Pre-Clinical Research Report

Journal of INTERNATIONAL MEDICAL RESEARCH

Journal of International Medical Research
48(12) 1–12

DOI: 10.1177/0300060520950934
journals.sagepub.com/home/imr
SAGE

Design and biomechanical study of slide-poking external fixator for hip fracture

Hua-Biao Chen*, Hong-Bo Wu*, Min Chen and Yu-Liang Huang

Abstract

Background: Femoral head collapse and coxa vara lead to internal fixator failure in elderly patients with hip fracture. External fixator application is an optimal choice; however, the existing methods have many disadvantages.

Methods: Type 31-A1.3 hip fracture models were developed in nine pairs of 1-year-old fresh bovine corpse femur specimens. Each left femur specimen was fixed by a dynamic hip screw (control group), and each right femur specimen was fixed by the slide-poking external fixator (experimental group). Vertical loading and torsion tests were then performed in both groups.

Results: In the vertical loading experiment, a 1000-N load was implemented. The mean vertical downward displacement of the femoral head in the experimental and control groups was 1.49322 ± 0.116280 and 2.13656 ± 0.166374 mm, respectively. In the torsion experiment, when the torsion was increased to 10.0 Nm, the mean torsion angle in the experimental and control groups was $7.9733^\circ \pm 1.65704^\circ$ and $15.4889^\circ \pm 0.73228^\circ$, respectively. The slide-poking external fixator was significantly more resistant to compression and rotation than the dynamic hip screw.

Conclusion: The slide-poking external fixator for hip fractures that was designed and developed in this study can provide sufficient stability to resist compression and rotation in hip fractures.

Keywords

Design, hip fracture, external fixator, dynamic hip screw, biomechanics, stability

Date received: 9 April 2020; accepted: 28 July 2020

Department of Orthopaedic Surgery, Affiliated Huizhou Central People Hospital, Sun Yat-Sen University and Guangdong Medical University, Huizhou City, Guangdong Province, People's Republic of China

*These authors contributed equally to this work.

Corresponding author:
Yu-Liang Huang, Department of Orthopaedic Surgery, Affiliated Huizhou Central People Hospital, Sun Yat-Sen University and Guangdong Medical University, 41 Goose Ridge North Road, Huizhou City, Guangdong Province 516001, People's Republic of China.
Email: huangyi113@163.com

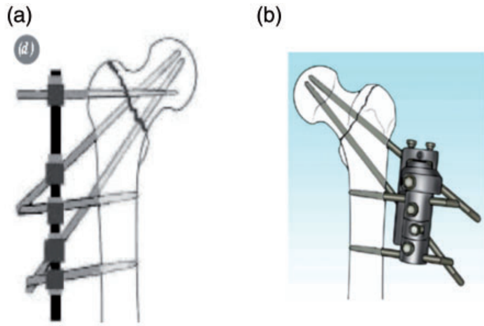

Figure 1. The two main types of external fixators used for treatment of hip fracture. (a) AO tubular external fixator. (b) Orthofix pertrochanteric fixator (Orthofix, Lewisville, TX, USA).

Introduction

Two surgical techniques are generally applied for treatment of hip fracture: external and internal fixation. Because of the frequent occurrence of osteoporosis in elderly patients, this population is at high risk of blood loss, varus deformity, limb shortening, and medial shift of the distal portion of the fracture[1] because of lag screw cut-out from the cortical bone.[2–4]

An external fixator has certain limitations, such as low patient satisfaction and a risk of pin infection. However, external fixation is optimal in certain cases. For example, elderly patients with osteoporosis-associated hip fracture are prone to develop failure of internal fixation; elderly patients also have greater risks associated with internal fixation because of concurrent limb thrombosis or severe organ dysfunction. Additionally, open fractures cannot be repaired by internal fixation. Finally, internal fixation devices are associated with a risk of osteomyelitis, tuberculosis, infection of the fixation device, and pathological fracture. Thus, external fixator application is an optimal choice in many cases.

