



Biomechanical Analysis of the External Fixation in a Lumbar Fracture Model: A Finite Element Study

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Abstract

Purpose This study aimed to investigate the biomechanical characteristics of the external spinal fixation for treating lumbar fracture through finite element analysis (FEA) and provide a theoretical basis for its further application.

Methods Two different models of L3 fracture fixed with the external spinal fixation and the internal fixation system respectively were constructed. The ROM, maximum stresses at L3, and the screws of the two models were measured under load control. Subsequently, the applied torque, the maximum stressed at L3, L1/2, L2/3, L3/4, L4/5 discs and the screws were analyzed under displacement control.

Results Under load control, the external fixation model reserved more ROM than the internal fixation model (40.4–48.0% vs 30.5–41.0%). Compared to the internal fixation model, the maximum stresses at L3 and screws in the external fixation model were increased. Under displacement control, the external fixation model required fewer moments (N·mm) than the internal fixation model (flexion: 7500 vs 12,294; extension: 7500 vs 9027). Further, the maximum stresses at L3 and the screws in the external fixation model were greater than those of the internal fixation model, while the maximum stresses at the upper and lower adjacent discs of fixed segments were less than the internal fixation model.

Conclusion Compared to the internal fixation system, the external fixation has a better stress distribution with the greater overall mobility. It theoretically reduces the stress concentration of the adjacent discs and the stress shielding of the fractured vertebral body.

Keywords External fixation · Finite element analysis · Lumbar fracture · Biomechanics

1 Introduction

Lumbar fractures are the most common clinical spinal fractures [1, 2]. The mild fractures cause pain and restricted mobility in the lower back, while the severe burst fractures may even protrude into the spinal canal, compressing the spinal cord causing corresponding nerve paralysis.

The posterior short-segment pedicle rod fixation is widely used in lumbar fractures and have achieved good clinical effects[3–6], however, a series of complications including adjacent segmental disc degeneration, loosening of the internal fixation and fracture non-union caused by stress shielding may still occur [7–9]. Thus, there is a demand of fixation that can stabilize the fracture segments and diminish the complications to treat lumbar fractures.

Provoked by the first application of the external spine skeletal fixation (ESSF) [10] and the Chinese traditional medicine concept of “Emphasizing both fasciae and bone, combining dynamic and static,” we developed an external spinal fixation and performed preliminary biomechanical test on it. Due to the advantages of minimal invasive, good reduction results, rapid rehabilitation, ease-to-use and non-destroy to spinal functional unit (FSU), it achieved a certain effect in clinic. The follow-up results of 50 patients who accepted the external spinal fixation were satisfied [11, 12].

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The finite element analysis (FEA) is an engineering technology used in stress analysis and is widely used in orthopedics and other medical fields. Moreover FEA is increasingly used in building lumbar fracture models for conducting mechanical analyses of various types of implants [13, 14].

This study investigated the biomechanical characteristics of the external spinal fixation in treating lumbar fractures compared to the pedicle screw rod internal fixation. The overall mobility, the maximum stresses at the screws and the fractured vertebral body, and the stress changes of the intervertebral disc in the fixed segment and its adjacent segments after fixation of a lumbar spine fracture were evaluated using FEM. We aimed to provide a comprehensive biomechanical analysis for the further application of the external spinal fixation.

2 Material and Methods

2.1 Construction of a Normal Lumbar Spine in FE Model

The data for normal lumbar spine model construction was obtained from a healthy volunteer (male, 26 years old, weight: 71 kg, Height: 173 cm) using the 64 slice spiral computed tomography (CT) scan. The obtained DICOM format files were subjected to thresholding segmentation, dynamic growth, mask editing, and Boolean manipulation using three dimensional (3D) reconstruction software Mimics 20.0 (Materialise, Belgium), and 3D contour models were extracted for each vertebra L1-L5. The 3D models were subsequently sliced one by one through Geomagic 12 (Geomagic Inc., USA) and underwent a series of image processing techniques, including smoothing, grinding, denoising, surface construction, and solidification. The optimized 3D model of the spine was further processed by SolidWorks 2015 (Dassault, France) to reconstruct the intervertebral disc (matrix and nucleus pulposus) and articular surface of the articular process and to complete the modeling of the normal model.

After reconstruction, vertebral body of the model was composed of cortical and cancellous bones and endplates. The intervertebral disc was composed of nucleus pulposus and annulus fibrosus (3:7), and the vertebral body was connected to adjacent discs. The articular cartilage thickness was set at 0.3 mm, and the friction coefficient of the upper and lower articular cartilage frictional contact was 0.1. Bonding contact was applied for the model except facet joint which used frictional contact. Seven paraspinal ligaments, including the anterior longitudinal ligament, posterior longitudinal ligament, ligamentum flavum, interspinous ligament, supraspinous ligament, capsular ligament, and transverse interspinous ligament, were simulated. The materials and

