

Oblique lateral interbody fusion combined with different internal fixation strategies for the treatment of lumbar spinal stenosis: a finite element analysis of postoperative spinal stability and stress distribution

Junqin Ye

Chinese PLA General Hospital

Jingbo Ma

Shanghai East Hospital

Qiang Jiang

PLA Navy General Hospital

Hanshuo Zhang

PLA Navy General Hospital

Rigbat Rozi

PLA Navy General Hospital

Jiaheng Han

PLA Navy General Hospital

Hongpeng Cui

PLA Navy General Hospital

Yu Ding

`dingyu@301hospital.com.cn`

PLA Navy General Hospital

Article

Keywords: Oblique lateral interbody fusion, lumbar spinal stenosis, interspinous process fixation, pedicle screw-rod fixation, finite element analysis, spinal biomechanics

Posted Date: August 26th, 2026

DOI: <https://doi.org/10.21203/rs.3.rs-10643445/v1>

License: © ⓘ This work is licensed under a Creative Commons Attribution 4.0 International License.
[Read Full License](#)

Additional Declarations: No competing interests reported.

**Oblique lateral interbody fusion combined with different internal
fixation strategies for the treatment of lumbar spinal stenosis: a finite
element analysis of postoperative spinal stability and stress
distribution**

Junqin Ye^{1,2†}, Jingbo Ma^{3,4†}, Qiang Jiang², Hanshuo Zhang², Rigbat Rozi², Jiaheng Han²,
Hongpeng Cui^{2,*} and Yu Ding^{2,**}

¹ Chinese PLA Medical School, Chinese PLA General Hospital, No. 28 Fuxing Road,
Haidian District, Beijing 100853, China

² Senior Department of Orthopedics of Traditional Chinese Medicine, The Sixth Medical
Center, Chinese PLA General Hospital, Beijing, China

³ Department of Orthopaedics, Shanghai East Hospital, School of Medicine, Tongji
University, Shanghai, China

⁴ Tongji University School of Medicine, Shanghai, China

† These authors contributed equally to this work and share first authorship.

* Corresponding author.

** Corresponding author at: Orthopedics of TCM Senior Department, The Sixth Medical
Center of PLA General Hospital, Beijing 100048, China.

E-mail addresses: cuihong-peng@163.com (Hongpeng Cui), dingyu@301hospital.com.cn (Yu
Ding).

Oblique lateral interbody fusion combined with different internal fixation strategies for the treatment of lumbar spinal stenosis: a finite element analysis of postoperative spinal stability and stress distribution

Abstract

Objective: To compare, using three-dimensional finite element analysis, the segmental stability, cage–endplate interface stress, and biomechanical responses of adjacent segments after oblique lateral interbody fusion (OLIF) combined with different supplemental fixation strategies for lumbar spinal stenosis, thereby providing biomechanical evidence to guide individualized selection of postoperative fixation strategies following OLIF.

Methods: A three-dimensional finite element model of the L1–S1 lumbar spine was constructed from computed tomography data obtained from a healthy adult and validated against previously published in vitro experimental data. The L4–L5 segment was designated as the target surgical level, and four models were developed: an intact model (M0), a stand-alone OLIF model (M1), an OLIF model with bilateral pedicle screw–rod fixation (M2), and an OLIF model with interspinous process fixation (ISPF; M3). With the inferior surface of S1 fully constrained, a 500-N axial load and a 10-N·m moment were applied to the superior surface of L1 to simulate six physiological motions: flexion, extension, left and right lateral bending, and left and right axial rotation. The range of motion (ROM) of the surgical and adjacent segments, maximum von Mises stress and displacement of the interbody cage, stresses on the inferior endplate of L4 and superior endplate of L5, and stress distribution within the internal fixation devices were compared across the models.

Results: Compared with the intact model, all three postoperative OLIF models reduced L4–L5 segmental ROM to varying degrees; however, their stabilizing effects differed markedly according to fixation configuration. M1 reduced ROM in most directions but showed relatively large residual motion during extension and under some lateral

bending conditions, accompanied by increased cage stress and L5 superior endplate stress. M2 provided the greatest multidirectional stability across all six loading conditions, reducing L4–L5 ROM by 95.97%–98.37%, and produced the largest reductions in cage stress, cage displacement, and L5 superior endplate stress. M3 also substantially improved segmental stability, reducing L4–L5 ROM by 80.86%–98.92%; its stabilizing effect was comparable to that of M2 in flexion and extension, although greater residual motion remained during lateral bending and axial rotation. Regarding cage stress, M1 showed the highest overall values, with the maximum occurring during left axial rotation. Cage stress ranged from 1.14 to 3.25 MPa in M2 and from 1.54 to 3.64 MPa in M3; both ranges were lower than those observed in M1, with M2 generally showing lower values than M3. L5 superior endplate stress was more sensitive to fixation strategy: it was highest in M1, most markedly reduced in M2, and also significantly decreased in M3 compared with M1. Regarding implant stress, the interspinous process fixation (ISPF) device showed pronounced stress concentration during lateral bending and axial rotation, reaching a maximum of 20.93 MPa, which indicated a direction-dependent biomechanical response. Adjacent-segment ROM showed limited overall variation, primarily reflecting changes in the motion distribution pattern of the lower lumbar spine.

Conclusion: Supplemental fixation strategies after OLIF produced distinct stiffness-dependent biomechanical effects. BPS provided the strongest multidirectional stability and the greatest reductions in cage and endplate stresses, but its high construct rigidity may raise concerns regarding stress shielding and adjacent-segment biomechanical compensation. In contrast, ISPF provided semi-rigid stabilization, reduced cage–endplate interface stress compared with stand-alone OLIF, and preserved a certain degree of controlled segmental micromotion. These characteristics suggest that ISPF may represent a biomechanically balanced option for selected patients. However, its weaker control of lateral bending and axial rotation and its relatively higher device stress indicate that its application should be individualized according to bone quality, spinous process morphology, and segmental stability.

Keywords: Oblique lateral interbody fusion; lumbar spinal stenosis; interspinous process fixation; pedicle screw–rod fixation; finite element analysis; spinal biomechanics

Introduction

Lumbar spinal stenosis (LSS) is a major degenerative lumbar disorder that causes chronic low back and leg pain, neurogenic intermittent claudication, and functional impairment, particularly in middle-aged and elderly patients. With population aging, the demand for surgical treatment of LSS continues to increase[1,2]. Conventional posterior fusion techniques, including posterior lumbar interbody fusion (PLIF) and transforaminal lumbar interbody fusion (TLIF), can achieve direct decompression and rigid fixation; however, they are often associated with paraspinal muscle detachment, disruption of posterior osseous structures, neural traction, and increased blood loss[3]. Therefore, achieving adequate decompression and segmental stability while minimizing surgical trauma and preserving physiological lumbar biomechanical transmission remains a key challenge in the surgical management of LSS[4].

Oblique lateral interbody fusion (OLIF) is a minimally invasive anterolateral fusion technique that accesses the intervertebral space through the retroperitoneal corridor along the anterior border of the psoas major. This approach avoids direct traversal of the psoas muscle and lumbar plexus region and permits disc preparation and cage implantation with minimal disruption of the posterior musculature, facet joints, laminae, and ligamentous complex. Because OLIF allows insertion of a sufficiently large interbody cage, it can effectively restore intervertebral and foraminal height. By restoring disc height and increasing tension in the posterior ligamentous complex, OLIF indirectly enlarges the spinal canal and neural foramina, thereby achieving indirect decompression[5]. Therefore, OLIF has increasingly become an important treatment option for degenerative lumbar diseases, mild-to-moderate degenerative spondylolisthesis, and selected cases of adult degenerative scoliosis.

However, postoperative outcomes after OLIF are not determined solely by the minimally invasive nature of the approach; the initial stability of the fused segment and

the biomechanical integrity of the cage–endplate interface are also critical. Stand-alone OLIF relies on anterior column cage support and the residual posterior elements to maintain stability. Although this approach maximizes the benefits of minimal invasiveness, reduced bone quality, insufficient endplate strength, excessive disc space distraction, suboptimal cage positioning, or pre-existing segmental instability may increase the risk of cage subsidence, endplate collapse, loss of disc height, reduced efficacy of indirect decompression, and even fusion failure[6]. In contrast, supplemental posterior fixation can substantially enhance segmental stability, reduce micromotion at the cage–endplate interface, and provide a more favorable mechanical environment for early osseous fusion. Therefore, whether additional fixation is required after OLIF and which fixation strategy should be selected remain key controversies in current clinical practice and biomechanical research[7].

Bilateral pedicle screw–rod fixation is widely regarded as one of the most reliable supplemental fixation strategies after OLIF. By establishing a combined anterior–posterior support system between the posterior bilateral pedicle screw–rod construct and the anterior interbody cage, this technique can substantially restrict segmental motion during flexion–extension, lateral bending, and axial rotation, while also sharing loads borne by the cage and endplates[8]. However, highly rigid fixation inevitably increases surgical trauma, fluoroscopic exposure, the risk of pedicle screw–related complications, and healthcare costs. Moreover, excessive restriction of mobility at the fused level may alter lumbosacral load-transmission patterns, shift motion and stress to adjacent segments, and thereby increase the risk of adjacent segment degeneration[9]. Therefore, supplemental fixation strategies in OLIF should not aim solely to maximize rigidity but should instead balance sufficient stability with the avoidance of excessive stiffness.

