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Biomechanical simulation of bed turning post-acetabular fracture fixation

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Before patients begin out-of-bed exercises following internal fixation surgery for acetabular fractures, turning over in bed serves as a crucial intervention to mitigate complications associated with prolonged bed rest. However, data on the safety of this maneuver post-surgery are limited, and the biomechanical evidence remains unclear. This study aims to introduce a novel loading protocol designed to preliminarily simulate the action of turning over in bed and to compare the biomechanical properties of two fixation methods for acetabular fractures under this new protocol. A RNJ-500 microcomputer-controlled electronic torsion tester was utilized to simulate the action of turning over in bed and to conduct a dynamic torsion loading test. Initially, the torque values and torsional stiffness of six intact pelvis specimens (Group A) were measured. A double-column acetabular fracture model was then created and stabilized using two different fixation methods: the Dynamic Anterior Plate-Screw System for the Quadrilateral plate (DAPSQ, Group B) and the traditional anterior reconstruction titanium plate plus a 1/3 tubular buttress plate (Group C). All specimens underwent cyclic torsion loading ranging from 2° to 8°. The medial displacement and strain values of the quadrilateral plate were recorded and analyzed. As the torsion angles increased from 2° to 8°, Groups A and B exhibited significantly higher torque values compared to Group C (all $P < 0.05$). Group C demonstrated notably lower torsional stiffness (1.51 ± 0.20) relative to Group A (2.33 ± 0.25, $P < 0.05$) and Group B (2.21 ± 0.29, $P < 0.05$). Additionally, the medial displacement of the quadrilateral plate was significantly reduced in Group B compared to Group C at all measured time points ($P < 0.05$). And Group C exhibited significantly higher maximum tensile and compressive strain than Group B (all $P < 0.05$). The DAPSQ plate with quadrilateral screws provides superior anti-rotational stability in a double-column acetabular fracture model under the newly established torsion loading protocol.

Keywords Acetabular fracture, Internal fixation, Screw, Plate, Biomechanics, Cadaver study

Acetabular fractures present a significant challenge for even the most seasoned orthopedic surgeons due to the intricate anatomy of the pelvis, the variability in fracture patterns, and the restricted surgical access available. Achieving a long-term functional hip joint necessitates anatomical reduction and stable fixation, which is currently the gold standard for managing displaced fractures[1]. Based on our clinical experience, complicated acetabular fractures especially double column fractures frequently involve the disruption of the quadrilateral plate (QLP), a thin and flat anatomical structure constituting the medial wall of the acetabulum[2,3].

At present, there is considerable debate regarding the optimal implant for the repair of acetabular fractures involving the QLP, and no widely recognized treatment guidelines have been established for these injuries. Initially, the most commonly reported fixation technique involved the use of pelvic brim plates combined with three or four long periarticular screws via the classic ilioinguinal approach[4,5]. However, this method carries a substantial risk of screw perforation into the articular cavity. In response, a variety of surgical implants, including

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cerclage wire-plate composites[6], buttress or spring plates[7,8], and both suprapectineal and infrapectineal plates[9,10], have been developed, demonstrating satisfactory clinical and biomechanical outcomes in the majority of studies. Despite these advances, each of these fixation methods has certain limitations that have impeded their widespread acceptance. To address these challenges, our group have designed a novel fixation implant (named Dynamic Anterior Plate-screw System for the Quadrilateral plate, DAPSQ [number ZL201621494131.0, CN PAT]) for acetabular fractures involving QLP, and several previous clinical trials and in vitro biomechanical experiments, including those utilizing cadaveric and finite element simulations, have confirmed that this system could provide robust biomechanical stability for the fixation of QLP fractures[3,11–15].

Moreover, in previous biomechanical studies, the most frequently simulated positions for intact pelvises were standing or sitting[13,15,16]. In addition to comparing biomechanical stability, numerous studies have sought to replicate the postoperative physiological conditions of patients as accurately as possible, aiming to demonstrate the feasibility of early rehabilitation exercises in standing or sitting positions. It is well recognized that early mobilization out of bed after surgery is crucial for reducing the risk of postoperative complications, such as pneumonia and deep vein thrombosis[17,18]. Consequently, reliable internal fixation methods must withstand the mechanical demands of early out-of-bed exercises. Before progressing to out-of-bed exercises, turning over in bed is one of the most effective strategies for minimizing the risk of pneumonia and preventing pressure sores[17]. Unlike the loading conditions in standing or sitting positions, the acetabulum primarily endures torsional forces during the process of turning over. However, to our knowledge, no studies have simulated the physiological conditions associated with turning over in bed.

Against this backdrop, our study aims to explore the following questions: (a) Can we develop a new loading protocol capable of simulating the biomechanics of turning over in bed? (b) Is it safe for patients to turn over in bed following acetabular fracture fixation? (c) Which fixation method, DAPSQ or a buttress plate, offers superior biomechanical resistance to rotational forces?

Materials and methods

This experimental study was carried out in accordance with the principles of the Declaration of Helsinki after obtaining the approval from the Institutional Review Board of General Hospital of Central Theater Command. The overall flowchart for this biomechanical study was shown in Fig. 1.

Specimens and instruments

Six intact-pelvis specimens, obtained from the Institute of Anatomy of Institute of Anatomy of Hubei University of Chinese Medicine, were used in this biomechanical investigation. Specimens presenting with arthritis, tuberculosis, malignancy, or imaging abnormalities were excluded from the study. All specimens were harvested from the fourth lumbar vertebra to the proximal one-third of both femoral shafts, with the pelvic ligaments, particularly the sacrospinous and sacrotuberous ligaments, remaining intact.

Standardized left-sided acetabular fractures, classified as Judet & Letournel double-column type[19], were created based on our previous study (Fig. 2A)[16]. The fracture models were randomly assigned to either Group B, involving DAPSQ and quadrilateral screw fixation, or Group C, which utilized suprapectineal reconstruction plates and 1/3 tube buttress plates[16], with each procedure conducted sequentially (Fig. 3).

In Group B (Fig. 3A), a pre-contoured, side-specific DAPSQ titanium plate (provided by Hua Sen Medical Instruments Co., Ltd., Changzhou, China) was selected and positioned along the superior arcuate line, with both ends extending toward the superior pubic ramus and iliac wing. Four 3.5 mm quadrilateral screws were employed to control medial displacement of the quadrilateral surface (QLP). The specific placement sequence and working principles of these screws have been detailed in our previous publications[3,11].

