

# Biomechanical Effects of Pedicle Screw Positioning on the Surgical Segment in Models After Oblique Lumbar Interbody Fusion: An in-silico Study

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**Background:** Bilateral pedicle screw (BPS) is the “gold standard” of fixation methods for patients with lumbar interbody fusion. Biomechanical deterioration initially triggers complications in the surgical segment. Studies proved that BPS positions and trajectory changes affect the local biomechanical environment. However, no study illustrates the biomechanical effect of insertional screw positions' change on the surgical segment.

**Methods:** Oblique lumbar interbody fusion (OLIF) with different BPS insertional positions has been simulated in a well-validated lumbo-sacral model. Fixation stability and stress responses on the surgical segment were evaluated under identical loading conditions.

**Results:** There is no clear variation tendency for the risk of BPS failure and the change of strain energy density of the grafted bone. However, shifting the insertional screw position close to the surgical segment will increase the range of motions (ROM) in the surgical segment and lead to stress concentration of bony structures, especially in the caudal side of the surgical segment.

**Conclusion:** Adjusting the insertional position of BPS close to the surgical segment in OLIF models will lead to stress concentration of bony structures and surgical segmental instability. Therefore, reducing BPS's fixation length was not recommended, which may increase the risk of segmental instability, non-union, and cage subsidence.

**Keywords:** oblique lumbar interbody fusion, bilateral pedicle screw, cage subsidence, insertional screw positions, non-union

## Introduction

Lumbar interbody fusion (LIF) operations have been widely used to treat lumbar degenerative diseases (LDD).<sup>1,2</sup> By transpedicular three-column fixation, bilateral pedicle screw (BPS) could provide the best fixation stability compared with other additional fixation devices.<sup>3,4</sup> Surgical segment complications (eg, cage subsidence, non-union, and BPS failures) will negatively affect LIF patients' prognosis.<sup>2,5</sup> Biomechanical deteriorations initially trigger these complications.<sup>6-8</sup> Studies reported that screw insertional positions and trajectories would affect the local biomechanical environment.<sup>9-11</sup>

Recently, the oblique lumbar interbody fusion (OLIF) has been rapidly promoted.<sup>1,2</sup> Compared with posterior approach LIF operations, the large footprint of OLIF cage is less likely to result in cage subsidence, but considering the indirect nerve structures' decompression is entirely rely on the distraction of disc space, the negative effect on patient's prognosis is pronounced in OLIF than other posterior approach LIF patients.<sup>1,5,7,12</sup> Percutaneous BPS fixation is widely used in OLIF operations. In this process, the insertional screw positions are highly adjustable under C-arm

radiography guidance. Insertional positions' changes will affect the fixation length of BPS, resulting in changes in surgical segment stiffness and local stress distribution that may impact the risk of complications on the surgical segment.<sup>3,11,13</sup> Thus, mechanical effects of insertional screw position's change on the surgical segment should be elucidated to optimize this surgical strategy and reduce complication risk. In this study, using a highly biomimetic lumbo-sacral model, BPS fixation with different insertional screw positions has been simulated, and mechanical indicators related to complications in the surgical segment have been computed and recorded. Published literature has not adequately clarified this issue to the best of our knowledge.

## Methods

### Model Construction

Our previously published studies have constructed and validated highly biomimetic in-vitro data based finite element (FE) model (L3-S1). Bone structures of this model include cortical, cancellous, and bony endplates (BEP).<sup>14,15</sup> The cortical thickness was set as 0.8 mm, the thickness, concave angle, and depth of superior and inferior BEPs were defined separately according to measured values from large sample anatomic studies. Nonbony components include the intervertebral disc (IVD) and facet cartilages. IVD consists of the nucleus core, the surrounding annulus, and cartilage endplates (CEP).<sup>14,15</sup> BEPs' outlines cover the entire IVD, and that of CEP was set on the cranial and caudal sides of the nucleus and inner part of the annulus. Based on the MRI data measurement, the nucleus's cross-sectional area accounted for 38% of the IVD.<sup>14,15</sup>