Compared with pedicle screw–rod fixation, interspinous process fixation (ISPF) does not require pedicle screw insertion. As a minimally invasive posterior supplemental fixation technique, ISPF can reduce disruption of the posterior musculature and osseous structures and offers potential advantages, including technical simplicity, lower invasiveness, and partial preservation of segmental micromotion[10].

ISPF primarily functions by reinforcing the posterior tension band, limiting abnormal interspinous motion, and improving stability during flexion–extension. Theoretically, it may serve as a semi-rigid supplemental fixation strategy between stand-alone OLIF and bilateral pedicle screw–rod fixation. However, because its biomechanical effect is mainly concentrated in the posterior column, its ability to control lateral bending and axial rotational loads may be limited. In addition, its fixation efficacy is influenced by spinous process morphology, bone quality, and the implant–bone interface contact condition[11]. Therefore, whether ISPF can simultaneously ensure segmental stability, optimize cage–endplate stress distribution, and provide biomechanical protection for adjacent segments after OLIF remains insufficiently supported by quantitative evidence.

Finite element analysis enables quantitative evaluation of different internal fixation configurations under standardized geometric, material, boundary, and loading conditions. It allows assessment of lumbar range of motion, cage stress, endplate stress, implant system stress, and the biomechanical environment of adjacent segments, making it an important method for investigating biomechanical mechanisms after spinal fusion surgery[12,13]. Previous studies have investigated OLIF combined with different pedicle screw constructs, lateral plates, and other supplemental fixation methods[13,14]. However, systematic comparisons among three strategies—stand-alone OLIF, OLIF with bilateral pedicle screw–rod fixation, and OLIF with interspinous process fixation—remain limited and warrant further investigation. Therefore, this study established a three-dimensional finite element model of the L1–S1 lumbar spine, with L4–L5 designated as the target surgical level. Four models were constructed: an intact model, a stand-alone OLIF model, an OLIF model with bilateral pedicle screw–rod fixation, and an OLIF model with interspinous process fixation. Six physiological loading conditions were simulated: flexion, extension, left and right lateral bending, and left and right axial rotation. The models were compared in terms of segmental and adjacent-level range of motion, cage stress, superior and inferior endplate stress, and implant system stress. This study systematically evaluated how different fixation strategies affect segmental stability, cage–endplate stress distribution, and the biomechanical environment of adjacent segments, thereby clarifying their roles

in stability reconstruction and load transfer after OLIF and providing a theoretical basis for individualized selection of postoperative fixation strategies.

1. Materials and Methods

1.1 Study Design

This study was designed as a three-dimensional finite element biomechanical analysis. A complete lumbar spine finite element model was constructed using computed tomography data of the L1–S1 segment from a healthy adult. After model validation, OLIF and different fixation configurations were simulated at the L4–L5 target segment. Under identical boundary and loading conditions, the range of motion and stress distribution of each model were compared across different physiological motion states.

1.2 Subject and Imaging Data Acquisition

A 33-year-old healthy male volunteer without any spinal disorders was recruited for this study. Lumbar spine imaging was performed using a 256-slice multidetector computed tomography (CT) scanner (Siemens, Germany). The slice thickness was set to 0.63 mm, yielding 501 image slices stored in Digital Imaging and Communications in Medicine (DICOM) format. Informed consent was obtained from the participant, including disclosure of radiation exposure-related information. Relevant demographic information, including age and sex, was recorded; name identifiers were not used to ensure confidentiality. This study was approved by the Ethics Committee of the Sixth Medical Center of the General Hospital of the People's Liberation Army (approval number: HZKY-PJ-2025-1).

1.3 Development of a Complete L1–S1 Finite Element Model

DICOM-format CT data were imported into Mimics software. The bony structures of L1, L2, L3, L4, L5, and S1 were segmented based on grayscale thresholds using thresholding, region-growing, hole-filling, and surface-smoothing techniques. The L1–S1 mask was further refined with the mask-editing tool to generate a preliminary three-dimensional bony model, which was exported in STL format.

The STL file of the L1–S1 model was then imported into Geomagic software. Surface artifacts and irregular structures were removed, and surface repair, smoothing,

and fitting were performed to generate a geometrically continuous and complete solid model. A uniform cortical bone shell with a thickness of 1.0 mm was defined, and the remaining internal region was assigned as cancellous bone. To simulate the cortical shell, the outer surface of each vertebral body was offset inward by 1.0 mm to generate the cancellous bone core, thereby ensuring a constant cortical thickness of 1.0 mm, consistent with standard lumbar finite element modeling practices[15]. After optimization of all bony structures, the model was exported in STEP format with all required geometric data.

The STEP model was imported into SolidWorks software and assembled using a fixed origin. Anatomical structures, including the endplates, nucleus pulposus, annulus fibrosus, and facet joint cartilage, were created using sketching, boss-extrusion, and cut operations. The intervertebral disc was modeled as the nucleus pulposus and annulus fibrosus, with the nucleus pulposus positioned slightly posterior to the disc center. The nucleus pulposus accounted for approximately 48.5% of the disc volume, consistent with the reported value of approximately 50% in the literature[16]. After interference checking, a standardized lumbar spine model was obtained.

The STEP file of the standardized lumbar spine model was imported into ANSYS software for finite element analysis (Figure 1). Material properties, including elastic modulus and Poisson's ratio, were assigned to each anatomical structure according to the parameters listed in Table 1[17]. Ligaments were simulated in the connection module using spring elements, with different stiffness values assigned to represent different ligamentous structures. An appropriate contact tolerance was specified to eliminate abnormal contact interactions; this value was generally set to 0.1 mm[18]. Contact definitions were assigned as follows: frictional contact with a coefficient of 0.1 between facet joint cartilages; frictional contact with a coefficient of 0.1 between the interbody cage and one endplate, and bonded contact with the contralateral endplate[19]; bonded contact between screws and bone and between rods and screws; and, for the interspinous process fixation device, frictional contact with a coefficient of 0.2 at the bone-implant interface according to device characteristics, while all other contacts were defined as bonded.

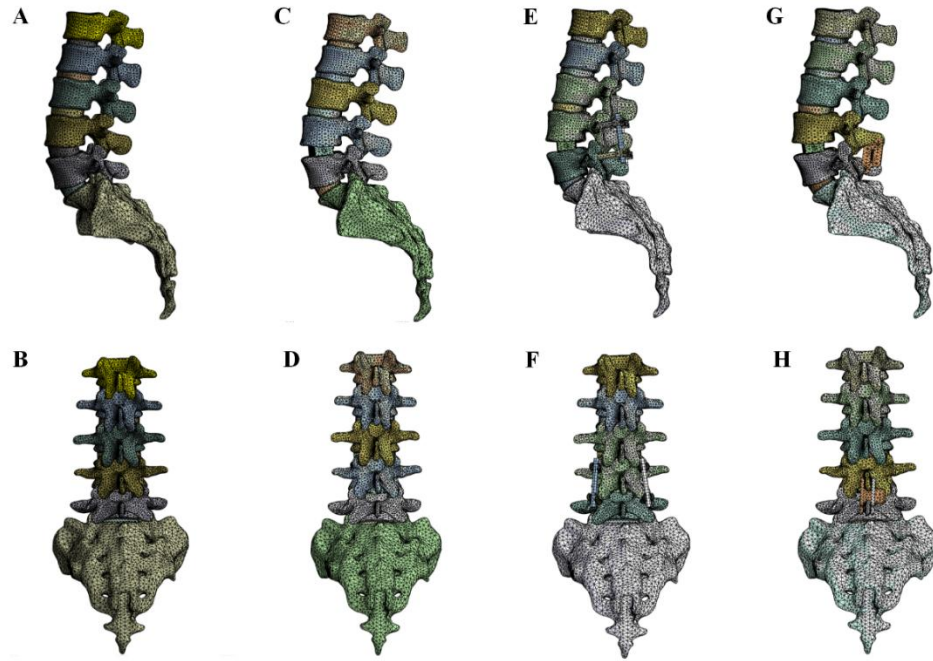


Fig. 1. Three-dimensional morphological comparison of reconstructed models: (A, B) illustrate the intact model in lateral and posterior views. (C, D) illustrate the SA model from lateral and posterior perspectives. (E, F) illustrate the BPSF model in lateral and posterior views. (G, H) illustrate the ISPF model in lateral and posterior views

Table 1.

Material parameters

Structure	Elastic Modulus (MPa)	Poisson's Ratio	Stiffness (N/mm)
Cortical bone	12000.00	0.30	/
Cancellous bone	100.00	0.20	/
Annulus fibrosus	4.20	0.45	/
Nucleus pulposus	1.00	0.49	/
Endplate	2000.00	0.20	/
Facet joint cartilage	25.00	0.25	/

Anterior longitudinal ligament	7.80	/	8.74
Posterior longitudinal ligament	10.00	/	5.83
Supraspinous ligament	8.00	/	15.38
Interspinous ligament	10.00	/	0.19
Ligamentum flavum	15.00	/	15.75
Intertransverse ligament	10.00	/	2.39
Cage	3600.00	0.30	/
Titanium alloy	110000.00	0.30	/

1.4 Meshing and Load Application

After the material properties and contact conditions were defined, a refined mesh was generated using tetrahedral elements. The mesh size was set to 1.0 mm for articular cartilage and 2.0 mm for the remaining structures[11], resulting in 311,474 elements and 177,555 nodes. The meshing details for each anatomical structure are shown in Figure 2.

