EXPERIMENTAL STUDY

Acta Orthop Traumatol Turc 2015;49(4):426–432
doi: 10.3944/AOTT.2015.14.0306

Biomechanical outcome of proximal femoral nail antirotation is superior to proximal femoral locking compression plate for reverse oblique intertrochanteric fractures: a biomechanical study of intertrochanteric fractures

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Objective: Reverse obliquity intertrochanteric fractures are a challenge for orthopedic surgeons. The optimal internal fixation for repairing this type of unstable intertrochanteric fractures remains controversial. This study aimed to compare the biomechanical properties in axial load and cyclical axial load of proximal femoral nail antirotation (PFNA) and proximal femoral locking compression plate (PFLCP) for fixation of reverse obliquity intertrochanteric fractures.

Methods: Sixteen embalmed cadaver femurs were sawed to simulate reverse obliquity intertrochanteric fracture and instrumented with PFNA or PFLCP. Axial loads and axial cyclic loads were applied to the femoral head by an Instron tester. If the implant-femur constructs did not fail, axial failure load was added to the remaining implant-femur constructs.

Results: Mean axial stiffness for PFNA was 21.10% greater than that of PFLCP. Cyclic axial loading caused significantly less (p=0.022) mean irreversible deformation in PFNA (3.43 mm) than in PFLCP (4.34 mm). Significantly less (p=0.002) mean total deformation was detected in PFNA (6.16 mm) than in PFLCP (8.67 mm).

Conclusion: For fixing reverse obliquity intertrochanteric fractures, PFNA is superior to PFLCP under axial load.

Keywords: Biomechanical testing; proximal femoral nail antirotation; proximal femoral locking compression plate; reverse obliquity intertrochanteric fractures.

Intertrochanteric fractures are common and result in significant morbidity and mortality, which bring great financial burden to society. The incidence of reverse obliquity intertrochanteric fractures, an unstable type of intertrochanteric fracture, has been reported variably. It has been estimated to account for approximately 2% of all hip fractures.[1] Two researchers[2,3] reported the incidence as 4.3% and 15%, respectively. Another re-

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Submitted: August 25, 2014 **Accepted:** January 3, 2015

Available online at
www.aott.org.tr
doi: 10.3944/AOTT.2015.14.0306
QR (Quick Response) Code

searcher[4] found an incidence of 23% in busy centralized trauma centers.

Treatment of reverse obliquity intertrochanteric fractures is still a challenge for orthopedic surgeons. In 2004, the AO/ASIF group developed proximal femoral nail antirotation (PFNA) to improve rotational and angular stability with 1 femoral neck helically shaped blade. Studies have shown that unstable proximal femoral fractures, including reverse obliquity intertrochanteric fractures, can be treated successfully with PFNA.[5–8]

More recently, locking plates developed especially for the proximal femur have become available and have gained increasing popularity for the management of complex proximal femur fractures. Biomechanical studies have shown locking plates to achieve stronger and stiffer fixation than other angularly stable implants for fixing subtrochanteric fractures.[9–11] A Chinese clinical study[12] reported that proximal femoral locking compression plate (PFLCP) can be a feasible alternative for the treatment of intertrochanteric fractures. PFLCP allows for use of multiple fixed 125° angle screws, and the placement of a locking plate does not require creating a large hole for the lag diameter screw, thereby reducing the amount of stress on the calcar aspect of a proximal femur fracture at the time of fixation.[9] Additionally, it does not generate inferolateral sliding tendency of the proximal fracture fragment of reverse obliquity intertrochanteric fractures.

To the best of our knowledge, no comparative study of intramedullary nails and proximal femoral locking compression plates exists,[13] leading us to undertake this biomechanical comparative study. To offer enough stiffness and stabilize the fracture are critical aspects of treatment of this kind of fractures. The purpose of this study was to evaluate the differences in biomechanical properties of PFNA and PFLCP for fixing reverse obliquity intertrochanteric fractures, investigating whether PFNA would provide superior properties in axial stiffness, irreversible deformation, total deformation, and failure load than PFLCP.

