Biomechanical Evaluation of Four Methods for Internal Fixation of Comminuted Subtrochanteric Fractures

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Abstract: Subtrochanteric fractures are common and result in significant morbidity and mortality. Various kinds of implants have been used to fix it. The aim of this study was to compare the biomechanical performance of PFN, DHS, DCS, and the PFLP in the treatment of subtrochanteric comminuted fractures.

A total of 32 antiseptic human femurs from 16 donors were randomly allocated to 4 groups for fixation with PFN, DHS, DCS, and PFLP. A 2-cm cylindrical bone fragment was removed 1 cm below the lesser trochanter to simulate OTA/AO 32-C3.2 post instrumentation fracture. All specimens in single-leg stance situation were preloaded 5 times at 100 N in the axial direction to eliminate the time effect of relaxation and settling, followed by cyclic testing at a rate of 1 Hz with stepwise increasing load. Keeping the valley load at a constant level of 100 N during the entire cyclic test, the peak load, starting at 200 N, was increased by 100 N at 300-cycle steps until a maximum of 1500 cycles or until failure of the bone-implant construct occurred. Each specimen was kept unloaded under 100 N compression for 30 minutes between the 300-cycle steps.

Femoral head displacement after 1500 cycles was 1.09 mm ± 0.13 for PFN, 1.78 mm ± 0.25 for DHS, 2.63 mm ± 0.46 for DCS, and 2.26 mm ± 0.16 for PFLP, with significant difference between any 2 implants (P < 0.01). The required load to reach 1-mm femoral head displacement was 563.04 N ± 158.34 for PFN, 485.73 N ± 147.27 for DHS, 258.44 N ± 97.23 for DCS, and 332.68 N ± 100.34 for PFLP. Significant differences were detected between any 2 implants (P < 0.001), except between DCS and PFLP and between DHS and PFN. The number of cycles until 1-mm femoral head displacement was 1458 ± 277 for PFN, 908 ± 184 for DHS, 369 ± 116 for DCS, and 603 ± 162 for PFLP. Significant differences were detected between any 2 implants (P < 0.01), except between DCS and PFLP.

From biomechanical point of view, comminuted subtrochanteric fractures OTA/AO 32-C3.2 revealed in the current test setup highest fixation strength with PFN, followed by DHS, PFLP, and DCS.

Abbreviations: CCD = caput-collum-diaphyseal, DCS = dynamic condylar screw, DHS = dynamic hip screw, PFLP = proximal femoral locking plate, PFN = proximal femoral nail.

Introduction
Subtrochanteric fractures are a kind of troublesome one, with a fracture line that could extend to pyriform fossa of the femur or the middle 1/3 of the diaphysis of femur. They account for about 10% to 34% of all hip fractures, with a complication rate ranging from 19% to 32%. Open reduction and internal fixation is an important method for treatment of subtrochanteric fractures, and the key to success with this method is how to achieve stable fixation and osseous healing in order to move and bear loads in the early postoperative phase of the injured hip. Taking into account this requirement, diverse methods for internal fixations emerged, such as PFN (proximal femoral nail), DHS (dynamic hip screw), and DCS (dynamic condylar screw).

PFN is an intramedullary implant, which could minimize invasiveness and offer stable fixation. However, in a previous study, it is reported that the outcomes with this intramedullary fixation are not satisfactory, because of the diversity of fracture patterns and anatomical morphology. In addition, screw cut-out of the neck, the so-called Z and reverse Z effects, secondary diaphyseal femur fractures, malunion or delayed union are not uncommon. DHS and DCS are extramedullary implants which, in comparison to the intramedullary nailing, might minimize problems such as injury to the superior gluteal nerve, associated abductor weakness, and heterotopic ossification. However, longer operative time, instability of the medial part of the subtrochanteric region and refracturing after removal of the implant were reported in some studies.

Proximal femoral locking plate (PFLP) is a new type of extramedullary implant for treatment of subtrochanteric fractures, developed in recent years. It allows us to use multiple 130° fixed-angle screws and placement of a locking plate without the necessity to drill a large hole as for the lag screws with DHS, PFN, or DCS, thus reducing the amount of stress on the calcar aspect of the proximal femur at the time of fixation.

DCS, initially designed for fixation of distal femoral fractures, have also been applied to fix proximal femoral fractures in recent years. DHS is used as “golden standard” for intertrochanteric fractures treatment. The main difference between these 2 implants is the angle of the lag screw with respect to the plate.

Several biomechanical studies on subtrochanteric fractures fixation were previously conducted. Crist et al and Floyd et al both compared different locked plating constructs with the standard 95° angled blade Plate; Forward et al investigated the biomechanical performance of a cephalomedullary nail, a proximal femoral locking plate, and a 95° angled blade Plate; Kim et al explored the fixation strengths of a locking plate, a...
long DCS, and a long proximal femoral nail. In these studies, the implants made mostly of stainless steel, and high quasi-static or cyclic loading was applied until failure of the bone-implant construct. To the best of our knowledge, there are no existing biomechanical studies for direct comparison between PFN, DCS, DHS, and PFLP, all made of titanium alloy.

