If the Adjustment of Insertional Pedicle Screw Positions Will Affect the Risk of Adjacent Segment Diseases Biomechanically? An in-silico Study

Chenyi Huang

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Zongchao Liu

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Huajun Hu

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Xueyan Chen

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Jingwen Chen

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Zhangchao Wei

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Zhongxin Fang

Xihua University

Zhipeng Xi

Jiangsu Province Hospital on Integration of Chinese and Western Medicine

Ping Cai

Affiliated Hospital of Nanjing University of Chinese Medicine

Jingchi Li (✉ Lijingchi9405@163.com)

Affiliated Traditional Chinese Medicine Hospital of Southwest Medical University

Research Article

Keywords: Fixation lengths, Moment arm, In-silico study, Posterior fixation system, Adjacent segment diseases

Posted Date: July 22nd, 2021

DOI: <https://doi.org/10.21203/rs.3.rs-714391/v1>

License: This work is licensed under a Creative Commons Attribution 4.0 International License.

[Read Full License](#)

Abstract

Background: The posterior fixation system (PFS) is extensively used in the treatment of spinal diseases. However, the fixation-induced pathological stress distribution pattern (e.g., stress concentration and motility compensation) will increase the risk of adjacent segment diseases (ASD). The accurate adjustment of insertional screw positions is possible under the guidance of the C-arm during percutaneous pedicle screw insertion. Although studies reported that the adjustment of insertional screw positions would affect the stiffness of the fusion segment, there were still no studies that clarified the issue that if the adjustment of insertional screw positions would affect the postoperative biomechanical environment and the risk of ASD.

Methods: To investigate if the change of screw positions and related moment arm in adjacent segments will affect mechanical indicators related to ASD, an intact lumbosacral model has been constructed and validated based on our anteriorly constructed model, the oblique lumbar interbody fusion fixed by bilateral PFS with different insertional positions has been simulated in the L4-L5 segment. Models were computed under completely identical loading conditions, including flexion, extension, bending, and rotation. Motility parameters, stress distribution in the intervertebral disc (IVD), and zygapophyseal joints (ZJ) in both cranial and caudal sides of functional units were recorded.

Results: Consist with anterior published studies, the change of fixation lengths has a more apparent biomechanical effect on the cranial than the caudal segment. Positive collections can be observed between the reduction of the moment arm in adjacent segments and the aggravation of motility compensation. Furthermore, which can also lead to stress concentration on ZJ facet cartilages. By contrast, no pronounced tendency of stress distribution on IVD can be observed with the change of moment arm.

Conclusions: during LIF operations fixed by percutaneous PFS, reducing the fixation stiffness by adjusting the insertional screw positions on the sagittal plane could alleviate the biomechanical deterioration and may be an effective method to reduce the risk of ASD (adjacent segmental instability in the short term and spinal stenosis in the long term).

Background

The posterior spinal fixation system (PFS) is extensively used in the treatment of spinal diseases to restore physiological alignment, maintain stability, and correct hypermotility [1, 2]. For example, during the lumbar interbody fusion surgery (LIF), PFS could construct three-column instant stability by transpedicular fixation [2, 3]. Although PFS was reported to be the gold standard of additional fixation technique in LIF surgery, the stiffness-increasing mechanism of PFS will lead to the fixation-induced pathological load transmission pattern (e.g., stress concentration and motility compensation in adjacent segments), and which have been repeatedly proved to be an initial trigger for adjacent segment diseases (ASD) [2–4].

Surgeons advocate removing PFS after solid interbody fusion, and its positive biomechanical effect on adjacent segments has been reported by in-silico studies [2, 3]. But this method has not been widely promoted in clinical practice, for it is difficult for patients without severe symptoms to accept a second surgical trauma. Additionally, different kinds of dynamic fixation systems (including different kinds of interspinous devices and dynamic, flexible rods) have been designed to alleviate postoperative biomechanical deterioration and reduce the risk of ASD and their biomechanical advantages have also been proved by in-silico and in-vitro studies [4–9]. However, a higher incidence rate of fixator failure and revision surgery inhibits these instrumentations' promotion [8, 10].

Hence, PFS is still the gold standard of the additional fixation device in LIF at the present stage. The optimization of surgical procedures during the use of PFS, rather than the replacement of PFS, may have a better clinical application prospect. The cortical bone trajectory (CBT) screw insertion method was proved to reduce the risk of ASD compared with the traditional trajectory screw fixation [11, 12]. However, the CBT screw has not been extensively promoted for its long and steep learning curve and limited application scenarios (CBT could not be inserted percutaneously) [11, 12]. During the modification of the traditional trajectory screw insertion technique, studies reported that the shift of the screw insertion point medially in the coronal plane could alleviate biomechanical deterioration and reduce the risk of ASD [13, 14]. Studies suggest that the biomechanical mechanism behind the relation between these screw insertion techniques and the risk of ASD should include not only biomechanical changes caused by different grades of the violation of zygapophyseal joints (ZJ) [15, 16], but also the change of fusion segmental stiffness and resulting overall lumbar biomechanical changes [2, 17].