The Orthofix pertrochanteric fixator (Orthofix, Lewisville, TX, USA)[5] and AO tubular external fixator (with more Steinmann pins)[6] are the main types of external fixators used for the treatment of hip fracture (Figure 1). However, these external fixators have two main disadvantages. First, they lack adequate slide-poking potential and are difficult to adjust in the event of varus deformity. Second, single-angle fixation makes satisfactory results difficult to obtain in complex fracture fixation.

We hypothesized that a slide-poking external fixator can be designed and produced to overcome the disadvantages of the existing external fixators and that its biomechanics might be superior to those of the dynamic hip screw (DHS) in the treatment of hip fracture.

Methods

Design and structure of the slide-poking external fixator

The slide-poking external fixator for hip fracture is composed of a sliding plate, universal joint, sliding device, top bolt screw setting, and bone screws. All parts of the fixator coordinate with one another to achieve fracture reduction and fixation (Figure 2).

The sliding plate is composed of two steel plates: the head and the body (Figure 3(a)). The head is the swollen, racket-shaped portion of the device and has a longitudinal

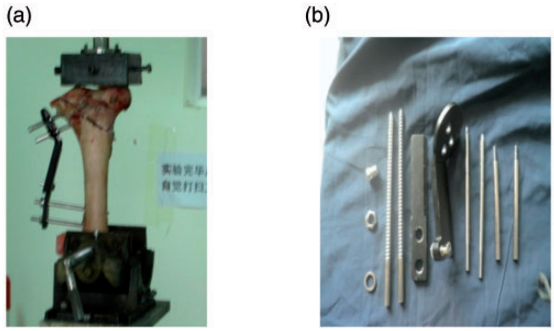

Figure 2. Slide-poking external fixator. (a) View of external fixator after assembly and fixation. (b) Parts of external fixator.

diameter of 4 cm, thickness of 0.8 cm, transverse diameter of 3 cm, and rough embossing on the back surface. The head contains five plate holes of 1-cm diameter in the plate surface. The angle of the plate hole is 137° to 145° in the coronal plane and 5° to 25° in forward rotation (anteversion) on the horizontal surface to physiologically comply with the neck–shaft angle and anteversion angle for flexible fixation of complex fractures. The body of the sliding plate is a straight handle with a length of 10 cm, thickness of 0.5 cm, width of 2 cm, and a smooth back surface. The palm side of the body has a chute groove with a depth of 0.4 cm, length of 8 cm, and width of 1.8 cm to accommodate the bolt of the far sheet plate. The connecting section of the head was at an angle of 160° to physiologically comply with the anatomical angle between the greater trochanter and the lateral femoral cortex. Under the bottom of the nest, there is a 0.3-cm-diameter shaft hole through the drive shaft.

The far sheet plate is divided into the bolt and the length of the fixed portion (Figure 3(b)). The bolt has a length of 4 cm, width of 1.8 cm, and thickness of 0.4 cm and can be inserted into the near sheet plate to facilitate sliding. A 0.1-cm-deep tooth groove is present lateral to the far sheet plate, corresponding to the gear of the primary plate; this groove coincides with the gear to facilitate sliding. The fixed portion is 4 cm long, 0.4 cm thick, and 1.8 cm wide; it has a toothless groove on the outside and an embossed back surface. Two plate holes with a 0.8-cm diameter in the vertical direction to the plate are present with a hole spacing of 2 cm.

The universal joint is shown in Figure 4 (a). It comprises the snap ring, nut, and washer. The snap ring is divided into the ball head and the cylinder, both of which are hollow and interlinked. The hollow screw part has a diameter of 0.74 cm, and the section of the far plate with fixing screws has a diameter of 0.5 cm. The maximum ball diameter and length are 1.5 and 1.0 cm, respectively, and the ball is divided into four 0.5-cm-thick blades by four slots. The bulb diameter becomes thin and transitional to the cylindrical portion with a 0.8-cm diameter, 2-cm length, and 0.1-cm thread length.