A static structural analysis was then performed. The inferior surface of S1 was fully constrained as a fixed support to restrict model motion, whereas the superior surface of L1 was defined as the loading interface for forces, moments, and coordinate systems. A 500-N compressive load was applied along the negative Z-axis, and a 10-N·m moment was applied in the corresponding directions to simulate physiological lumbar loading during daily motion. To facilitate calculation of segmental range of motion, remote points were introduced into the model for solution analysis, and command scripts were used in the solution directory to obtain accurate displacement and motion measurements.

1.5 Mesh Convergence Verification and Selection

To ensure that the calculated stress results were independent of element size, a

systematic mesh-refinement convergence analysis was performed on the intact L4–L5 segment model under a 10-N·m flexion loading condition. Seven mesh configurations with different element densities were constructed, ranging from approximately 80,000 to 380,000 elements. For each mesh configuration, the peak von Mises stress within the intervertebral disc was recorded, and the relative differences between adjacent mesh densities were calculated to determine whether further mesh refinement would substantially alter key biomechanical outcomes. To quantitatively evaluate differences between adjacent mesh configurations, the relative change rate ($\Delta\%$) was calculated using the following formula:

$$\Delta\% = \frac{|\sigma_n - \sigma_{n-1}|}{\sigma_{n-1}} \times 100\%$$

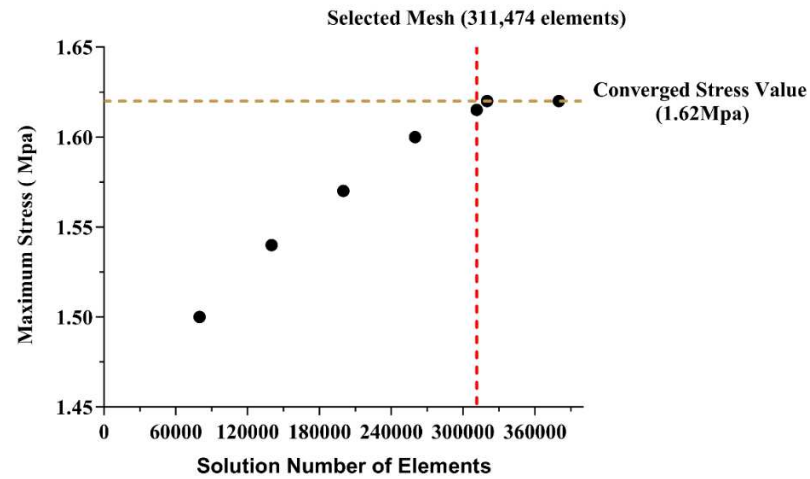


Fig. 2. Mesh convergence analysis for the L4-L5 segment under flexion

This study adopted a commonly used mesh convergence criterion in finite element analysis, in which a variation of less than 5% in key stress parameters between two successive mesh refinements was considered to indicate mesh-independent results and convergence.

As shown in Figure 2, the peak von Mises stress in the intervertebral disc gradually stabilized with progressive mesh refinement. When the number of elements reached 311,474, the stress results entered a plateau phase, with the peak value stabilizing at approximately 1.615 MPa. Compared with the mesh containing approximately 260,000 elements, the stress variation was 0.94%, which was below the predefined 5% threshold

and indicated that convergence had been achieved. Therefore, the mesh containing 311,474 elements and 177,555 nodes was selected as the final discretization for all subsequent simulations to balance computational accuracy and efficiency.

1.6 OLIF Surgical Model Construction

A total of four finite element models were constructed, comprising one intact control model and three postoperative OLIF fixation configurations: the intact control model (M0), stand-alone OLIF model (M1), OLIF with bilateral pedicle screw–rod fixation (M2), and OLIF with interspinous process fixation (M3). The geometric parameters of all implants in the postoperative models were defined according to commonly used clinical specifications and actual device dimensions (Figure 3). The OLIF interbody cage measured 40 mm in length, 12 mm in width, and 10 mm in height, with a 6° lordotic angle. The pedicle screws measured 6.5 mm in diameter and 50 mm in length, and the connecting rods had a diameter of 5.5 mm. The interspinous process fixation device was selected with a 14-mm distraction size according to the L4–L5 interspinous distance to achieve appropriate interspinous distraction[20]. All implants were assembled according to clinically relevant insertion trajectories and anatomical positions.

M0: The intact L1–S1 lumbar spine finite element model was used as the normal control model without surgical intervention, representing biomechanical behavior under physiological conditions.

M1: A stand-alone OLIF model was constructed. The annulus fibrosus, nucleus pulposus, and corresponding central cartilaginous endplates at the L4–L5 segment were removed to simulate disc preparation. Subsequently, a single OLIF interbody cage was inserted from left to right into the L4–L5 disc space, achieving contact fixation with the inferior endplate of L4 and the superior endplate of L5, without additional posterior or lateral instrumentation.

M2: An OLIF model with bilateral pedicle screw–rod fixation was constructed. Cage insertion was identical to that in M1. In addition, pedicle screws were inserted bilaterally into L4 and L5 and connected with bilateral rods to simulate OLIF combined with posterior bilateral pedicle screw–rod fixation at the L4–L5 segment.

M3: An OLIF model with interspinous process fixation was constructed. Cage placement was identical to that in M1. The L4–L5 interspinous distance was measured to determine the implant size, and the device was inserted laterally through a sleeve to achieve posterior supplemental fixation between the L4 and L5 spinous processes.

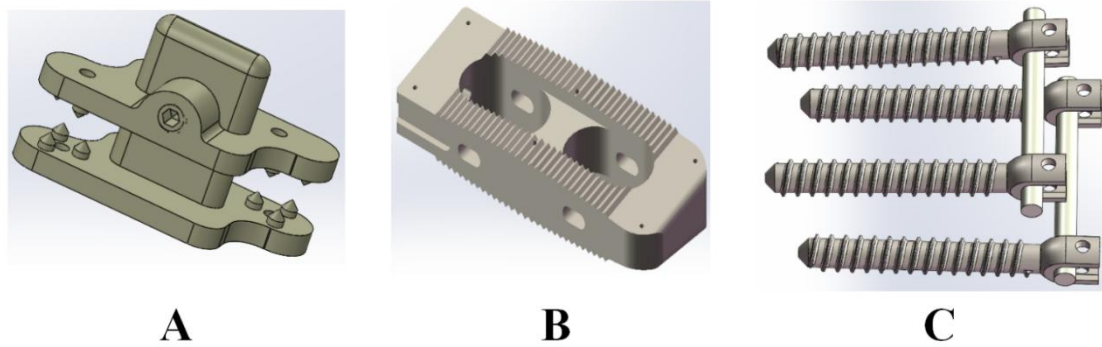


Fig. 3. Schematic representations of different surgical implants used in this study: (A) Interspinous process fixation device; (B) oblique lateral interbody fusion (OLIF) cage; (C) bilateral pedicle screw fixation system

1.7 Outcome Measures

The range of motion (ROM) of the L3–L4, L4–L5, and L5–S1 segments was recorded under six loading conditions, with particular emphasis on segmental stability at the L4–L5 surgical level and compensatory motion changes in the adjacent L3–L4 and L5–S1 segments. The maximum von Mises stress and deformation of the L4–L5 interbody cage were extracted. The maximum von Mises stresses of the inferior endplate of L4 and the superior endplate of L5 were also obtained. In addition, the maximum von Mises stresses of the pedicle screw–rod system and the interspinous process fixation device were recorded.

2. Results

2.1 Validation of the Intact Lumbar Spine Model

The range of motion (ROM) of the intact L4–L5 three-dimensional finite element model developed in this study was compared with in vitro experimental data reported by Yamamoto Shim, and Chen[21-23]. The results showed that the ROM values were comparable to those reported in previous studies and generally fell within the standard

deviation ranges of the published experimental data, indicating that the established intact L4–L5 lumbar finite element model was valid, reliable, and suitable for subsequent biomechanical analysis (Figure 4).