In Group C (Fig. 3B), a 1/3 tube plate was slightly over-contoured with an arched angle at its proximal third to span the pelvic brim. Once properly seated, two screws were inserted at its uppermost holes to buttress the QLP displacement. Following the placement of the buttress plate, a 12-hole suprapectineal reconstruction plate was aligned along the superior side of the pelvic brim, with three screws used at each end to stabilize the anterior column of the acetabulum and reinforce the buttress plate's effect on the quadrilateral surface.

Furthermore, according to the reduction principles for double-column fractures, it is crucial that fractures involving the iliac wing in both groups achieve satisfactory reduction and fixation with a 5- or 6-hole reconstruction plate and four screws before proceeding.

Specimen loading protocol

Biomechanical tests were performed at the Key Laboratory of Mechanics, Wuhan University of Technology. The experimental procedure consisted of the following steps (Fig. 1). First, the torque values and torsional stiffness of the intact pelvic specimens (Group A) were measured. Subsequently, a double-column acetabular fracture model was created, and two fixation methods were tested in a randomized sequence. In addition to the parameters mentioned above, additional measures, including medial displacement and the strain value of QLP, were introduced to quantitatively assess and compare the biomechanical stability of the different fixation techniques.

A torsion device (RNJ-500 microcomputer-controlled electronic torsion tester, Model ETT502, Capacity 500 N·m, Power 1.5 kW, Power Supply 220 V, WANCE, Shenzhen, China) was used to perform the dynamic torsion loading test (Fig. 4). All fracture models were secured in molds with denture-based polymer materials and allowed to harden for 24 h at room temperature. Custom-designed pelvis fixtures were employed to secure the cadaveric femoral shafts into bilateral stainless steel-sleeves with 8-point locked screws, minimizing movement at the metal-bone interface. Both sleeves were anchored to a rigid steel base, and a precisely aligned deep-groove linear guide was positioned beneath the sleeves to adjust the spacing of both lower limbs. A cylindrical steel post,

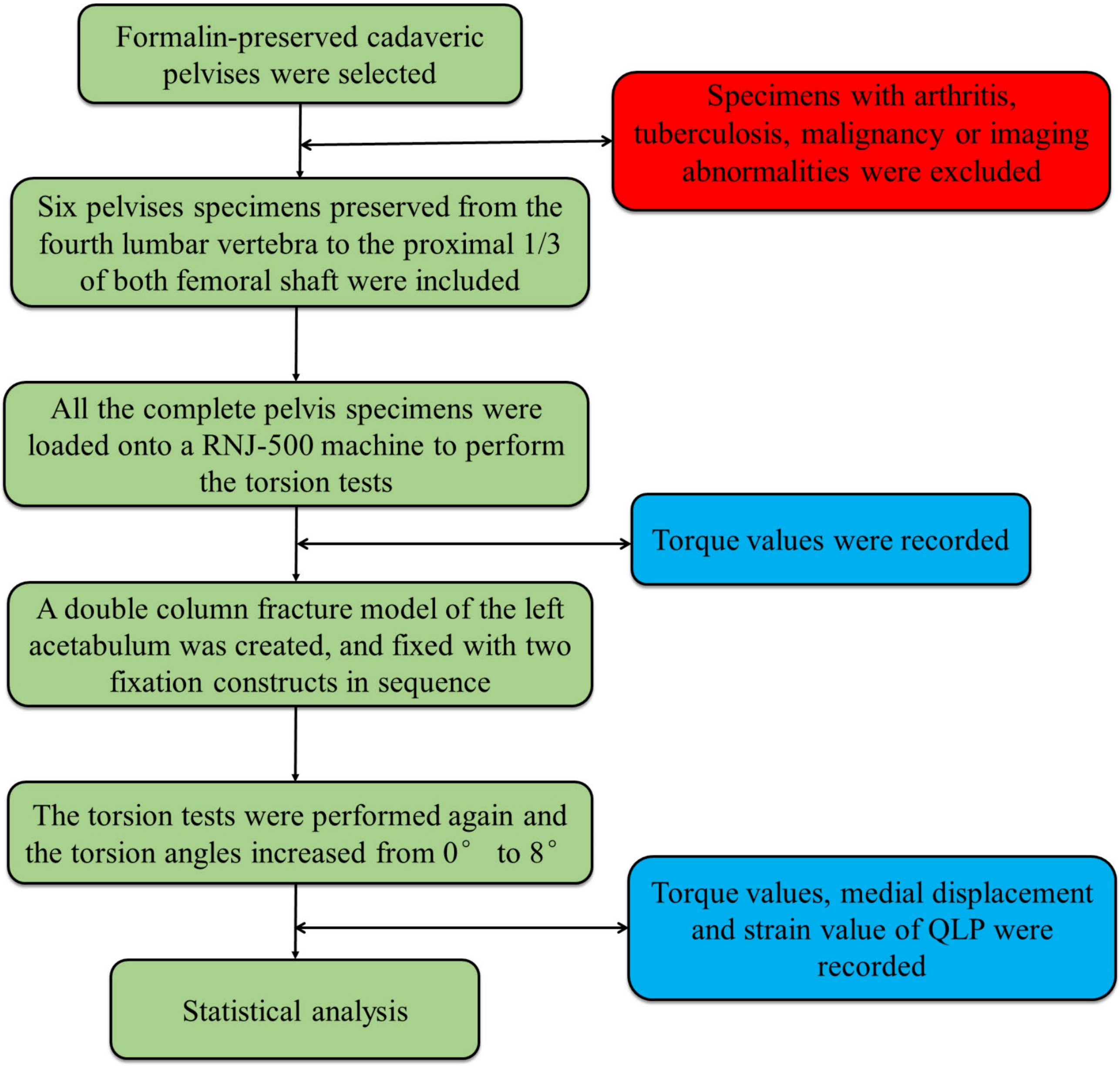

Fig. 1. The overall flowchart for this biomechanical study.

centrally located on the base, was fixed with a right-locked bearing, while the opposite end of the specimen was secured to the left roller bearing of the testing machine with a designated clamp (Fig. 5). The specimens were positioned supine, ensuring the bilateral anterior superior iliac crests were parallel to the pubic tubercles, with the femoral head oriented at 30° of abduction and 15° of internal rotation. Upon completion of this setup, a torsional load was applied while the femurs remained fixed.

Torsion angles ranging from 0° to 8° were applied incrementally in the left torsional direction (towards the fractured side) (Fig. 5). The torsional tests were conducted at a rate of 1° per minute, with a maximum torque limit of 30 N·m. Prior to formal testing, each specimen underwent preconditioning with three cycles (0° to 4°) at a rate of 1° per minute to mitigate creep. The testing machine was calibrated to ensure that a torsion angle of 0° corresponded to 0 N·m of torque. Each testing mode was conducted in three cycles, and the mean values were used for analysis.