Table 1 Material properties of the finite element model

Component name	Young's modulus(MPa)	Poisson's ratio	Cross-sectional area(mm ²)
Cortical bone	12,000	0.3	–
Cancellous bone	100	0.3	–
Injured cancellous bone	10	0.3	–
Cartilage	10	0.4	–
Bony endplate	1000	0.4	–
Nucleus pulposus	1	0.499	–
Annulus fibrosus	4.2	0.3	–
ALL	20	0.3	63.7
PLL	20	0.3	20
LF	19.5	0.3	40
ISL	11.6	0.3	40
SSL	15	0.3	30
TL	58.7	0.3	3.6
CL	32.9	0.3	60
Instruments	110,000	0.3	–

ALL anterior longitudinal ligament, PLL posterior longitudinal ligament, LF ligamentum flavum, ISL interspinous ligament, SSL supraspinous ligament, TL transverse ligament, CL capsular ligament

Table 2 Parameters for mesh convergence

Size of mesh (mm)	Element	Node	Stress on vertebra(MPa)
0.5	2,229,305	3,157,044	20.09
1	693,194	1,030,241	20.08
1.5	311,639	493,005	19.19
2	279,010	302,980	18.15
3	90,017	173,045	16.86

characteristics of the above correlation models were chosen based on previous studies [15, 16], as shown in Table 1.

The normal model was meshed by the size of 3, 2, 1.5, 1 and 0.5. The equivalent stress of vertebral cortical bone was taken as the reference. The change with 5% suggested that the mesh was converged. According to the results of mesh convergence as shown in Table 2, the mesh size of 1 mm was selected for the subsequent analyse in this study.

The overall model was meshed with hexahedrons for the intervertebral discs and facet joints and tetrahedrons for the other components, with a mesh size of 1 mm. The final L1-L5 model was completed by applying 693,194 elements and 1,030,241 nodes using the Ansys workbench 18.0 (Ansys, USA). Then, the constructed L1-L5 full model was validated by comparing it with previous study data.

The internal fixation system's pedicle screws and connection rods had commonly used diameter specifications, 6.5

and 6 mm, respectively, while those for the new spinal external fixation pedicle screws and connection rods were 5.5 and 5 mm, respectively. Models of the internal and external fixation systems were constructed using SolidWorks 2015 (Dassault, France) (Fig. 1G and I).

Pedicle screws were placed parallel to the upper endplates in both fixation models, and the depth of screw placement into the vertebral body was 60%. The pedicle screw holders of the internal fixation system were partially adherent to the bone surface, and the connecting rods were fixed inside the screw holder U-groove. In contrast, connecting rods of the external fixation system were fixed within the pyridial apparatus secured at 65.0 and 70.0 mm from the talar surface of the L2 and L4 segments, respectively. A line parallel to the superior endplate was made from the anterior margin to the posterior margin of the L3 vertebral body at 15% from the superior margin, then an oblique line was made from the intersection of this line and the posterior margin of the vertebral body to the anterior margin of the vertebral body at 55% from the superior margin, which was in a triangular shape to cut the part of the vertebral body between the two lines to establish the fracture reduction model [17, 18]. Finally, the internal fixation model included 694,386 elements and 1,058,953 nodes, while the external fixation model included 739,105 elements and 1,121,711 nodes.

2.2 Loading and Boundary Conditions

2.2.1 Load Control

The upper edge of the L1 vertebral body was set as the loading plane without accepting any restraint, and the lower edge of the L5 vertebral body had zero degrees of freedom. Then a 500 N preload was applied to the superior surface of the L1 vertebral body and an additional movement moment of 7.5 N·m. The maximum activities of the normal model, internal fixation, and external fixation models were measured by observing the activity of each motion segment under the six conditions, including the forward flexion, extension, left and right flexion, and left and right axial rotation. Then the maximum stress of the fractured vertebral body (L3 vertebral body) and pedicle screw of internal and external fixation models were analyzed.

2.2.2 Displacement Control

Azusa Nishizawa et al. reported that the forward flexion and backward extension have the greatest influence on the intervertebral disc pressure when changing the body position (forward flexion, extension, left and right lateral flexion, and left and right rotation), and also the most frequent