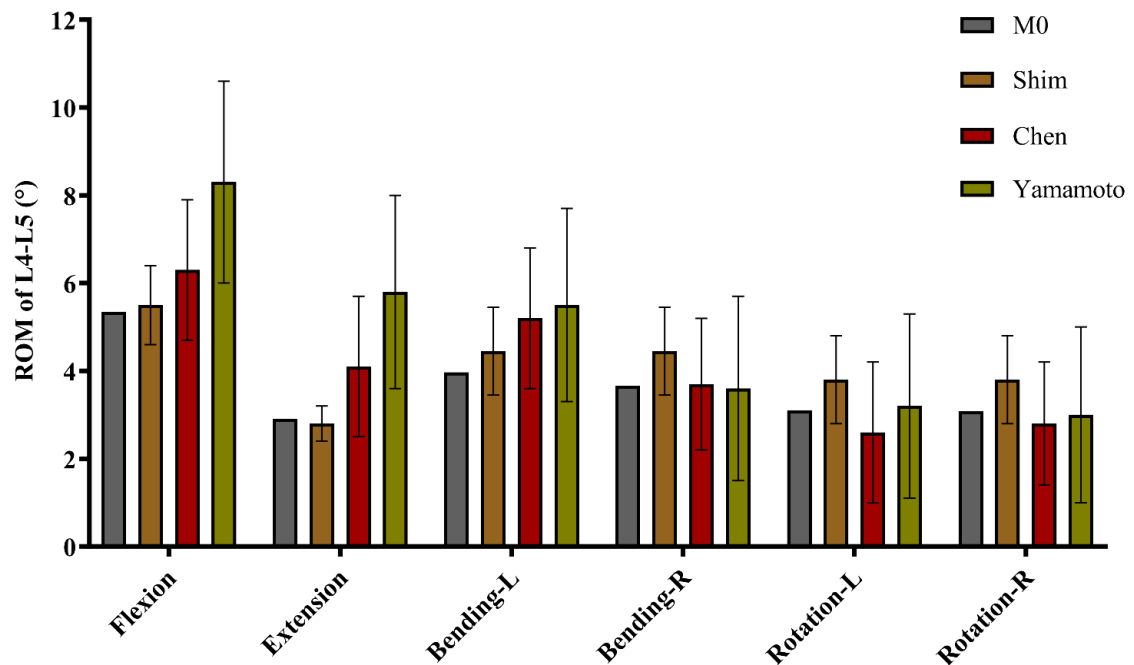


Fig. 4. Verification of the FE Model Validity and Comparison of L4-L5 Segment ROMs with Published Experimental Data

2.2 Changes in Segmental Range of Motion

Changes in the range of motion (ROM) of each model under the six loading conditions are shown in Table 2. Compared with the intact model (M0), all three surgical models altered lumbar segmental motion to varying degrees, with the most pronounced changes occurring at the L4–L5 surgical segment (Figure 5).

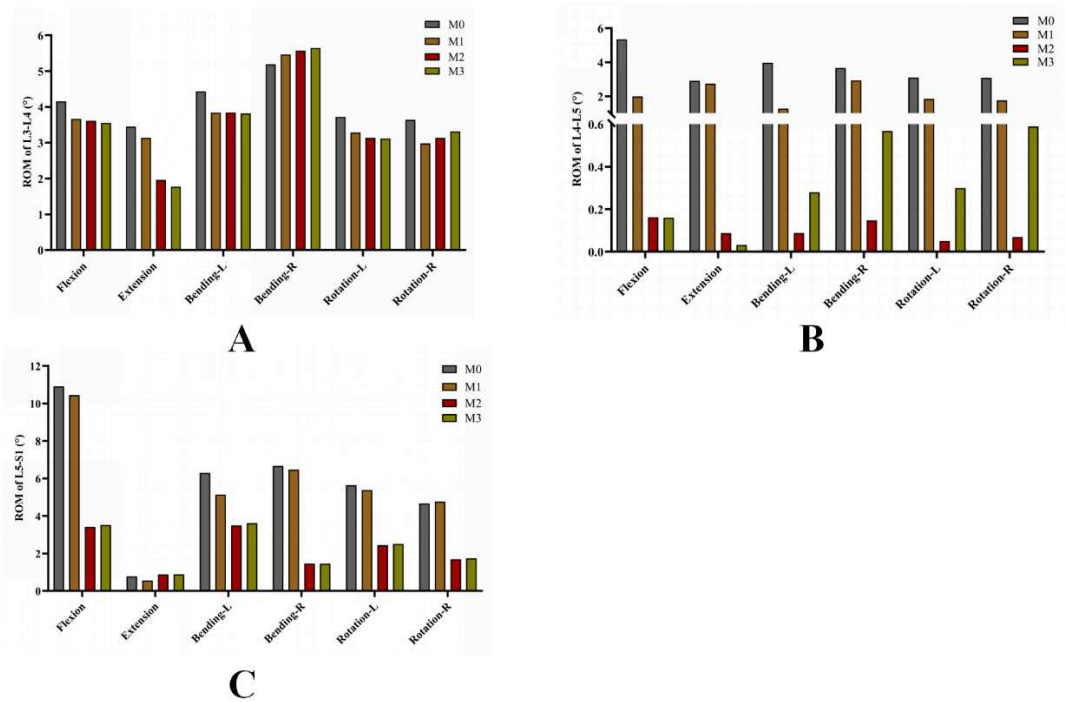


Fig. 5. Comparison of motion range of L3-S1 segments under different loading conditions among model groups: (A) ROM at L3–L4; (B) ROM at L4–L5; (C) ROM at L5–S1. M0 represents the intact lumbar spine (L1–S1) finite element model; M1 represents the stand-alone OLIF model (SA); M2 represents OLIF combined with bilateral pedicle screw fixation (BPSF); and M3 represents OLIF combined with interspinous process fixation (ISPF)

Table 2.

Comparison of motion range of L3-S1 segments under different loading conditions among model groups(°)

Model	ROM of L3-L4 segment (°)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M0	4.15	3.45	4.43	5.58	3.72	3.64
M1	3.66 (-11.80%)	3.14 (-9.03%)	3.84 (-13.35%)	5.46 (-2.04%)	3.28 (-11.76%)	2.98 (-18.08%)
M2	3.61 (-13.04%)	1.96 (-43.10%)	3.84 (-13.33%)	5.57 (-0.12%)	3.13 (-15.74%)	3.14 (-13.82%)
M3	3.55 (-14.49%)	1.77 (-48.68%)	3.82 (-13.90%)	5.65 (+ 1.22%)	3.17 (-16.21%)	3.32 (-8.82%)

Model	ROM of L4-L5 segment (°)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M0	5.34	2.91	3.97	3.66	3.09	3.08
M1	1.98 (-62.90%)	2.74 (-6.06%)	1.27 (-67.92%)	2.93 (-19.93%)	1.84 (-40.56%)	1.75 (-43.07%)
M2	0.16 (-96.97%)	0.09 (-96.99%)	0.09 (-97.78%)	0.15 (-95.97%)	0.05 (-98.37%)	0.07 (-97.76%)
M3	0.16 (-97.00%)	0.03 (-98.92%)	0.28 (-92.94%)	0.57 (-84.46%)	0.30 (-90.32%)	0.59 (-80.86%)

Model	ROM of L5-S1 segment (°)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M0	10.91	0.82	6.3	6.67	5.64	4.67
M1	10.44 (-4.26%)	0.55 (-33.04%)	5.14 (-18.46%)	6.47 (-3.00%)	5.38 (-4.61%)	4.76 (+2.00%)
M2	3.41 (-68.72%)	0.88 (+6.80%)	3.49 (-44.59%)	1.46 (-78.12%)	2.44 (-56.77%)	1.69 (-63.78%)
M3	3.52 (-67.75%)	0.88 (+7.31%)	3.62 (-42.62%)	1.45 (-78.21%)	2.51 (-55.52%)	1.74 (-62.76%)

At the L4–L5 level, the stand-alone OLIF model (M1) reduced ROM in most directions; however, its stabilizing effect showed marked directional dependence. In M1, ROM decreased by 62.90% in flexion and 67.92% in left lateral bending, whereas smaller reductions were observed in extension and right lateral bending, at 6.06% and 19.93%, respectively. The OLIF model with bilateral pedicle screw–rod fixation (M2) demonstrated the greatest stability in all directions, with L4–L5 ROM reductions ranging from 95.97% to 98.37%. The OLIF model with interspinous process fixation (M3) also substantially reduced segmental ROM, with reductions ranging from 80.86% to 98.92%. Compared with M2, M3 provided comparable stability in flexion and extension but retained greater residual motion during lateral bending and axial rotation (Figure 6).

At the adjacent L3–L4 segment, overall ROM changes were minimal, with only a

slight increase observed during right lateral bending. At the L5–S1 segment, ROM decreased in most directions in both the M2 and M3 models, whereas a slight increase was observed during extension, suggesting that supplemental fixation altered the motion distribution pattern of the lower lumbar spine.

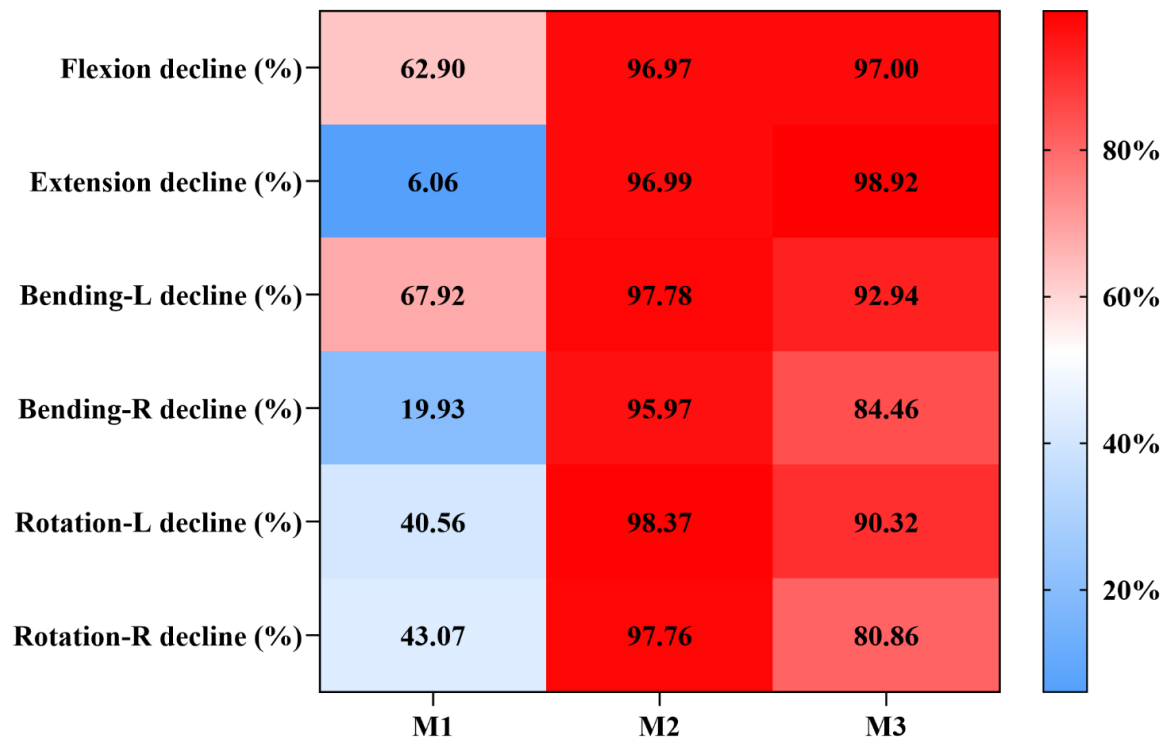


Fig. 6. Percentage reduction in range of motion (ROM) at the L4–L5 segment under different loading conditions.