Materials and methods

Our study was an *in vitro* study to test the properties of PFNA and PFLCP for treatment of reverse obliquity intertrochanteric fractures by biomechanical methods.

Sixteen embalmed intact adult cadaver femurs were obtained and assigned randomly to either the PFNA or PFLCP groups. All femurs were embalmed for 6–12 months. Ages ranged from 35–56 years (mean 43.5 years). The femurs were stripped of all soft tissues and radiographed to ensure that there were no abnormal specimens that would later affect results. A 4 cm^2 region of each femoral head was screened with the use of dual energy X-ray absorptiometry (DEXA, Hologic, Bedford, MA, USA) to determine whether the femurs were osteoporotic. The length, diameter between medial and lateral, neck-shaft angle, and anteversion angle of the femurs were measured. The femurs were kept frozen at -20°C and then thawed at room temperature before device implantation and mechanical tests. This experiment was approved by the ethics committee of Tianjin Medical University.

Instron 8874 dynamic multidimensional biomechanical fatigue testing machine (Instron Corporation, Norwood, MA, USA) was applied to test the specimens. The Instron 8874 testing machine features up to ±25.000 N axial force capacity and ±100 Nm torque capacity. Containing a patented Dynacell load cell, it allows for precision and accuracy of all measurements while compensating for inertia and reducing dynamic load and operator errors. PFNA and PFLCP (donated by Da Bo Yingjing Medical Instrument Corporation, Xiamen, China), both widely used in China, were tested in this experiment. All implants were made of titanium alloy, and the angle of proximal screws of PFLCP was 125°, while the angle of proximal nail of PFNA was 130°. The elasticity modulus of the implants was equal.

Osteotomies were created in all femurs at an angle of 33° running inferolaterally from the lesser trochanter to simulate reverse obliquity intertrochanteric fracture.[14] All fracture lines were similar. After the fracture model was created, all femurs were instrumented following the recommendations of the respective implant manufacturers under radiological image intensifier control. All fixations were performed by the same person in the same manner. There was no gap between the fracture fragments. The implant-femur constructs were then tested by X-ray to ensure optimal implant position was achieved. Each construct group after osteotomy and their radiographs (Figure 1) indicated the final position of the implants in the femurs.

Each specimen was potted in a metal tube using polymethylmethacrylate (PMMA) to simulate single-leg stance of the corpus femur adducted 15° at the frontal plane while vertical at the sagittal plane, maintaining medial rotation of 5°~10°[15] before testing. They were instrumented on the biomechanical testing machine in the single-leg stance position.

The load was applied to the head of the femur through a custom solid cup made of PMMA. The superior surface of the solid cup was fixed on the load cell to resist horizontal slide of the femoral head.

