability of the proximal segment of a Vancouver B1 periprosthetic fracture is improved with the use of tangentially directed bicortical locking screws as compared to traditional locked unicortical screw and cable techniques. Also in this biomechanical study of unstable Vancouver B1 fractures, the addition of cerclage cables to a unicortical screw construct offered no significant improvement in mechanical performance in either axial or torsional loading compared to unicortical screws alone.

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References

1. Holley K, Zelken J, Padgett D, et al. Periprosthetic fractures of the femur after hip arthroplasty: an analysis of 99 patients. *HSS J. Musculoskelet. J. Hosp. Spec. Surg.* 2007; 3:190–197.
2. Kurtz SM, Lau E, Ong K, et al. Future young patient demand for primary and revision joint replacement: national projections from 2010 to 2030. *Clin. Orthop.* 2009; 467:2606–2612. [PubMed: 19360453]
3. Brew CJ, Wilson LJ, Whitehouse SL, et al. Cement-in-cement revision for selected Vancouver Type B1 femoral periprosthetic fractures: a biomechanical analysis. *J. Arthroplasty.* 2013; 28:521–525. [PubMed: 23146585]
4. Choi JK, Gardner TR, Yoon E, et al. The effect of fixation technique on the stiffness of comminuted Vancouver B1 periprosthetic femur fractures. *J. Arthroplasty.* 2010; 25:124–128. [PubMed: 20558031]
5. Dennis MG, Simon JA, Kummer FJ, et al. Fixation of periprosthetic femoral shaft fractures occurring at the tip of the stem: A biomechanical study of 5 techniques. *J. Arthroplasty.* 2000; 15:523–528. [PubMed: 10884215]
6. Dubov A, Kim SYR, Shah S, et al. The biomechanics of plate repair of periprosthetic femur fractures near the tip of a total hip implant: the effect of cable-screw position. *Proc. Inst. Mech. Eng. [H].* 2011; 225:857–865.
7. Fulkerson E, Koval K, Preston CF, et al. Fixation of periprosthetic femoral shaft fractures associated with cemented femoral stems: a biomechanical comparison of locked plating and conventional cable plates. *J. Orthop. Trauma.* 2006; 20:89–93. [PubMed: 16462560]

8. Konstantinidis L, Hauschild O, Beckmann NA, et al. Treatment of periprosthetic femoral fractures with two different minimal invasive angle-stable plates: Biomechanical comparison studies on cadaveric bones. *Injury*. 2010; 41:1256–1261. [PubMed: 21288467]
9. Lenz M, Gueorguiev B, Joseph S, et al. Angulated locking plate in periprosthetic proximal femur fractures: biomechanical testing of a new prototype plate. *Arch. Orthop. Trauma Surg*. 2012; 132:1437–1444. [PubMed: 22707211]
10. Lenz M, Perren SM, Richards RG, et al. Biomechanical performance of different cable and wire cerclage configurations. *Int. Orthop*. 2013; 37:125–130. [PubMed: 23142812]
11. Lenz M, Perren SM, Gueorguiev B, et al. A biomechanical study on proximal plate fixation techniques in periprosthetic femur fractures. *Injury*. 2013
12. Lenz M, Windolf M, Mückley T, et al. The locking attachment plate for proximal fixation of periprosthetic femur fractures--a biomechanical comparison of two techniques. *Int. Orthop*. 2012; 36:1915–1921. [PubMed: 22638607]
13. Lever JP, Zdero R, Nousiainen MT, et al. The biomechanical analysis of three plating fixation systems for periprosthetic femoral fracture near the tip of a total hip arthroplasty. *J. Orthop. Surg*. 2010; 5:45.
14. Moazen M, Jones AC, Jin Z, et al. Periprosthetic fracture fixation of the femur following total hip arthroplasty: A review of biomechanical testing. *Clin. Biomech*. 2011; 26:13–22.
15. Shah S, Kim SYR, Dubov A, et al. The biomechanics of plate fixation of periprosthetic femoral fractures near the tip of a total hip implant: cables, screws, or both? *Proc. Inst. Mech. Eng. [H]*. 2011; 225:845–856.
16. Talbot M, Zdero R, Schemitsch EH. Cyclic loading of periprosthetic fracture fixation constructs. *J. Trauma*. 2008; 64:1308–1312. [PubMed: 18469655]
17. Wähnert D, Schröder R, Schulze M, et al. Biomechanical comparison of two angular stable plate constructions for periprosthetic femur fracture fixation. *Int. Orthop*. 2013
18. Wilson D, Frei H, Masri BA, et al. A biomechanical study comparing cortical onlay allograft struts and plates in the treatment of periprosthetic femoral fractures. *Clin. Biomech*. 2005; 20:70–76.
19. Zdero R, Walker R, Waddell JP, et al. Biomechanical Evaluation of Periprosthetic Femoral Fracture Fixation. *J Bone Jt. Surg Am*. 2008; 90:1068–1077.
20. Bergmann G, Graichen F, Rohlmann A, et al. Realistic loads for testing hip implants. *Biomed. Mater. Eng*. 2010; 20:65–75. [PubMed: 20592444]
21. Pletka JD, Marsland D, Belkoff SM, et al. Biomechanical comparison of 2 different locking plate fixation methods in vancouver b1 periprosthetic femur fractures. *Geriatr. Orthop. Surg. Rehabil*. 2011; 2:51–55. [PubMed: 23569670]
22. Bryant GK, Morshed S, Agel J, et al. Isolated locked compression plating for Vancouver Type B1 periprosthetic femoral fractures. *Injury*. 2009; 40:1180–1186. [PubMed: 19539924]
23. Old AB, McGrory BJ, White RR, et al. Fixation of Vancouver B1 peri-prosthetic fractures by broad metal plates without the application of strut allografts. *J. Bone Joint Surg. Br*. 2006; 88:1425–1429. [PubMed: 17075084]
24. Ricci WM, Borrelli J Jr. Operative management of periprosthetic femur fractures in the elderly using biological fracture reduction and fixation techniques. *Injury*. 2007; 38(Suppl 3):S53–58. [PubMed: 17723793]
25. Smith SD, Jansson KS, Philippon MJ, et al. Fracture mechanics of the femoral neck in a composite bone model: effects of platen geometry. *J. Biomech*. 2014; 47:602–606. [PubMed: 24315288]
26. Bitsakos C, Kerner J, Fisher I, et al. The effect of muscle loading on the simulation of bone remodelling in the proximal femur. *J. Biomech*. 2005; 38:133–139. [PubMed: 15519348]
27. de Winter JCF. Using the Student's t-test with extremely small sample sizes. *Practical Assessment, Research & Evaluation*. 2013; 18(10)
28. Marsh JL, Slongo TF, Agel J, et al. Fracture and Dislocation Classification Compendium - 2007: Orthopaedic Trauma Association Classification, Database and Outcomes Committee. *J Orthop Trauma*. 2007; 21(Supplement 10):S1–S133. [PubMed: 18277234]