2.3 Cage Stress and Displacement

The maximum von Mises stress and displacement of the interbody cage are presented in Table 3 and Figure 7. The M1 model showed the highest overall cage stress, with a maximum value of 6.51 MPa occurring during left axial rotation. Compared with M1, both M2 and M3 reduced cage stress under all loading conditions. M2 produced the greatest reduction, with cage stress ranging from 1.14 to 3.25 MPa, whereas M3 showed a stress range of 1.54 to 3.64 MPa, with values generally higher than those in M2 but lower than those in M1.

Cage displacement showed a similar trend. In the M1 model, the maximum cage displacement occurred during flexion, reaching 10.01 mm. Both M2 and M3 markedly

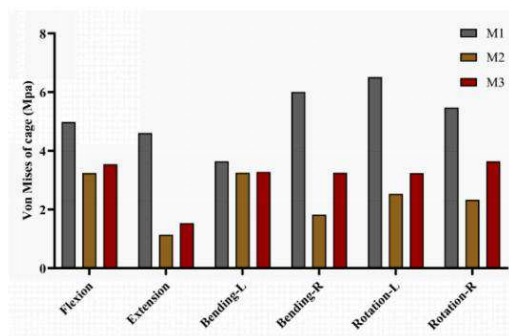
reduced cage displacement across all loading directions, with displacement values ranging from 1.41 to 3.93 mm in M2 and from 1.50 to 4.15 mm in M3. Overall, bilateral pedicle screw–rod fixation provided the strongest control of cage stress and displacement, followed by ISPF. The cage displacement contour maps for each model are shown in Figure 8.

Table 3.

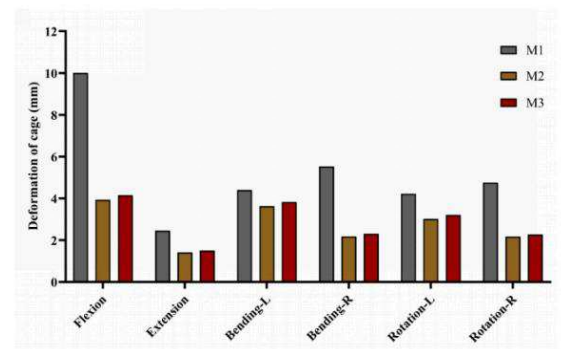
von Mises stress and deformation of the cage under six loading conditions in different models (M1-M3)

Model	von Mises of the cage (MPa)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M1	4.98	4.61	3.64	6.00	6.51	5.47
M2	3.24	1.14	3.25	1.82	2.54	2.33
M3	3.54	1.54	3.28	3.25	3.23	3.64

Model	Deformation of the cage (mm)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M1	10.01	2.45	4.40	5.53	4.22	4.76
M2	3.93	1.41	3.63	2.18	3.02	2.17
M3	4.15	1.50	3.82	2.31	3.21	2.27



A



B

Fig. 7. (A) von Mises stress of the cage, (B) Deformation of the cage under six different loading conditions in different models (M1-M3)

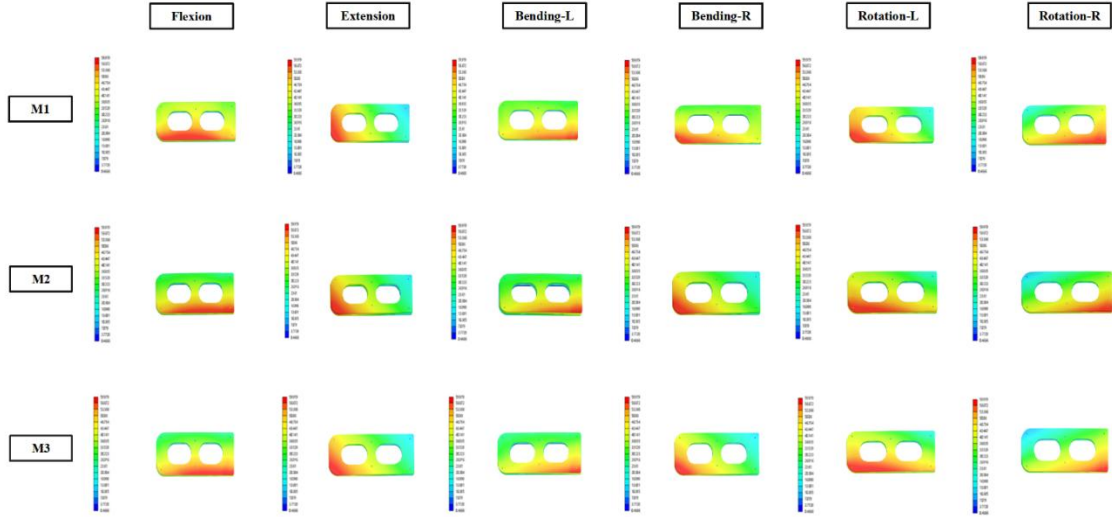


Fig. 8. Deformation distribution of the cage under six loading conditions in different models (M1-M3)

2.4 Endplate Stress

The stresses on the inferior endplate of L4 and the superior endplate of L5 are presented in Table 4 and Figure 9. The L4 inferior endplate stress showed relatively small differences among the three surgical models. Compared with M1, both M2 and M3 showed an overall decreasing trend, although the magnitude of reduction was limited.

In contrast, changes in L5 superior endplate stress were more pronounced. The M1 model showed the highest L5 superior endplate stress under all loading conditions, with a maximum value of 10.63 MPa occurring during left axial rotation. M2 markedly reduced L5 superior endplate stress, with values ranging from 1.78 to 4.86 MPa. M3 also reduced L5 superior endplate stress, with values ranging from 2.29 to 6.22 MPa; however, these values were generally higher than those in M2. These findings suggest that supplemental fixation can reduce stress concentration at the cage–endplate interface, with bilateral pedicle screw–rod fixation showing the most pronounced effect. Stress contour maps of the L4 inferior and L5 superior endplates are shown in Figures

10 and 11.

Table 4.

von Mises stress distribution on the inferior endplate of L4 and superior endplate of L5 under six loading conditions in different models (M1-M3)

Model	Stress on the Inferior Endplate of L4 (MPa)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M1	5.36	5.17	5.35	5.80	4.87	5.17
M2	5.39	2.72	4.25	4.19	4.03	4.21
M3	5.01	3.17	4.83	4.23	4.81	3.74

Model	Stress on the Superior Endplate of L5 (MPa)					
	Flexion	Extension	Bending-L	Bending-R	Rotation-L	Rotation-R
M1	10.17	7.89	6.29	9.34	10.63	5.75
M2	3.98	1.78	4.86	3.30	4.01	3.31
M3	6.22	2.29	5.04	4.88	5.30	5.10

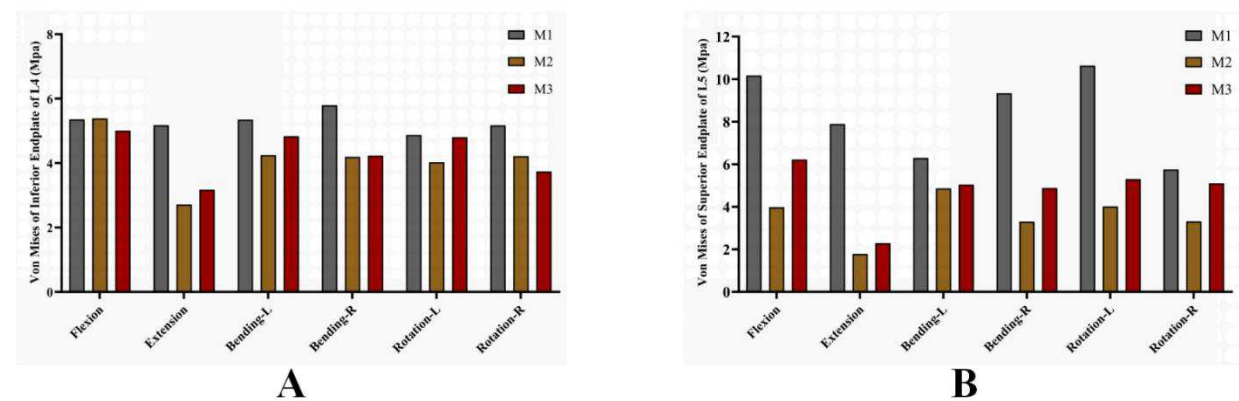


Fig. 9. Comparison of von Mises stress at the (A) L4 inferior endplate and (B) L5 superior endplate under six loading conditions in different models (M1-M3)