The nuts, shown in Figure 4(b), are round muffs designed for hexagonal holes. The hole diameter is 0.8 cm, and the distance between the inside and outside is 0.4 cm. The body length is 0.5 cm, and the inside contains a threaded groove with a depth of 0.1 cm, riveted with cylindrical portion threads. Conical protrusions are present in the side of the nut extending to

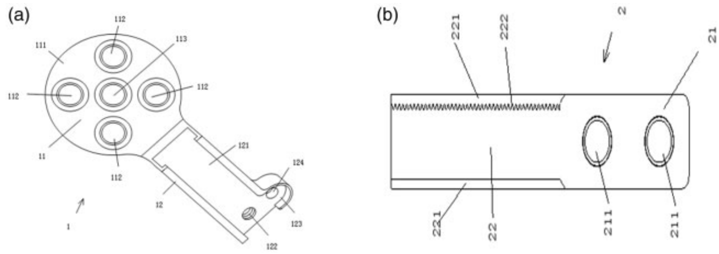

Figure 3. Sliding plate. (a) Head and body. 11–Head. 12–Body. 111–Ball fossa. (b) Far sheet plate. 21–Fixed part. 112, 113–Universal joint holes. 22–Bolt. 211–Bone screw hole. 114–Embossed surface. 121–Chute. 221–Gave way trough. 222–Tooth groove. 122–Hole of top cone screw. 123–Gear nest. 124–Shaft hole.

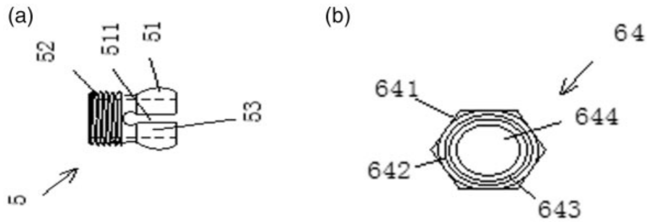

Figure 4. Universal joint and nut. (a) Universal joint. 51–Ball. 52–Cylindrical portion. 53–Center hole. 511–Radial groove. (b) Nut. 64–Joint nut. 641–Outside six-party. 642–Spherical flange. 643–Flange bore. 644–Internally threaded cap portion.

the plate; they are smooth on both sides, and the inner bore diameter is only 0.1 cm thick to coincide with the gasket at different angles. The gasket (Figure 5) is composed of an inner and outer ring with an inner diameter of 0.8 cm, outer diameter of 1.1 cm, and length of 0.1 cm. The gasket has three small point-like protrusions on the plate surface that are consistent with the embossed surface of the plate to prevent rotation and sliding.

The sliding device consists of a gear, drive shaft screw, and drive shaft (Figure 5). The drive shaft comprises the coarse handle on the dorsal aspect of the plate and the thin handle on the volar aspect of the plate. The drive shaft is actually formed by the handle with a diameter of 0.5 cm, length of 1.5 cm, and triangular shape, and it has a baffle to extend from the phase at the shank to prevent the drive shaft from sliding toward the volar aspect of the plate.

The thin shank is located in the gear nest. It has a length of 1 cm and diameter of 0.2 cm, and it is smooth and cylindrical to help the gear slide. A threaded screw is present at the volar end of the primary plate, and this screw can rivet with the shaft screws (Figure 5(b)). The gear has a diameter of 1 cm and thickness of 0.5 cm, thus adapting to gear nest. The gear thread is 0.1 cm deep and the diameter of the gear hollow is 0.2 cm, coinciding with the thin shank (Figure 6(a)). The shaft screw is a nut with a diameter of 0.2 cm, and through the thread of the thin shank end, the gears can be limited in the gear nest (Figure 6(b)).

The top bolt screw setting is composed of screw holes located in the middle of the back side of the near plate and the nest gear level. The diameter of each screw hole is 0.5 cm through the full thickness of the plate. The full depth of the groove rivets

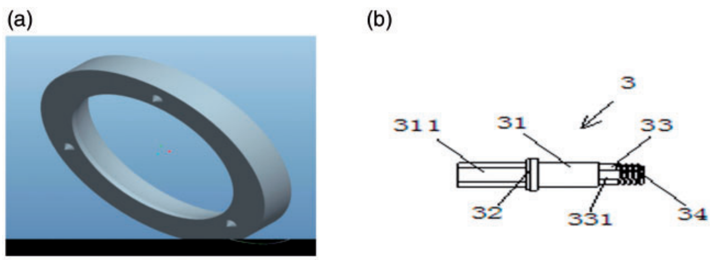

Figure 5. Gasket and drive shaft. (a) Point-like projections of gasket. (b) 3–Drive shaft. 31–Shaft. 32–Apron. 311–Crude. 331–Gear shaft. 34–Screw thread.