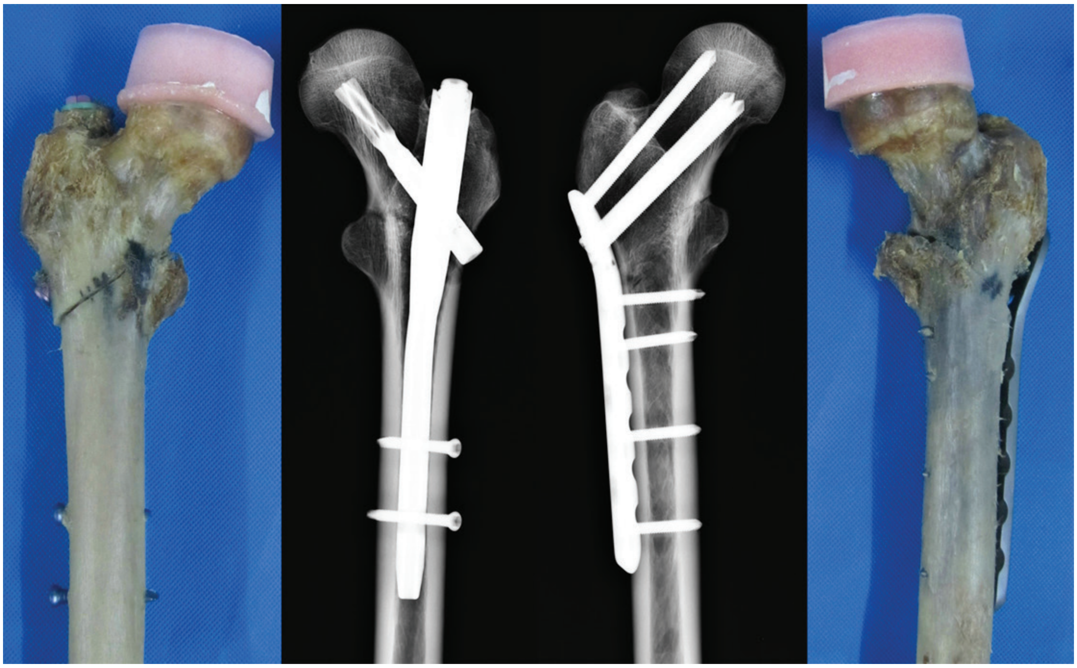

Fig. 1. Implant-femur constructs and their radiographs. [Color figure can be viewed in the online issue, which is available at www.aott.org.tr]

For axial loading test, preload was circulated 3 times under the same velocity (10 mm/min) and the same maximum load (100 N) before the official test in order to stabilize the construct. Prior to official testing, position of the load cell after preloading was manually adjusted to ensure the load cell was in contact with the solid cup and the compression load of the load cell was less than 1 N. This position of the load cell was set as the base line to record femoral head sink displacement. The construct was then loaded in compression at a loading rate of 10 mm/min from the base line position. Testing was stopped when 1,800 N was reached, prior to any visual loss of fixation.

Cyclic axial loading was performed following axial loading. The load form consisted of 100–300 N for 100 cycles, 100–400 N for 100 cycles, 100–500 N for 100 cycles ... 100–1,800 N for 100 cycles. Testing was conducted in a frequency control mode at 0.5 Hz. After this loading, if the implant-femur construct did not fail—failure was defined as new fracture seen on the femur or acute change in load-displacement curve indicating rapid change in displacement and loss of construct stability—the next cyclic axial loading was performed. Axial load was loaded on the construct from 100–1,800 N at 1 Hz for 3,000 cycles.

After cyclic loading, if the implant-femur constructs did not fail, they were then loaded to failure in single-leg stance position under axial loading through the solid cup at the rate of 10 mm/min. Failure was defined as the same as that in the cyclic loading test in our study. Failure load of each implant-femur construct was recorded. Femurs were regularly moistened during tests to avoid desiccation. However, no muscle loads were simulated in the single-leg stance model during the above testing procedures.

Load cell data were recorded using Bluehill 2 software (Instron Corporation, Norwood, MA, USA) and MAX software (Instron Corporation, Norwood, MA, USA). For axial testing, a load-displacement curve was plotted for each construct, and stiffness was calculated as the slope of the linear portion of the curve. For axial cyclic testing, total deformation was calculated by subtracting the amount of displacement present at the start of the first cycle (300 N) from the displacement observed at the end of cyclic loading. Irreversible deformation was calculated by subtracting the initial displacement from displacement present after reaching the point of load removal.

SPSS version 16.0 software (SPSS Inc., Chicago, IL, USA) was used to analyze the data. Statistical analyses were performed by using independent Student's t-test. The level of statistical significance was defined as $p<0.05$.

Results

After randomly grouping, there were no statistically significant differences among the construct groups regard-

Table 1. Comparison of measured parameters of femurs in each group.

Internal fixation	T-score	Length (mm)	Diameter# (mm)	Neck-shaft angle (°)	Anteversion angle (°)
PFNA	-1.23±0.40	400.77±19.04	21.83±2.03	129.64±6.06	10.35±2.70
PFLCP	-1.37±0.47	404.25±24.25	22.097±1.58	132.24±4.87	14.04±3.85
p	0.527	0.754	0.779	0.361	0.045

#Diameter between medial and lateral side of femoral shaft, measured at the middle of the shaft.

ing mean T-score, length, or diameter between medial and lateral side of femoral shaft and neck-shaft angle, with the exception of the anteversion angle of the femurs (p=0.045) (Table 1).

