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Original Article

Three-dimensional finite element analysis of unilateral and bilateral pedicle screw fixation with intervertebral body fusion for degenerative lumbar instability

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Abstract: This study aims to establish three-dimensional finite element models of lumbar instability, and compares the biomechanics of unilateral and bilateral pedicle screw fixation with intervertebral body fusion. Simulated physiological loads with axial compression, anterior flexion, posterior extension, left lateral bending, right lateral bending, left-handed rotation, and right-handed rotation were applied in unilateral and bilateral fixation models. The stresses of pedicle screws, intervertebral fusion cages, and adjacent intervertebral discs, and the displacement of the vertebral body in two models were recorded and compared. Under 7 kinds of loads, the stresses of screws in a unilateral fixation model were higher than in a bilateral fixation model, but there were no significant differences ($P>0.05$). Under loads with left and right lateral bending, and left- and right-handed rotation, the stresses of intervertebral fusion cages in a unilateral fixation model were significantly higher than in a bilateral fixation model ($P<0.05$). Under 6 kinds of loads except for axial compression, the stresses of adjacent upper and lower intervertebral discs in a bilateral fixation model were clearly greater than in a unilateral fixed model ($P<0.05$), and the stress of upper intervertebral discs was clearly greater than that of lower intervertebral discs ($P<0.05$). There was no significant difference in L4 vertebral body displacement between the two models under different loads ($P>0.05$). Unilateral pedicle screw fixation with intervertebral body fusion can be used for treatment of lumbar instability. It can provide initial stability for spinal fusion, and reduce the effect on adjacent segment degeneration.

Keywords: Lumbar instability, pedicle screw fixation, biomechanics, finite element analysis

Introduction

Degenerative lumbar instability causes lumbosacral pain and significantly affects routine work and daily life. Surgical treatment is often needed. Complete decompression, reduction and internal fixation, and bone graft fusion have become the main treatment methods for this condition, with a high spinal fusion rate [1]. However, McAfee et al. [2] reported that overly rigid internal fixation of the spine can cause stress shielding in the bone graft area, which leads to osteoporosis and bone graft absorption, resulting in a decreased bone fusion rate. An appropriate amount of stress is beneficial for fusion in the bone graft area. The clinical application of unilateral pedicle screw fixation was first reported by Kabins et al. in 1992, with development of minimally invasive techniques in spinal surgery [3]. Suk et al. [4] compared unilateral and bilateral lumbar pedicle screw fixation, and found that there was no difference in fusion rate and complications between the two methods, but the operative time, hospitalization time, and treatment cost were statistically different. Increasingly often, authors perform unilateral pedicle screw fixation surgery, and have confirmed good treatment results [5, 6]. In addition, experiments on specimens proved that unilateral pedicle screw fixation can obtain a good biomechanical environment [7, 8]. However, experiments on specimens have certain limitations.

A three-dimensional finite element model can accurately simulate physiological activity in the lumbosacral region, and the corresponding biomechanical analysis has been reliable [9-11].

Table 1. Stress of pedicle screw under different loads (mean ± SD, Mpa, n=30)

Load	L4			L5		
	Unilateral fixation	Bilateral fixation	*P*	Unilateral fixation	Bilateral fixation	*P*
Axial compression	3.54±1.823	3.16±1.343	0.3618	3.07±1.031	2.78±0.944	0.2605
Anterior flexion	14.74±3.925	14.40±6.982	2.0452	12.23±3.744	11.45±3.841	0.4290
Posterior extension	10.89±5.417	10.69±4.574	0.9777	9.72±4.136	8.58±3.589	0.2589
Left lateral bending	12.42±5.225	10.74±5.445	0.2276	10.89±5.417	10.40±4.132	0.6951
Right lateral bending	12.42±5.239	9.85±4.466	0.1023	9.84±4.096	8.29±5.217	0.2013
Left-handed rotation	11.29±3.470	10.01±3.729	0.1740	8.78±1.228	8.23±3.434	2.0452
Right -handed rotation	10.68±3.827	8.43±4.852	0.0508	8.99±2.749	7.96±1.967	2.0452

However, research on three-dimensional finite element analysis of unilateral pedicle screw fixation is lacking. In this study, such models were established for L3-S1 vertebral segments. These models were used to mimic unilateral and bilateral pedicle screw fixation. Physiological loads with axial compression, anterior flexion, posterior extension, left lateral bending, right lateral bending, left-handed rotation, and right-handed rotation were applied in unilateral and bilateral fixation models. The stresses of pedicle screws, intervertebral fusion cages, and adjacent intervertebral discs, and the displacement of the L4 vertebral body in two models were investigated. The purpose of the present study was to develop a biomechanical support for treatment of lumbar instability using unilateral pedicle screw fixation with intervertebral body fusion.