The L4-L5 segment was selected to simulate the anterior psoas approach oblique lumbar interbody fusion (OLIF) fixed by BPS with different insertional screw positions. In the surgical simulation processes, lateral parts of the annulus, all of the nucleus, and CEPs were removed, and a polyether-ether-ketone (PEEK) OLIF cage (18 mm width and 50 mm length) filled with grafted bone was inserted into the interbody space.<sup>16,17</sup> Cage and BEP were assumed to be completely matched. The lordotic angle and disc height of postoperative models were identical to the preoperative model.<sup>13,17,18</sup>

During the simulation of BPS fixation, four identical titanium alloy pedicle screws (diameter: 6mm, length: 45 mm) were inserted into L4-L5 vertebral bodies. The axes of screws in the transverse plane were parallel to the pedicle axis, which in the sagittal plane was parallel to the BEP in the cranial side.<sup>19,20</sup> Screw threads were preserved, and the screw compaction effect was simulated by adjusting the material property of cancellous around the thread.<sup>13,20,21</sup> The connection between the screw tulip and the nut was simplified to increase the computational efficiency. By adjusting insertional positions in different vertebral bodies, five models with different fixation lengths of BPS were simulated (Figure 1).

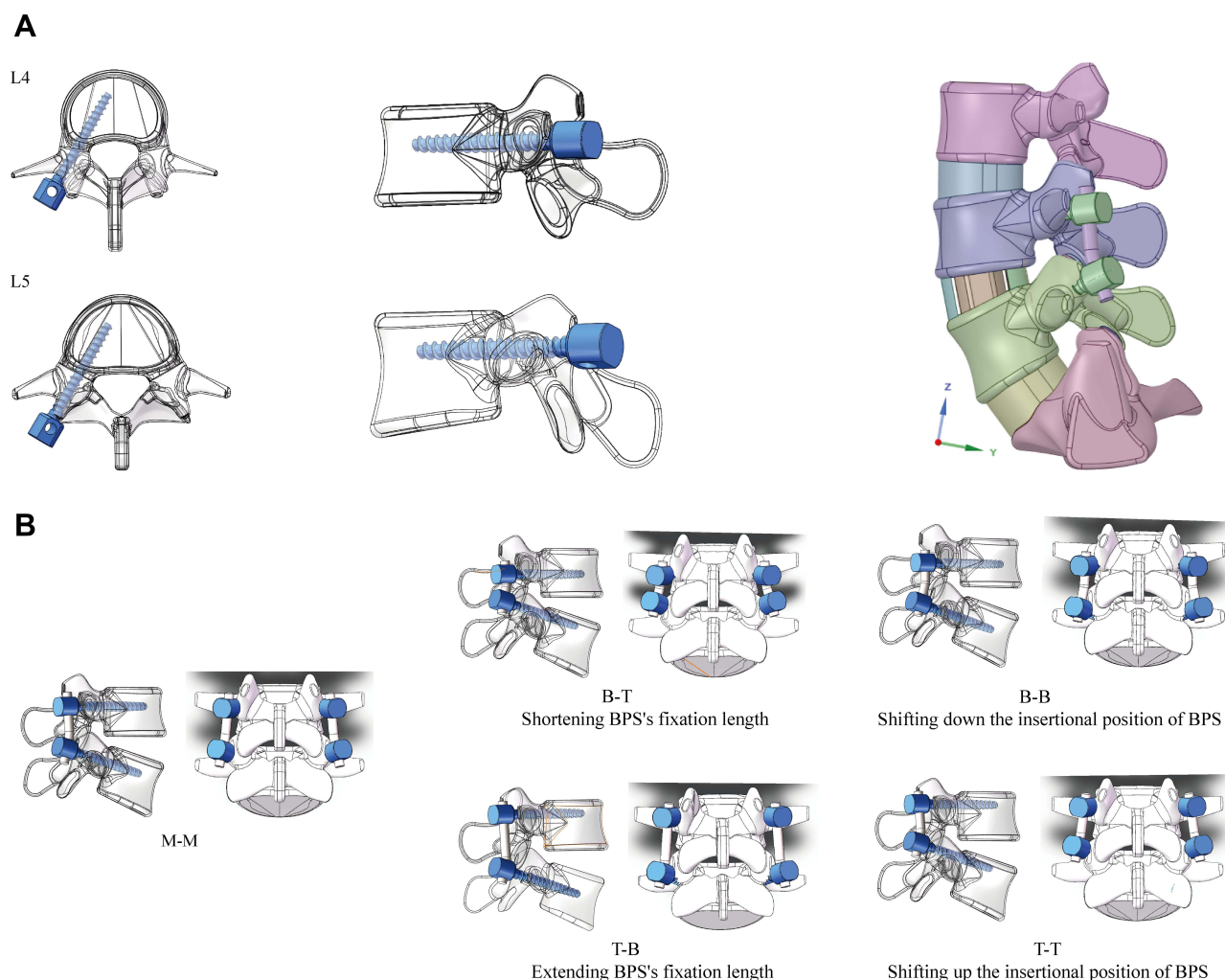
### Boundary and Loading Conditions

Hybrid meshes with different sizes were set in different components. Mesh refinement was set in structures with low thickness and large deformation (eg, BEP, facet cartilage, and posterior parts of the annulus).<sup>14,15</sup> In the definition of material properties, cortical and cancellous bone were defined by anisotropic law,<sup>22,23</sup> other components of these models were defined by isotropic law. The annulus was assumed to be hypoelastic material, and the nucleus was set as a semi-fluid incompressible material.<sup>24,25</sup> Ligaments and facet capsules were defined as cable elements. The frictional coefficient between facet cartilages, cage and BEP, and screw-cancellous interfaces were set as 0, 0.8, and 0.2, respectively.<sup>26,27</sup>