Currently, as a minimally invasive surgical method, the percutaneous pedicle screw insertion technique is widely promoted [18, 19]. In which insertional screw positions can be accurately adjusted under the guidance of the C-arm, but there were no studies elucidate the biomechanical changes with the adjustment of screw insertion positions in the sagittal plane. Considering the adjustment of screw insertion positions will affect the moment arm of adjacent segments and which has been proved to have significant locally biomechanical impacts [2, 17]. and long segment LIF with the expansion of fixation length has been proved to be a risk factor of ASD [20, 21]. We believe that the optimization of screw insertion positions in the sagittal plane (i.e., reduce the fixation strength of PFS) could prolong the moment arm in adjacent segments and reduce the risk of ASD. In this study, the biomechanical significance for the adjustment of pedicle screw insertion positions in the sagittal plane has been investigated by in-silico analysis. To the best of our knowledge, published literature has not adequately clarified this issue.

Methods

Model construction

A previously constructed intact lumbosacral model (L3 to S1) has been used as a template in the model construction process [15, 22, 23]. Bone structures include cortical, cancellous, and bony endplates (BEP).

The cortical thickness was set as 0.8 mm, and BEP's thickness was defined according to measured values from large sample anatomic studies [22, 24, 25]. Nonbony components include the intervertebral disc (IVD) and ZJ cartilages. IVD consist of the nucleus core, the surrounding annulus, and cartilage endplates (CEP) on the cranial and caudal sides of the nucleus and inner part of the annulus (Fig. 1) [26, 27].

Boundary and loading conditions

Models were computed under completely identical loading conditions, including flexion, extension, bending, and rotation, and sizes of the moment in the mechanical indicators computation process were set to be consistent with the validation of the range of motion (ROM). They were set to be symmetric in the sagittal plane to increase their computational efficiency by allowing the unilateral calculation of the bending and axial rotation loading conditions (Fig. 1) [22, 28, 29]. Hybrid elements (including tetrahedron and hexahedron) with different mesh sizes were established in different components, and smaller mesh sizes were used in structures with low thickness and large deformation [15, 22]. In the definition of material properties, cortical and cancellous bone were set as anisotropic materials [30, 31], other parts of the model were defined by isotropic law [28, 31, 32]. The annulus was assumed to be hypoelastic [31, 33], and the nucleus was set as an incompressible 'semi-fluid pad' [28, 29]. Ligaments structures and capsules of ZJ were defined as cable elements in the pre-processing step of FEA [23, 34]. Contact elements defined facet cartilages of ZJ, and its frictional coefficient was set as zero [35, 36].

Model calibration and validation

All freedom degrees were fixed under the inferior surfaces of current models, and moments were applied on their superior surfaces [28, 37]. The stiffness of ligaments under different loading conditions was calibrated to reduce the difference between the computed ROM in the L4-L5 segment with that of in-vitro studies [38, 39]. A mesh convergency test was performed on the intact model by evaluating intradiscal pressure (IDP) change with different mesh sizes. The model was considered to be converged if the change of computed IDP was less than 3% [40, 41]. To ensure computational credibility, multi-indicators model validation has been accomplished by comparing the computed ROM, IDP, the disc compression (DC), and the facet contact force with values from in-vitro studies under different sizes and directions load [9, 42–44].

Surgical simulations and ASD's risk evaluation

The L4-L5 segment has been selected to simulate the oblique lumbar interbody fusion (OLIF) fixed by bilateral PFS with different insertional positions for the incidence rate of lumbar degenerative diseases in this segment was higher than that of the L3-L4 segment, and the L5-S1 segment was not suitable for OLIF generally [18, 19]. Lateral parts of the annulus, all of the nucleus, and CEPs in the surgical segment were removed, and a PEEK OLIF cage (18 mm long and 50 mm wide) filled with grafted bony material was inserted into interbody space [18, 32]. It was assumed that the disc height and lordotic angle of disc space were not affected by cage insertion, and the outline between cage and BEP was assumed to be perfectly matched (Fig. 2) [44, 45]. Considering ASD was a typical long-term complication, the boundary

conditions have been defined to simulate solid interbody fusion. In which, the contact type between grafted bone and BEP was set to be “bounded” (completely constrains the motion under all degrees of freedom), and the frictional coefficient in surfaces between cage and BEP was 0.8 [46, 47].

During the simulation of titanium alloy (Ti6Al4V) PFS fixation with different insertion positions, bilateral pedicle screws (were inserted into L4 and L5 vertebral bodies. The axis of screws on the cross-section was set to be parallel to the pedicle axis, and the screw axis was parallel to which of corresponding cranial BEP [6, 19]. The connection between the screw tulip and the nut was simplified to reduce the computational burden [2, 6, 48]. Five postoperative models with different insertional screw positions have been constructed, and the screw compaction effect was simulated by adjusting the material property of bony tissue around the screw thread (Fig. 2) [49, 50]. Motility parameters, stress distribution in IVD, and ZJ in both cranial and caudal sides of functional units were recorded to evaluate the risk of ASD.