The aim of the present study was to evaluate the biomechanical behavior of comminuted subtrochanteric fractures treated with 4 different methods for internal fixation and offer some reference for clinical work.

MATERIALS AND METHODS

The present study was approved by the ethics committee of the Tianjin Medical University.

Specimens Preparation

A total of 32 human femurs were obtained from 16 donors with mean age of 65.3 years (range 56–73 years) and embalmed with 10% formalin for 6 to 12 months before their use. All femurs were stripped of soft tissues and radiographed to ensure no existing abnormalities that could affect the study results. The femurs were randomly divided into 4 groups with 8 specimens each for instrumentation with 240-mm PFN, 5-hole 130° DHS, 9-hole 95° DCS, and 6-hole PFLP (Da Bo Yingjing Medical instrument Corporation, Haicang industrial zone, Xiamen, China), followed by measurement of the length, average diaphyseal diameter (in the middle of the shaft), caput-collum-diaphyseal (CCD) angle, and anteversion angle of each specimen.

The specimens in each group were instrumented with the respective implant according to the producer’s guidelines. The used implants are illustrated in Figure 1, with further information given in Table 1. An unstable subtrochanteric OTA/AO 32-C3.2 fracture was simulated postinstrumentation by circumferentially removing a 2-cm segment of bone tissue beginning 1 cm below the lesser trochanter.

The distal part of each specimen was potted in a metal tube using polymethylmethacrylate to simulate single-leg stance with 15° adduction (lateral angulation) of the femur in the frontal plane and vertical position in the sagittal plane (Figure 2).

Biomechanical Testing

Biomechanical testing was performed on an Instron 8874 servohydraulic testing machine (Instron Corporation, 825 University Avenue, Norwood, MA, USA) with a 25kN load cell in a setup shown in Figure 2. The femoral head was axially loaded via a polymethylmethacrylate cup, attached to the machine actuator and simulating the acetabulum.

The loading protocol of each specimen comprised preloading and main parts. Axial compression preload was circulated 5 times from 0 to 100 N at a rate of 10 mm/min before the main test in order to eliminate time effect such as specimen relaxation and settling after preloading, each specimen was then cyclically tested at 1 Hz under stepwise incrementally increasing axial sinusoidal load applied to the femoral head. Keeping the valley load at a constant level of 100 N during the entire cyclic test, the peak load, starting at 200 N, was increased by 100 N at 300-cycle steps until a maximum of 1500 cycles or until failure of the bone-implant construct occurred. The specimen was kept unloaded under 100 N compression 30 minutes long between the 300-cycle steps.

Data Acquisition and Analysis

Machine data in terms of axial displacement and load were recorded during biomechanical testing at a rate of 20Hz from the machine controller. The following parameters of interest were identified and evaluated: specimen’s length, average

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FIGURE 1. Implants for internal fixation tested in the present study: (a) DCS, (b) PFN, (c) DHS, (d) PFLP. DCS = dynamic condylar screw; DHS = dynamic hip screw; PFLP = proximal femoral locking plate; PFN = proximal femoral nail.
RESULTS

Morphological Parameters

The results for the morphological parameters of interest are given in Table 2. No significant difference between the groups was found with regard to length, diameter, and CCD angle, $P = 0.671$, $P = 0.845$, and $P = 0.773$, respectively. Regarding the anteversion angle, significant difference was found between the DHS and PFLP ($P < 0.001$), as well as between the DHS and DCS ($P < 0.001$), with no further significant differences detected.

Femoral Head Axial Displacement

The femoral head axial displacement progressively increased during the cyclic test as the load and cycles increased. The biggest displacement after 1500 cycles was detected for DCS (2.63 mm), followed by PFLP (2.26 mm), DHS (1.78 mm), and PFN (1.09 mm) (Figure 3). Significant differences were detected between any 2 implants ($P < 0.01$).

Load to 1 mm Femoral Head Axial Displacement

The highest load leading to 1 mm femoral head axial displacement was registered for PFN (563.04 N), followed by DHS (485.73 N), PFLP (332.68 N), and DCS (258.44 N) (Figure 4). Significant differences were detected between any 2 implants ($P < 0.001$), except between DCS and PFLP and between DHS and PFN.

Cycles to 1 mm Femoral Head Axial Displacement

The highest cycles to 1 mm femoral head axial displacement were detected for PFN (1458), followed by DHS (908), PFLP (603), and DCS (369) (Figure 5). Significant differences were detected between any 2 implants ($P < 0.01$), except between DCS and PFLP.