Inferior surfaces of models were fixed entirely, all moments in different directions were applied on the superior endplate of models.<sup>14,15</sup> Mechanical indicators were computed under four different loading conditions, including 8 Nm flexion, 6Nm extension, 6Nm bending, and 4Nm rotation.<sup>18,28</sup> In this process, models are set to be symmetrical along the sagittal plane to increase their computational efficiency by allowing the unilateral calculation of the bending and axial rotation loading conditions.<sup>14,15</sup>

### Model Calibration and Validation

ROMs in the L4-L5 segment were seen as the reference of model calibration. The stiffness of ligaments under different loading conditions was calibrated to reduce the difference between the computed and in-vitro measured values.<sup>29,30</sup> We performed a mesh convergence test on the calibrated intact model by evaluating the change of intradiscal pressure (IDP)



**Figure 1** Diagrams of surgical simulations in this study. **(A)** BPS fixation 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). **(B)** Screw trajectories in the transverse and sagittal planes, and cage's position in postoperative models.

with different mesh sizes. The model was considered converged if the change of computed IDP was less than 3%. Then, to ensure computational credibility, multi-indicator model validation has been accomplished.<sup>31,32</sup> In this process, the computed ROM and DC were compared with measured values from in-vitro studies reported by Renner et al, IDP was compared with values measured by Schilling et al, and which of FCF were also compared with Wilson et al's study.<sup>33–35</sup>

## Results

### Multi-Indicator Model Validation

Computational results of multi-indicator model validation show that the intact model with calibrated ligamentum stiffness could make good representations of the actual mechanical environment. These mechanical indicators computed by the calibrated intact model were within  $\pm 1$  standard deviation of the average values measured by in-vitro studies (Figure 2).

### Mechanical Effects of Insertional Screw Position's Changes

Mechanical indicators were computed to evaluate the risk of complications in the surgical segment. We measured the surgical segment range of motions (ROM) to investigate the fixation stability and the risk of non-union. Positive correlations between PFS's fixation length and stiffness were recorded (Figure 3). Shifting BPS to the caudal side can