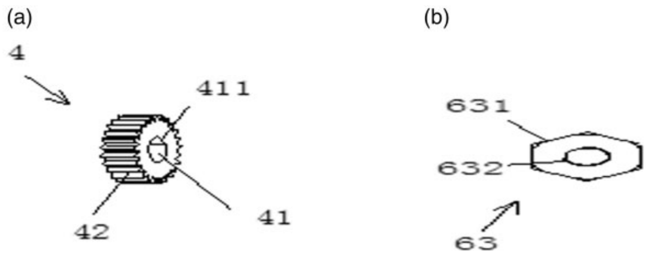

Figure 6. (a) 4–Shank. 41–Shaft hole. 42–Tooth pattern. 411–Platform. (b) 63–Gear. 631–Hexagon. 632–Inside thread.

with the top bolt screw thread. The diameter and length of the top bolt screw are 0.5 and 0.4 cm, respectively. The top bolt screw is designed as a hexagonal recess to facilitate hexagonal nut screwing and withstand the locking plate (Figure 7).

The bone screws comprise cannulated screws and bone needles (Figure 8). The mean values of the femoral neck measurements in the general Chinese population have been reported as follows[7]: the angle of the femoral neck shaft is 135°, the neck length is 95.94 ± 7.69 mm, the length of the upper edge of the femoral neck is 84.31 ± 7.06 mm, the length of the lower edge of the femoral neck is 100.39 ± 8.99 mm, and the cross-sectional diameter of the proximal femur is 3 to 5 cm (excluding hip area). The length of the spicules of the bone needle is 9 cm, the length of the threaded portion is 6 cm with dense thread, and the diameter is 0.5 cm (Figure 8(a)). The length of the hollow nail is 16 cm, the length of the threaded part is 13 cm, the thread angle is 45°, and the diameter of 0.75 cm, (Figure 8(b)).

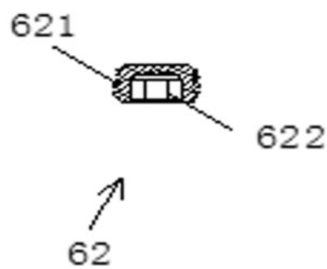

Figure 7. Top bolt screw setting. 621–External thread screw of top taper. 622–Hexagon socket.

Biomechanical experiments

Test specimens. We used nine pairs of femur specimens from <1-year-old calf carcasses. No differences in bone density were present among all 18 specimens.

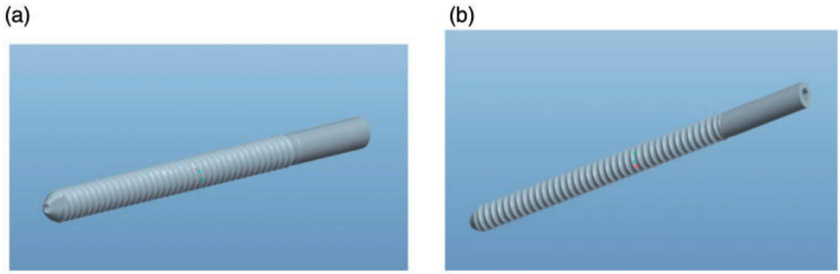

Figure 8. Bone screws. (a) Bone needle. (b) Hollow nail.

Handling and preparation of the specimens. All specimens were sealed within double plastic bags to retain moisture and cryopreserved at −20°C. They were thawed at room temperature for 12 hours before the experiment. Each carcass was confirmed to have no rheumatism, bone disease, fractures, deformities, or anatomical variations by visual inspection and X-ray examination. The soft tissues on the surfaces of all nine pairs of matching calf femurs were cleanly removed using scalpel blades and periosteal strippers. The femoral condyle was amputated, and its proximal length was 30 cm.