Results

Multi-indicators model validation

Well-validated computational results can be recorded in the intact model. Specifically, the values of computed ROM and DC under were compared with which in in-vitro studies reported by Renner et, al [44], values of IDP were compared with the study published by Schilling et, al [9], and which of FCF were also compared with Wilson et al.'s study [43]. All of these indicators computed by the intact model were within ± 1 standard deviation of the average values reported by the above-mentioned *in-vitro* studies, proving that current models could make a good representation of real biomechanical situations (Fig. 3).

Changes in mechanical indicators related to ASD

Overall ROM, ROM in different segments (including the surgical and adjacent segments), and the proportion of different segmental ROM to the overall value have been computed and recorded to evaluate the motility compensation. Positive relations can be observed between PFS's fixation length and fixational stiffness except for the axial rotation condition. Specifically, the change of fixation length will slightly affect the overall ROM (the variation range was smaller than 5% except for model 2 (model with shortest fixation length) under the flexion loading condition). By contrast, the change of ROM in the fusion segment was dramatically under most loading conditions. Meanwhile, pathological motility compensation could be amplified and alleviated with the increase and decrease of moment arms in these segments, especially under the flexion condition in the cranial and bending in the caudal segment (Fig. 4).

FCF was not recorded under the flexion loading condition for ZJ cartilages that were not in contact. Based on the same reason, FCF ON the opposite side to the bending condition and the rotation side could not be recorded. In other words, FCF under left lateral bending is observed on left-side cartilages, while FCF under left axial rotation is observed on right-side cartilages. The change of insertional screw positions can lead to the change of FCF. Generally, the reduction of moment arm in adjacent segments will decrease FCF and vice versa. The variation tendency of the cranial side was more pronounced than

the caudal one (Fig. 5). To investigate the risk of disc degeneration, we calculate IDP, maximum values of annulus shear and equivalent stress, and the average strain energy density. Inconsistent with the variation tendency of ROM and FCF, no apparent tendency of these mechanical indicators can be observed with the change of insertional screw positions, especially under the rotation condition (Fig. 6).

Discussion

This work evaluated biomechanical deterioration and the related risk of ASD after OLIF fixed by PFS with different insertional screw positions (Fig. 2). An intact lumbosacral model and corresponding OLIF models were constructed, and biomechanical indicators closely related to ASD were computed and evaluated. The importance of the biomechanical environment for achieving positive postoperative clinical outcomes has been repeatedly demonstrated [2, 17, 48]. Thus, investigations on the biomechanical changes caused by different insertional screw positions and adjacent segments moment arms are of great significance for optimal operative strategy.

OLIF, rather than other LIF operations, has been selected for the following reasons. The percutaneous pedicle screw insertion was accomplished under C-arm fluoroscopy in OLIF, and the adjustment of insertion positions is feasible in this operation. By contrast, the selection of screw insertion positions in other lumbar fusion operations (e.g., transforaminal and posterior lumbar interbody fusion) was based on identifying anatomic structures [13, 14]. Considering the prevalence of anatomic variations and the hypertrophy of the articular process during the pathological process of spinal stenosis [50, 51], it's difficult to accurately judge and adjust the exact insertional screw position under the freehand pedicle insertion process. Furthermore, for the same reason, the promotion of the optimized insertional screw positions elucidated by this study may also be limited in LIF fixed by percutaneous pedicle screw.

The deterioration of the biomechanical environment caused by inappropriate surgery may be continuously amplified and lead to a devastating prognosis [20, 22, 52]. Therefore, the optimization of a surgical technique based on a biomechanical in-silico study are of great significance. There are three common pathological changes of ASD, including disc degeneration, ZJ degenerative osteoarthritis and spinal stenosis, and segmental instability [20, 21]. Herein, mechanical indicators related to these pathological changes have been computed. In the disc degeneration process, the endplate calcification and the annulus tears are two common phenotypes in the lumbar spine [53, 54]. IVD is an avascular structure. Therefore, trans-cartilage endplate diffusion is the primary nutrition and metabolism pathway of IVD [53, 55, 56]. Aberrant increase of average strain energy density is a compensatory reaction to endplate stress concentration, and this change was proved to be a predictor of the calcium deposition and osteogenesis process [57, 58]. The resulting occlusion of IVD metabolism could accelerate disc degeneration [55, 56].

The annulus-driven phenotype is the most common reason for disc degeneration in the lower lumbar spine [54, 55]. Different kinds of stress concentration on the annulus, especially on the post and post-lateral parts of the annulus, were proved to be related to different types of annulus tears [37, 59].

Meanwhile, the aberrant increase of IDP could also increase the risk of annulus failure [31, 60]; therefore, annulus stress distribution and IDP are critical indicators in related mechanical studies [31, 59]. Simultaneously annulus tears and increased intradiscal pressure would promote disc herniation, a common type of ASD. The in-growth of blood vessels along annulus tears will promote locally the inflammatory response, leading to extracellular matrix catabolism and further disc degeneration [56, 61]. The in-growth of pain-sensing nerve fibers is also the primary reason for postoperative pain recurrence in ASD [61, 62].