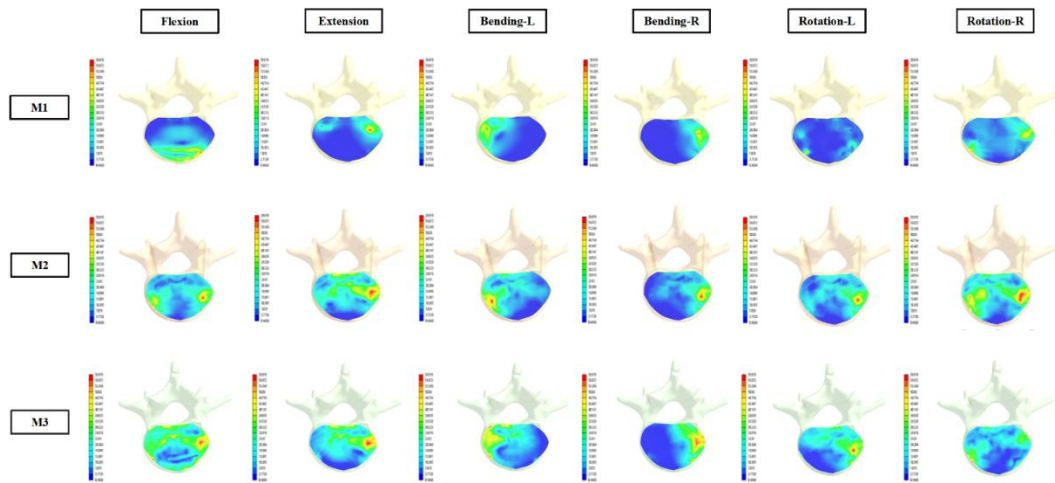


Fig. 10. Finite element analysis of von Mises stress distribution on the inferior endplate of L4 under six loading conditions in different models (M1-M3)

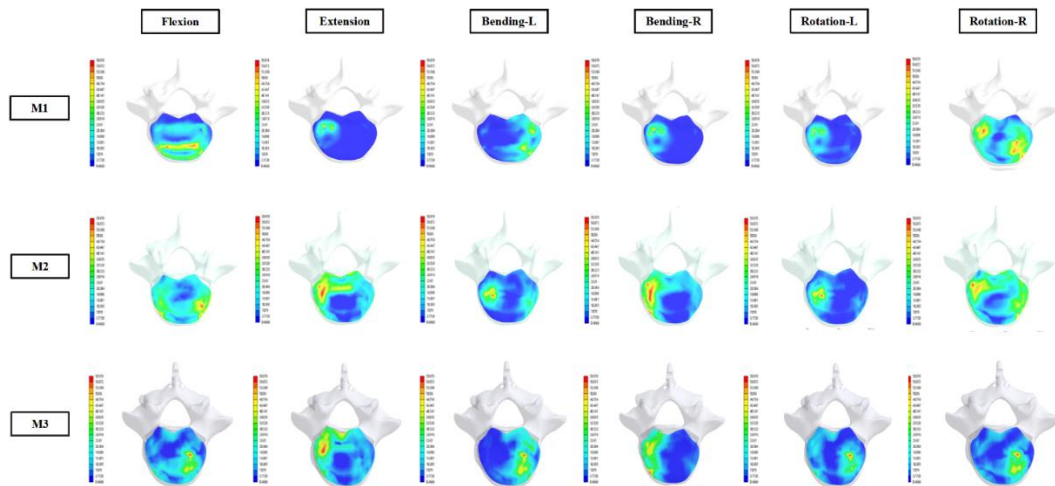


Fig. 11. Finite element analysis of von Mises stress distribution on the superior endplate of L5 under six loading conditions in different models (M1-M3)

2.5 Stress of Internal Fixation Devices

The maximum von Mises stress contour maps of the internal fixation devices are shown in Figure 12. In the bilateral pedicle screw–rod fixation (BPS) model, the stress of the internal fixation system ranged from 5.07 to 15.47 MPa, with the maximum value occurring during left axial rotation. In the interspinous process fixation (ISPF) model, device stress ranged from 5.50 to 20.93 MPa, with the maximum value observed during

right axial rotation.

Compared with BPS, ISPF showed higher device stresses during flexion, lateral bending, and right axial rotation. Stress values were particularly elevated during right lateral bending and right axial rotation, reaching 15.37 MPa and 20.93 MPa, respectively. These results indicate that the stress distribution of ISPF is direction-dependent, with a greater tendency toward stress concentration during lateral bending and axial rotation.

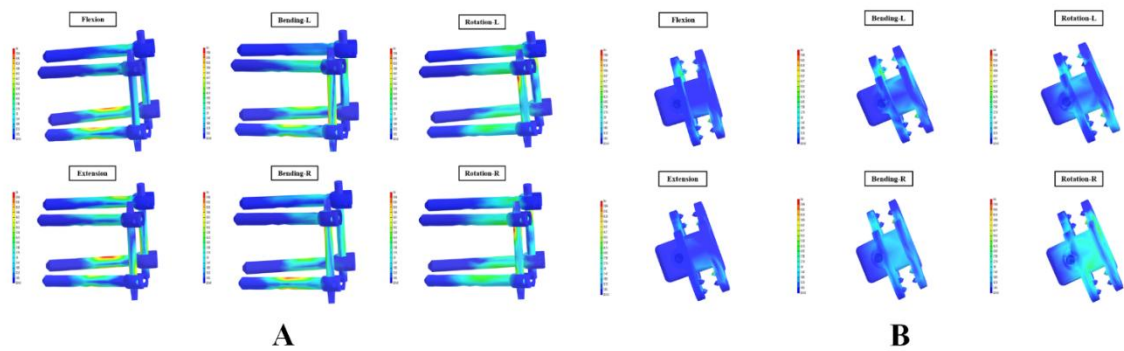


Fig. 12. Finite element analysis of von Mises stress distribution in internal fixation systems under six loading conditions: (A) bilateral pedicle screw-rod system (BPS); (B) interspinous process fixation (ISPF)

2.6 Overall Trends

The stand-alone OLIF model (M1) reduced ROM but increased cage and L5 superior endplate stress. The OLIF model with bilateral pedicle screw-rod fixation (M2) achieved the highest stability with the greatest stress reduction across all conditions. The OLIF model with interspinous process fixation (M3) provided intermediate stability with direction-dependent limitations and higher device stress. Collectively, the three constructs exhibited a graded biomechanical response from minimal stabilization to rigid fixation.

3 Discussion

Oblique lateral interbody fusion (OLIF) can reduce surgical trauma and improve postoperative outcomes after spinal fusion[24] . However, insufficient initial stability remains a concern, especially in patients with poor bone quality, because inadequate

fixation may increase cage micromotion, endplate stress, cage subsidence, and intervertebral displacement[25,26]. Therefore, supplemental fixation is frequently used to enhance postoperative stability after OLIF.

The principal finding of this study is not simply that OLIF combined with bilateral pedicle screw–rod fixation (BPS) provides the greatest stability, but that different fixation strategies form a continuous biomechanical spectrum from stand-alone OLIF to semi-rigid interspinous process fixation (ISPF) and rigid pedicle screw–rod fixation [19] . With increasing fixation stiffness, segmental ROM and cage–endplate interface stress generally decreased; however, implant load distribution, surgical invasiveness, and potential healthcare costs also changed[3] . Stand-alone OLIF reduced L4–L5 ROM but showed direction-dependent stability, with higher cage stress, greater cage displacement, and increased L5 superior endplate stress[27] . BPS provided the strongest multidirectional stability and the most pronounced reduction in cage and endplate stresses [9] . In contrast, ISPF showed intermediate biomechanical behavior: it provided stability comparable to BPS during flexion–extension but retained greater residual motion during lateral bending and axial rotation, with relatively higher device stress under certain loading conditions[11].

Successful fusion requires not only sufficient initial fixation but also an appropriate mechanical environment at the fusion interface. Excessive micromotion may impair osseointegration and increase the risk of cage subsidence, whereas complete elimination of mechanical loading may contribute to stress shielding. Therefore, the goal of supplemental fixation after OLIF should not be maximal rigidity alone, but a balance among segmental stability, cage–endplate load sharing, subsidence risk, and surgical invasiveness[8,9,28] .

The present findings regarding BPS are consistent with previous finite element studies. Fang et al. reported that BPS substantially improved segmental stability and reduced endplate stress after OLIF, thereby potentially decreasing the risk of cage subsidence [8] . Other studies have also shown that BPS generally provides the greatest reductions in ROM, cage stress, and endplate stress among supplemental fixation constructs [9,29] . In this study, M2 demonstrated the lowest L4–L5 ROM across all

six loading conditions and markedly reduced cage stress, cage displacement, and L5 superior endplate stress. These results suggest that BPS creates a highly stable early mechanical environment for fusion. However, this advantage does not mean that BPS should be applied indiscriminately. Rigid posterior fixation may increase surgical trauma, fluoroscopic exposure, screw-related risks, and medical cost, and may also alter load transmission at adjacent segments[8]. Thus, BPS may be more suitable for patients requiring strong mechanical stabilization, such as those with poor bone quality, marked instability, or high subsidence risk, although its optimal indications require further clinical validation [9] .