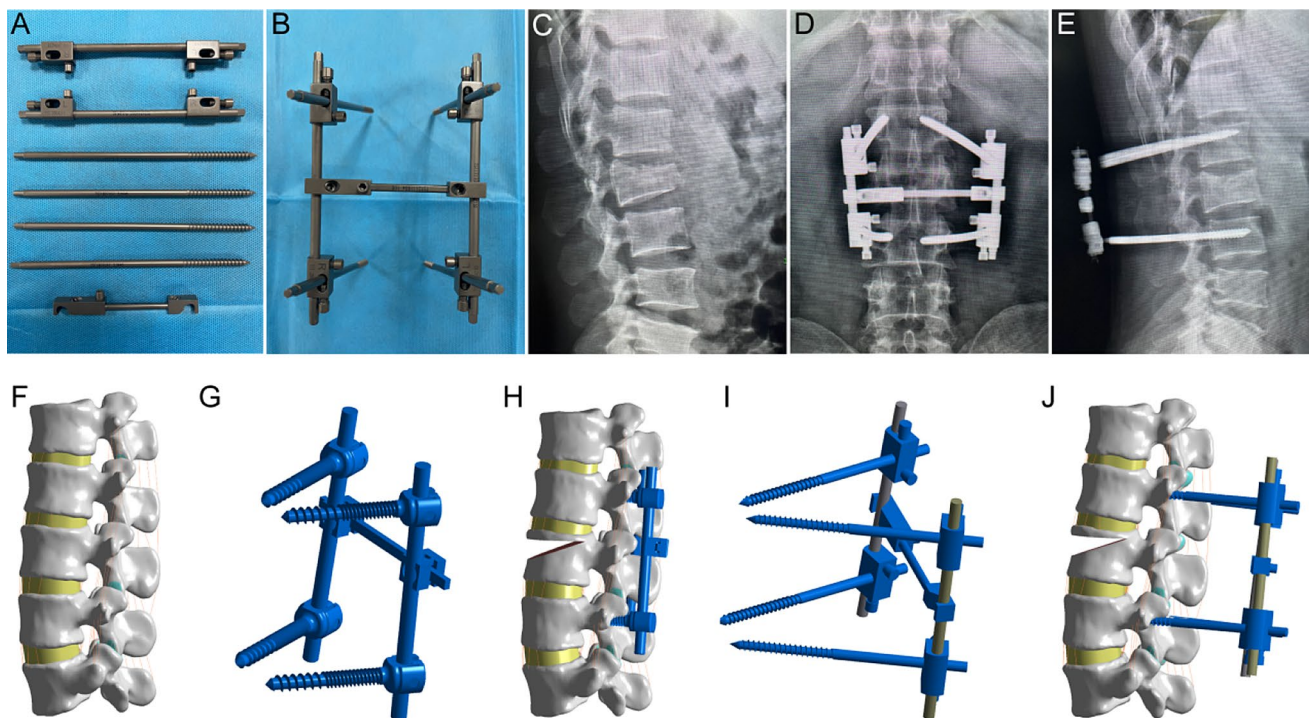


Fig. 1 Clinical application of the external fixation and model establishment **A** Components of the external fixation. **B** The overall diagram of the external fixation. **C–E** X-Ray images of clinical application of the external fixation. **F** Model of normal lumbar spine. **G**

Model of traditional internal fixation. **H** Model of L3 fractured fixed with the internal fixation. **I** Model of external fixation. **J** Model of L3 fractured fixed with the external fixation

two directions involving activities of daily life [19]. Thus, we studied the forward flexion and extension movements considering their significance on disc pressure and daily life activities. The maximum ROM of the external fixation model during the forward flexion and backward extension was obtained using the previously described load control method and was taken as the target value. For the internal fixation model, a 500 N preload was imposed to the upper surface of the L1 vertebral body with a constraint of 0 degrees of freedom to the lower surface of the L5 vertebral body. Further, it was allowed to reach the desired ROM target value under both forward flexion and backward extension by gradually increasing the additional movement moment. The additional movement moment that both models need to apply when reaching the same ROM was compared. The maximum stresses of L3, pedicle screws, discs of the fixed, and their adjacent segments were analyzed in both internal and external fixation models.

3 Results

3.1 Model Validation

In this study, the mobility of each segment of the 3D FE model of the lumbar spine was compared with previous study findings [20–24] and was within the range of reported literature (Fig. 2), confirming the validity of this model.

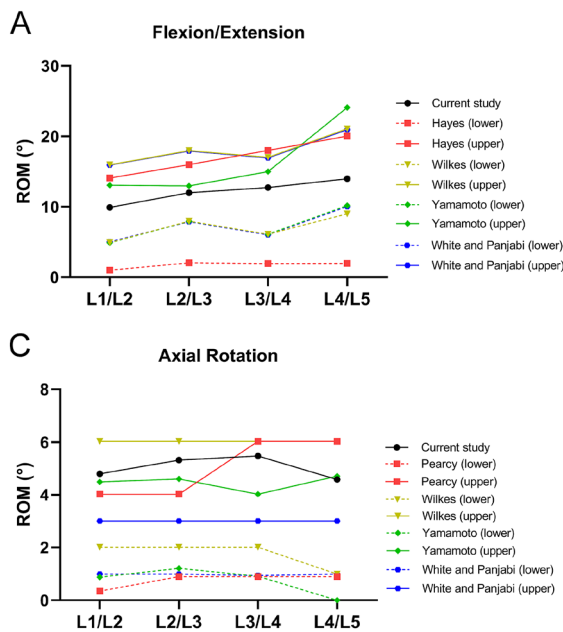


Fig. 2 Validation of the lumbar spine model. **A** Comparison of flexion/extension ROM between current study and previous literature. **B** Comparison of lateral bending ROM between current study and pre-