Methods

General data

Experiment software included Mimics 13.0 (Materialise, Belgium), Hypermesh (Altair, USA), and Abaqus (Abaqus, USA). Major apparatus included 32-row spiral computed tomography (CT) (to collect original Digital Imaging and Communications in Medicine [DICOM] data; scanning thickness, 0.625 mm; GE, USA) and an internal fixator (prepared by molding, using products of Shandong Weigao Medical Equipment, China: pedicle screws, 6.5 mm × 50 mm; connecting rods, 6 mm × 80 mm; and intervertebral fusion cages, 10 mm × 15 mm × 22 mm).

CT data were collected from a healthy male (27 years old, 172 cm) with no lumbar trauma or lumbocrural pain history. Lumbar vertebral and intervertebral disc disease was excluded by X-ray and CT examination, and related anatomic parameters were in the normal range. Consent was obtained from the research subject.

Comparison experiments were performed in the Spine Laboratory, Affiliated Lianyungang Hospital of Xuzhou Medical College, China, and the Biomechanics Laboratory of Shanghai Jiao Tong University, China, from January 2011 to December 2011.

This study was conducted in accordance with the declaration of Helsinki, and with approval from the Ethics Committee of the First People's Hospital of Lianyungang Hospital Affiliated with Xuzhou Medical College. Written informed consent was obtained from all participants.

Establishment of finite element models of unilateral and lateral pedicle screw fixation

The ultrathin CT data (DICOM format) of L3-S1 lumbar segments were read into Mimics software to establish an L3-S1 three-dimensional finite element model. The STL format data for pedicle screws, connecting rods, and intervertebral fusion cages produced by Auto CAD software were read into Mimics. Pedicle screw fixation with posterior intervertebral body fusion was mimicked by translation and rotation. Unilateral embedding of 2 screws (right lateral) and bilateral embedding of 4 screws were performed.

The model was input in Hypermesh software, followed by surface mesh optimization of each part. Then, non-endemic assembly of each part of the model was performed to obtain a common interface. The unilateral fixed model included 10-node solid element models, such as L3-S1 vertebrae, 2 intervertebral discs, 2 screws, 1 intervertebral fusion, and 1 connect-