It is worth noting that the pathological change in ASD was not limited to IVD. The degenerative osteoarthritis, hypertrophy of the articular process, and resulting spinal canal stenosis were also essential triggers for the recompression of nerve structures, symptoms recurrence, and revision surgery [47, 63]. In other words, ZJ degeneration should also be taken into consideration in studies related to ASD, and which can be well reflected by evaluating the change of FCF [2, 22, 31]. Additionally, as mentioned above, postoperative pathological motility compensation and resulting spinal instability is also a basic form of ASD [20, 47], which could be reflected by the variation of ROM and its proportion [6, 22, 52]. Therefore, ROM can be used as an indicator not only for model calibration and validation but also for assessing ASD's risk. Moreover, close interactions were observed amongst different biomechanical indicators, and the interaction between segmental instability and spinal canal stenosis was also clearly elucidated. Reactive hyperplasia of the articular process and ligamentum structures caused by segmental instability was the main reason for spinal stenosis over a long period [50, 64]. In a word, by computing these biomechanical indicators, the risk of ASD could be investigated systematically.

Based on the current computational results, slight changes in stress concentration on the disc can be observed in cranial and caudal IVDs. Therefore, we can deduce that the tendency of disc degeneration acceleration may not be changed obviously with the change of moment arms (Fig. 6). By contrast, the fixation stiffness in the surgical segment and motility compensation in adjacent segments could be distinctly affected by the change of fixation length. It is worth noting that pronounced motility compensation can be recorded when the absolute value of the fixation length increased, and when the moment arm decreased is caused by the PFS shifts towards the measured side (Fig. 4). Meanwhile, although the range of variations is higher in the cranial than the caudal segment, the overall variation tendency of FCF is still consistent with which of the motility compensation (Fig. 5). Considering above mentioned interaction between segmental instability and spinal canal stenosis [50, 64], the optimization of insertional screw positions (i.e., reduce the fixation length) could optimize the biomechanical environment and reduce the risk of adjacent segmental instability in the short term and spinal stenosis in the long term.

Limitations

The most important limitation of the current in-silico study is the definition of ligaments. In this study, the mechanical effect of ligaments can only be acted on artificially selected positions rather than their entire original surfaces, for these ligaments were defined as cable elements in this study, and the potential risk

of mechanical indicators distortions should be taken into considerations. However, we believe that the computational results elucidated by current models are still reliable for the following reasons. The definition of cable ligaments has been widely accepted in the same kind of in-silico spinal studies [28, 29, 37], and the multi-indicators model validation has guaranteed the credibility of the current models. Additionally, no attach positions of cable elements are defined on structures with computed indicators (e.g., annulus, CEPs, and facet cartilages of ZJ); this determines that even if there is computational distortion, it can be excluded from the indicator's computation. The definition of ligaments should still be optimized in future in-silico studies.

Conclusion

Collectively, computed indicators in this study elucidated that during LIF operations fixed by percutaneous PFS, reducing the fixation stiffness by adjusting the insertional screw positions on the sagittal plane could alleviate motility compensation and stress concentration on ZJ cartilages, especially on the cranial segment. Thus, this mechanical effect may be an effective method to reduce the risk of ASD (adjacent segmental instability in the short term and spinal stenosis in the long term).

Abbreviations

ASD: Adjacent segment diseases,

BEP: Bony endplate,

CBT: Cortical bone trajectory,

CEP: Cartilage endplate,

DC: Disc compression,

FCF: Facet contact force,

IDP: Intradiscal pressure,

IVD: Intervertebral disc,

LIF: Lumbar interbody fusion,

OLIF: Oblique lumbar interbody fusion,

PFS: Posterior fixation system,

ROM: Range of motion,

ZJ: Zygapophyseal joint.

Declarations

Ethics approval and consent to participate

Approval for the current study protocol (including the lumbar CT scan) was obtained from the ethics committees of Jiangsu Province Hospital on Integration of Chinese and Western Medicine (2019LWKY015). We confirm that the subject signed the informed consent and submitted it to the ethics committee for review before the examination, and all methods were carried out in accordance with relevant guidelines and regulations.

Consent for publication

Not Applicable

Availability of data and materials

All the data of the manuscript are presented in the paper.

Competing interests

The authors declare that they have no competing interests.

Funding

This study was funded by the Key project of Jiangsu province social development (BE2019765). The fund provided by the above project are used for the conference affairs and travel expenses of studying courses related to three-dimensional spinal modeling and finite element analysis and the publication charges of this manuscript.

Authors' contributions

JCL, PC, CYH and ZCL contributed to the conception and design of the study, CYH, JCL, ZCL and ZXF contributed to the model construction, HJH, XYC, JWC and ZCW contributed to the literature review, PC, HJH and ZPX drawn figures, ZCL, ZXF, JWC and ZCW contributed to the data collection, CYH, JCL, PC and ZCL contributed to the manuscript preparation. All authors read and approved the final manuscript.