Mean axial stiffness for PFNA was 21.10% greater than that of the PFLCP (459.72 vs. 362.73 N/mm; p=0.001). No catastrophic failure was noted in either group after axial loading. One implant-femur construct in the PFLCP group was damaged during cyclic axial loading at 1,567 cycles, leaving 7 specimens in the PFLCP group on which to conduct statistic analysis. No catastrophic failure was noted in the PFNA group after cyclic axial loading. Cyclic axial loading caused significantly less mean irreversible deformation (p=0.022) in the PFNA group (3.43 mm) than in the PFLCP group (4.34 mm). Significantly less mean total deformation (p=0.002) was detected in the PFNA group (6.16 mm) than in PFLCP group (8.67 mm) (Table 2).

Failure loads of the implant-femur constructs were significantly different between the 2 groups (PFLCP: N=7, 2742±02 N; PFNA: N=8, 4119±677 N) (p<0.0001) (Figure 2). In the PFLCP group, failure modes were all fractured at the proximal part of the distal fracture fragment (Figure 3). Direction of the fracture line indicated that the implant-femur construct was bent as a result of axial failure load. In the PFNA group, 2 suffered diaphyseal fracture, 3 fractured around the proximal screw tail on the lateral side of the greater trochanter (the fracture line running horizontally from the lateral side of the greater trochanter to the medial side), 2 suffered lateral sliding of the blade, 1 did not have visual fracture, but the load-displacement curve became flat at the failure load.

Table 2. Stiffness and deformation of PFNA and PFLCP constructs.

	Implant	
	PFNA	PFLCP
Axial stiffness (N/mm)	n=8	n=8
Mean	459.72	362.73
SD	49.13	44.56
Difference	96.99	
p*	0.001	
Total deformation (mm)	n=8	n=7
Mean	6.16	8.67
SD	1.09	1.35
Difference	2.51	
p*	0.002	
Irreversible deformation (mm)	n=8	n=7
Mean	3.43	4.34
SD	0.63	0.71
Difference	0.91	
p*	0.022	

*Independent Student t-test; n: Number of specimens for analysis.

Discussion

In our biomechanical study, with the exception of the anteversion angle, there was no statistically significant difference regarding morphological characters and T-score among the femurs in different construct groups, which increased the comparability among the femurs. However, different anteversion angles could affect the value of the strain and the strain distribution at the proximal part of the femur.[16] Furthermore, reduced anteversion could result in decreased bone strength with an increased risk of fracture,[17] and low bone density (BMD) has been recognized as a reliable predictor of osteoporotic frac-

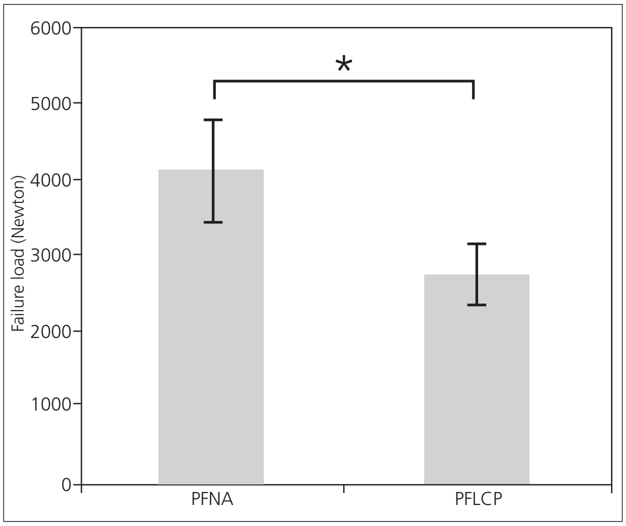

Fig. 2. Comparison of failure loads between the PFLCP and PFNA groups (Newton). *Significantly different.