Outcome variables
Construct evaluation
Construct failure was defined as either the relative step opening of fracture lines exceeding 2 mm, visual breakage of the screws, plates and specimen models, or femoral head dislocation resulting in not being able to support any torsion load[20].

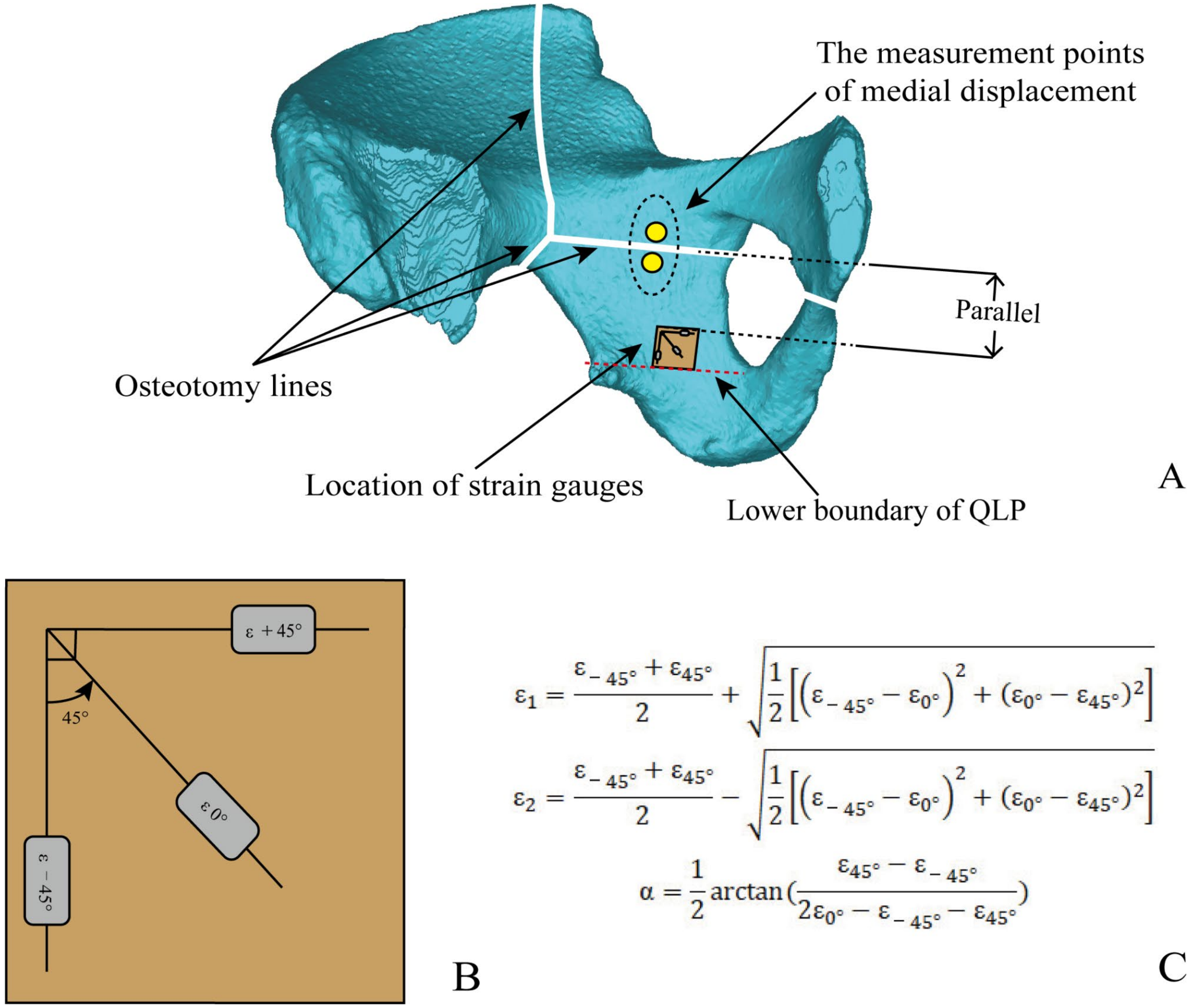

Fig. 2. (A) Double column acetabular fracture model. The yellow dots represent the measurement site of medial displacement. (B) Three sensors including ε_{+45°, ε_{0°, ε_{-45° arranged at an angle of 45°. (C) The formula to calculate the maximum tensile strain (ε_1), the maximum compressive strain (ε_2) and the direction of maximum tensile strain (α). Figures were created by Microsoft PowerPoint, URL: https://creativecommons.org/licenses/by/4.0/.

Torsion angles-torque

The torque value of each pelvic specimen was automatically recorded and measured by the sensor of the RNJ-500 torsion tester. Torque values at the torsion angles of 2°, 4°, 6° and 8° were recorded and compared across the three groups.

Torsional Stiffness

Torsional stiffness was defined as the slope of the torque versus the torsion angles curve and calculated with the formula: GJp = Mn/ϕ (GJp means torsional stiffness, Mn means torque value, ϕ means torsion angle).

Medial displacement of QLP

As previously described by Wu et al.[16], medial displacement amounts of QLP were measured with a multifunctional digital dial gauge (Deli love Monitoring Technology, Beijing, China). Figure 2A has shown the location of the measurement site at the midpoint of QLP fracture line with one pair marker. The medial displacement of QLP under the torsion angles of 2°, 4°, 6° and 8° were recorded and compared between Groups A and B.

Strain values

Strain values were measured using strain gauges positioned 2 cm above the midpoint of lower boundary of QLP. As illustrated in Fig. 2A and B, three sensors (including ε_{+45°, ε_{0°, ε_{-45°) arranged at an angle of 45° were

