A biomechanical comparison of static versus dynamic lag screw modes for cephalomedullary nails used to fix unstable peritrochanteric fractures

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BACKGROUND: The gamma nail has an option to statically lock its lag screw (static mode) or to allow its lag screw to move within the nail (dynamic mode). The purpose of this study was to compare the biomechanical stiffness of static and dynamic lag screw modes for a cephalomedullary nail used to fix an unstable peritrochanteric fracture. Unstable four-part peritrochanteric fractures were created in 30 synthetic femurs and fixed with Long Gamma 3 Nails. Mechanical tests were conducted for axial, lateral, and torsional stiffness with intact femurs, femur-nail constructs with static lag screw mode, and femur-nail constructs with dynamic lag screw mode. A paired Student’s t test was used for all statistical comparisons between test groups. Axial and torsional stiffness of intact femurs was significantly greater than femur-nail constructs (p < 0.01 all comparisons), whereas lateral stiffness was significantly less (p < 0.01 all comparisons). Axial stiffness of the femur-nail construct was significantly greater (p < 0.01) in static mode (484.3 N/mm ± 80.2 N/mm) than in dynamic mode (424.1 N/mm ± 78.0 N/mm). Lateral stiffness was significantly greater (p < 0.01) in static mode (113.9 N/mm ± 8.4 N/mm) than in dynamic mode (109.5 N/mm ± 8.8 N/mm). Torsional stiffness was significantly greater (p = 0.02) in dynamic mode (114.5 N/mm ± 28.2 N/mm) than in static mode (111.7 N/mm ± 27.0 N/mm).

RESULTS: There is a 60 N/mm (12.4%) reduction in axial stiffness when the lag screw is in dynamic mode. Given the statistically significant reduction in axial and lateral stiffness with use of the dynamic mode, static lag screw mode should be further explored clinically for treatment of unstable peritrochanteric fractures. (J Trauma. 2012;72: E65–E70. Copyright © 2012 by Lippincott Williams & Wilkins)

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KEY WORDS: Cephalomedullary nail; gamma nail; hip fracture; biomechanical.

Cephalomedullary nails have clinically and biomechanically proven to be excellent implants for the fixation of unstable trochanteric hip fractures. Cephalomedullary nails may be inserted through a small incision with minimal blood loss and short operative times and may result in lower rates of malunion and fracture collapse than extramedullary sliding hip screws. Reduction of displaced fractures may be more technically challenging with the use of a cephalomedullary nail. Improved implant designs and surgical techniques have led to a reduction in perioperative complications that were once associated with cephalomedullary nails (e.g., femur fracture and lag screw cut-out). It is important to continue to identify factors that may influence the biomechanics of the cephalomedullary nail when used to fix an unstable hip fracture. One such factor, which is determined at the time of surgery, is dynamization of the lag screw. Cephalomedullary nails rely on a large lag screw that provides fixation into the femoral head. There is an option to statically lock the lag screw (static mode) or to allow the lag screw to move within the nail to compress the intertrochanteric fracture (dynamic mode). Dynamic mode, or dynamization of the lag screw, allows for controlled compression of the intertrochanteric fracture line as the lag screw is allowed to move linearly within the aperture of the nail during weight bearing. The lag screw is prevented from rotating by interlocking of the set screw (Fig. 1) with grooves on the lag screw. Static mode involves fully tightening the set screw onto the lag screw so that linear movement of the lag screw is prevented by friction.

Although the dynamic lag screw mode allows for fracture site compression, there is concern that this mode may significantly reduce femur-nail construct stiffness when used to treat unstable hip fractures. As such, many surgeons prefer the use of the static lag screw mode when treating these types of fractures. To date, there are no biomechanical or clinical studies comparing static and
dynamic lag screw modes for cephalomedullary nails. The purpose of this study was to determine whether there is a significant difference in axial, lateral, and/or torsional stiffness between static and dynamic modes when a cephalomedullary nail is used to fix an unstable peritrochanteric hip fracture.