3.2 Load Control

3.2.1 Range of Motion (ROM)

The maximum ROM of the normal, external fixation, and internal fixation models, under the six operating conditions, are shown in Fig. 3. The ROM of the internal fixation model was less than those of the other two groups, reaching only 30.5–41.0% of the normal model under each operating condition. The external fixation model had a greater ROM than the internal fixation model and reached 40.4–48.0% of the normal model. ROM of the external fixation model under the six conditions were 12.02° (flexion), 8.31° (extension),

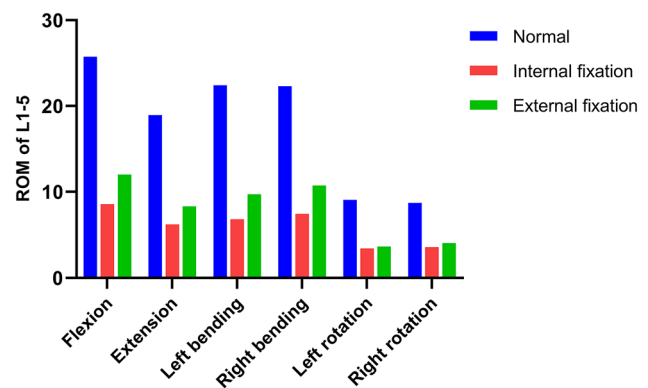
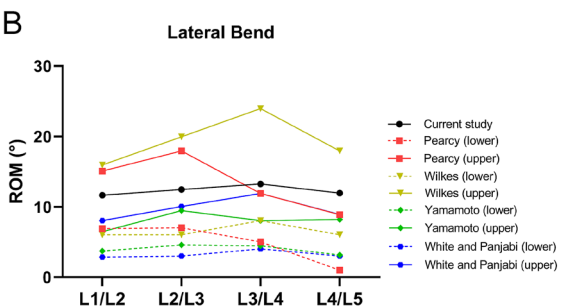


Fig. 3 Restriction of two fixations on lumbar mobility under the six conditions



vious literature. **C** Comparison of axial rotation ROM between current study and previous literature

9.72° (left bending), 10.73° (right bending), 3.67° (left rotation) and 4.07° (right rotation). ROM of the internal fixation model under the six conditions were 8.56° (flexion), 6.24° (extension), 6.84° (left bending), 7.45° (right bending), 3.43° (left rotation) and 3.58° (right rotation).

3.2.2 Von Mises Stress of the Fractured Vertebra

The maximum stresses of the fractured vertebral bodies of the external fixation and internal fixation models under the six operating conditions are shown in Fig. 4. The maximum stresses on the L3 body in the external fixation model were larger than those of the internal fixation model. In the external fixation model, the maximum stresses on L3 under flexion, extension, left bending, right bending, left rotation and right rotation were 1.34, 2.21, 1.71, 1.89, 2.88, 2.71 MPa respectively. In the internal fixation model, the maximum stresses on L3 under flexion, extension, left bending, right bending, left rotation and right rotation were 0.60, 0.49, 0.65, 0.75, 1.19, 0.97 MPa respectively.

3.2.3 Von Mises Stress of the Pedicle Screws

The stress cloud plots and maximum stress values of pedicle screws of the external and internal fixation models under the six operating conditions are shown in Fig. 5. The maximum stress values of pedicle screws of the external and internal fixation models appeared at the junction between the screw and the vertebral body. The maximum stress value of the screws of the external fixation model was larger than that of the internal fixation model under all six operating conditions. It was prominent in the forward flexion, backward extension, and left and right side flexion. In the external fixation model, the maximum stresses on screws under flexion, extension, left bending, right bending, left rotation and right rotation were 230.66, 273.21, 259.40, 266.98, 236.48, 235.43 MPa respectively. In the internal fixation model,

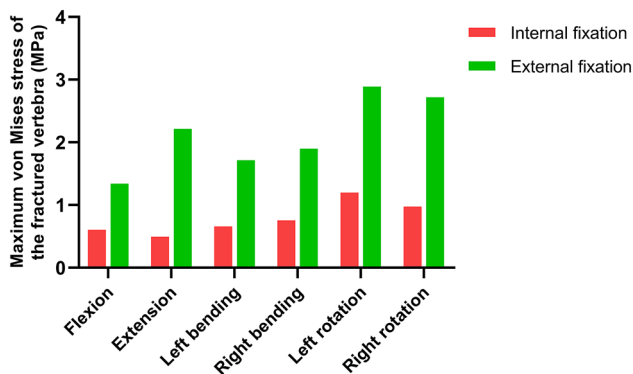


Fig. 4 Maximum von mises stress on fractured vertebra body (L3) in the internal and external fixations under the six conditions

the maximum stresses on screws under flexion, extension, left bending, right bending, left rotation and right rotation were 129.41, 142.72, 127.42, 115.30, 212.38, 193.89 MPa respectively.