Specimen grouping. Each of the 18 specimens was subjected to a longitudinal loading experiment to eliminate creep deformation before fixation. The right femur specimens were then fixed by the slide-poking external fixation device (experimental group, $n = 9$), and the left femur specimens were fixed by a DHS (control group, $n = 9$). Longitudinal and torsional loading experiments were conducted in both groups.

Preparation of fracture models. To implement accurate fracture reduction and firm fixation, fixed installation of the specimen was performed, followed by removal. Next, an AO type 31-A1.3 fracture model was developed by sawing the femur with a handsaw from 1 cm distal to the tip of the greater trochanter to slightly dorsal to the lesser trochanter along the hip line. The width of the fracture line was approximately 1 cm to eliminate the stress load between the fracture pieces. Finally, the fixed installation was readjusted (Figure 9).

Installation of slide-poking external fixator and DHS. Two hollow needles were inserted into the femoral neck according to the standard method, and the plate was positioned at a distance of 7 cm from the femur, representing the position of the slide-poking external fixator because of the effect of thick soft tissue around the femur (Figure 9(a)). The DHS was installed in the proximal femur in accordance with the standard protocol of DHS implantation (Figure 9(b)).

Fixation of specimens and fixtures in loading experiments. The contact portion of the biomechanical testing device was formed into the shape of an acetabulum and then placed in contact with the femoral head of the specimen, and no distraction occurred when the load was applied. The distal femur was fixed to a homemade jig with denture powder, and the specimen was maintained at an angle of 25° to the vertical in the coronal plane and in the neutral position in the sagittal plane to simulate the posture of the body while standing on one leg.

Fixation of specimens and fixtures in torsion experiments. Torsion of 15° was ensured to maintain consistency with the vertical axis of the human body while casting and fixing

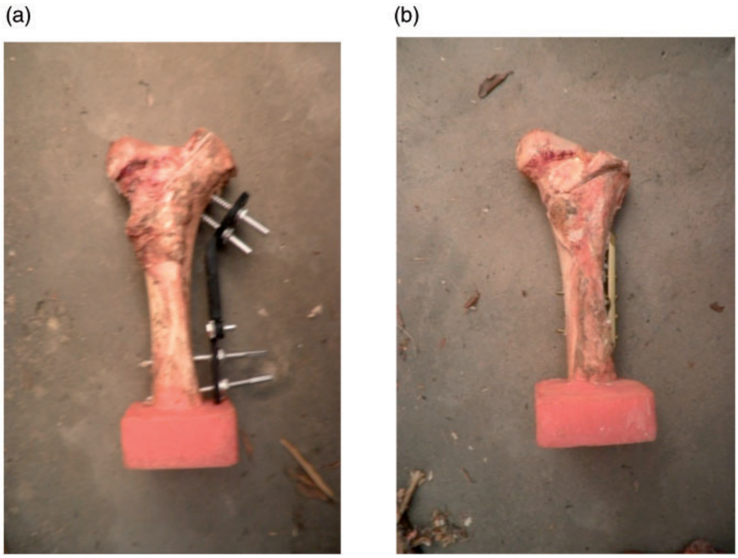

Figure 9. Establishment of fracture model. (a) External fixator. (b) Dynamic hip screw.

the sample with the denture powder. A torsion test was then performed by turning the specimen upside down, ensuring that the femoral head was below the shaft so that the jigs could grip the femoral head.

Preload testing. Vertical preload tests were performed on all 18 unfixed specimens by placing them on the biomechanical test instrument platforms. A downward vertical load was applied with a 300-N preload, completely unloading three times to eliminate creep deformation of the femur specimen (Figure 10).

Figure 10. Preloading test.

Vertical loading and torsional loading experiments before specimen fixation. Gradually increasing vertical loads (100 N, 200 N, 300 N, ..., 1000 N) were applied three times each. The sinking displacement of the femoral head was recorded, and the mean values were calculated. Next, gradually increasing torsion loads (1, 2, 3, ... ,10 Nm) were applied three times each. The specimens were placed upside down, with the femoral head below the shaft. The specimens were fixed with clamps and connected and fixed to the torsional testing machines, and torsion testing was then carried out by maintaining the proximal femur in a stationary position, moving the device in a clockwise (left femur) or counterclockwise (right femur) direction to reverse the direction of the distal femur. The corresponding angles of twist in the experimental and control groups were recorded, and the mean values were calculated.