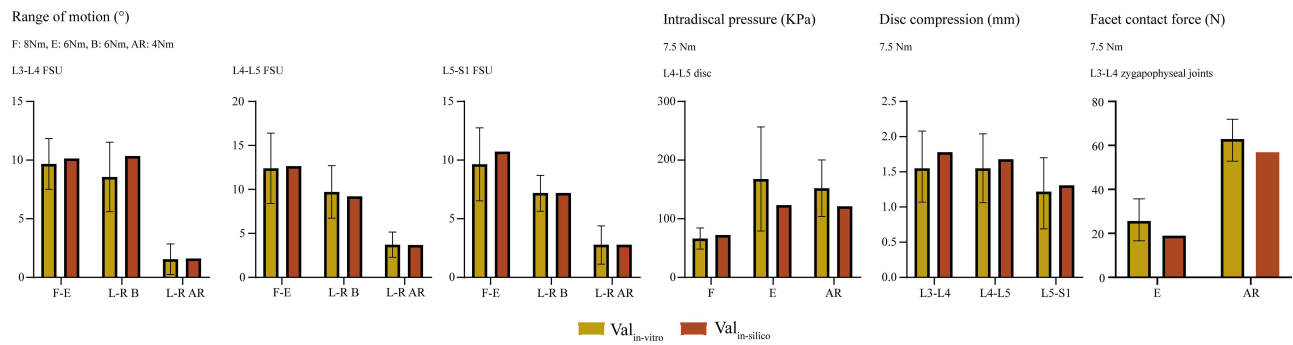


Figure 2 Multi-indicator model validation.

also increase surgical segment stiffness under the same fixation length. The grafted bone’s strain energy density (SED) was also computed to identify the non-union risk, and the maximum von-Mises stress on BPS was computed to investigate the risk of BPS failure (Figure 3). With the change of fixed length and insertional screw positions, no apparent variation tendency can be observed for SED. By contrast, the maximum stress of BPS increased whether the screw was shifted upward or downward from the middle of the pedicle.

Additionally, maximum stress on the BEP and subendplate cancellous bone (cancellous bone within 5mm of the BEP, SCB) was also recorded to evaluate the risk of cage subsidence.<sup>36,37</sup> The evident variation tendency of this indicator can be observed with the insertional screw position’s change. Specifically, when the screw position is close to the surgical segment (shifting down of screw on the cranial side and shifting up of screw on the caudal side), stress concentration occurs in these bony structures (Figure 4). The most significant change can be observed in the inferior SCB. When the insertional position of the caudal screw shifted up, the maximum stress of inferior BEP and SCB increased dramatically under all loading conditions. More importantly, the maximum stress is close to or even higher than the average yield strength of cancellous bone (2.37 Mpa) under some loading conditions (Figure 5).<sup>19,38</sup>