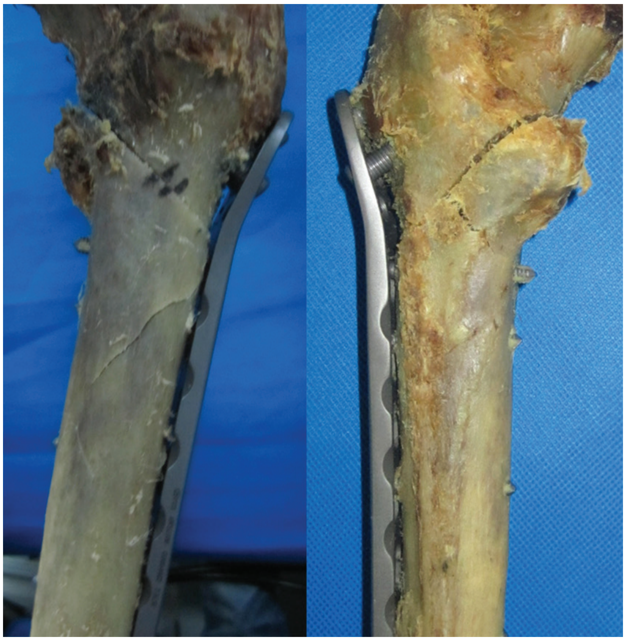

Fig. 3. Failure mode of the implant-femur construct in the PFLCP group during the axial damage loading. Left picture (anterior side of the femur): The fracture line runs inferomedially from the lateral side of the diaphysis behind the fracture model. Right picture (posterior side of the femur): The fracture line runs inferomedially from the middle of the fracture model line. [Color figure can be viewed in the online issue, which is available at www.aott.org.tr]

ture risk.[18] Thus, different anteversion and BMD were the limitations of our study. All implants used in our experiment were made of titanium alloy. Compared to stainless steel, titanium has impressive biocompatibility, high corrosion resistance, and specific mechanical properties. The elastic modulus of titanium is much closer to that of bone, which reduces the local stress concentration in the bone[19] and could reduce the rate of nonunion fractures.[20]

PFNA was developed by the AO/ASIF in 2004. The main design characteristic of the implant is the use of a single blade with a large surface area. Insertion of the blade compacts the cancellous bone. These characteristics provide optimal anchoring and stability when the implant is inserted into cancellous bone. Its shape provides an increased contact-area between bone and implant, rendering better purchase of the blade in the femoral head compared to a screw, therefore preventing or at least delaying rotation induced cutout,[6] which was the most critical complication of the intramedullary nail for fixing intertrochanteric fractures.[21,22] PFNA is central fixation which can reduce the force on the femoral head/neck stabilization implant by positioning the intramedullary device close to the weight-bearing axis of the femur.[23] Nail in the medullary and proximal part of femur could resist the proximal fracture fragment sliding laterally and the distal fracture fragment sliding medially, which overcomes the harmful force of reverse obliquity intertrochanteric fractures. Many clinical studies[6,8,24] have proven that PFNA is as effective as other implants in the treatment of intertrochanteric fractures.

As a new implant created to fix proximal femoral fractures, PFLCP achieved a high failure rate in the treatment of trochanteric fractures in the Western world.[25,26] However, other biomechanical studies on subtrochanteric femoral fracture models show that PFLCP provides greater or equivalent stability compared to other fixation techniques.[10,11] Nonetheless, no biomechanical study comparing PFLCP and PFNA was found for reverse obliquity intertrochanteric fractures. Only 1 biomechanical study comparing intramedullary nail and PFLCP for subtrochanteric fractures was found, which concludes that intramedullary nail construct is biomechanically superior to PFLCP construct.[27] This finding is similar to that reported in our study. PFLCP used in our study is widely used in China for intertrochanteric fractures.[12,28] This kind of PFLCP differs from what is used in the Western world. The proximal screws of PFLCP in the Western world are angular-stable, and the material is stainless steel. The proximal screws of PFLCP in China are parallel to each other, and the material is titanium alloy. One study showed that PFLCP can be a feasible alternative to the treatment of pertrochanteric fractures.[12] In this study, the overall technical complication rate for PFLCP treatment was only 2.7%, and rate of breakage of the implant was as low as 1.0%. However, no clinical studies have reported on PFLCP use for treating intertrochanteric fractures in the Western world. In our opinion, the strain concentration of the proximal femoral head screws of Western types of PFLCP was more severe than that of Chinese PFLCP as a result of the angular-stable function. Hence, 1 case report[29] showed that 4 intertrochanteric fracture cases treated with Western type PFLCP suffered breakage or loosening of the proximal femoral head locking screws. Additionally, 1 biomechanical study[27] reported that Western type PFLCP bended at the fracture site under the proximal femoral head locking screws after cyclical axial loading.