Longitudinal loading and torsion testing after fixation. The fracture models in the experimental and control groups were osteotomized and fixed. Biomechanical testing was then carried out by the same compression and torsion methods described above (Figures 11 and 12).

Statistical analysis. All data are expressed as mean ± standard deviation. The experimental data were analyzed with a paired *t*-test using the statistical software SPSS version 16.0 (SPSS Inc., Chicago, IL, USA). A P value of <0.05 was considered statistically significant.

Ethics. This study was performed in accordance with the ethical standards in the 1964 Declaration of Helsinki and relevant regulations of the US Health Insurance Portability and Accountability Act (HIPAA). Ethics approval was not required because this was not an animal or human study.

Results

Before fixation, the mean vertical downward displacement of the femoral head with a 1000-N load was 1.23300 ± 0.27331 mm in the experimental group and 1.20300 ± 0.63633 mm in the control group. The mean reverse torsion angle with application of 10.0 Nm of reverse torsion was 1.13333° ± 0.40000° in the experimental group and 1.10000° ± 0.48734° in the control group. The biomechanical indexes are compared between the two groups in Table 1. No significant differences were found between the two groups. After fixation, the mean vertical downward displacement of the femoral head with a 1000-N load was 1.49322 ± 0.116280 mm in the experimental group and 2.13656 ± 0.166374 mm in the control group. The mean torsion angle with application of 10.0 Nm of torsion was 7.9733° ± 1.65704° in the experimental group and 15.4889° ± 0.73228° in the control group. The post-fixation biomechanical indexes are compared between the two groups in Table 2; significant

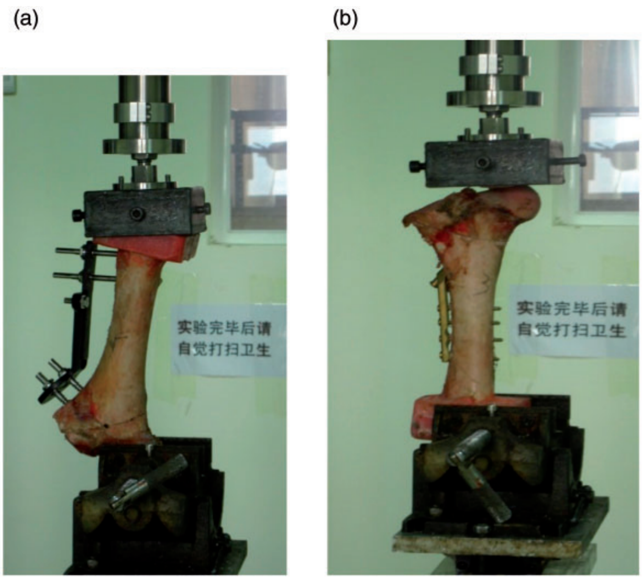

Figure 11. Vertical loading test. (a) External fixator. (b) Dynamic hip screw.

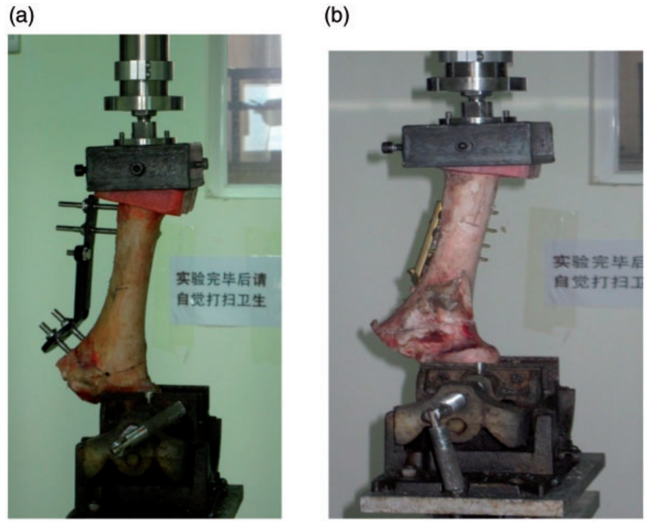

Figure 12. Torsional loading test. (a) External fixator. (b) Dynamic hip screw.