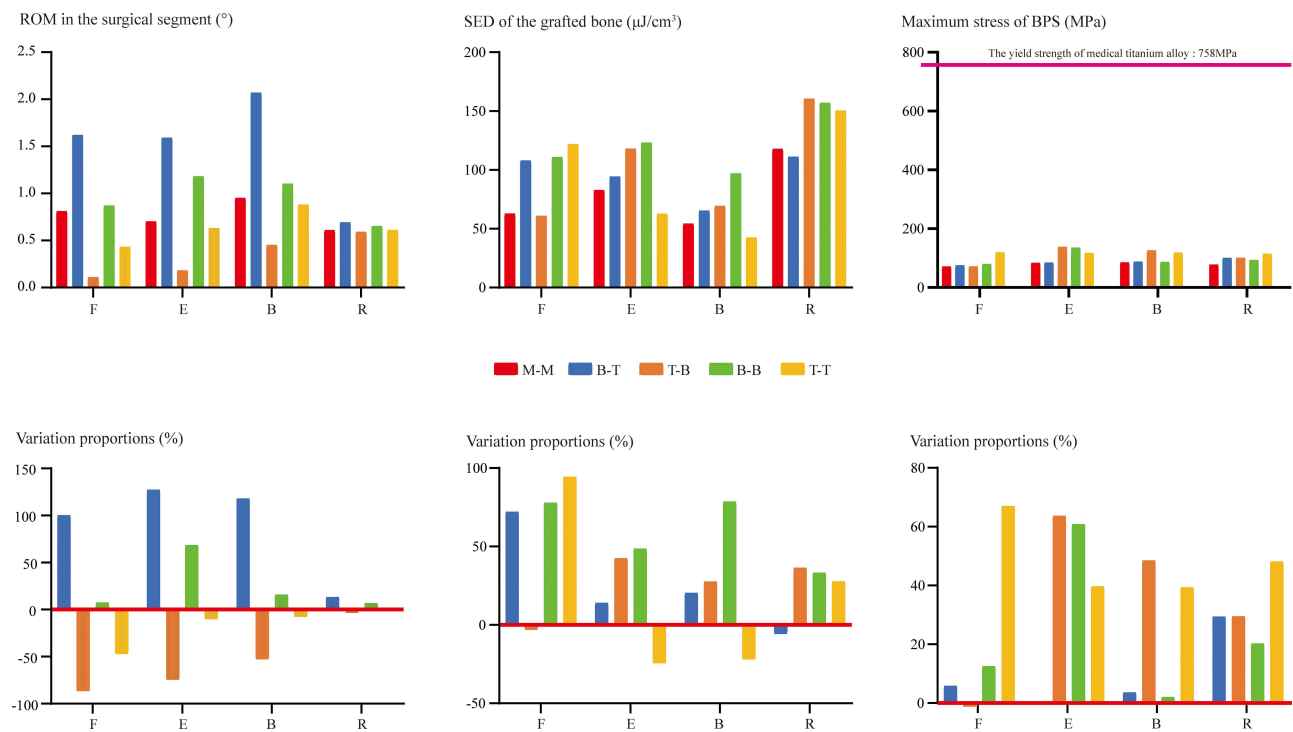
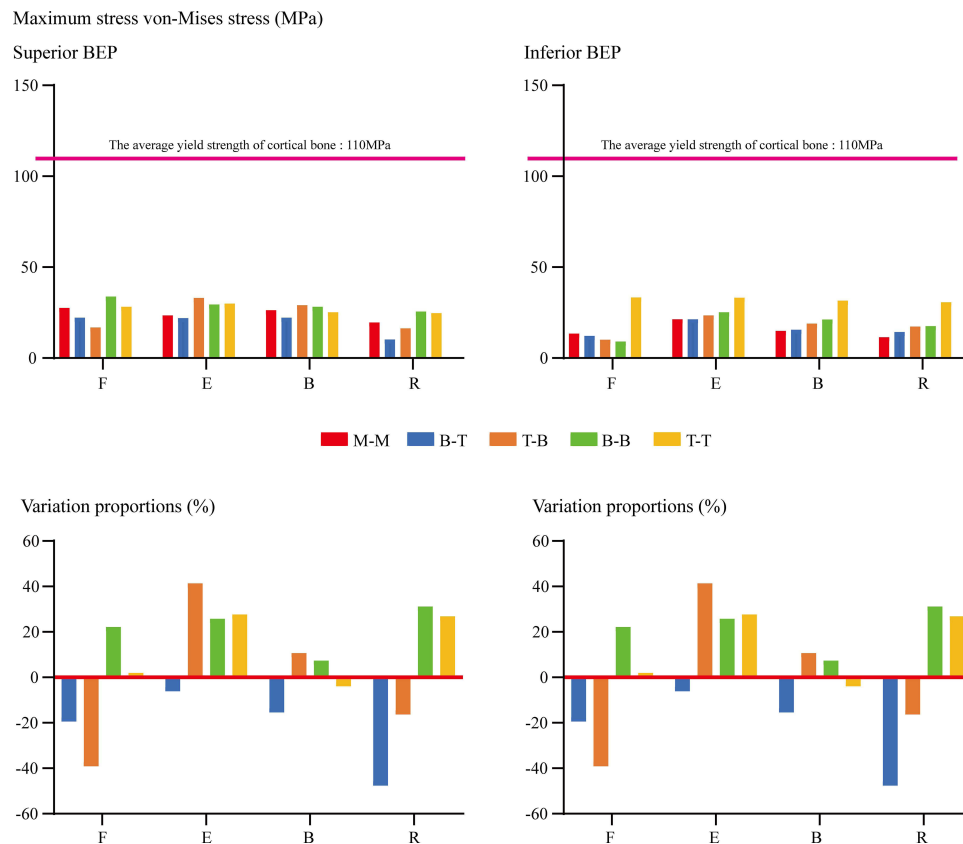


Figure 3 ROM in the surgical segment, SED of the grafted bone, and the maximum von-Mises stress on BPS. Abbreviations: F, flexion; E, extension; B, bending; R, axial rotation.



**Figure 4** The maximum von-Mises stress on BEP.

## Discussion

Biomechanical deteriorations initially trigger surgical segment complications.<sup>6–8</sup> Relations between screw positions and biomechanical environments have been widely reported.<sup>9–11</sup> This study has numerically investigated the effects of screw insertional position changes, and biomechanical indicators related to surgical segment complications (ie, cage subsidence, non-union, segmental instability, and fixation failure) have been computed and recorded. Modifying the surgical strategy based on these computational results may be an effective method to optimize patients' prognoses.