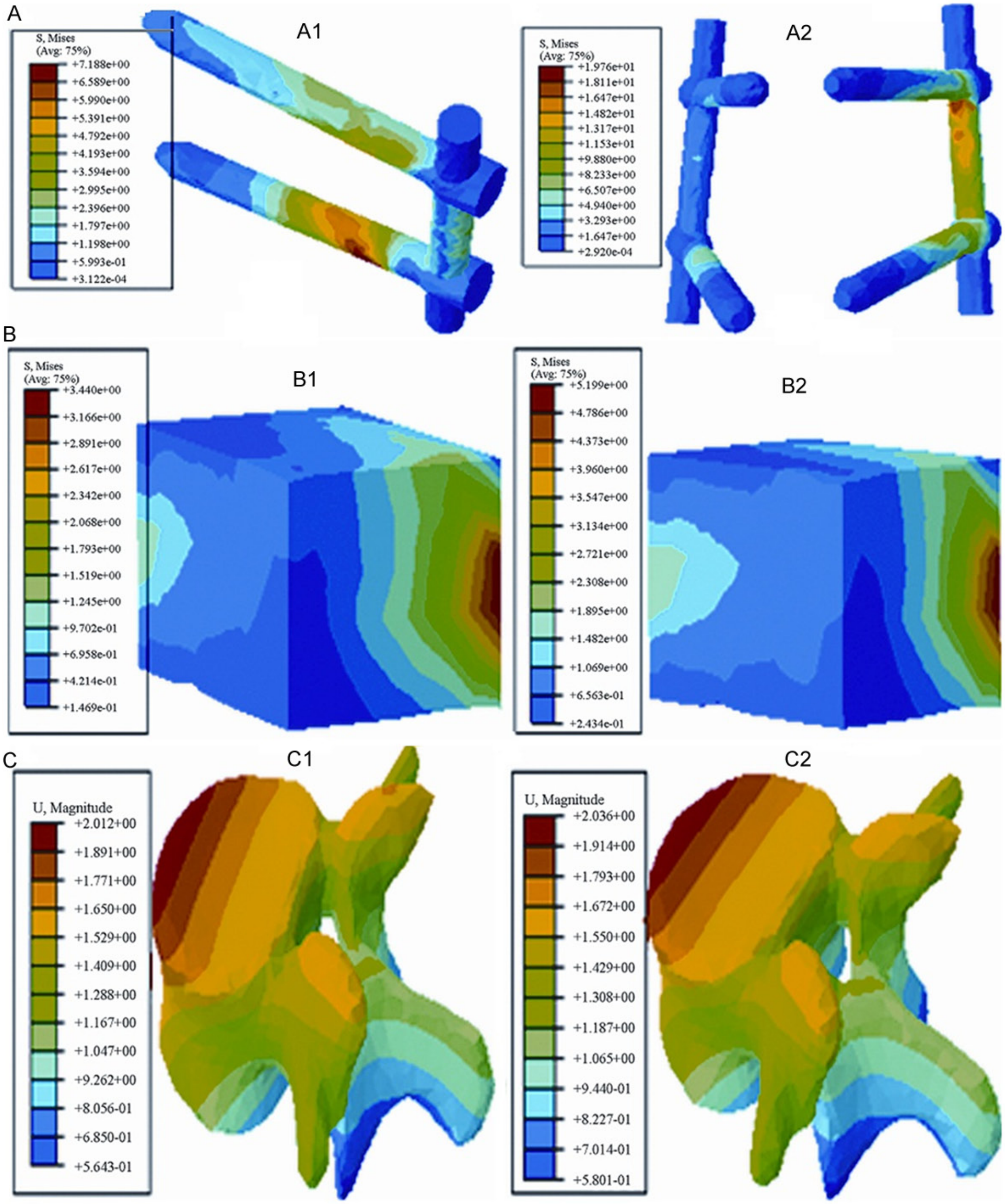

Figure 1. A. Stress nephogram of pedicle screw under load with axial compression. A1, unilateral fixation; A2, bilateral fixation. B. Stress nephogram of intervertebral fusion cage under load with left lateral bending anteroposterior view; B1, unilateral fixation; B2, bilateral fixation. C. Displacement nephogram of L4 vertebral body under load with anterior flexion. C1, unilateral fixation; C2, bilateral fixation.

ing rod, with a total of 14,984 nodes and 53,420 units. The bilateral fixation model included 10-node solid element models, such as L3-S1 vertebrae, 2 intervertebral discs, 4 screws, 1 intervertebral fusion, and 2 connecting rods, with a total of 16,800 nodes and 58,883 units.

Based on CT data for bone structure, the material function of the finite element analysis (FEA)

Table 2. Stress of intervertebral fusion cage under different loads (mean ± SD, Mpa, n=30)

Model	Axial compression	Anterior flexion	Posterior extension	Left lateral bending	Right lateral bending	Left-handed rotation	Right -handed rotation
Unilateral fixation	0.88±0.63	2.12±0.42	1.03±0.45	1.67±1.04	0.95±0.57	3.33±1.05	3.37±0.98
Bilateral fixation	0.76±0.42	2.07±0.48	0.85±0.42	1.11±0.72	0.76±0.61	2.55±0.81	2.46±0.89
P	0.3994	0.6692	0.1147	0.0184	0.0472	0.0021	0.0004

Table 3. Stress of adjacent intervertebral discs under different loads (mean ± SD, Mpa, n=48)

Load	Upper intervertebral disc			Lower intervertebral disc		
	Unilateral fixation	Bilateral fixation	*P*	Unilateral fixation	Bilateral fixation	*P*
Axial compression	0.0814±0.0286	0.0815±0.0288	0.115	0.0353±0.0124	0.0355±0.0123	0.108
Anterior flexion	0.1959±0.0726	0.1965±0.0729	0.000	0.1136±0.0410	0.1149±0.0401	0.000
Posterior extension	0.0417±0.0147	0.0433±0.0145	0.043	0.0325±0.0154	0.0329±0.0148	0.000
Left lateral bending	0.0835±0.0357	0.0839±0.0361	0.014	0.0423±0.0162	0.0438±0.0163	0.000
Right lateral bending	0.0794±0.0304	0.0798±0.0303	0.011	0.0273±0.0094	0.0289±0.0088	0.000
Left-handed rotation	0.1068±0.0319	0.1075±0.0321	0.016	0.0427±0.0179	0.0438±0.0173	0.000
Right -handed rotation	0.0972±0.0307	0.0979±0.0307	0.000	0.0354±0.0107	0.0363±0.0108	0.000