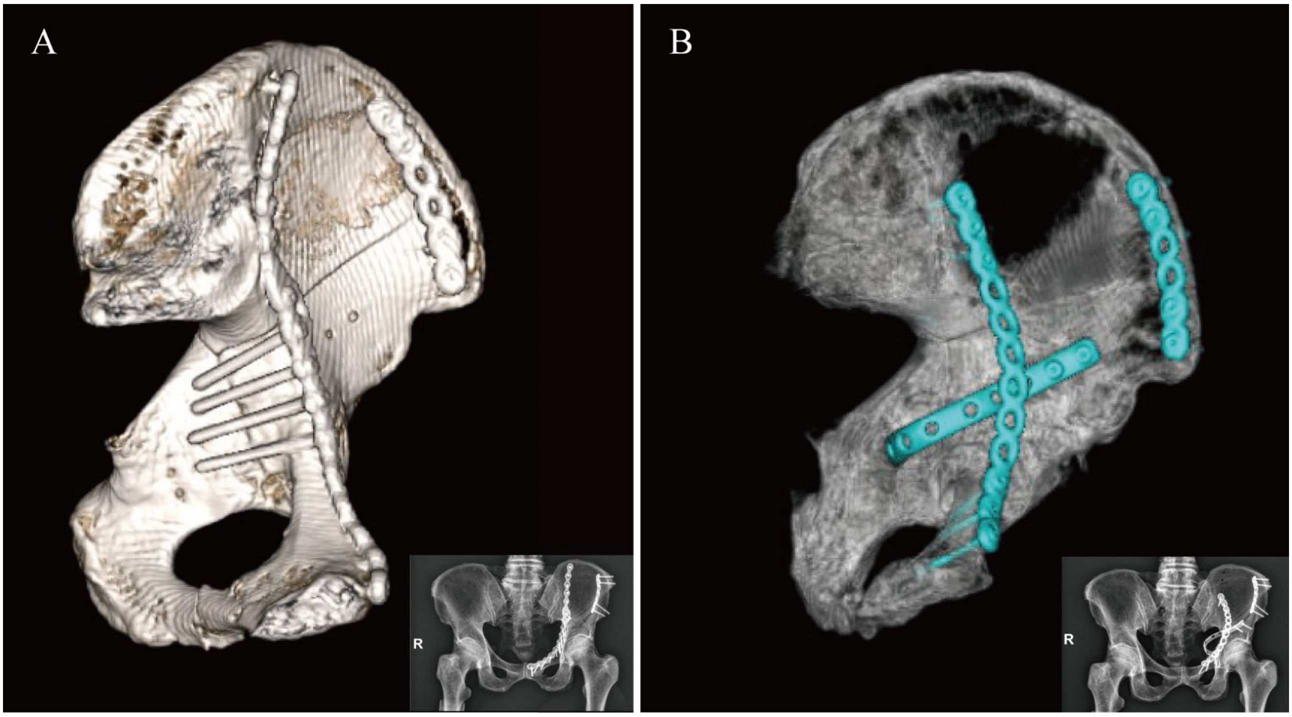

Fig. 3. Double column acetabular fracture fixed with Dynamic Anterior Plate-screw System for the Quadrilateral plate (A) and traditional anterior reconstruction titanium plate plus 1/3 tube buttress-plate (B). Figures were created by Microsoft PowerPoint, URL: https://creativecommons.org/licenses/by/4.0/.

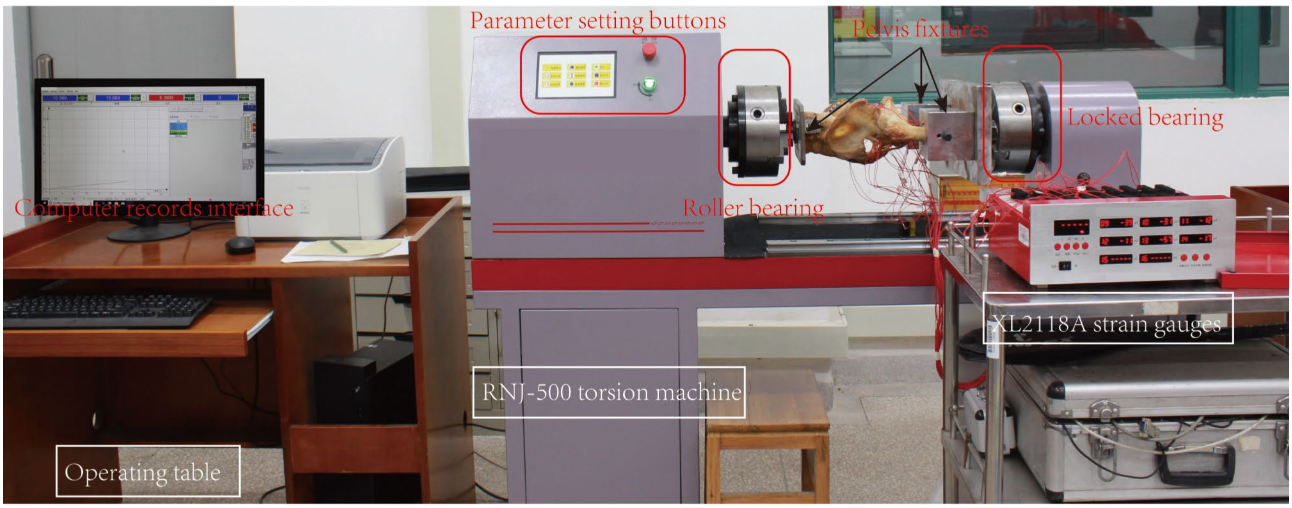

Fig. 4. The physical map of the entire experimental setup.

used. It is noteworthy that the direction of ε_{+45° strain gauge was aligned parallel to the direction of the QLP fracture lines. This configuration of strain gauges allowed for the monitoring of strains in various directions. Once measurements were obtained from the three strain gauges at each time point, the maximum tensile strain (positive values), maximum compressive strain (negative values), and the direction of maximum tensile strain could be calculated using the formula shown in Fig. 2C. Of these, ε_1 represents the maximum tensile stress, ε_2 represents the maximum compressive strain and α represents the direction of maximum tensile strain. All of these parameters were compared between Groups B and C.

Statistical analysis

Statistical analysis was performed using SPSS software (version 19.0; SPSS, Chicago, IL, USA). Data have been tested for normality using the Shapiro-Wilk normality test. Quantitative data were presented as mean values