Figure 1.
Custom fixture used to hold synthetic femur in a reproducible position, guide an oblique femoral neck cut, and direct initial drilling of the intramedullary canal.

Figure 2.
Fluoroscopic images of representative constructs from each group.

Figure 3. Schematics of mechanical loading setups for (A) axial and (B) torsional loadings.

Figure 4.

Maximum force reached, for both torsional (A) and axial (B) loadings. Bicortical constructs typically did not fail during testing; the above results for those two construct types were limited by force capacity and space limitations in the testing setup. Horizontal bars at the top indicate pairs of constructs having significantly different means ($p < 0.05$). Error bars indicate standard deviation, and diamonds represent the data from each individual specimen tested.

Figure 5.

Stiffness (up to 500 N load) for both torsional and axial loadings. Horizontal bars at the top indicate pairs of constructs having significantly different means ($p < 0.05$). Error bars indicate standard deviation, and diamonds represent the data from each individual specimen tested.

Construct preparation methods, and observed failure mechanisms as determined by visual inspection of specimens following mechanical loading and by video recorded during loading.

Table 1

Construct	Construct Preparation Method	Failure Mechanism	
		Axial Loading	Torsional Loading
(1) Cables	Threaded cerclage positioning pins were placed in holes 2, 4, & 6. Cerclage cables (1.7 mm) were tensioned to 50 kg using a cable tensioner and fixated with crimps.	Proximal cable failure and loosening of femur relative to plate	Loosening of femur relative to plate
(2) Unicortical (locking screws)	5 mm dia. ×12 mm long locked screws were placed at holes 1, 3, 4, & 6. All locking screws were tightened using a torque limiting screw driver.	Catastrophic femur fracture (cracking stemming from unicortical screw insertion sites)	Femur fracture with little or no displacement (cracking stemming from unicortical screw insertion sites)
(3) Cables + Unicortical (locking screws)	1.7mm cables were placed at holes 2 & 5 using the same manner as above, with 5mm dia. ×12 mm long locked unicortical screws placed at holes 1, 3, 4, & 6.	Femur fracture with little or no displacement (cracking stemming from unicortical screw insertion sites)	Femur fracture with little or no displacement (cracking stemming from unicortical screw insertion sites)
(4) Bicortical SS LCP	5mm dia. ×12 mm long unicortical screws were placed at holes 1, 3, 4, & 6. A 4 hole Synthes 3.5mm Locking Attachment Plate (LAP) for LCP 4.5/5.0 was fastened at holes 2 & 5 with four 3.5mm locking screws placed under fluoroscopic guidance into each attachment plate (8 additional screws).	Test machine limit reached before catastrophic failure; Femur fracture with no displacement (cracking typically stemming from unicortical screw insertion sites)	Plastic deformation of plate
(5) Bicortical Ti NCB	NCB 9-hole plate was applied under fluoroscopic guidance using 5mm NCB screws placed bicortically at holes 1, 2, 4, 5, 7, & 8. The screws were then fixed in place with NCB locking caps using a torque limiting screw driver.	Test machine limit reached before catastrophic failure; Femur fracture with no displacement	2 of 3 specimens: Fracture of cement and/or bone-cement interface at implant neck