DISCUSSION

In the present study, we applied cyclic discontinuous stepwise incremental loading during biomechanical testing to simulate gradually increasing weight-bearing in patients after treatment of subtrochanteric fractures. We chose to use embalmed femurs because of their similar properties to the fresh-frozen ones. There were some morphology differences among the embalmed femurs, which would affect the accuracy of the testing results. However, we tried our best to select femurs with similar length, diameter, CCD angle, and anteversion angle to reduce this influence. The most important finding of our research is that according to the present study design PFN proved to be the most stable implant for treatment of comminuted subtrochanteric fractures, followed by DHS, PFLP, and DCS.

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**TABLE 1. Description of the Used Implants for Fixation**

| Implants | Materials | Proximal Screws | Distal Screws |
|----------|-----------|-----------------|---------------|
| PFN      | Titanium alloy | Lag screw | Interlocking screw |
| DHS      | Titanium alloy | Lag screw | Interlocking screw |
| DCS      | Titanium alloy | Lag screw | Bicortical screw |
| PFLP*    | Titanium alloy | Locking screw | Locking screw |

PFLP = proximal femoral locking plate; PFN = proximal femoral nail.

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The proximal screws were arrayed as isosceles triangle with the tip toward the greater trochanter.

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FIGURE 2. Setup with a specimen mounted for biomechanical testing.
There are 4 main factors affecting the healing process of subtrochanteric fractures. First, the subtrochanteric region is with high stress concentration, resulting from both axial and bending forces generated from the eccentric loading applied to the femoral head. This leads to high compressive stress on the medial part of the subtrochanteric region, especially its medial-posterior cortex, and tensile stress on the lateral cortex of this region. A previous biomechanical study\textsuperscript{16} demonstrated that an 890 N man could generate compressive stresses of 8.3 MPa along the medial aspect of the proximal femur from 2.5 to 7.5 cm distal to the lesser trochanter. Frankel and Burstein\textsuperscript{17} showed that the hip joint reaction forces could reach 3 times of body weight upon muscle contraction. The stress on the femoral head could reach 4.9 times of body weight under slow walking.\textsuperscript{18} Second, because the subtrochanteric region is with compact bone, the blood supply of this region is less rich than cancellous bone. If the soft tissues are damaged excessively during the surgery, the healing process may be delayed. Third, many muscles adhere to the femur in the subtrochanteric region, resulting in complex loading circumstances within this region. Froimson\textsuperscript{19} described the deforming forces associated with subtrochanteric fractures as flexion and external rotation of the proximal fragment (caused by the iliopsoas and glutei) and adduction and shortening of the distal fragment (caused by the adductors and hamstrings). Last, keeping the fractured bone segments stationary relative to each other and allowing some micromovements (relative stability) can promote union of the fracture. Numerous studies have proved that micromovements could promote the formation and calcification of callus, and accelerate the fracture healing process.\textsuperscript{20,21} Those 4 factors lead to the difficulty for healing of the subtrochanteric fractures, especially the last one.

Our biomechanical study used a model with circumferentially removed 2 cm segment of bone starting 1 cm below the lesser trochanter in order to cause deficiency of the medial cortex. A medial buttress is important in order to minimize implant stress and prevent fatigue failure.\textsuperscript{22} Our aim was using this fracture model to compare the biomechanical performance of different internal fixations for subtrochanteric fractures under dynamic loading conditions. After 1500 cycles, the axial displacement of the femoral head with PFN fixation accounted for 61.2 percent compared to DHS, 48.2% in comparison to PFLP, and 41.4% versus DCS. Moreover, the cycles to 1 mm femoral head axial displacement after PFN treatment were also significantly bigger than all other 3 internal fixations and the load was significantly higher than DCS and PFLP, demonstrating that the biomechanical performance of PFN was superior to the other 3 fixation methods. Our conclusion was similar to previous published biomechanical studies.\textsuperscript{10–12,23–25} Pajarinen et al\textsuperscript{26} reported a series of 108 subtrochanteric fractures treated with PFN and indicated that patients treated with PFN could weight bear and take functional recovery training earlier. PFN is designed for intramedullary fixation, so its bending lever arm is shorter than that of extramedullary fixation devices and it

| Group/Implant | Length (mm) | Diameter \textsuperscript{a} (mm) | CCD Angle (°) | Anteversion Angle (°) |
|--------------|-------------|-------------------------------|--------------|-----------------------|
| PFN          | 401.8 ± 6.5 | 24.9 ± 0.9                    | 131.4 ± 2.3  | 13.6 ± 1.5            |
| DHS          | 413.2 ± 11.9| 21.3 ± 1.5                    | 129.5 ± 3.2  | 15.1 ± 1.1            |
| DCS          | 408.4 ± 9.1 | 22.7 ± 1.2                    | 132.7 ± 1.7  | 12.5 ± 1.2            |
| PFLP         | 418.6 ± 13.1| 23.8 ± 1.1                    | 134.4 ± 2.4  | 11.3 ± 0.9            |
| \textit{P} value | 0.671        | 0.845                         | 0.773        | <0.001                |

CCD = caput-collum-diaphyseal; DCS = dynamic condylar screw; DHS = dynamic hip screw; PFLP = proximal femoral locking plate; PFN = proximal femoral nail.