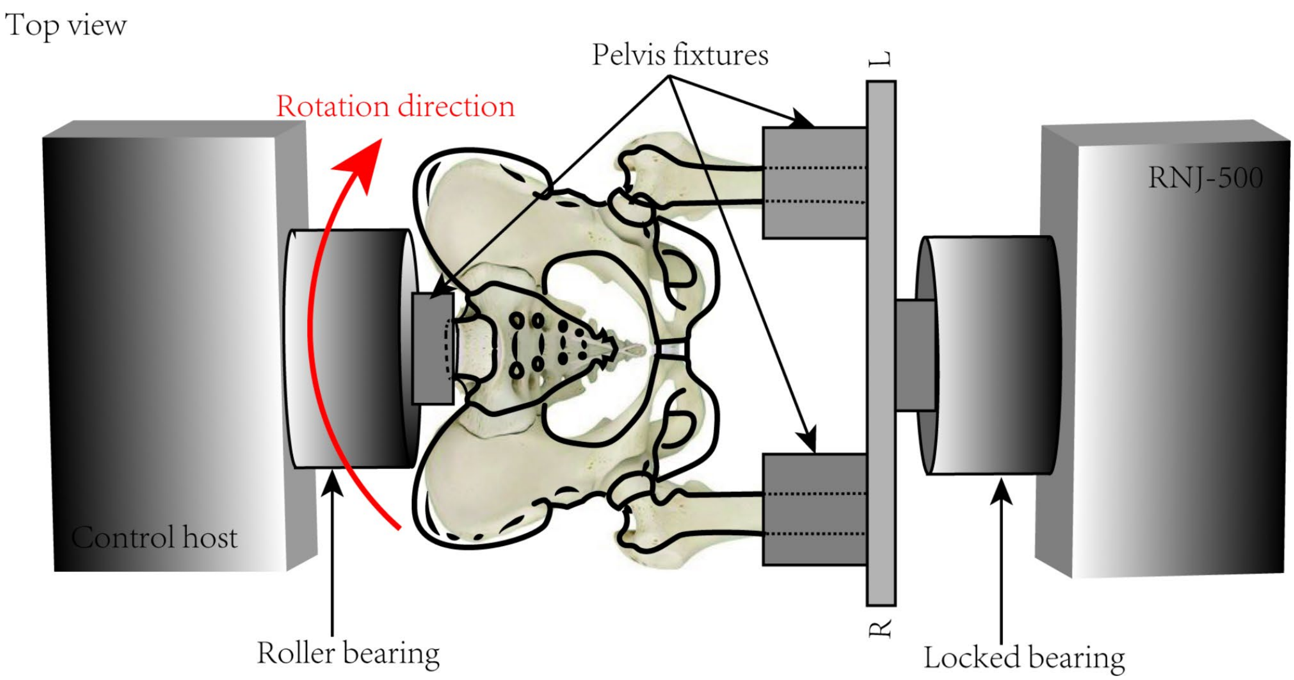

Fig. 5. Schematic indicating the specimens were fixed to the RNJ-500 microcomputer-controlled electronic torsion tester from the top view. Figures were created by Microsoft PowerPoint, URL: https://creativecommons.org/licenses/by/4.0/.

with standard deviations (mean ± SD) and the enumeration data were expressed as percentage (%). Comparisons between two paired groups were performed using a paired samples *t* test, while comparisons among multiple groups were conducted by the one-way analysis of variance (ANOVA) and homogeneity test of variance. For all analyses, statistical significance was accepted at the $P < 0.05$ level and indicated in figures as *$P < 0.05$, **$P < 0.01$, and ***$P < 0.001$.

Results

Torque Value

During the whole torsion test, the torsion angles-torque curves were recorded automatically by the computer workstation. Representative curve was shown in Fig. 6. As torsion angles increased from 2° to 8°, the torque value of each pelvic specimen increased approximately linearly. In Fig. 7A, we presented the torque values of three groups measured at each torsion angle. Over the four measurement points from 2° to 8°, torque values were significantly higher in Group A and Group B than in Group C (all $P < 0.05$). Whereas the torque values in Group B were slightly lower compared with Group A, the difference was not statistically significantly ($P > 0.05$).

Torsional Stiffness

The torsional stiffness for Groups A, B, and C were (2.33 ± 0.25) N·m/°, (2.21 ± 0.29) N·m/°, and (1.51 ± 0.20) N·m/°, respectively. The torsional stiffness value for the intact pelvis was defined as 100%. The torsional stiffness of the fracture model fixed with DAPSQ was 95.3% of the intact pelvis, while the construct with suprapectineal reconstruction plate and 1/3 tube buttress plate showed only 65.1% of the intact pelvis (as shown in Fig. 7B. $P < 0.05$ for comparison of Groups A and C, $P < 0.05$ for comparison of Groups B and C, and $P > 0.05$ for comparison of Groups A and B).

Medial displacement of QLP

As shown in Fig. 7F, the graph illustrated the changing trend of medial displacement of QLP in Groups B and C at different torsion angles. Each black dot represented the experimental data of an individual specimen. Throughout the whole torsion test, no breakage or migration of the internal fixation device was observed. The medial displacement of QLP in Groups B and C showed an increasing trend with the torsion angles increased from 2° to 8°, but all the values were within 2 mm range. And the medial displacement was significantly lower in Group B than in Group C at each evaluation time point ($P < 0.05$).

The Maximum tensile and compressive strain

The maximum tensile and compressive strain were calculated according to the formula in Fig. 2C. Over the range of measured torsion angles, the strain value increased with torsional loading. The maximum tensile values were larger than absolute values of compressive strain in the two groups at different time points, which indicated that the measurement site of strain in QLP was mainly subjected to tensile stress. The results in Fig. 7D and E

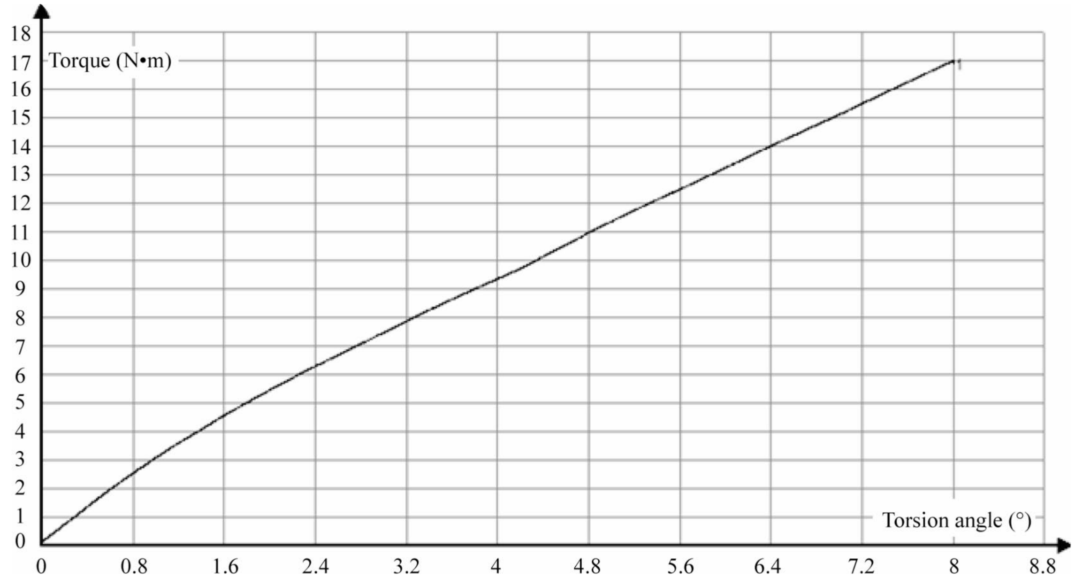

Fig. 6. The torsion angle- torque curve.