MATERIALS AND METHODS

General Approach

Thirty intact synthetic femurs were mechanically tested to obtain axial, lateral, and torsional stiffness baseline values. Unstable four-part intertrochanteric fractures were then created, reduced, and repaired using a cephalomedullary nail equipped with a lag screw in the same 30 femurs. With the lag screw in static mode (i.e., lag screw position was fixed in the cephalomedullary nail proximal hole) and then in dynamic mode (i.e., lag screw was permitted to slide in the cephalomedullary nail proximal hole), the same mechanical tests were repeated on all 30 femur-nail constructs. All intact and instrumented specimens were tested for axial, lateral, and torsional stiffness using an Instron 8874 mechanical tester (Instron, Canton, MA). Load cell characteristics included a force capacity of \( \pm 25 \) kN, a resolution of 0.1 N, and an accuracy of \( \pm 0.5\% \). Statistical comparisons were performed.

Femur Selection

The study used 30 large, left, fourth generation composite femurs (Model #3406; Pacific Research Laboratories, Vashon, WA) of 485 mm length and 16 mm intramedullary canal diameter. Cortical material (density \( = 1.64 \text{ g/cm}^3 \)) for the synthetic femurs was made from epoxy resin matrix with e-glass fibers, whereas cancellous material (density \( = 0.20 \text{ g/cm}^3 \)) was composed of polyurethane foam matrix containing e-glass fibers. The long axes of the synthetic femurs were oriented vertically in both the coronal and sagittal planes. The femoral condyles were shaved medially and distally and then inserted 50 mm into steel square-tube chambers that were filled with anchoring cement.

Axial Stiffness Tests

Each specimen was oriented in 15 degrees of adduction in the coronal plane and aligned vertically in the sagittal plane to mimic one-legged stance (Fig. 2). Adduction angles of 15 degrees to 25 degrees have been shown to simulate physiologic loading of the proximal femur in one-legged stance phase of gait and has been used in previous biomechanical studies. Distally, each specimen was fixed rigidly in a vice. Proximally, the femoral head was inserted into an acetabulum-type cup that was cut out of a stainless steel cylindrical block. The femoral head was free to rotate inside the cup. A vertical force was applied at the apex of the femoral head using displacement control (wave form = linear ramp up/down, max displacement = 1 mm, load rate = 5 mm/min, and preload = 50 N).

Lateral Stiffness Tests

Each specimen was fixed horizontally onto a test jig, and the femoral head was able to slide under a force application plate. An aluminum support triangle with rounded tip was placed at 200 mm from the top of the square-tube cement chamber to minimize long-axis bending (Fig. 2). A vertical force was then applied to each femoral head using displacement control (wave form = linear ramp up/down, max displacement = 2 mm, load rate = 5 mm/min, and preload = 50 N) to generate bending in the coronal plane.

Torsional Stiffness Tests

Each specimen was positioned horizontally onto a test jig, and the femoral head was able to slide underneath a force application plate. An aluminum triangle support with rounded tip was placed 25 mm distal to the lesser trochanter to
minimize long-axis bending (Fig. 2). An offset vertical force was applied to the anterior side of the femur head using displacement control (wave form = linear ramp up/down, max displacement = 1 mm, load rate = 5 mm/min, and preload = 50 N), thereby generating torsion around the long axis.

**Calculation of Stiffness**

The slope of each force-displacement curve was used to calculate stiffness. Stiffness for each test was obtained from an average of three trials. Specimens were kept within the linear elastic region to prevent any permanent damage. The mean linearity coefficient for each test ranged from $r^2 = 0.92$ to 0.99, indicating specimens were tested within the linear elastic range.