module in Mimics software was applied to automatically define the density [12], Young's modulus [13], and Poisson's ratio [14] of the spinal bone structure by 10 grades. Young's modulus with maximum grade was defined as 12,000 MPa (cortical bone). Poisson's ratio for all bone structures was defined as 0.29 [15]. The materials of intervertebral discs, pedicle screws, intervertebral fusion cages, and added ligaments were designed using HyperMesh software [16]. The contacts between vertebral bodies and intervertebral discs, fusion cages, and screws were defined as Tie, and the contact between small joints was defined as finite sliding.

Loading and calculation

Abaqus software was used for loading and calculation. The lower surface of the S1 vertebral body was fixed. Axial compression (50 kg, 2/3 of adult weight) was applied to the upper surface of the L3 vertebral body, and 500 N compression was uniformly distributed on the upper surface of the entire lumbar vertebral body. In addition, under 500 N axial compression, 15 Nm of torque in anterior flexion, posterior extension, left lateral bending, right lateral bending, left-handed rotation, and right-handed rotation was applied to the upper surface of the L3 vertebral body to simulate physiological conditions. According to previous research data [17, 18], the stretching force of the anterior longitudinal ligament in posterior extension for lumbar activities with a normal intervertebral disc is 57.037 N. In anterior flexion, the stretching forces of the posterior longitudinal ligament, ligamentum flavum, and supraspinous ligament are 19.578 N, 9.976 N, and 13.29 N, respectively. Seven levels of torque under physiological conditions, such as lumbar upright, posterior extension, left lateral bending, right lateral bending, left-handed rotation, and right-handed rotation, were applied.

The purpose of this study was to compare the stresses of pedicle screws, intervertebral fusion cages, and adjacent intervertebral discs, and the displacement of the L4 vertebral body in two models. Intervertebral fusion could play the greatest role in micromodification of screws or rods, so the different effects of unilateral and bilateral screws or cages could be seen. Therefore, 500 N axial compression and 15 Nm of torque with anterior flexion, posterior extension, left lateral bending, right lateral bending, left-handed rotation, and right-handed rotation were selected, as in previous research [19].

Observation indices

Stress of pedicle screw. In clinical use, fatigue fracture of pedicle screws often occurs at the screw root. This was confirmed by Kim [20]. In this study, only the stress of the pedicle screw root was evaluated. Because of right lateral fix-