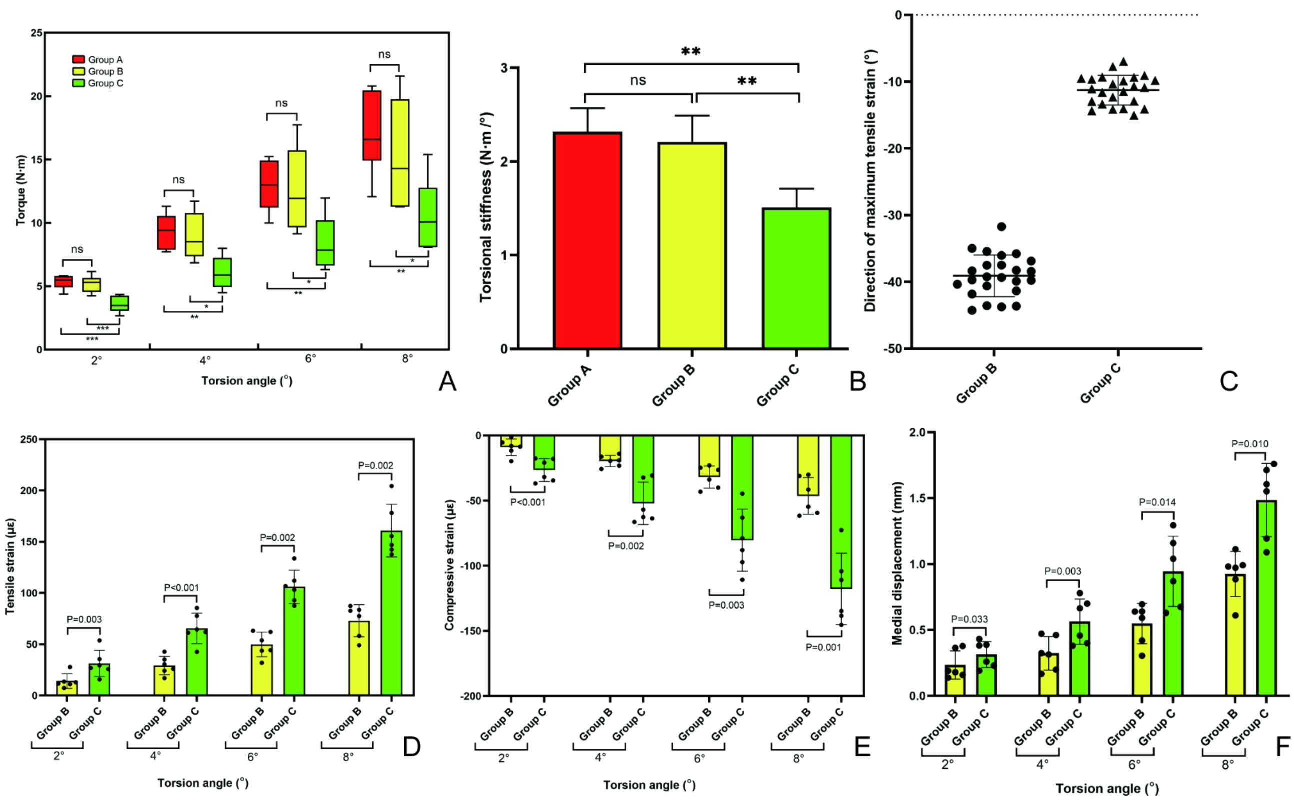

Fig. 7. The comparison of torque value, torsional stiffness, medial displacement, strain value and direction in different groups. (A) shows the torque values of three groups measured from 2° to 8°. (B) shows the torsional stiffness in three groups. (C) shows the distribution of maximum tensile strain direction in Groups B and C. (D-E) shows the maximum tensile and compressive strain of Groups B and C. (F) shows the medial displacement of QLP in Groups B and C. *$P<0.05$, **$P<0.01$, and ***$P<0.001$.

showed that Group C was found to have significantly higher the maximum tensile and compressive strain than Group B (all $P<0.05$).