**Creation of Fracture and Fixation With Cephalomedullary Nail**

Using an industrial band saw and an ad hoc positioning jig, an unstable four-part intertrochanteric fracture pattern was created having two fracture lines. The first fracture line was intertrochanteric, whereas the second fracture line was a wedge with a 20-mm medial defect to simulate loss of the medial calcar (Fig. 3). This unstable peritrochanteric fracture pattern is classified by the OTA/AO fracture compendium as subtrochanteric multifragmentary with extension into the greater trochanter (31-A3.3.1).16

Reduction of the fracture was performed and a cephalomedullary nail (Long Gamma 3 nail, Left 420 mm × 11 mm, 120°; Stryker, Mahwah, NJ) was used for fixation of the fracture. The Gamma 3 Targeting Device (Stryker) was used to ensure appropriate position of the guide wire for the lag screw. Position of the guide wire was confirmed using fluoroscopy before insertion of the lag screw. The lag screw was placed, and fluoroscopy was used to ensure that position and tip-apex distance was similar for all 30 femur-nail constructs. One distal locking screw was placed in the static locking hole. All 30 femur-nail constructs were tested for axial, lateral, and torsional stiffness with the lag screw in static mode (i.e., the set screw was fully tightened). These tests were repeated with the lag screw in dynamic mode (i.e., the set screw was fully tightened and then untightened by a 1/4 to 1/2 turn).

**Statistical Analysis**

All statistical calculations were performed using SPSS version 16.0 (SPSS Inc., Chicago, IL). Statistical comparisons were performed using a paired $t$ test with significance set at $p < 0.05$. Statistical comparisons were performed for axial, lateral, and torsional stiffness for each of the following pairings: (1) intact femur versus femur-nail construct in static lag screw mode, (2) intact femur versus femur-nail construct in dynamic lag screw mode, and (3) femur-nail construct in static lag screw mode versus femur-nail construct in dynamic lag screw mode.

A two-tailed post hoc power analysis was performed to determine whether there was an adequate number of specimens per group and therefore adequate power to detect all actual statistical differences using a paired $t$ test for comparisons, i.e., avoiding type II error. Alpha error was set at 0.05, effect size was calculated with mean difference and SD of the mean difference, and adequate power was defined as 80% or greater (i.e., beta error $<0.2$).
RESULTS

Axial Stiffness

The axial stiffness tests on the intact femurs, femur-nail constructs in static mode, and femur-nail constructs in dynamic mode yielded means (and SD) of 1900.7 N/mm ± 149.2 N/mm, 484.3 N/mm ± 80.2 N/mm, and 424.1 N/mm ± 78.0 N/mm, respectively (Fig. 4). Axial stiffness of the intact femurs was significantly greater than the femur-nail constructs in static and dynamic modes ($p < 0.01$ for both comparisons). For the femur-nail constructs, axial stiffness was significantly greater with the lag screw in static mode than in dynamic mode ($p < 0.01$).

Lateral Stiffness

The lateral stiffness tests on the intact femurs, femur-nail constructs in static mode, and femur-nail constructs in dynamic mode yielded means of 89.8 N/mm ± 6.5 N/mm, 113.9 N/mm ± 8.4 N/mm, and 109.5 N/mm ± 8.8 N/mm, respectively (Fig. 5). Lateral stiffness of the intact femurs was significantly less than the femur-nail constructs in static and dynamic modes ($p < 0.01$ for both comparisons). For the femur-nail constructs, lateral stiffness was significantly greater with the lag screw in static mode than in dynamic mode ($p < 0.01$).

Torsional Stiffness

The torsional stiffness tests on the intact femurs, femur-nail constructs in static mode, and femur-nail constructs in dynamic mode yielded means of 258.4 N/mm ± 17.3 N/mm, 111.7 N/mm ± 27.0 N/mm, and 114.5 N/mm ± 28.2 N/mm, respectively (Fig. 6). Torsional stiffness of the intact femurs was significantly greater than the femur-nail constructs in static and dynamic modes ($p < 0.01$ for both comparisons). For the femur-nail constructs, torsional stiffness was significantly less with the lag screw in static mode than in dynamic mode ($p < 0.02$).

