A biomechanical comparison of the two- and four-hole side-plate dynamic hip screw in an osteoporotic composite femur model

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Abstract

Objectives: The objectives of this study were (1) to compare the axial and torsional stiffness of a dynamic hip screw with a two- and four-hole side-plate in a synthetic model of a healed and stable intertrochanteric femur fracture and (2) to evaluate the load to failure, as well as propensity to peri-implant fracture. Methods: Fourth-generation synthetic composite femur models, simulating osteoporotic bone, were implanted with 135° dynamic hip screws (DHS) with either a two- or four-hole side-plate with or without a stable intertrochanteric fracture. Specimens were cyclically loaded up to a nondestructive load to determine the axial and torsional stiffness. Constructs were then loaded to failure in axial compression emulating physiologic forces. Failure load and location of the peri-implant fractures were recorded. Results: Axial and torsional stiffness did not differ significantly between the two- and four-hole constructs in either model. Likewise, there was no significant difference in the load to failure. In the intact femurs, failure occurred either at the end of the plate at the distal screw or through the lag screw hole. Conclusion: The results of this study demonstrate that DHS constructs with a two- or four-hole side-plate are biomechanically comparable with regard to axial and torsional stiffness and load to failure in an osteoporotic composite femur model. In a healed intertrochanteric fracture model, a two-hole construct did not appear to be more prone to peri-implant fracture. To date, a biomechanical comparison of these two implants with regard to torsional forces has not been reported.

Keywords

biomechanical, dynamic hip screw, hip fracture, intertrochanteric, sliding hip screw

Introduction

Hip fracture is a significant cause of impairment in the geriatric population. The annual incidence in the United States is approximately 296,000.1 Intertrochanteric fractures comprise around half of these fractures. Intertrochanteric hip fractures are extracapsular proximal femur fractures involving the area between the lesser and greater trochanters.2 Surgery is indicated for nearly all patients as nonoperative treatment has historically resulted in high morbidity and mortality rates.3 With few exceptions, the majority of these fractures are managed with internal fixation.2 The goals of surgical treatment are stable fixation and early mobilization. Though multiple implants have been developed for the stabilization of intertrochanteric fractures, the sliding hip screw has been regarded as the standard treatment for several decades.4 With their development in the 1990s, intramedullary devices have been gaining in popularity and are now preferred for certain fracture patterns.5
The dynamic hip screw (DHS) first patented by Ernst Pohl in Germany in 1951 had a plate to barrel angle of 135° with a two-hole side-plate. The first clinical description of the use of this implant in the English literature was published in 1955 and also featured a two-hole side-plate. With time, however, the four-hole side-plate became the standard choice of implant in intertrochanteric fractures. Nevertheless, multiple reports have demonstrated satisfactory outcomes and minimal complications with the use of a two-hole side-plate in stable intertrochanteric fractures.

Although a large number of biomechanical experiments assessing internal fixation of proximal femur fractures can be found in the orthopedic literature, very few have focused on the length of the side-plate. McLoughlin et al. performed a biomechanical study in cadaveric femurs comparing the strength and stiffness of a DHS with a two-hole and a four-hole side-plate. The femurs with an unstable fracture were axially loaded after instrumentation, and strain, fragment migration, and load to failure data were recorded. The authors found the two-hole side-plate to be biomechanically comparable to the four-hole side-plate, with paradoxically less fracture movement in the two-hole side-plate.

The torsional stiffness of a DHS with a two- or four-hole side-plate, however, was not assessed in this biomechanical study and has not been reported in the orthopedic literature. We hypothesized that axial stiffness would be comparable between a two- and four-hole side-plate but the two-hole would be less stiff in torsion compared to a four-hole. Predisposition to peri-implant fracture below the side-plate and fracture patterns has also not been evaluated. Peri-implant fractures have been documented below the tip of a short intramedullary nail. The end of the two-hole side-plate is closer to the subtrochanteric region than the four-hole implant. Therefore, we hypothesized that a two-hole side-plate would be more prone to peri-implant fracture and consequently fail at lower loads in a physiologic testing model.

**Materials and methods**

A total of 18 composite femurs with a customized density were obtained (fourth-generation, sawbones; Pacific Research Laboratories, Vashon, Washington, USA). The fourth-generation composite femurs were used because their biomechanical properties are similar to human cadaveric femur and have a decreased variability between specimens. To simulate osteoporotic trabecular bone, as referenced in Marmor et al., Sokol et al., and Sommers et al., these femurs were produced with lower density solid polyurethane cores at 10 pounds per cubic feet. The central canal was also widened to 18 mm, which thinned the diaphyseal cortex to 3.7 mm.