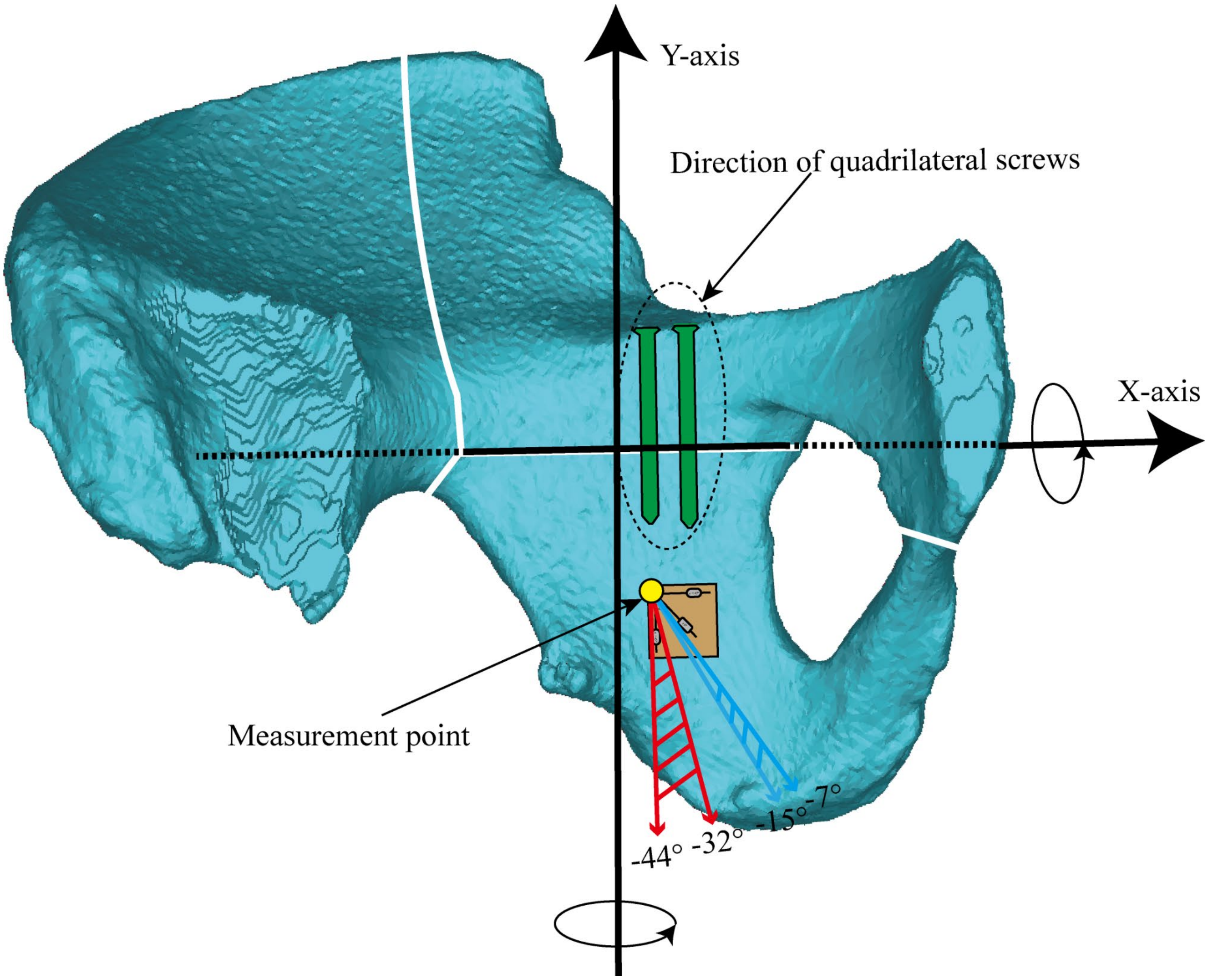

Fig. 8. The distribution of the maximum tensile strain directions in a coordinate system. Figures were created by Microsoft PowerPoint, URL: https://creativecommons.org/licenses/by/4.0/.

Direction of maximum tensile strain

The direction of maximum tensile strain was also calculated according to the formula in Fig. 2C. The distribution of maximum tensile strain direction was shown in Fig. 7C. The directions of maximum tensile strain in Group B were located between − 31.72° and − 44.27°, with an average angle of −39.10°. In Group C, it was located within − 6.96° and − 15.03°, in the predominantly − 11.3° direction. For a more intuitive comparison, reference coordinate axes were established by taking the fracture line of QLP as X-axis and the midpoint of QLP fracture line as coordinate origin. Figure 8 showed the distribution of the maximum tensile strain directions in this coordinate system. It could be seen clearly that the maximum principal tensile strain directions in Group B had a good agreement with the directions of the quadrilateral screws.

Discussion

As an irregular anatomical structure situated along the central axis of the human body, the pelvis exhibits a complex pattern of stress distribution and transmission. Functionally, it serves as a bridge between the spine and the lower limbs, facilitating the transfer of upper body weight to the lower extremities while also transmitting ground reaction forces upwards through the spine[21]. Under normal physiological conditions, standing and sitting are the most prevalent postures. In a standing position, the weight of the trunk is transmitted to both hip joints via the sacroiliac joint. Conversely, in a seated position, the direction of gravitational force changes, being transmitted primarily through the sacroiliac joint to the bilateral ischial tuberosities. Consequently, biomechanical analyses of various internal fixation systems frequently simulate these two postural states[15,16]. However, a crucial aspect of pelvic mobility is often overlooked during routine activities, the rotational movements, particularly when turning around or shifting position in bed. Previous biomechanical studies have acknowledged the presence of torque forces during such rotational motions, leading some researchers to introduce additional torsional loading alongside a compressive preload[22–24]. Furthermore, one study has specifically compared the biomechanical properties of intact pelvic specimens under static standing conditions with those under standing combined with rotational forces. The findings indicated that torsional loads significantly increased the stress on the sacroiliac joint and the anterior column of the acetabulum[25].

The primary objective in treating displaced acetabular fractures is to restore the anatomical structure of the acetabulum, alleviate pain, and enable early mobilization. Postoperatively, patients are generally encouraged to engage in lower-extremity functional exercises in bed for the first 2–4 weeks. This is followed by a transition from wheelchair use to partial weight-bearing exercises with crutches at 4–8 weeks, and then progressing to full weight-bearing at 8–12 weeks[3]. Therefore, the initial 2–4 week bed rest period is crucial not only for bone healing and remodeling but also for preventing postoperative complications. Prolonged immobilization can lead to significant morbidities such as deep vein thrombosis, pneumonia, and urinary tract infections, especially in elderly patients[18]. Additionally, extended bed rest can result in muscle atrophy and impair bone metabolism[26]. Multiple studies have suggested that timely repositioning in bed following fixation surgery can reduce the incidence of pressure sores and pulmonary infections[27,28], promote incision healing, and prevent venous thrombotic events by enhancing blood circulation. Despite these findings, data on the safety of turning over in bed following surgical repair of acetabular fractures remain limited, and the biomechanical implications are not well-defined. Moreover, to the best of our knowledge, no other biomechanical studies have simulated the torsional state of an intact pelvis in the supine position.

The act of turning over in bed involves transitioning the patient's body from a supine to a lateral position. Typically, this movement begins with the upper part of the trunk initiating rotation, which then drives the lower limbs to follow suit. During this process, the pelvis transitions from a mechanically stable state to an unstable one, akin to the start-up of a machine that requires a certain amount of torque[29]. In the present study, in order to simulate the action of turning over in bed, RNJ-500 microcomputer-controlled electronic torsion tester was used to generate the torsional loading. Key parameters, including torque, torsional stiffness, displacement, and strain, were measured across the three groups. The results indicated that as the torsion angles increased from 2° to 8°, both the torque values and torsional stiffness were significantly higher in Groups A and B compared to Group C, while no significant difference was observed between Groups A and B. The torsional stiffness of the fracture model fixed with the DAPSQ was 95.3% of that of the intact pelvis, whereas the configuration using a suprapectineal reconstruction plate and a 1/3 tubular buttress plate exhibited only 65.1% of the torsional stiffness of the intact pelvis. Torsional stiffness is defined as the slope of the torque versus torsion angle curve, representing the average torque per unit of torsion angle[30]. A higher torsional stiffness indicates a greater ability to resist torsional loading, reflecting stronger torsional resistance of the pelvis. These findings suggest that DAPSQ fixation provide superior anti-rotational stability in models of double-column acetabular fractures. Furthermore, the observation that the torsional stiffness of Group B was 95.3% of that of the intact pelvis suggests that acetabular fractures treated with DAPSQ fixation have biomechanically returned to a normal or near-normal state. Thus, it appears safe for patients with these fractures to turn over in bed following DAPSQ fixation. Moreover, it is noted that various factors can influence postoperative mobility. These factors include the precise technique employed during the fixation, the quality of reduction, the patient's bone quality, soft tissue healing, and the presence of any comorbidities or complications that could affect recovery. Additionally, the method of patient repositioning, as well as postoperative pain management and rehabilitation protocols, play significant roles in the safety and effectiveness of mobilization after surgery.