Acknowledgements

We acknowledge Mr. Xiaoyu Zhang for the guidance of figures drawing.

References

1. Xu M, Yang J, Lieberman IH, Haddas R: **Finite element method-based study of pedicle screw-bone connection in pullout test and physiological spinal loads.** *Medical engineering & physics* 2019, **67**:11-21.

2. Hsieh YY, Chen CH, Tsuang FY, Wu LC, Lin SC, Chiang CJ: **Removal of fixation construct could mitigate adjacent segment stress after lumbosacral fusion: A finite element analysis.** *Clinical biomechanics (Bristol, Avon)* 2017, **43**:115-120.
3. Tsuang FY, Tsai JC, Lai DM: **Effect of lordosis on adjacent levels after lumbar interbody fusion, before and after removal of the spinal fixator: a finite element analysis.** *BMC musculoskeletal disorders* 2019, **20**(1):470.
4. Galbusera F, Bellini CM, Anasetti F, Ciavarro C, Lovi A, Brayda-Bruno M: **Rigid and flexible spinal stabilization devices: a biomechanical comparison.** *Medical engineering & physics* 2011, **33**(4):490-496.
5. Shen H, Fogel GR, Zhu J, Liao Z, Liu W: **Biomechanical Analysis of Different Lumbar Interspinous Process Devices: A Finite Element Study.** *World neurosurgery* 2019, **127**:e1112-e1119.
6. Chuang WH, Lin SC, Chen SH, Wang CW, Tsai WC, Chen YJ, Hwang JR: **Biomechanical effects of disc degeneration and hybrid fixation on the transition and adjacent lumbar segments: trade-off between junctional problem, motion preservation, and load protection.** *Spine* 2012, **37**(24):E1488-1497.
7. Niosi CA, Wilson DC, Zhu Q, Keynan O, Wilson DR, Oxland TR: **The effect of dynamic posterior stabilization on facet joint contact forces: an in vitro investigation.** *Spine* 2008, **33**(1):19-26.
8. Lee CH, Kim YE, Lee HJ, Kim DG, Kim CH: **Biomechanical effects of hybrid stabilization on the risk of proximal adjacent-segment degeneration following lumbar spinal fusion using an interspinous device or a pedicle screw-based dynamic fixator.** *Journal of neurosurgery Spine* 2017, **27**(6):643-649.
9. Schilling C, Krüger S, Grupp TM, Duda GN, Blömer W, Rohlmann A: **The effect of design parameters of dynamic pedicle screw systems on kinematics and load bearing: an in vitro study.** *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2011, **20**(2):297-307.
10. Lee MJ, Lindsey JD, Bransford RJ: **Pedicle screw-based posterior dynamic stabilization in the lumbar spine.** *The Journal of the American Academy of Orthopaedic Surgeons* 2010, **18**(10):581-588.
11. Sakaura H, Ikegami D, Fujimori T, Sugiura T, Mukai Y, Hosono N, Fuji T: **Early cephalad adjacent segment degeneration after posterior lumbar interbody fusion: a comparative study between cortical bone trajectory screw fixation and traditional trajectory screw fixation.** *Journal of neurosurgery Spine* 2019, **32**(2):155-159.
12. Matsukawa K, Yato Y, Hynes RA, Imabayashi H, Hosogane N, Asazuma T, Matsui T, Kobayashi Y, Nemoto K: **Cortical Bone Trajectory for Thoracic Pedicle Screws: A Technical Note.** *Clinical spine surgery* 2017, **30**(5):E497-e504.
13. He B, Yan L, Guo H, Liu T, Wang X, Hao D: **The difference in superior adjacent segment pathology after lumbar posterolateral fusion by using 2 different pedicle screw insertion techniques in 9-year minimum follow-up.** *Spine* 2014, **39**(14):1093-1098.
14. Kim HJ, Chun HJ, Kang KT, Moon SH, Kim HS, Park JO, Moon ES, Kim BR, Sohn JS, Ko YN *et al*: **The biomechanical effect of pedicle screws' insertion angle and position on the superior adjacent segment in 1 segment lumbar fusion.** *Spine* 2012, **37**(19):1637-1644.