Table 1. Comparison of relative biomechanical index in two pre-fixated groups and the analysis of statistical data.

Group	Vertical displacement of femoral head with 1000-N load (mm)	Torsion angle when twisted to 10 Nm
Control group (before DHS fixation) (n = 9)	1.20300 ± 0.63633	1.10000 ± 0.48734
Experimental group (before external fixator fixation) (n = 9)	1.23300 ± 0.27331	1.13333 ± 0.40000
t	−1.148	−1.864
P	0.280	0.095

Data are presented as mean ± standard deviation.
DHS, dynamic hip screw.

differences were observed between the two groups.

Discussion

Advantages of the slide-poking external fixator

The slide-poking external fixator developed in this study has several important advantages. First, linking of the hollow nails into the femoral neck and proximal plate is a universal mechanism effectuated by adjustments through universal joints to achieve adequate slide-poking magnitudes of the hollow nails. Hence, this external fixator is more effective than other external fixators. Second, more universal joint holes are present, and the specific installation location of the universal joints can be selected during surgery according to the fracture requirements. The angle and direction of the universal joint and hollow nail can vary, and multiple angles and bone screws can be fixed at the same time to adapt to a variety of complex fracture

Table 2. Comparison of relative biomechanical index in two post-fixated groups and the analysis of statistical data.

Group	Vertical displacement of femoral head with 1000-N load (mm)	Torsion angle when twisted to 10 Nm
Control group (after DHS fixation) (n = 9)	2.13656 ± 0.166374	15.4889 ± 0.73228
Experimental group (after external fixator fixation) (n = 9)	1.49322 ± 0.116280	7.9733 ± 1.65704
t	7.981	3.700
P	0.000	0.005

Data are presented as mean ± standard deviation.
DHS, dynamic hip screw.

situations. Third, the bone screws across the plate can be extended; they are pressured by the large pincer grip and the two ends of the fractures, which ensures proximal outreach of the fracture, prevents the occurrence of hip varus deformity, and ensures tight compression of the fracture fragments, thus promoting fracture healing. Fourth, the far plate can slide in the groove of the proximal plate, after which the plates are locked by the top bolt to ensure the smooth progress of poking and abduction of the hollow nails. Fifth, the designed bone screws (including both the cannulated screws and bone needles) with their long thread portion, larger diameter of the screw thread angle than previous external fixators, and denser thread pitch than previous external fixators, greatly improves the gripping force of the screws to the bone tissues, the load of screw failure, and the strength of the external fixator. Finally, by sliding of the far plate relative to the proximal plate and the adjustment mechanism of the universal joint, the system achieves its maximal outreach poking characteristic to avoid the occurrence of hip varus.

Biomechanical evaluation

Under an identical load, the greater displacement of the femoral head led to lower resistance to compression of the fixation. In addition, at the same torque, the smaller torsion angle resulted in a larger anti-rotation capability of the fixation. Our biomechanical experiments showed that the mean vertical downward displacement of the femoral head with a 1000-N load before fixation was 1.23300 ± 0.27331 mm in the external fixator group and 1.20300 ± 0.63633 mm in the DHS group. Additionally, the mean torsion angle with 10 Nm of torque before fixation was 1.13333° ± 0.40000° in the external fixator group and 1.10000° ± 0.48734° in the DHS group. The paired t-test revealed a P value of >0.05 (i.e., no significant difference between the two groups), thereby indicating no significant difference in the biomechanical properties of the bone between the two specimens.

After the specimens were fixed in the external fixator group, the mean vertical downward displacement of the femoral head during loading of 1000 N was 1.49322 ± 0.116280, and the mean torsion angle with 10 Nm of torque was 7.9733° ± 1.65704°. After the specimens were fixed in the DHS group, the mean vertical downward displacement of the femoral head during loading of 1000 N was 2.13656 ± 0.166374 mm, and the mean torsion angle with 10 Nm of torque was 15.4889° ± 0.73228°. The paired t-test showed that the differences between the two groups were significant ($P < 0.05$), indicating that upon resistance to the compression (bending) and anti-rotation forces, significant differences were detected between the slide-poking external fixator and DHS; specifically, the slide-poking external fixator was superior to the DHS in resistance to the compression and anti-rotation forces. Qin et al.[8] reported a new type of two-head automatic pressure external fixator for hip fracture and showed that the strength, stiffness, and twist mechanical function of the femora in the experimental group were obviously superior to those in the DHS group. However, the external fixator lacked the slide-poking mechanism. The present report is the first to describe the use of a slide-poking external fixator for treatment of hip fracture.