3.3 Displacement Control

3.3.1 Applied Moment of Motion

We set the maximum ROM (12.02° for forward flexion and 8.31° for backward extension) obtained using the load control method as the target displacement. For 12.02° flexion, the external fixation model required 7500 N·mm and the internal fixation model required 12294 N·mm. For 8.31° extension, the external fixation model required 7500 N·mm and the internal fixation model required 9027 N·mm. The results show that the required motor moment of the internal fixation model was significantly greater than that of the external fixation model (Fig. 6A).

3.3.2 Von Mises Stress of the Pedicle Screws

The stress distribution of the pedicle screws of the two models under displacement control was consistent with that of the load control. In the external fixation model, the maximum stresses on screws under flexion and extension were 230.66 and 273.21 MPa. In the internal fixation model, the maximum stresses on screws under flexion and extension were 179.20 and 199.51 MPa. Although the maximum stress value of the external fixation model pedicle screw was greater than that of the internal fixation model under both load and displacement control, the stress gap of the two fixations was significantly decreased under the displacement control. (Fig. 6B).

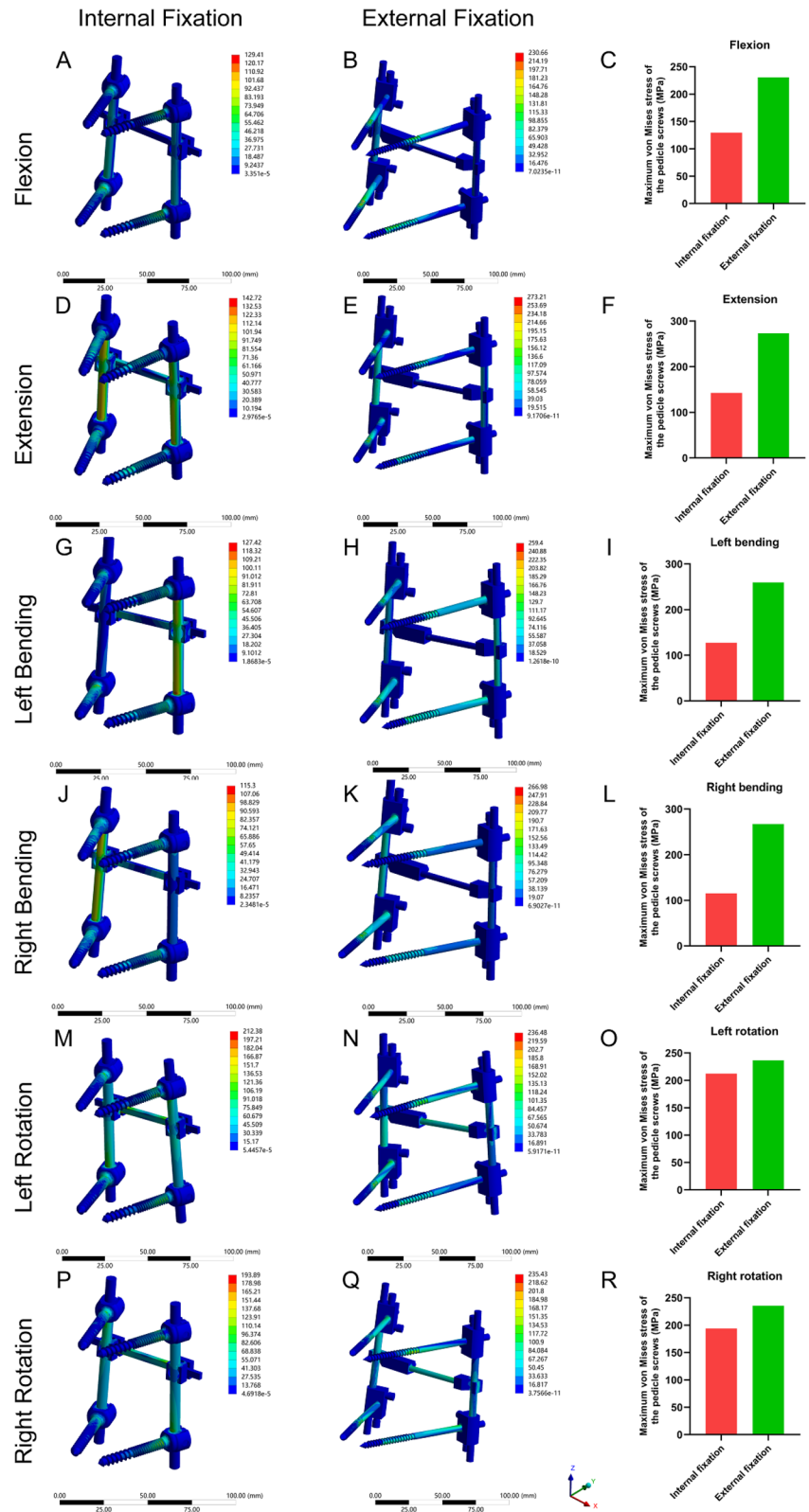
3.3.3 Von Mises Stress of the Fractured Vertebra

The maximum stresses on the fractured vertebral bodies of the external fixation and internal fixation models under displacement control are shown in Fig. 6C. In the external fixation model, the maximum stresses on L3 under flexion and extension were 1.34 and 2.21 MPa. In the internal fixation model, the maximum stresses on L3 under flexion and extension were 0.89 and 0.57 MPa. The maximum stress of the fractured vertebral body in the external fixation model was greater than that of the internal fixation model.