15. Li J, Zhang X, Xu W, Xi Z, Xie L: **Reducing the extent of facetectomy may decrease morbidity in failed back surgery syndrome.** *BMC musculoskeletal disorders* 2019, **20**(1):369.
16. Cardoso MJ, Dmitriev AE, Helgeson M, Lehman RA, Kuklo TR, Rosner MK: **Does superior-segment facet violation or laminectomy destabilize the adjacent level in lumbar transpedicular fixation? An in vitro human cadaveric assessment.** *Spine* 2008, **33**(26):2868-2873.
17. La Barbera L, Costa F, Villa T: **ISO 12189 standard for the preclinical evaluation of posterior spinal stabilization devices-II: A parametric comparative study.** *Proceedings of the Institution of Mechanical Engineers Part H, Journal of engineering in medicine* 2016, **230**(2):134-144.
18. Guo HZ, Tang YC, Guo DQ, Luo PJ, Li YX, Mo GY, Ma YH, Peng JC, Liang D, Zhang SC: **Stability Evaluation of Oblique Lumbar Interbody Fusion Constructs with Various Fixation Options: A Finite Element Analysis Based on Three-Dimensional Scanning Models.** *World neurosurgery* 2020, **138**:e530-e538.
19. Quillo-Olvera J, Lin GX, Jo HJ, Kim JS: **Complications on minimally invasive oblique lumbar interbody fusion at L2-L5 levels: a review of the literature and surgical strategies.** *Annals of translational medicine* 2018, **6**(6):101.
20. Wang H, Ma L, Yang D, Wang T, Liu S, Yang S, Ding W: **Incidence and risk factors of adjacent segment disease following posterior decompression and instrumented fusion for degenerative lumbar disorders.** *Medicine* 2017, **96**(5):e6032.
21. Park P, Garton HJ, Gala VC, Hoff JT, McGillicuddy JE: **Adjacent segment disease after lumbar or lumbosacral fusion: review of the literature.** *Spine* 2004, **29**(17):1938-1944.
22. Li J, Xu C, Zhang X, Xi Z, Liu M, Fang Z, Wang N, Xie L, Song Y: **TELD with limited foraminoplasty has potential biomechanical advantages over TELD with large annuloplasty: an in-silico study.** *BMC musculoskeletal disorders* 2021, **22**(1):616.
23. Li J, Xu W, Zhang X, Xi Z, Xie L: **Biomechanical role of osteoporosis affects the incidence of adjacent segment disease after percutaneous transforaminal endoscopic discectomy.** *Journal of orthopaedic surgery and research* 2019, **14**(1):131.
24. Zhao FD, Pollintine P, Hole BD, Adams MA, Dolan P: **Vertebral fractures usually affect the cranial endplate because it is thinner and supported by less-dense trabecular bone.** *Bone* 2009, **44**(2):372-379.
25. Liu J, Hao L, Suyou L, Shan Z, Maiwulanjiang M, Li S, Wang C, Fan S, Zhao F: **Biomechanical properties of lumbar endplates and their correlation with MRI findings of lumbar degeneration.** *Journal of biomechanics* 2016, **49**(4):586-593.
26. DeLucca JF, Cortes DH, Jacobs NT, Vresilovic EJ, Duncan RL, Elliott DM: **Human cartilage endplate permeability varies with degeneration and intervertebral disc site.** *Journal of biomechanics* 2016, **49**(4):550-557.
27. Jacobs NT, Cortes DH, Peloquin JM, Vresilovic EJ, Elliott DM: **Validation and application of an intervertebral disc finite element model utilizing independently constructed tissue-level constitutive**

- formulations that are nonlinear, anisotropic, and time-dependent. *Journal of biomechanics* 2014, **47**(11):2540-2546.
28. Dreischarf M, Zander T, Shirazi-Adl A, Puttlitz CM, Adam CJ, Chen CS, Goel VK, Kiapour A, Kim YH, Labus KM *et al*: **Comparison of eight published static finite element models of the intact lumbar spine: predictive power of models improves when combined together.** *Journal of biomechanics* 2014, **47**(8):1757-1766.
 29. Chuang WH, Kuo YJ, Lin SC, Wang CW, Chen SH, Chen YJ, Hwang JR: **Comparison among load-, ROM-, and displacement-controlled methods used in the lumbosacral nonlinear finite-element analysis.** *Spine* 2013, **38**(5):E276-285.
 30. Morgan EF, Bayraktar HH, Keaveny TM: **Trabecular bone modulus-density relationships depend on anatomic site.** *Journal of biomechanics* 2003, **36**(7):897-904.
 31. Tsouknidas A, Sarigiannidis SO, Anagnostidis K, Michailidis N, Ahuja S: **Assessment of stress patterns on a spinal motion segment in healthy versus osteoporotic bony models with or without disc degeneration: a finite element analysis.** *The spine journal : official journal of the North American Spine Society* 2015, **15**(3 Suppl):S17-s22.
 32. Kim HJ, Chun HJ, Moon SH, Kang KT, Kim HS, Park JO, Moon ES, Sohn JS, Lee HM: **Analysis of biomechanical changes after removal of instrumentation in lumbar arthrodesis by finite element analysis.** *Medical & biological engineering & computing* 2010, **48**(7):703-709.
 33. Wu HC, Yao RF: **Mechanical behavior of the human annulus fibrosus.** *Journal of biomechanics* 1976, **9**(1):1-7.
 34. Chen C, Yuchi CX, Gao Z, Ma X, Zhao D, Li JW, Xu B, Zhang CQ, Wang Z, Du CF *et al*: **Comparative analysis of the biomechanics of the adjacent segments after minimally invasive cervical surgeries versus anterior cervical discectomy and fusion: A finite element study.** *Journal of orthopaedic translation* 2020, **23**:107-112.
 35. Du CF, Yang N, Guo JC, Huang YP, Zhang C: **Biomechanical response of lumbar facet joints under follower preload: a finite element study.** *BMC musculoskeletal disorders* 2016, **17**:126.
 36. Woldtvedt DJ, Womack W, Gadowski BC, Schuldt D, Puttlitz CM: **Finite element lumbar spine facet contact parameter predictions are affected by the cartilage thickness distribution and initial joint gap size.** *Journal of biomechanical engineering* 2011, **133**(6):061009.
 37. Schmidt H, Galbusera F, Rohlmann A, Shirazi-Adl A: **What have we learned from finite element model studies of lumbar intervertebral discs in the past four decades?** *Journal of biomechanics* 2013, **46**(14):2342-2355.
 38. Schmidt H, Heuer F, Simon U, Kettler A, Rohlmann A, Claes L, Wilke HJ: **Application of a new calibration method for a three-dimensional finite element model of a human lumbar annulus fibrosus.** *Clinical biomechanics (Bristol, Avon)* 2006, **21**(4):337-344.
 39. Schmidt H, Heuer F, Drumm J, Klezl Z, Claes L, Wilke HJ: **Application of a calibration method provides more realistic results for a finite element model of a lumbar spinal segment.** *Clinical biomechanics (Bristol, Avon)* 2007, **22**(4):377-384.