Femurs were divided into two groups: intact, which simulated a healed intertrochanteric fracture, and stable intertrochanteric fracture groups. In each group, half of samples were plated with a two-hole construct, and the other half with a four-hole construct.

All implants were placed by the same surgeon (DR) with fluoroscopic guidance using a standard technique. Lag screws were inserted in the center–center position, and fluoroscopy was used to confirm a tip–apex distance less than 25 mm as described by Baumgaertner et al. (Figure 1). To mimic a stable trochanteric fracture (AO/OTA type 31-A1), the proximal femur was first pre-drilled for the lag screw. An osteotomy from the center of the greater trochanter to the apex of the lesser trochanter was then created using a band saw, and the implants were then inserted using the standard technique (Figure 2). The implants used in all experiments were provided by DePuy Synthes (West Chester, Pennsylvania, USA) and consisted...
of two- and four-hole side-plates with a 135° angle and a 95 mm lag screw. All cortical screws placed in the shaft were 4.5 mm large fragment screws that were 40 mm in length. Several femurs from those initially obtained were sacrificed to develop the experimental models.

**Biomechanical testing**

Nondestructive cyclic loading was performed initially. Fourteen femurs were assigned; however, one was lost during trial testing. Thus, seven intact (three 2-hole and four 4-hole) and six osteotomized femurs were tested in axial compression and torsion. Femurs were secured distally in a mold made of a low melting point metal alloy (Cerrobend, Chicago, Illinois, USA) in a fashion that when the metal base was secured into the testing machine, the femurs were oriented vertically in the sagittal plane with 8° of adduction. The head of the femur was also secured in metal mold and gripped with a clamp attached to a universal material testing machine (E10000; Instron, Canton, Massachusetts, USA; Figure 3). Specimens were loaded for 10 cycles in axial compression at 700 N and then for 10 cycles in external rotation at 3 N m.

To compare the propensity to peri-implant fracture of a two- and four-hole DHS, a custom testing apparatus was constructed as described by Tsai et al. to produce a subtrochanteric fracture in a cadaveric femur. Femurs were clamped distally in a mold made of metal alloy in a similar manner as previous specimens. The actuator with a load cell attached applied a vertical load to a steel cross-bar. A nylon strap (mean breaking strength 18 kN), acting as an abductor tendon, was secured to the distal end of the bar. A polyethylene cup was placed over the femoral head.

The distance from the load application point to the center of the femoral head was 2.5 times the distance from the center of the femoral head to the abductor strap. The strap itself was secured to the proximal femur using a padded hose clamp. A rubber strip was used to pad the hose clamp to prevent impinging into the cortical layer, which was lightly fastened around the femur (Figure 4). Fourteen specimens (eight in the intact group and six in the fracture group) were then loaded to failure by applying an axial load at a rate of 1 mm/s.

**Data analysis**

The load and displacement data were recorded. The slope of the linear portion of the force–displacement curve was used to calculate axial stiffness. Torsional stiffness was determined from the moment–rotation curve in a similar fashion. The average of the last three cycles from each test was used in statistical analysis. The maximum load encountered during failure test was recorded as the failure load, along with the location and mechanism of failure. Two-way analysis of variance tests (factors: femur condition (intact, fractured) and DHS type (two- and four-hole)) were performed to detect the differences among the constructs in stiffness and failure load. For all comparisons, statistical significance was set at a p value <0.05.

**Results**

In the nondestructive stiffness tests, the mean axial stiffness of the intact femurs plated with a two- and four-hole DHS
was $1519 \pm 357$ N/mm and $1548 \pm 137$ N/mm, respectively. In the stable fracture model, the mean axial stiffness for the two- and four-hole groups was $751 \pm 124$ N/mm and $760 \pm 210$ N/mm, respectively. Torsional stiffness of intact femurs plated with two- and four-hole DHS was $8.05 \pm 1.57$ Nm/deg and $7.67 \pm 1.26$ Nm/deg, respectively. In femurs with an intertrochanteric osteotomy, the torsional stiffness was $3.58 \pm 1.18$ Nm/deg and $2.95 \pm 0.44$ Nm/deg for the two-and four-hole groups, respectively (Figure 5). The DHS type did not have any significant effect on either axial ($p = 0.88$) or torsional ($p = 0.47$) stiffness. There were no significant interactions between the femur condition and DHS type for axial ($p = 0.94$) and torsional ($p = 0.86$) stiffness.