Post Hoc Power Analysis

A two-tailed post hoc power analysis with $\alpha = 0.05$ was conducted for the paired t test comparisons. For the comparison of intact femurs to femur-nail constructs with lag screw in static mode, the powers were 100%, 100%, and 100% for axial, lateral, and torsional stiffness, respectively. For the comparison of intact femurs to femur-nail constructs with lag screw in dynamic mode, the powers were 100%, 100%, and 100% for axial, lateral, and torsional stiffness, respectively. For the comparison femur-nail constructs with lag screw in static mode to dynamic mode, the powers were 100%, 93.3%, and 63.3% for axial, lateral, and torsional stiffness, respectively.

Figure 5. Graph of mean lateral stiffness for intact femurs, femur-nail constructs with lag screw in static mode, and femur-nail constructs with lag screw in dynamic mode. Values represent mean ± 1 SD.

Figure 6. Graph of mean torsional stiffness for intact femurs, femur-nail constructs with lag screw in static mode, and femur-nail constructs with lag screw in dynamic mode. Values represent mean ± 1 SD.
stiffness, respectively. Using 80% as the minimum for a good study design that avoids statistical type II error, results showed that eight of nine statistical pairwise comparisons were more than adequately powered, whereas only one pairwise comparison was slightly underpowered.

**DISCUSSION**

Dynamic lag screws were first employed with extramedullary sliding hip screw implants to allow for controlled compression of intertrochanteric hip fractures. This effectively changes a shear force into a compressive force at the primary fracture site that runs along the intertrochanteric line. Fractures of the lateral wall, subtrochanteric fractures, and reverse oblique intertrochanteric fractures render the sliding hip screw biomechanically ineffective. With these types of fractures, there is no lateral buttress to prevent continued collapse of the lag screw, and, therefore, femoral shaft medialization occurs. Femoral shaft medialization distorts the normal proximal femoral anatomy, reduces tension in the abductor muscles, and causes postoperative pain.

Does changing the lag screw mode from static to dynamic in a cephalomedullary nail cause axial stiffness to more closely approximate a sliding hip screw? The results of this study show that there was a 12.4% (60.2 N/mm) reduction in axial stiffness of the cephalomedullary nail when the lag screw was changed from static to dynamic mode. Two previous biomechanical studies were identified that compared axial stiffness of a sliding hip screw implant to a cephalomedullary nail implant when used to treat an unstable peritrochanteric hip fracture. Kuzyk et al. showed a 13.2% and Mahomed et al. showed a 9.5% reduction in axial stiffness between the sliding hip screw and cephalomedullary femur-nail constructs. These values are similar to the change in axial stiffness when the lag screw was changed from static to dynamic mode in this study, suggesting that use of the dynamic mode in a cephalomedullary nail renders its axial stiffness similar to that of a sliding hip screw.

Principles from physics can explain the relative performance of the constructs during axial, lateral, and torsional load application. Axial and lateral stiffness were both reduced with a change from static to dynamic lag screw mode. Torsional stiffness, on the other hand, remained relatively unchanged with only a small difference noted. When testing axial and lateral stiffness, the force on the femoral head was applied such that a component of this applied force was directed parallel to the lag screw, thereby causing micromotion (Fig. 2). Such micromotion of the lag screw has the effect of behaving like a mechanical instability, resulting in a reduced mechanical stiffness. The force applied during torsional testing, however, was perpendicular to the lag screw; therefore there was no component of the force parallel to the lag screw that would cause such micromotion. Thus, axial and lateral stiffness decreased by 12.4% (60.2 N/mm) and 3.9% (4.4 N/mm), respectively, with dynamization of the lag screw, but torsional stiffness remained relatively unchanged with only a 2.4% difference (2.8 N/mm). We suspect that this difference between static and dynamic torsional stiffness might not be clinically relevant, but further investigation would need to be performed to conclusively demonstrate its importance.