3.3.4 Von Mises Stress of the Intervertebral Disc

The maximum stresses on the intervertebral discs of the fixed segments (L2/L3 and L3/4) in both models were significantly decreased than discs of adjacent segments (L1/L2 and L4/L5). In the external fixation model, the maximum

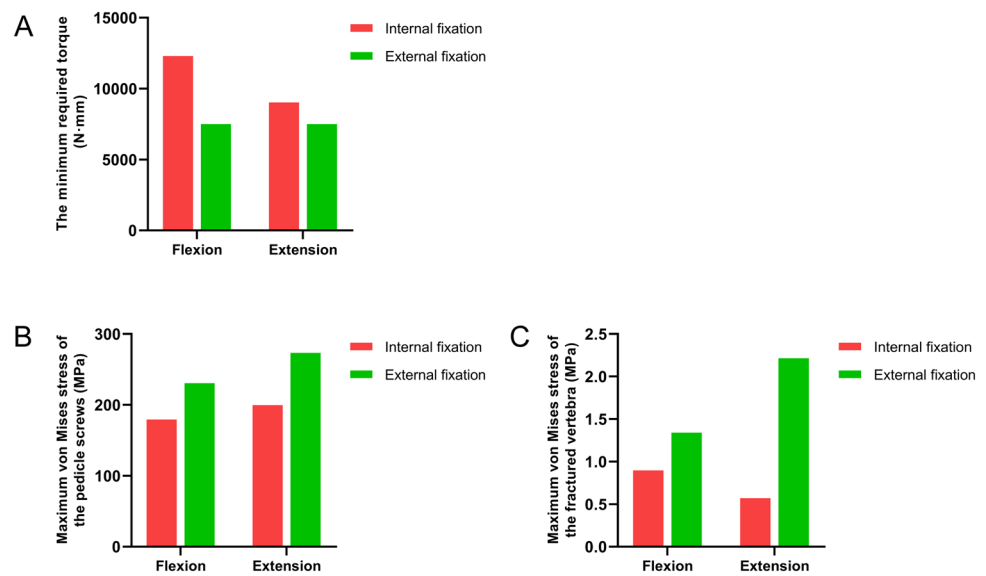
Fig. 5 Cloud plots and histogram of the Maximum von mises stress on pedicle screws of the internal and external fixations under the six conditions



stresses on L1/2 disc under flexion and extension were 1.73 and 2.25 MPa; the maximum stresses on L4/5 disc under flexion and extension were 1.43 and 2.13 MPa. In

the internal fixation model, the maximum stresses on L1/2 disc under flexion and extension were 1.90 and 2.61 MPa; the maximum stresses on L4/5 disc under flexion and extension were 1.91 and 2.23 MPa.

Fig. 6 Comparison of the internal and external fixations under the displacement control. **A** Required torque for the two fixations to reach the ROM of flexion 12.02° and extension 8.31° . **B** Maximum von mises stress on fractured vertebra body (L3) in the internal and external fixations under the displacement control. **C** Maximum von mises stress on pedicle screws of the internal and external fixations under the displacement control



Under displacement control, the maximum stress of the intervertebral discs in adjacent segments (L1/L2 and L4/L5) of the external fixation model was less than that of the internal fixation model, whereas the maximum stress of the intervertebral discs in fixed segments (L2/L3 and L3/4) of the external fixation model was greater than that of the internal fixation model. The smaller difference between the fixed segments and the adjacent segments in the external fixation model indicates a better stress distribution from L1/2 to L4/5 (Fig. 7B and C). The L1/L2 and L4/L5 segmental disc stress cloud diagrams of the two models are shown in Fig. 7A.