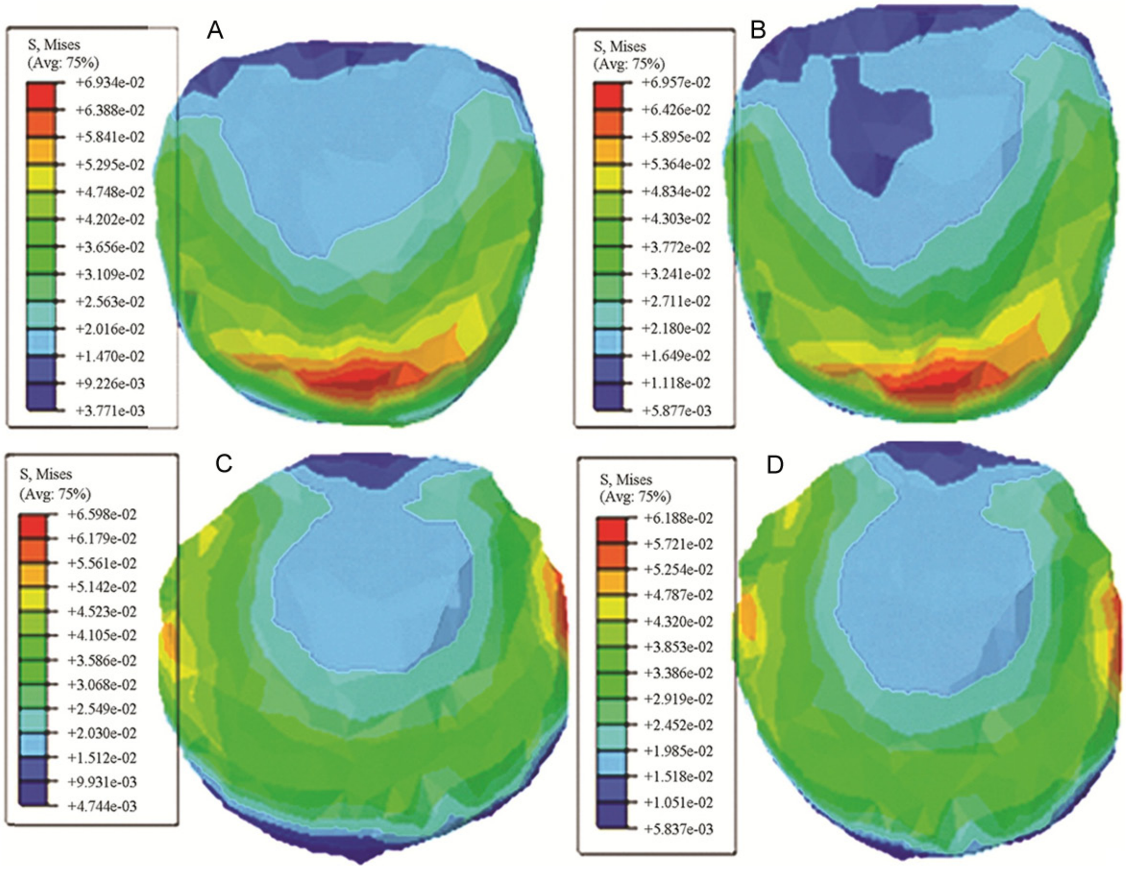

Figure 2. Stress nephogram of intervertebral disc. A. Unilateral fixation, upper intervertebral disc under load with posterior extension; B. Bilateral fixation, upper intervertebral disc under load with posterior extension; C. Unilateral fixation, lower intervertebral disc under load with right lateral bending; D. Bilateral fixation, lower intervertebral disc under load with right lateral bending.

ation, the stress of the screw on the right L4-5 vertebral body was studied. A total 30 nodes were equably selected at the screw root, and node stress was calculated under 7 kinds of loads.

Stress of intervertebral fusion cage. The intervertebral fusion cage was divided into upper and lower parts, with 15 nodes in each part. The node stress was calculated under 7 kinds of loads.

Stress of adjacent intervertebral discs. The L3-4 and L5-S1 intervertebral discs were divided into 4 areas (front left, front right, rear left, and rear right), using the center as boundary. Twelve nodes were equably selected. The node stress was calculated under the above loads.

L4 vertebral body displacement. The maximum displacements of the L4 vertebral body under the above loads were determined, and the stabilities of the two models were evaluated.

Statistical analysis

Data in each model were expressed using statistical tables. Analysis was performed using SPSS 16.0 software, and a *t*-test was used to analyze the differences in stress distribution under different loads in the two models. $P<0.05$ was considered as statistically significant.

Results

Stress of pedicle screw

Under loads with axial compression, anterior flexion, posterior extension, left lateral bending, right lateral bending, left-handed rotation, and right-handed rotation, the stresses of screws in the unilateral fixation model were greater than in the bilateral fixation model, but the difference was not statistically significant ($P>0.05$) (**Table 1**; **Figure 1**).

Stress of intervertebral fusion cage

Under loads with left and right lateral bending, and left- and right-handed rotation, the stresses of intervertebral fusion cages in the unilateral fixation model were significantly greater than in the bilateral fixation model ($P<0.05$). Under three other kinds of load, there was no significant difference in stress between the two models ($P>0.05$) (**Table 2**; **Figure 1**).

Stresses of adjacent intervertebral discs

Under loads with axial compression, the stresses of adjacent upper and lower intervertebral discs in the two models were not significantly different ($P>0.05$). Under six other kinds of load, the stresses of adjacent upper and lower intervertebral discs in the bilateral fixation model were clearly greater than in the unilateral fixation model ($P<0.05$), and the stress of the upper intervertebral disc was clearly greater than for the lower intervertebral disc ($P<0.05$) (**Table 3**; **Figure 2**).

Vertebral body displacement

The L4 vertebral body displacements in the two models were the greatest under load with anterior flexion, at 2.012 mm and 2.036 mm, respectively. There was no significant difference in maximum displacement between the two models under different loads ($P>0.05$) (**Figure 1**).