40. Fan W, Guo LX, Zhang M: **Biomechanical analysis of lumbar interbody fusion supplemented with various posterior stabilization systems.** *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2021.
41. Ottardi C, Galbusera F, Luca A, Prosdocimo L, Sasso M, Brayda-Bruno M, Villa T: **Finite element analysis of the lumbar destabilization following pedicle subtraction osteotomy.** *Medical engineering & physics* 2016, **38**(5):506-509.
42. Kim HJ, Kang KT, Son J, Lee CK, Chang BS, Yeom JS: **The influence of facet joint orientation and tropism on the stress at the adjacent segment after lumbar fusion surgery: a biomechanical analysis.** *The spine journal : official journal of the North American Spine Society* 2015, **15**(8):1841-1847.
43. Wilson DC, Niosi CA, Zhu QA, Oxland TR, Wilson DR: **Accuracy and repeatability of a new method for measuring facet loads in the lumbar spine.** *Journal of biomechanics* 2006, **39**(2):348-353.
44. Renner SM, Natarajan RN, Patwardhan AG, Havey RM, Voronov LI, Guo BY, Andersson GB, An HS: **Novel model to analyze the effect of a large compressive follower pre-load on range of motions in a lumbar spine.** *Journal of biomechanics* 2007, **40**(6):1326-1332.
45. Zhao X, Du L, Xie Y, Zhao J: **Effect of Lumbar Lordosis on the Adjacent Segment in Transforaminal Lumbar Interbody Fusion: A Finite Element Analysis.** *World neurosurgery* 2018, **114**:e114-e120.
46. Polikeit A, Ferguson SJ, Nolte LP, Orr TE: **Factors influencing stresses in the lumbar spine after the insertion of intervertebral cages: finite element analysis.** *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2003, **12**(4):413-420.
47. Zhong ZM, Deviren V, Tay B, Burch S, Berven SH: **Adjacent segment disease after instrumented fusion for adult lumbar spondylolisthesis: Incidence and risk factors.** *Clinical neurology and neurosurgery* 2017, **156**:29-34.
48. He L, Xiang Q, Yang Y, Tsai TY, Yu Y, Cheng L: **The anterior and traverse cage can provide optimal biomechanical performance for both traditional and percutaneous endoscopic transforaminal lumbar interbody fusion.** *Computers in biology and medicine* 2021, **131**:104291.
49. Wilke HJ, Wenger K, Claes L: **Testing criteria for spinal implants: recommendations for the standardization of in vitro stability testing of spinal implants.** *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 1998, **7**(2):148-154.
50. O'Leary SA, Paschos NK, Link JM, Klineberg EO, Hu JC, Athanasiou KA: **Facet Joints of the Spine: Structure-Function Relationships, Problems and Treatments, and the Potential for Regeneration.** *Annual review of biomedical engineering* 2018, **20**:145-170.
51. O'Leary SA, Link JM, Klineberg EO, Hu JC, Athanasiou KA: **Characterization of facet joint cartilage properties in the human and interspecies comparisons.** *Acta biomaterialia* 2017, **54**:367-376.
52. Wang B, Hua W, Ke W, Lu S, Li X, Zeng X, Yang C: **Biomechanical Evaluation of Transforaminal Lumbar Interbody Fusion and Oblique Lumbar Interbody Fusion on the Adjacent Segment: A Finite**