4 Discussion

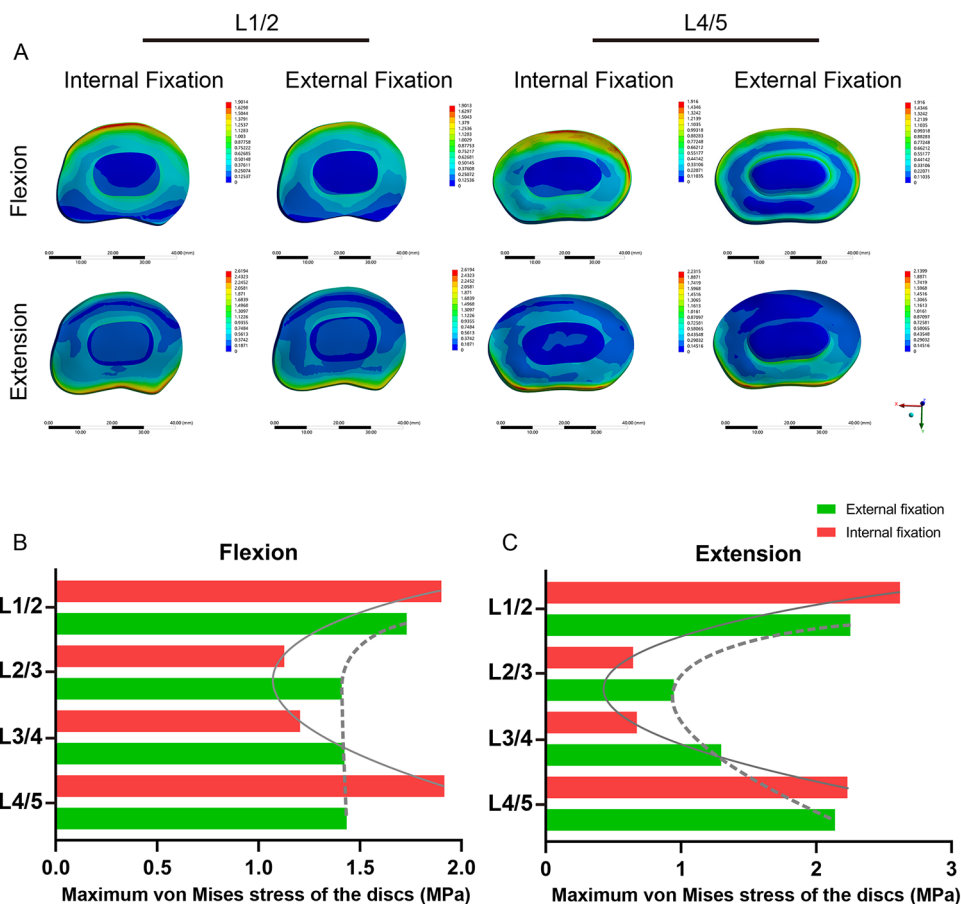
Lumbar fracture is a common spinal trauma [1, 2]. The internal and external fixation were both appropriate therapeutic strategy for lumbar fracture [5, 6, 11]. However, the differences of biomechanical characteristics between internal and external fixation were rarely reported. In this study, we established a finite model of L3 fracture and simulated the conditions under internal and external fixation to analyzed the ROM of lumbar spine, the maximum stress on fractured vertebra and screws. We found that the external fixation could reduce the stress shielding and improve the stress distribution on the premise of ROM restriction and fixation stability. This study aimed to analyze and validate the biomechanical mechanism underlying the phenomenon in clinical application of the external spinal fixation system via the 3D FE simulation and analysis. The findings provided more theoretical data for the future clinical application of the external fixation.

Finite element analysis is one of the most popular methods for predicting the biomechanical properties of

orthopedic implants. We constructed a lumbar finite model with 693,194 elements and 1,030,241 nodes. The finite model of L3 fracture was constructed according to the previous reports [17, 18]. Validation results showed that the ROM of the presenting finite model was not obvious different from the reported models [20–24]. Therefore, we simulated the internal and external fixation on the presenting L3 fracture model. There were 694,386 elements and 1,058,953 nodes in the internal fixation model and 739,105 elements and 1,121,711 nodes in the external fixation model.

The core concept of the external spinal fixation is that the vertebral body is fixed extracorporeally using the Schanz screws, and the screws can be distracted, pulled, and pryed using the extracorporeal prydial kit as a fulcrum to correct, reduce, and fix the fractured vertebral body. Because the Schanz screw with the external prydial kit is longer than the force arm of the conventional internal fixation pedicle screws, the overall system acts like a micro-dynamic elastic fixation. Results showed that the external fixation restricted the ROM of lumbar spine to 40.4–48.0% and the internal fixation restricted the ROM of lumbar spine to 30.5–41.0%, which indicated that the external fixation preserved more ROM of lumbar spine. The mechanism of the external spinal fixation is compatible with the Chinese traditional medicine concept “Emphasizing both fasciae and bone, combining dynamic and static.” “Emphasizing both fasciae and bone” is mainly manifested in that the installation of the external fixation is based on the percutaneous pedicle screw technique which avoids open operation and extensive dissection of paravertebral muscle and minimizes the damage to soft tissue. More importantly, the external fixation can be removed at outpatient after fracture healed without removal operation. “Combining dynamic and static” is mainly manifested in that the effects of elastic fixation can be achieved through the

Fig. 7 Stress distribution of lumbar discs in two fixations under the displacement control. **A** Cloud plots of adjacent discs (L1/2 and L4/5) in the internal and external fixations. **B** Maximum von mises stress of lumbar discs (L1/2–L4/5) under the flexion condition. **C** Maximum von mises stress of lumbar discs (L1/2–L4/5) under the extension condition



deformation of screws in a static screw rod fixation system, allowing local micro-movement to stimulate callus formation of the fractured body [25, 26].