\textsuperscript{a}Average diaphyseal diameter, measured in the middle of the shaft.

FIGURE 3. Femoral head sink displacement during cyclic testing of the 4 implant systems in terms of mean and standard deviation.

FIGURE 4. Loads needed to reach 1 mm head sink displacement during testing of the 4 implant systems in terms of mean and standard deviation (* represents \textit{P} < 0.05 between the 2 implants).
could bear more compressive stresses of the medial aspect so as to reduce the incidence of varus malunion. Moreover, with its stronger brace force, the PFN can better prevent collapse of the medial cortex of subtrochanteric region, thus reducing the incidence of failure rate and displacements of the fracture fragments.27,28 The ability to place a percutaneous device may decrease surgical time, and studies have documented significantly less intraoperative blood loss by using intramedullary devices versus plate constructs.29 In addition, it is not necessary to strip the periosteum during nailing, which helps preserving the blood circulation of the injured part and enhances fracture healing.30 Studies reported that the healing rate of subtrochanteric fractures fixed by intramedullary nail was 95%.31 Nowadays, in order to preserve blood supply at the fracture site and promote healing, more and more orthopedists advocate minimally invasive surgery.32,33 Intramedullary nail caters to this concept and becomes a good choice of treatment for subtrochanteric fractures.

Because of the 8 mm diameter of the proximal screws of DCS and DHS, orthopedists should drill a big hole on the proximal part of the femur, causing a major damage. DCS is indicated to fix the condylar fractures of the femur, and this big hole is not a problem for the distal femur.34 With many years of clinical application, DCS was gradually used for subtrochanteric fractures, but its biomechanical performance was not always as good as those of DHS24 and PFLP.11 In our biomechanical study, after 1500 cycles, the femoral head axial displacement was 147.8% compared to DHS and PFN. The proximal angle of DHS was 130°, which makes the one part of the force acting on the femoral head to disintegrate to the lateral side of femur through the sliding screw and the other part of the force acting on the medial cortex of the subtrochanter region vertically. This makes the medial compressive stress of DHS less than that of DCS. Moreover, extramedullary fixation acts as a tension band in subtrochanteric fractures when the medial cortex is intact. Once the medial cortex is missing or the quality of the medial cortex is not good, the body weight and the forces caused by muscles in contraction will act on the plate, which will increase the incidence of fatigue breakage of the weak part of this internal fixation. When using DHS or DCS to fix subtrochanteric fractures, orthopedists should make a long incision and strip more soft tissues and periosteal, which will reduce the blood supply of the fracture part and impede the healing process.

PFLP as an internal fixation for proximal femoral fractures has emerged for several years35 as improvement of the distal femoral locking plate utilized to fix femoral condylar fractures. As a result, PFLP is more suitable for the anatomy of the proximal femur. Its locking screws could firmly joint the plate and proximal and distal parts of the femur as a whole. Compared to DHS and DCS, the pull-out strength of the locking screws is much higher than this of common screws, which reduces the incidence of screw migration and loosening. Moreover, PFLP allows for use of multiple screws, and the placement of this locking plate also does not require a large hole for the lag diameter screw, which reduces the amount of stress on the calcaneal aspect of a proximal femur fracture at the time of fixation.35 Furthermore, PFLP is designed for less invasive placement, which is to the benefit of fracture healing. In our biomechanical study, after 1500 cycles, the femoral head axial displacement of PFLP was more than that of PFN and DHS, but less than that of DCS, and the differences were significant. The cycles to 1 mm femoral head axial displacement of PFLP were more than DCS, but the difference was not significant. Our findings were not similar to Kim’s study.11

In terms of study limitations, human fresh-frozen cadaveric bone which is an ideal test material was not used in the present study. We did not test the biomechanical behavior of the internal fixations under enough long cyclic loading. This was another weak point of this study. The third one was that the samples in this study were a little small. This could have affected the reliability of the results. The last one was that we didn’t compare anteroposterior and mediolateral translation or rotation of the fracture surfaces in all 3 planes. Further study should be conducted to get more reliable results.

CONCLUSIONS

PFN was found to be the most stable implant for treatment of comminuted subtrochanteric fractures with the current test setup. Among the extramedullary fixations, the biomechanical behavior with DHS was better than that of PFLP and DCS, and PFLP performed better than DCS. However, due to the limitations of this study, further studies are necessary to confirm the current findings.

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