- Element Analysis.** *World neurosurgery* 2019, **126**:e819-e824.
53. Adams MA, Freeman BJ, Morrison HP, Nelson IW, Dolan P: **Mechanical initiation of intervertebral disc degeneration.** *Spine* 2000, **25**(13):1625-1636.
 54. Adams MA, Dolan P: **Intervertebral disc degeneration: evidence for two distinct phenotypes.** *Journal of anatomy* 2012, **221**(6):497-506.
 55. Adams MA, Roughley PJ: **What is intervertebral disc degeneration, and what causes it?** *Spine* 2006, **31**(18):2151-2161.
 56. Vergroesen PP, Kingma I, Emanuel KS, Hoogendoorn RJ, Welting TJ, van Royen BJ, van Dieën JH, Smit TH: **Mechanics and biology in intervertebral disc degeneration: a vicious circle.** *Osteoarthritis and cartilage* 2015, **23**(7):1057-1070.
 57. Von Forell GA, Nelson TG, Samartzis D, Bowden AE: **Changes in vertebral strain energy correlate with increased presence of Schmorl's nodes in multi-level lumbar disk degeneration.** *Journal of biomechanical engineering* 2014, **136**(6):061002.
 58. Ganbat D, Kim YH, Kim K, Jin YJ, Park WM: **Effect of mechanical loading on heterotopic ossification in cervical total disc replacement: a three-dimensional finite element analysis.** *Biomechanics and modeling in mechanobiology* 2016, **15**(5):1191-1199.
 59. Qasim M, Natarajan RN, An HS, Andersson GB: **Damage accumulation location under cyclic loading in the lumbar disc shifts from inner annulus lamellae to peripheral annulus with increasing disc degeneration.** *Journal of biomechanics* 2014, **47**(1):24-31.
 60. Pezowicz CA, Schechtman H, Robertson PA, Broom ND: **Mechanisms of anular failure resulting from excessive intradiscal pressure: a microstructural-micromechanical investigation.** *Spine* 2006, **31**(25):2891-2903.
 61. Stefanakis M, Al-Abbasi M, Harding I, Pollintine P, Dolan P, Tarlton J, Adams MA: **Annulus fissures are mechanically and chemically conducive to the ingrowth of nerves and blood vessels.** *Spine* 2012, **37**(22):1883-1891.
 62. García-Cosamalón J, del Valle ME, Calavia MG, García-Suárez O, López-Muñiz A, Otero J, Vega JA: **Intervertebral disc, sensory nerves and neurotrophins: who is who in discogenic pain?** *Journal of anatomy* 2010, **217**(1):1-15.
 63. Ravindra VM, Senglaub SS, Rattani A, Dewan MC, Härtl R, Bisson E, Park KB, Shrime MG: **Degenerative Lumbar Spine Disease: Estimating Global Incidence and Worldwide Volume.** *Global spine journal* 2018, **8**(8):784-794.
 64. Szkoda-Poliszuk K, Żak M, Pezowicz C: **Finite element analysis of the influence of three-joint spinal complex on the change of the intervertebral disc bulge and height.** *International journal for numerical methods in biomedical engineering* 2018, **34**(9):e3107.

Figures

Figure 1

Moments applied in the current models: (a) Flexion-Extension; (b) Left bending right; (c) Left axial rotation.

Figure 2

Diagrams of OLIF fixed by PFS with different fixation lengths and screw positions. M-M: Screws were inserted into the middle positions of both cranial and caudal vertebral bodies; B-T: Screws were inserted into the bottom of the cranial and the top of the caudal vertebral bodies (Shortest fixation length of PFS); T-B: Screws were inserted into the top of the cranial and the bottom of the caudal vertebral bodies (Longest fixation length of PFS); BB: Screws were inserted into the bottom of both cranial and caudal

vertebral bodies (The downward movement of PFS); TT: Screws were inserted into the top of both cranial and caudal vertebral bodies (The upward movement of PFS).

Figure 3

Multi-indicators model validation under different loading conditions. Valin-vitro = Indicators measured by published in-vitro studies; Valin-silico = Indicators computed by the current in-silico study; F-E = Flexion-Extension; L-R = Left-Lateral; B = Bending; AR = Axial rotation.

Figure 4

Comparison of ROM between surgical models with different insertional screw positions. F = Flexion; E = Extension; B = Bending; AR = Axial rotation.

Figure 5

Comparison of FCF between surgical models with different insertional screw positions. F = Flexion; E = Extension; B = Bending; AR = Axial rotation.

L3-L4 (Cranial segment)

Average strain energy density ($\mu\text{J}/\text{mm}^3$)

Intradiscal pressure (KPa)

Maximum annulus equivalent stress (MPa)

Maximum annulus shear stress (MPa)

Amplitude of changes (%)

Amplitude of changes (%)

Amplitude of changes (%)

Amplitude of changes (%)

L5-S1 (Caudal segment)

Average strain energy density ($\mu\text{J}/\text{mm}^3$)

Intradiscal pressure (KPa)

Maximum annulus equivalent stress (MPa)

Maximum annulus equivalent stress (MPa)

Amplitude of changes (%)

Amplitude of changes (%)

Amplitude of changes (%)

Amplitude of changes (%)

■ M-M ■ B-T ■ T-B ■ B-B ■ T-T

Figure 6

Comparison of indicators related to disc degeneration between surgical models with different insertional screw positions. F = Flexion; E = Extension; B = Bending; AR = Axial rotation.