In our study, axial stiffness of PFLCP was lower than that of PFNA, and the total displacement and irreversible displacement were higher than that of PFNA. Because of the locking screws, fracture fragment was limited to slide between each other. To summarize the reasons why PFNA was superior to PFLCP for fixing reverse obliquity intertrochanteric fractures, first, PFL-

CP is fixed to the lateral cortex, whereas PFNA is fixed within the medullary canal; second, the moment of inertia of PFNA is less than that of PFLCP within the proximal femur; third, the thickness of PFLCP is 5 mm, whereas the thickness of PFNA is 16 mm within the plane of the applied moment.

PFNA had the much higher failure load than PFLCP in our study. This also could be attributed to the intramedullary fixation of PFNA.[30,31] Central fixation has a shorter bending moment than extramedullary fixation and could bear more compressive stress on the femoral head. The proximal loads on the femur are transferred via the device to the diaphysis. Thus, the structure of PFNA is load sharing, and the bone will consequently deform considerably less than when it is bearing the load fixed by PFLCP. In addition, the thickness of PFNA is much greater than that of PFLCP, so the stiffness of PFNA is higher than that of PFLCP, and PFNA could bear more compressive loads than PFLCP.

Many clinical studies have reported that PFLCP in the Western world has a high failure rate. Glassner et al. reported hardware failure in 4 cases using PFLCP, with broken plates in 2 cases and broken screws in the other 2.[32] Streubel et al. reported cumulative failure rates of 33% at 12-month follow-up, with failures occurring as varus collapse with screw cutout, proximal screw breakage, screw loosening with varus deformity, and plate fracture.[26] Zha et al. reported 1 implant breakage and no varus angulation in unstable intertrochanteric fracture fixed with PFLCP used in China.[12] Though the studies above do not share similar outcomes, we think that there was compression stress concentration on the fracture site, causing the fracture of the implant and the varus of femur. In the present study, we detected that the fracture line was running inferomedially on the proximal part of the distal fracture fragment, as in Figure 3 after axial damage loading, indicating that varus stress (tension stress on lateral side, compression stress on medial side) was concentrated on this level.

In the present study, the implant-femur constructs did not occur as cutout failure mode, which was attributed to the PFNA blade. Biomechanical tests have demonstrated that the blade has a significantly higher cutout resistance than commonly used screw systems.[33] Other failure modes detected in our study have all been reported by previous clinical studies.[6,8,24] One construct had no visual fracture, but the load-displacement curve became flat at the failure load. This might be attributed to the cutting of the blade in the femoral head or blade loosening.

There are some limitations in our study. First, fresh human cadaveric bone was not used. Second, osteotomy was performed using a saw, producing flat bony interfaces, whereas fractures in patients usually have irregular surfaces. Third, all soft tissues and ligaments were removed to produce a standardized osteotomy, unlike in the clinical situation. Fourth, we did not measure the fracture gap displacement, which could be explored in future studies. Fifth, there was no gap between the fracture lines, and we did not test the 3-part reverse oblique fracture, though we will test these fracture models in the future. Sixth, we did not test other positions of the femur, such as abduction, adduction, and flexion; nor did we test the proximal and distal strains on the implants.

In fixing reverse obliquity intertrochanteric fracture, PFNA could provide more stability than PFLCP under axial load because the PFNA construct is significantly stiffer, has a smaller total and irreversible displacement than the PFLCP construct, and could bear more failure loads.

Acknowledgements

The authors are grateful to the foundation items of the National Natural Science Foundation of China (81102607), the Science and Technology Research Projects of the Tianjin Municipal Health Bureau (11KG137, 12KG120), and the Key Technology R&D program of Tianjin (13ZCZDSY01700).

Conflics of Interest: No conflicts declared.

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