During load to failure testing, the mean load at failure of an intact femur with a two-hole DHS was $990 \pm 124$ N and $1011 \pm 263$ N for the four-hole DHS group. In the fracture model, the mean load at failure of the two- and four-hole groups was $644 \pm 106$ N and $650 \pm 103$ N, respectively (Figure 6). The femur condition significantly affected the failure load ($p = 0.04$) but DHS type did not have any significant effect ($p = 0.89$). Osteotomized femurs failed at a lower load. There was no significant interaction between the femur condition and DHS type ($p = 0.94$). Femurs failed at one of two locations. Two of four femurs in the two-hole group failed below the plate at the most distal screw, producing a subtrochanteric peri-implant fracture, while the other two failed around the lag screw hole (Figure 7). In the four-hole group, failure occurred below the plate in only one femur, while the other three also failed at the lag screw hole. All of the osteotomized femurs failed at the lag screw hole.

**Discussion**

The DHS remains the gold standard for the fixation of stable intertrochanteric hip fractures. Despite a large body of literature evaluating proximal femur fracture fixation, the length of the side-plate has only been addressed in a few studies. Utilizing a healed intertrochanteric fracture model, Reich et al. measured the tensile forces on the
side-plate screws and determined that more than four screws was not necessary. In another biomechanical study evaluating screw tension, Yian et al. showed that adding a fourth screw did not substantially change the tension on the proximal three screws. McLoughlin et al. concluded that a two- and four-hole DHS constructs were biomechanically comparable in axial loading.

When the femoral head is loaded in stance phase, internal rotation secondary to femoral anteversion is restricted by the coupling of the femoral condyles to the ground, which leads to a torsional force acting on the femur. In anterior loading type activities, such as stair climbing, this torsion can reach up to 2.2% body weight-meters. Biomechanical comparison of torque in the fixation of intertrochanteric femur fractures using a DHS with either a two- or four-hole side-plate has not been reported in the orthopedic literature.

The susceptibility to peri-implant fracture below the side-plate in DHS constructs has also not been evaluated in the orthopedic literature. It has been documented that the presence of an implant or drill-hole in the subtrochanteric region of the femur may lead to an iatrogenic or peri-implant subtrochanteric fracture after femoral neck fixation. There is also some evidence that the tip of a proximal femur, and a biomechanical comparison between these two implants has not yet been reported in the orthopedic literature.

In this study, no significant difference in torsional and axial stiffness between a two- and four-hole DHS was found in both intact femurs and a stable intertrochanteric fracture model. Load to failure experiments also did not show any significant difference between a two-hole and a four-hole DHS in intact and osteotomized femurs. Although the testing conditions differed, our study appears to corroborate the conclusion of McLoughlin et al. that the two different side-plate lengths are biomechanically comparable.

A synthetic femur model was used in this study, as these femurs have been validated to be biomechanically similar to human bone with less interspecimen variability. The femurs used in our experiments were customized to more closely approximate osteoporotic bone, which is the underlying cause of geriatric hip fractures. Our femurs failed at lower loads than reported in McLoughlin et al. This may partly be explained by the lower density of our femurs.

The testing apparatus used in our study was designed to more closely approximate in vivo forces. Although the McLoughlin study used a cantilever type mechanism with an abductor strap for cyclical testing, this was replaced with a fixed acetabular cup with the abductor strap removed for failure testing. It is possible that load to failure testing utilizing the cantilever mechanism with an abductor strap leads to lower loads at failure due to a larger lever arm and different distribution of forces.

During failure testing, no implant failure or bending was observed in either the intact or fracture models. In the intact model, failure occurred at either the lag screw hole or below the plate at the most distal shaft screw. It is difficult to conclude that the two-hole side-plate DHS is not more prone to peri-implant fracture than a four-hole, as our testing apparatus did not consistently reproduce a peri-implant subtrochanteric fracture. However, it does appear that the plate, the distal-most screw, and the lag screw hole are weak points in the construct.

We recognize several limitations of the current study. The number of samples per group was small; however, a power analysis showed that between 2000 and 8000 samples would be needed to show a difference at 75% statistical power, indicating a small difference between the means. Although validated to be biomechanically similar to human bone, the synthetic surrogate femurs may behave differently than a human healed osteoporotic intertrochanteric fracture under excessive loading. A healed fracture with a DHS in place may have stress shielding present, which is not reproduced in the foam models. The axial stiffness testing was not performed using the cantilever mechanism in order to maintain similar conditions as the torsional stiffness testing.

The results of this study demonstrate that DHSs with a two- and four-hole side-plate are biomechanically similar with regard to axial and torsional stiffness, as well as load to failure, in an osteoporotic synthetic femur model. Torsion is a significant component of the forces acting on the proximal femur, and a biomechanical comparison between these two implants has not yet been reported in the orthopedic literature. Our results also suggest that a two-hole DHS is not more prone to peri-implant fracture due to the proximity of its distal screw to the subtrochanteric region, but there should be further consideration to evaluate this in a biomechanical testing model that will consistently reproduce a peri-implant fracture distal to the plate.

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