Discussion

In this study, three-dimensional finite element models were established from CT data of a patient's thoracolumbar spine, using Mimics software. Hypermesh software was used for geometric surface data smoothing, grid deletion, and optimization, which fully guaranteed the geometric accuracy of the simulacrum, and reduced analysis error due to geometry data loss. The parameters used in this study have been adopted by most researchers; this ensures the comparability of results to a certain extent. The L3-S1 finite element models can accurately simulate physiological activity in the lumbosacral region, in terms of maintenance of geometry, accuracy of material properties, and simulated mechanical characteristics. The lower surface of the S1 vertebral body was fixed, and compressions from different directions were applied to the upper surface of the L3 vertebral body. Axial displacements with 500 N and 2500 N were 0.3 and 1.5 mm, respectively. This is consistent with other experimental models [21]. The activities of pure flexion and extension, deformation degree under load, activities of left and right lateral bending, and left- and right-handed rotation torques are consistent with previous reports [22]. This confirms that the three-dimensional finite element models can simulate thoracolumbar physiological movement, and can be used for further experiments.

In various simulated movement states, the stress of pedicle screws in unilateral fixation models is greater than in bilateral fixation models, but the difference is not statistically significant ($P>0.05$). From a biomechanical viewpoint, this suggests that unilateral fixation can provide sufficient strength for reconstruction of spinal stability, as with bilateral fixation. This is consistent with existing research results [10, 11, 23]. The strength and stiffness of unilateral pedicle screw fixation are less than with bilateral fixation, but greater than in normal lumbar vertebrae. This can provide sufficient lumbar stability and a satisfactory mechanical environment for fusion.

Experimental results show that under loads with axial compression, anterior flexion and posterior extension, there is no significant difference in intervertebral fusion cage stress between the two models, suggesting that unilateral fixation does not affect the stress of the intervertebral fusion cage. This is consistent with the results of previous research [7, 8, 24] and clinical experiments [5, 6, 25, 26]. However, under four other kinds of loads, the intervertebral fusion cage stress in the unilateral fixation model is greater than in the bilateral fixation model. As these movements are not permitted for three months postoperatively, these differences do not affect the clinical outcome.

Most authors believe that the compensatory increase of activity and stress of adjacent segments after spinal internal fixation are important causes of adjacent segment degeneration [27]. In this study, under six kinds of loads except for axial compression, the stresses of adjacent upper and lower intervertebral discs in the bilateral fixation model are significantly greater than in the unilateral fixed model. Therefore, lumbar fusion can clearly affect the biomechanics of adjacent intervertebral discs. Unilateral fixation is better for relieving the stress of adjacent intervertebral discs than bilateral fixation. Accelerated degeneration of adjacent intervertebral discs after rigid internal fixation may be the result of kinematic changes and increased intradiscal pressure [28]. The adjacent segment stress concentration, compensatory activity increase, and stability loss are the most direct and major biomechanical changes leading to accelerated degeneration. The intradiscal pressure may also increase due to internal fixation. In this study, unilateral fixation has less strength, which causes decreased compensatory activity in adjacent segments, leading to reduction of intervertebral disc pressure. Thus, unilateral fixation can reduce the effect of internal fixation on adjacent segment degeneration. Rigid internal fixation can also reportedly accelerate adjacent segment degeneration. Proper control of internal fixation strength can reduce the stress shielding effect. In theory, this can provide an ideal biomechanical environment for the bone graft area, and is beneficial for intervertebral bone fusion [29, 30].

Results show that under different movement conditions, L4 vertebral body displacement in the unilateral fixation model is greater than in the bilateral fixation model (i.e., the stability of the unilateral fixation model is less than that of the bilateral fixation model), but the difference is not statistically significant ($P>0.05$). This suggests that unilateral fixation can provide initial stability, as with bilateral fixation. This is consistent with the in vitro biomechanical test results of Chen et al. [23], showing that the strength of a single intervertebral fusion cage in unilateral fixation is less than with bilateral fixation, but clearly better than in normal lumbar vertebrae without internal fixation. A single cage in unilateral fixation can provide sufficient strength for intervertebral fusion.

Our research is based on a three-dimensional finite element model without any consideration for the impact of the muscles and soft tissue; therefore, there must be differences between the model and the human body. Since this model can be modified and many human specimens (including soft tissue such as ligaments and muscles) can be analyzed further, the model can closely simulate the actual human body.

Conclusions

Unilateral pedicle screw fixation with intervertebral body fusion can obtain initial stability, as with bilateral fixation. In addition, it can reduce the stress shielding of the internal fixator and the effect on adjacent segment degeneration. For degenerative lumbar instability with a complete anatomical structure on the nonoperative side, unilateral pedicle screw fixation with implantation of a single cage can provide effective spinal segmental stability.

Disclosure of conflict of interest

None.

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