The biomechanical value of ISPF lies in its semi-rigid nature. Previous studies have suggested that interspinous process devices can effectively limit segmental motion, especially during extension, but have relatively limited ability to control lateral bending and axial rotation [6] . The present results support this view. M3 substantially reduced L4–L5 ROM and decreased cage and endplate stresses compared with stand-alone OLIF, indicating that ISPF provided additional posterior stability and partial load sharing. During flexion–extension, its stabilizing effect was close to that of BPS, which may be attributed to reinforcement of the posterior tension band and restriction of abnormal interspinous motion. Unlike BPS, ISPF does not require pedicle screw insertion and may therefore reduce screw-related surgical trauma while preserving a certain degree of controlled micromotion. This intermediate behavior suggests that ISPF may offer a favorable balance between stability and physiological load transmission in selected patients (PMID: 41618409 ; PMID: 38057738)[20,30] .

However, ISPF should not be regarded as a complete substitute for BPS. In this study, M3 provided less restriction than M2 during lateral bending and axial rotation, and ISPF device stress increased notably during right lateral bending and right axial rotation. This indicates possible load concentration at the spinous process–implant interface. Recent studies evaluating multiple OLIF fixation constructs have mainly focused on ROM, cage stress, and endplate stress, whereas stress concentration within semi-rigid fixation devices has received less attention[29]. The increased device stress observed in M3 suggests that the implant actively participates in postoperative load

sharing rather than acting as a passive stabilizer. Whether this represents effective load sharing or a potential fatigue-related risk requires further evaluation through spinous process stress analysis, cyclic fatigue testing, and clinical follow-up [31] . Therefore, ISPF may be more appropriate for patients with adequate bone quality, intact spinous process morphology, and mild-to-moderate instability, whereas caution is needed in patients with obvious rotational instability, lateral instability, osteoporosis, or weakened spinous processes[7,11,32].

Stand-alone OLIF preserves the minimally invasive advantages of OLIF by avoiding posterior muscle injury, screw-related complications, and additional posterior instrumentation[33] . Nevertheless, the present study showed that anterior column support alone did not provide sufficient stability in all directions. M1 retained relatively large motion during extension, right lateral bending, and axial rotation and produced higher cage and L5 superior endplate stresses. These findings suggest that the potential risk of stand-alone OLIF is not limited to increased ROM, but also involves greater compressive, shear, and torsional loading at the cage–endplate interface, which may contribute to endplate injury and cage subsidence[27,34] . Clinical studies and systematic reviews have also identified osteoporosis, endplate injury, excessive disc distraction, and advanced age as risk factors for cage subsidence after OLIF[6,35] . Accordingly, stand-alone OLIF may be more suitable for carefully selected patients with good bone quality, intact endplates, and no obvious multidirectional instability.

Notably, L5 superior endplate stress was more sensitive to fixation strategy than L4 inferior endplate stress. This asymmetric stress response may be related to the oblique cage trajectory, differences in superior and inferior endplate geometry, local contact area, and loading direction. Previous studies have shown that endplate morphology is associated with cage subsidence after OLIF, supporting the importance of local cage–endplate stress distribution [36] . In the present study, BPS produced the greatest reduction in L5 superior endplate stress, whereas ISPF also reduced this stress compared with stand-alone OLIF, suggesting that semi-rigid posterior fixation may contribute to cage–endplate load sharing.

Adjacent-segment responses after OLIF remain controversial. Some studies have reported that rigid fixation may increase adjacent-segment motion or intervertebral disc stress, whereas others have found limited and direction-dependent changes [7,11,32] . In this study, L3–L4 ROM changed only minimally, and L5–S1 ROM did not show a typical compensatory increase; instead, it slightly decreased in most directions. This result should be interpreted cautiously as a direction-dependent redistribution of motion rather than direct evidence of adjacent-segment protection or reduced adjacent segment degeneration risk[5,6] . Future studies should include intervertebral disc stress, facet joint loading, and more physiologically representative loading conditions to clarify adjacent-segment biomechanics after different OLIF fixation strategies[37].

This study has several limitations. First, the finite element model was based on a single lumbar spine geometry and therefore cannot fully represent variations in age, bone mineral density, degeneration grade, endplate morphology, and spinous process anatomy. Second, bone and soft tissues were simplified as homogeneous, isotropic, and linearly elastic materials, which may not fully reflect the nonlinear and heterogeneous behavior of biological tissues. Third, active muscle forces, fatigue failure, bone fusion, and long-term remodeling were not included. In addition, the cage–endplate interface was modeled using idealized contact conditions, without simulating postoperative osseointegration or interface evolution. Therefore, the present results mainly reflect early postoperative initial stability rather than long-term fusion outcomes. Future studies should incorporate different bone densities, cage positions, spinous process morphologies, and interface friction conditions, and should be validated by cadaveric experiments, fatigue testing, and prospective clinical follow-up.

4 Conclusion

This study demonstrated that supplemental fixation strategies after OLIF produce distinct biomechanical effects related to fixation rigidity. Stand-alone OLIF reduced L4–L5 segmental ROM, but its stabilizing effect was direction-dependent and was associated with increased cage–endplate interface loading. OLIF with BPS provided the strongest multidirectional stability and produced the greatest reductions in cage and

endplate stresses; however, its high construct rigidity may alter intersegmental load transfer, reduce mechanical stimulation at the fusion region, and increase concerns regarding stress shielding and adjacent-segment biomechanical compensation.

In contrast, OLIF with ISPF substantially improved segmental stability and reduced cage–endplate interface stress compared with stand-alone OLIF. As a semi-rigid fixation strategy, ISPF preserved a certain degree of controlled segmental micromotion while providing meaningful supplemental stability, which may help reduce excessive stress shielding and maintain a more favorable load-sharing environment for interbody fusion. These characteristics suggest that ISPF may represent a biomechanically balanced alternative for selected patients. Nevertheless, ISPF provided less stability than BPS during lateral bending and axial rotation and exhibited relatively higher device stress, indicating that its application should be individualized according to bone quality, spinous process morphology, and segmental stability. Overall, fixation after OLIF should not simply pursue maximal rigidity, but should balance initial stability, fusion-interface mechanics, adjacent-segment biomechanical effects, and surgical invasiveness.

Acknowledgements

We thank all colleagues in the Sixth Medical Center of People's Liberation Army General Hospital.

Author contributions

Junqin Ye and Jingbo Ma contributed equally to this work and share first authorship. Junqin Ye: Conceptualization, methodology, data curation, and drafting of the original manuscript. Jingbo Ma: Conceptualization, methodology, investigation, and drafting of the original manuscript. Qiang Jiang: Data acquisition, clinical investigation, and validation. Hanshuo Zhang: Formal analysis and data interpretation. Rigbat Rozi: Data collection and resources support. Jiaheng Han: Statistical analysis and visualization. Hongpeng Cui: Supervision, funding acquisition, critical revision of the manuscript. Yu Ding: Project administration, overall supervision, and final approval of the manuscript. Hongpeng Cui and Yu Ding are corresponding authors.

Funding

This project was supported by the National Natural Science Foundation of China (No. 82274637) and the “Sancai–Yidui” Science and Technology Innovation Team Program. The authors express their gratitude for this financial support.

Data availability

The datasets used during the current study are available from the corresponding author on reasonable request.

Declarations

Ethics approval and consent to participate

All procedures performed were in accordance with the ethical standards of the institutional review board and with the 1964 Helsinki Declaration and its later amendments or comparable ethical standards. The study was approved by the Ethics Review Committee of the Sixth Medical Center of the General Hospital of the Chinese People’s Liberation Army (No. HZKY-PJ-2025-1). Written informed consent to participate was obtained from the volunteer whose CT and MRI data were used for the model construction.

5. References

- 1 Katz, J. N., Zimmerman, Z. E., Mass, H. & Makhni, M. C. Diagnosis and Management of Lumbar Spinal Stenosis: A Review. *Jama* **327**, 1688-1699, doi:10.1001/jama.2022.5921 (2022).
- 2 Deyo, R. A. *et al.* Trends, major medical complications, and charges associated with surgery for lumbar spinal stenosis in older adults. *Jama* **303**, 1259-1265, doi:10.1001/jama.2010.338 (2010).
- 3 Mobbs, R. J., Phan, K., Malham, G., Seex, K. & Rao, P. J. Lumbar interbody fusion: techniques, indications and comparison of interbody fusion options including PLIF, TLIF, MI-TLIF, OLIF/ATP, LLIF and ALIF. *J Spine Surg* **1**, 2-18, doi:10.3978/j.issn.2414-469X.2015.10.05 (2015).
- 4 Skovrlj, B., Gilligan, J., Cutler, H. S. & Qureshi, S. A. Minimally invasive procedures on the lumbar spine. *World J Clin Cases* **3**, 1-9, doi:10.12998/wjcc.v3.i1.1 (2015).
- 5 Quillo-Olvera, J., Lin, G. X., Jo, H. J. & Kim, J. S. Complications on minimally invasive oblique lumbar interbody fusion at L2-L5 levels: a review of the literature and surgical strategies. *Ann Transl Med* **6**, 101, doi:10.21037/atm.2018.01.22 (2018).
- 6 Shen, S., You, X., Ren, Y. & Ye, S. Risk Factors of Cage Subsidence Following Oblique Lumbar Interbody Fusion: A Meta-analysis and Systematic Review. *World Neurosurg* **183**, 180-186, doi:10.1016/j.wneu.2023.12.110 (2024).
- 7 Zhang, S. *et al.* Oblique lateral interbody fusion with internal fixations in the treatment for cross-segment degenerative lumbar spine disease (L2-3 and L4-5) finite element analysis. *Sci Rep* **13**, 17116, doi:10.1038/s41598-023-43399-x (2023).
- 8 Fang, G. *et al.* Biomechanical Comparison of Stand-Alone and Bilateral Pedicle Screw Fixation