The selection of torsion angles was informed by previous studies[25]. Given the limited range of motion of the sacroiliac joint when the lower limbs are straightened, the torsion angle was kept within 10° during the experiment to avoid excessive torsion that could damage the ligaments of the pelvis and sacroiliac joint. Several studies have shown that, in a standing position, torsion angles exceeding 15° can cause irreversible damage to the pelvic and sacroiliac ligaments[25]. Therefore, the torsion angles chosen in our study fall within a safe and appropriate range. As the torsion angles increased from 2° to 8°, the torque values of each pelvic specimen increased approximately linearly. This suggests that the elastic deformation remained within the yield strength and maximum load-bearing capacity of the pelvic specimens, making it suitable for repeated loading. Moreover, as previously mentioned[3], the success of the DAPSQ technique critically depends on the dynamic compression and partial fixation provided by quadrilateral screws. In a prior cadaveric study[15], we compared the biomechanical stability of DAPSQ with that of a 1/3 buttress plate under classical standing simulation. The results indicated that the DAPSQ combined with quadrilateral screws provided superior biomechanical stability compared to the traditional anterior reconstruction titanium plate and the 1/3 tubular buttress plate. Similarly, Lei et al.[12] tested three different fixation methods for anterior column and posterior hemi-transverse acetabular fractures under double-limb standing using finite element analysis. Their findings also demonstrated that DAPSQ with quadrilateral screws had results comparable to double-column plating and was superior to the anterior column plate combined with posterior screws. In the present study, we further examined the biomechanical properties of the DAPSQ under a new loading protocol. Consistent with our previous findings, the medial displacement of the QLP in Group B was significantly lower than in Group C, demonstrating that the DAPSQ plate with quadrilateral screws provides enhanced anti-rotational stability in a double-column acetabular fracture model.

It is worth mentioning that the requirements for instrumentation in acetabular fractures differ from those for limb fractures. Generally, provided the patient's condition allows, surgery for acetabular fractures should be performed within 3 to 7 days post-injury, or ideally within 2 weeks. Delays beyond this timeframe can significantly increase surgical complexity[3,31]. This timeframe suggests that acetabular fractures demonstrate relatively rapid healing, with the fractured ends stabilizing within 2–3 weeks. As bone healing progresses, the stability of most internal fixation devices for acetabular fractures improves markedly. Thus, the duration for which these internal fixators must bear weight is relatively brief, and their mechanical stability is predominantly evident in the initial 2 weeks. However, most patients are advised to remain on bed rest during this period, which may contribute to the satisfactory clinical outcomes often observed with these fixators. Consequently, the new loading protocol we propose gains added significance.

The measurement of strain values reflects the ability of the measurement point to share the torsional load. The larger the strain shared in the measurement point of the pelvis, the less for the internal fixation, which means internal fixators were less reliable. Our study found that, across the measured range of torsion angles, strain values increased with torsional loading. The maximum tensile strains were higher than the absolute values of compressive strain in both groups, indicating that the strain measurement sites on the QLP primarily experienced tensile stress. Pairwise comparisons revealed that Group C exhibited significantly higher maximum tensile and compressive strains compared to Group B. These results suggest that the fracture fragments fixed with the QLP and quadrilateral screws demonstrated significantly better mechanical resistance to torsional loading compared to those fixed with the 1/3 tubular buttress plate. These findings can be explained as follows. First, excellent internal fixation instruments require the stress distribution to be as uniform as possible, and should not be concentrated to a certain point. The high stress concentration might lead to fatigue breakage of the plates. Nevertheless, the fixation pattern of 1/3 tube buttress plate could only provide a single point medial support for the QLP fracture fragments resulted in a concentrated high-stress pattern in the region[16]. In a stark contrast, the quadrilateral screws were distributed throughout the QLP region, and the stress distribution was more dispersive. Second, as we had described previously, unlike the end of 1/3 tube buttress plate fit snugly to quadrilateral surface, 1/3 to 1/2 transverse diameter of the quadrilateral screws was inserted into the bone of QLP, which can not only block the inward displacement of QLP, but also play an important role in lifting and pulling the fracture fragments of posterior column[3]. To demonstrate the existence of the lifting and pulling forces of quadrilateral screws[3], we validated the above conjecture by analyzing the direction of the maximum tensile strain. The directions of maximum tensile strain in Group B were located between − 31.72° and − 44.27°, with an average angle of −39.10°. As illustrated in Fig. 8, the directions of the maximum principal tensile strains in Group B closely aligned with the orientations of the quadrilateral screws. These observations indirectly suggest that the quadrilateral screws exert substantial lifting and pulling forces on the fracture fragments of the posterior column.

Limitations

Despite the novelty of our study, there are several limitations that merit discussion. First, due to restrictions including the number of specimens and imperfect devices, we did only simulate the situation in which patients turn over to the fractured side. While in actual clinical work, patients were often encouraged to turn over to the healthy side after acetabular fracture surgery to avoid the injured side under compression. Although we are cognizant of these considerations, we suppose that the experimental design of turning over to the fractured side is more representative. It is clear that further refinement of the current experimental protocols is necessary. Second, to attain high-quality outcome measurement data and given the reproducibility, major muscle groups in the acetabulum and spine have been removed and the movements in bilateral lower limbs have been limited, which might ignore the important effects of muscle activation and the strength of the lower limbs during the process of turning up. Third, torsion loads in the in vivo situation may be different from the load applied in vitro. However, this study has been carefully designed to minimize such variation from specimen selection, model establishment and parameter setting.

Conclusions

In this biomechanical study, we introduced a novel loading protocol designed to preliminarily simulate the action of turning over in bed. Our findings indicate that DAPSQ with quadrilateral screws for unstable acetabular fractures involving the QLP offers superior biomechanical anti-rotational stability compared to the traditional anterior reconstruction titanium plate and 1/3 tubular buttress plate.

Data availability

The datasets used and/or analysed during the current study available from the corresponding author on reasonable request.

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Author contributions
Haiyang Wu, Qixiao Shen, Xianhua Cai: Conceptualization, Methodology, Data Curation, Formal analysis, Resources, Investigation, Writing - Original Draft; Haiyang Wu, Zaijie Sun, Xuejian Wu, and Cheng Li: Conceptualization, Methodology, Formal analysis, Resources, Investigation, Writing- Review & Editing.

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Declarations

Competing interests
The authors declare no competing interests.

Ethics review committee
The study had been performed in accordance with the Declaration of Helsinki and had been approved by ethics committee of General Hospital of Central Theater Command. Informed consent was obtained from the legal guardians of these specimens.

Additional information
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