The slide-poking external fixator enhances the stability of fracture reconstruction in terms of the distribution of rotational and axial compression loads. Therefore, we suggest that the clinical stability of the slide-poking external fixator might be superior to that of the DHS. External fixators, which are minimally invasive and involve the placement of screws outside of the body, can also reduce blood loss, wound infections, and other complications. These results require confirmation in prospective clinical studies of patients undergoing treatment with the slide-poking external fixator.

Notably, external fixators for hip fracture are used in very limited circumstances; e.g., elderly patients with osteoporosis-associated hip fracture who are prone to develop internal fixation failure, patients with a higher risk of internal fixation failure (such as those with limb thrombosis or severe organ dysfunction), patients with open fractures, and patients who develop osteomyelitis, tuberculosis, infection of the internal fixator, or pathological fractures. In these few cases, compared with other external fixators, our external fixator has the advantages of external pry and re-locking, and it shows sufficient biomechanical stability. Thus, it can serve as a useful supplement in the surgical treatment of intertrochanteric fractures when an external fixator is a viable treatment choice.

Limitations

This study had several limitations. First, the experiment only involved AO type 31-A1.3 fractures; severe type 31-A2.2, 31-A2, and 31-A3 fractures were not examined. Second, the bone was surrounded by thick soft tissue. The bar part of the external fixator was locked in a different position than in our experimental setting, which might lead to different results. Third, osteoporosis was not considered in this study; therefore, the results are not necessarily applicable to clinical practice. However, we plan to perform animal experiments followed by human specimen experiments and then clinical experiments to verify our findings.

Conclusion

The slide-poking external fixator for treatment of hip fracture that was designed and produced in this study can provide sufficient stability and resistance to the compression and rotation in hip fractures.

Declaration of conflicting interest

The authors declare that there is no conflict of interest.

Funding

This research received no specific grant from any funding agency in the public, commercial, or not-for-profit sectors.

References

1. Kwak DK, Kim WH, Lee SJ, et al. Biomechanical comparison of three different intramedullary nails for fixation of unstable basicervical intertrochanteric fractures of the

proximal femur: experimental studies. *Biomed Res Int* 2018; 2018: 7618079.
2. Alcivar E. A new method of external fixation for proximal fractures of the femur. *Injury* 2001; 32: Sd107–Sd1014.
3. Ehlinger M, Favreau H, Eichler D, et al. Early mechanical complications following fixation of proximal femur fractures: from prevention to treatment. *Orthop Traumatol Surg Res* 2020; 106: S79–S87.
4. Gomez M, Marc C, Talha A, et al. Fast track care for pertrochanteric hip fractures: how does it impact length of stay and complications? *Orthop Traumatol Surg Res* 2019; 105: 979–984.
5. Moroni A, Faldini C, Pegreffi F, et al. Dynamic hip screw compared with external fixation for treatment of osteoporotic pertrochanteric fractures. A prospective, randomized study. *J Bone Joint Surg Am* 2005; 87: 753–759.
6. Wang XK, Ye JD, Gu FS, et al. Numerical simulation research to both the external fixation surgery scheme of intertrochanteric fracture and the healing process, and its clinical application. *Biomed Mater Eng* 2014; 24: 625–632.
7. Zhao JX, Su XY, Zhao Z, et al. [Radiographic analysis of treatment of intertrochanteric fractures using proximal femoral nails]. *Beijing Da Xue Xue Bao Yi Xue Ban* 2015; 47: 263–268.
8. Qin YX, Sheng GQ, Wang YJ, et al. [Biomechanical study of new type two-head automatic pressure external fixator (TAPEF) for the treatment of intertrochanteric fracture]. *Zhongguo Gu Shang* 2008; 21: 771–773.