Axial and torsional stiffness both decreased significantly from intact femur to femur-nail construct values. This is because of the creation of an unstable fracture (Fig. 3) that reduces both axial and torsional stiffness far greater than introduction of the stainless steel nail increases stiffness in these two planes. This has been noted in previous studies. Interestingly, lateral stiffness increased significantly from intact femur to femur-nail values (Fig. 6). This increase in lateral stiffness indicates that introduction of the nail increases stiffness greater than creation of the fracture reduces stiffness in the lateral plane. No previously published biomechanical studies have documented change in lateral stiffness from intact femur values. These findings would suggest that methods of improving implant fixation should be directed toward design features that improve stiffness in the axial and torsional modes.

Several potential limitations of this investigation should be noted related to synthetic bones, physiologic loading, and statistical methodology. First, some may consider the use of synthetic femurs to be a weakness of this study because material properties of synthetic bones may more closely simulate normal rather than osteoporotic bone. However, synthetic bones have a number of advantages over cadaveric bones for biomechanical testing. Geometric and material properties of synthetic bones (e.g., cortical thickness, canal size, neck-shaft angle, and bone mineral density) vary only slightly between specimens, whereas these same properties vary significantly among cadaveric bone specimens. Preservation and storage methods used for cadaveric bones (i.e., formalin, freezing, etc.) result in dehydration and degradation of material properties over time. Synthetic bones have been used successfully in previous biomechanical investigations on femurs, have shown excellent agreement in axial and torsional stiffness when compared with cadaveric femurs, and have demonstrated excellent concurrence of cortical and cancellous screw purchase results when compared with human bone. Finally, the relative intrastudy mechanical performance of static versus dynamic locking mode femur-nail constructs at present in synthetic femurs would be expected to be similar for an intrastudy mechanical comparison using human femurs. For these reasons, the use of synthetic femurs, rather than human femurs, was deemed appropriate in this investigation.

Second, accurate representation of the forces acting on the hip is difficult for any biomechanical study because of the presence of soft tissue and the fact that there are a number of muscles that exert force on the proximal femur and pull on the fracture fragments in different directions. Based on previous studies, axial loading of the femur in 15 degrees of adduction was an accurate summation of the major forces acting on the proximal femur during the one-legged stance phase of gait. Consequently, axial stiffness values for this study are similar to values recorded in previous biomechanical studies involving the gamma nail. Axial stiffness values for this study were 424.1 N/mm and 484.3 N/mm for femur-nail constructs, whereas previous studies have recorded val-
ues ranging from 432.0 N/mm² to cadaveric femurs to 656.4 N/mm² in synthetic femurs. These studies did not vary the mode of the lag screw; rather they compared the biomechanics of the gamma nail to other implants.

Third, the study was adequately powered for all axial and lateral stiffness comparisons and for torsion comparisons of intact femur to femur-nail constructs. However, the study was slightly underpowered for the torsional stiffness comparison of static versus dynamic modes. This was caused by the large between-sample variation for torsional testing of the femur-nail constructs (Fig. 6). Much of this between-sample variation was effectively removed by use of a paired samples t-test to compare specimens, and therefore, a significant difference was detected. However, the large amount of between-sample variation suggests that another factor (e.g., lag screw position in the femoral head or position of the distal locking screw) might have a greater impact on torsional stiffness than changing the lag screw from static to dynamic mode.

In conclusion, given the statistically significant reduction in axial and lateral stiffness when the cephalomedullary nail was used with the lag screw in dynamic mode, we suggest that the static lag screw mode should be further explored biomechanically and clinically with regard to treating unstable peritrochanteric fractures with a cephalomedullary nail. Dynamic lag screw mode may be used when using a cephalomedullary nail to treat a stable hip fracture pattern as axial stiffness in dynamic mode resembles that of an extramedullary sliding hip screw. To date, this is the only study in the literature that has experimentally examined the mechanical characteristics of static versus dynamic locking modes for femur-nail constructs that use a cephalomedullary nail to address this particular fracture pattern.

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DISCLOSURE

30 Gamma Nails donated by Stryker for research.

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