There is a consensus that appropriate stress stimulation is vital for fracture healing and stress shielding may lead to delay in fracture healing or even nonunion [25–28]. We analyzed the maximum stress of the fractured vertebral body in these two fixation models under both load and displacement control. Under load control, the maximum stresses of L3 in the internal fixation were 0.60 MPa (flexion), 0.49 MPa (extension), 0.65 MPa (left bending), 0.75 MPa (right bending), 1.19 MPa (left rotation), 0.97 MPa (right rotation) and the maximum stresses of L3 in the external fixation were 1.34 MPa (flexion), 2.21 MPa (extension), 1.71 MPa (left bending), 1.89 MPa (right bending), 2.88 MPa (left rotation), 2.71 MPa (right rotation). Under displacement control, the maximum stresses of L3 in the internal fixation were 0.89 MPa (flexion) and 0.57 MPa (extension), and the maximum stresses of L3 in the external fixation were 1.34 MPa (flexion) and 2.21 MPa (extension). The maximum stress on the fractured vertebral body of the external fixation was greater than that of the internal fixation under all the conditions. Accordingly, we deduced that the increased stress of the fractured vertebral body in the external spinal fixation

might promote fracture healing (some cases such as severe osteoporosis should be excluded). In contrast, given the characteristic of rigid fixation, the internal spinal fixation system can cause the stress shielding of the fractured vertebral body, further it can induce the bone defect, unstructured bone trabecular and delayed healing in the fractured segment [9]. These findings partly revealed the mechanism underlying the clinical phenomenon that patients with lumbar fracture can achieve clinical healing within three months via treatment of the external spinal fixation system [11].

Adjacent Segment Degeneration (ASD) following lumbar fusion is due to the compensatory enlarged ROM and overloaded stress in the discs and articular process joints of adjacent segments because the rigid fixation limits the motion of the fixed segment and alters the motional mode of the lumbar spine [29, 30]. We found that the maximum stresses on the adjacent discs of the external fixation were smaller than those of the internal fixation (L1/2: 1.73 vs 1.90 MPa in flexion, 2.25 vs 2.61 MPa in extension; L4/5: 1.43 vs 1.91 MPa in flexion, 2.13 vs 2.23 MPa in extension). Further, the maximum stress gap between the discs within fixed segments and adjacent segments was decreased in external fixation. In the external fixation, the stresses were relatively and evenly distributed to the whole lumbar spine. Additionally, the fixation time of the

spinal external fixation system is seldom beyond three months generally. Therefore, the better stress distribution and the shorter fixation time make the external fixation has the capacity to reduce the risk for developing ASD postoperatively.

Screw breakage is a severe complication of spinal fixation surgery [31]. It is closely related to the maximum stress on the screws or rods. In this study, we analyzed the maximum stress of the two models and found that the maximum stresses were both concentrated on the pedicle screws. Under load control, the maximum stresses of screw in the internal fixation were 129.41 MPa (flexion), 142.72 MPa (extension), 127.42 MPa (left bending), 115.30 MPa (right bending), 212.38 MPa (left rotation), 193.89 MPa (right rotation) and the maximum stresses of screw in the external fixation were 230.66 MPa (flexion), 273.21 MPa (extension), 259.40 MPa (left bending), 266.98 MPa (right bending), 236.48 MPa (left rotation), 235.43 MPa (right rotation). Under displacement control, the maximum stresses of screw in the internal fixation were 179.20 MPa (flexion) and 199.51 MPa (extension), and the maximum stresses of screw in the external fixation were 230.66 MPa (flexion) and 273.21 MPa (extension). The maximum stress of external fixation system is larger than that of internal fixation system, which might be caused by the longer force arm of the external fixation pedicle screws. Although the stress of the external fixation screw is larger, the existing maximum stress is much lower than the threshold of static fracture of the material (924 MPa) [18]. In addition, the shorter fixation period of external fixation (3–4 months) also contributes to decreasing the risk of screw breakage. Intriguingly, the internal fixation required a greater torque (12294 N·mm for flexion, 9027 N·mm for extension) to achieve the same ROM (12.02° flexion and 8.31° extension) than the external spinal fixation (7500 N·mm for flexion and extension) under the displacement control, which suggested that patients who adopt the internal fixation may need a larger force produced by lumbar-back muscles than those who adopt the external fixation to complete a same action, adversely affecting postoperative recovery.

There are some limitations in our study. First, the finite element model of the lumbar spine cannot fully reflect the real situation of patients. Animal experiment should be considered to better understand the effectiveness the external fixation. Second, because of the percutaneous surgical procedure, the external fixation are inapplicable for some situation where the open surgery is necessary.

5 Conclusion

In conclusion, we find that the external spinal fixation shows the better biomechanical characteristics in ROM preservation and stress distribution, which is beneficial to

postoperative rehabilitation. Although the maximum stress of the external fixation is greater than that of internal fixation, it is much lower than the breaking threshold of the screw. Results also indicate that the external spinal fixation can theoretically prevent ASD through decreasing the stresses of the adjacent discs. Therefore, the external spinal fixation might be a better alternative for lumbar fracture.

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Declarations

Conflict of Interest The authors declare that they have no competing interests.

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