- for Oblique Lumbar Interbody Fusion Surgery-A Finite Element Analysis. *World Neurosurg* **141**, e204-e212, doi:10.1016/j.wneu.2020.05.245 (2020).
- 9 Xu, Z. *et al.* Biomechanical evaluation of different oblique lumbar interbody fusion constructs: a finite element analysis. *BMC Musculoskelet Disord* **25**, 97, doi:10.1186/s12891-024-07204-8 (2024).
- 10 Doulgeris, J. J., Aghayev, K., Gonzalez-Blohm, S. A., Lee, W. E., 3rd & Vrionis, F. D. Biomechanical comparison of an interspinous fusion device and bilateral pedicle screw system as additional fixation for lateral lumbar interbody fusion. *Clin Biomech (Bristol)* **30**, 205-210, doi:10.1016/j.clinbiomech.2014.10.003 (2015).
- 11 Liu, Z., Zhang, S., Li, J. & Tang, H. Biomechanical comparison of different interspinous process devices in the treatment of lumbar spinal stenosis: a finite element analysis. *BMC Musculoskelet Disord* **23**, 585, doi:10.1186/s12891-022-05543-y (2022).
- 12 Zhang, S. *et al.* Oblique lateral interbody fusion combined with different internal fixations for the treatment of degenerative lumbar spine disease: a finite element analysis. *BMC Musculoskelet Disord* **23**, 206, doi:10.1186/s12891-022-05150-x (2022).
- 13 Cai, X. Y., Bian, H. M., Chen, C., Ma, X. L. & Yang, Q. Biomechanical study of oblique lumbar interbody fusion (OLIF) augmented with different types of instrumentation: a finite element analysis. *J Orthop Surg Res* **17**, 269, doi:10.1186/s13018-022-03143-z (2022).
- 14 Pan, C. C., Lee, C. H., Chen, K. H., Yen, Y. C. & Su, K. C. Comparative Biomechanical Analysis of Unilateral, Bilateral, and Lateral Pedicle Screw Implantation in Oblique Lumbar Interbody Fusion: A Finite Element Study. *Bioengineering (Basel)* **10**, doi:10.3390/bioengineering10111238 (2023).
- 15 Ananth Swaminathan, S. *et al.* Impact of Habitual Flexion on Bone Formation After Spinal Fusion Surgery: An In Silico Study. *JOR Spine* **8**, e70075, doi:10.1002/jsp2.70075 (2025).
- 16 Cao, L., Liu, Y., Mei, W., Xu, J. & Zhan, S. Biomechanical changes of degenerated adjacent segment and intact lumbar spine after lumbosacral topping-off surgery: a three-dimensional finite element analysis. *BMC Musculoskelet Disord* **21**, 104, doi:10.1186/s12891-020-3128-5 (2020).
- 17 Guo, W. *et al.* Biomechanical evaluation of oblique lateral interbody fusion with various fixation methods for degenerative lumbar scoliosis: a finite element analysis considering different bone densities. *Front Bioeng Biotechnol* **13**, 1562268, doi:10.3389/fbioe.2025.1562268 (2025).
- 18 Shi, Y., Xie, Y. Z., Zhou, Q., Yu, Y. & Fan, X. H. The biomechanical effect of the relevant segments after facet-disectomy in different diameters under posterior lumbar percutaneous endoscopes: a three-dimensional finite element analysis. *J Orthop Surg Res* **16**, 593, doi:10.1186/s13018-021-02733-7 (2021).
- 19 Li, Z. *et al.* Biomechanical comparison of oblique lateral interbody fusion with interspinous versus other fixation techniques: A finite element study. *Medicine (Baltimore)* **105**, e48346, doi:10.1097/md.00000000000048346 (2026).
- 20 Chen, M. *et al.* Biomechanical characteristics of a novel interspinous distraction fusion device in the treatment of lumbar degenerative diseases: a finite element analysis. *BMC Musculoskelet Disord* **24**, 944, doi:10.1186/s12891-023-07066-6 (2023).
- 21 Yamamoto, I., Panjabi, M. M., Crisco, T. & Oxland, T. Three-dimensional movements of the whole lumbar spine and lumbosacral joint. *Spine (Phila Pa 1976)* **14**, 1256-1260, doi:10.1097/00007632-198911000-00020 (1989).

- 22 Shim, C. S. *et al.* Biomechanical evaluation of an interspinous stabilizing device, Locker. *Spine (Phila Pa 1976)* **33**, E820-827, doi:10.1097/BRS.0b013e3181894fb1 (2008).
- 23 Chen, C. S., Cheng, C. K., Liu, C. L. & Lo, W. H. Stress analysis of the disc adjacent to interbody fusion in lumbar spine. *Med Eng Phys* **23**, 483-491, doi:10.1016/s1350-4533(01)00076-5 (2001).
- 24 Zhang, Q. Y., Tan, J., Huang, K. & Xie, H. Q. Minimally invasive transforaminal lumbar interbody fusion versus oblique lateral interbody fusion for lumbar degenerative disease: a meta-analysis. *BMC Musculoskelet Disord* **22**, 802, doi:10.1186/s12891-021-04687-7 (2021).
- 25 Chang, S. Y. *et al.* Can Indirect Decompression Reduce Adjacent Segment Degeneration and the Associated Reoperation Rate After Lumbar Interbody Fusion? A Systemic Review and Meta-analysis. *World Neurosurg* **153**, e435-e445, doi:10.1016/j.wneu.2021.06.134 (2021).
- 26 Ma, X. *et al.* Oblique lateral interbody fusion combined with unilateral versus bilateral posterior fixation in patients with osteoporosis. *J Orthop Surg Res* **18**, 776, doi:10.1186/s13018-023-04262-x (2023).
- 27 Guo, H. Z. *et al.* Stability Evaluation of Oblique Lumbar Interbody Fusion Constructs with Various Fixation Options: A Finite Element Analysis Based on Three-Dimensional Scanning Models. *World Neurosurg* **138**, e530-e538, doi:10.1016/j.wneu.2020.02.180 (2020).
- 28 Song, C. *et al.* Biomechanical Evaluation of Oblique Lumbar Interbody Fusion with Various Fixation Options: A Finite Element Analysis. *Orthop Surg* **13**, 517-529, doi:10.1111/os.12877 (2021).
- 29 Yu, Q. *et al.* Finite element analysis of biomechanical investigation on diverse internal fixation techniques in oblique lumbar interbody fusion. *BMC Musculoskelet Disord* **25**, 804, doi:10.1186/s12891-024-07887-z (2024).
- 30 Ma, J., Li, T., Shen, N., Rozi, R. & Ding, Y. Interspinous process fixation versus posterior lumbar interbody fusion following decompression for single-level grade I degenerative spondylolisthesis: a retrospective propensity score-matched study. *J Orthop Surg Res* **21**, doi:10.1186/s13018-026-06683-w (2026).
- 31 Shen, H., Ju, C., Gao, T., Zhu, J. & Liu, W. A Biomechanical Evaluation of a Novel Interspinous Process Device: In Vitro Flexibility Assessment and Finite Element Analysis. *Bioengineering (Basel)* **12**, doi:10.3390/bioengineering12040384 (2025).
- 32 Du, C. F. *et al.* Does oblique lumbar interbody fusion promote adjacent degeneration in degenerative disc disease: A finite element analysis. *Comput Biol Med* **128**, 104122, doi:10.1016/j.compbiomed.2020.104122 (2021).
- 33 Wang, W. *et al.* Oblique lateral interbody fusion stand-alone vs. combined with percutaneous pedicle screw fixation in the treatment of discogenic low back pain. *Front Surg* **9**, 1013431, doi:10.3389/fsurg.2022.1013431 (2022).
- 34 Li, X. *et al.* Biomechanics of Cage for Stand-alone Oblique Lumbar Interbody Fusion. *Clin Spine Surg*, doi:10.1097/bsd.0000000000001949 (2025).
- 35 Kotheeranurak, V. *et al.* Subsidence of Interbody Cage Following Oblique Lateral Interbody Fusion: An Analysis and Potential Risk Factors. *Global Spine J* **13**, 1981-1991, doi:10.1177/21925682211067210 (2023).
- 36 Hu, Z. *et al.* The Influence of Endplate Morphology on Cage Subsidence in Patients With Stand-Alone Oblique Lateral Lumbar Interbody Fusion (OLIF). *Global Spine J* **13**, 97-103, doi:10.1177/2192568221992098 (2023).

- 37 Wang, Y. *et al.* Biomechanical Analysis of Adjacent Segments after Spinal Fusion Surgery Using a Geometrically Parametric Patient-Specific Finite Element Model. *J Vis Exp*, doi:10.3791/66247 (2024).