Segmental instability and non-union could trigger segmental instability and the recurrence of low back pain. The SED of the grafted bone was proved to be a credible predictor of osteogenesis by mechanical and clinical studies,<sup>39,40</sup> fixation stability of the surgical segment was also important for the osteointegration process,<sup>41,42</sup> and the surgical segment's ROM a reliable indicator when evaluating fixation stability.<sup>43–45</sup> In this study, the SED of grafted bone did not show a clear variation tendency with the change of screw position. By contrast, the surgical segment's ROM increased with the reduction of fixation length. Therefore, we can deduce that reducing fixation length by changing insertional screw positions may increase the risk of segmental instability and non-union.

Besides, aberrant stress concentration in the BPS may lead to its failure, and in turn, negatively affect the surgical segment's stability and impede interbody bone integration.<sup>20,46</sup> The increase of maximum stress can be observed when shifting up and down the BPS from the middle of the pedicle. Variations in the length of screws that were not inserted into the vertebra could explain this phenomenon, which may be related to the length of screws that do not insert into the vertebral body. Specifically, mechanical studies show that uniform stress distribution is achieved when the screw is fully inserted into the vertebral body,<sup>20,47,48</sup> no matter which direction the insertional screw position shifts from the middle position of the pedicle, the length of screws that do not insert into the vertebral body will be increased (Figure 6). This may be an initial trigger of stress concentration in BPS. Considering the maximum stress's value was far less than the

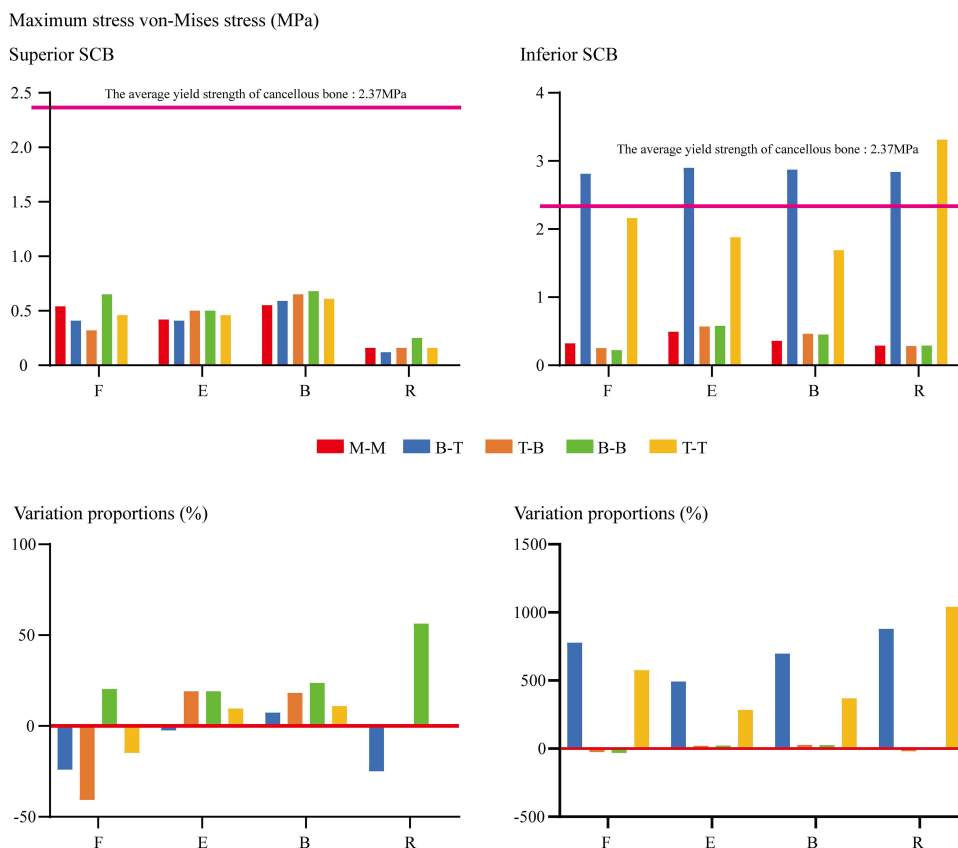
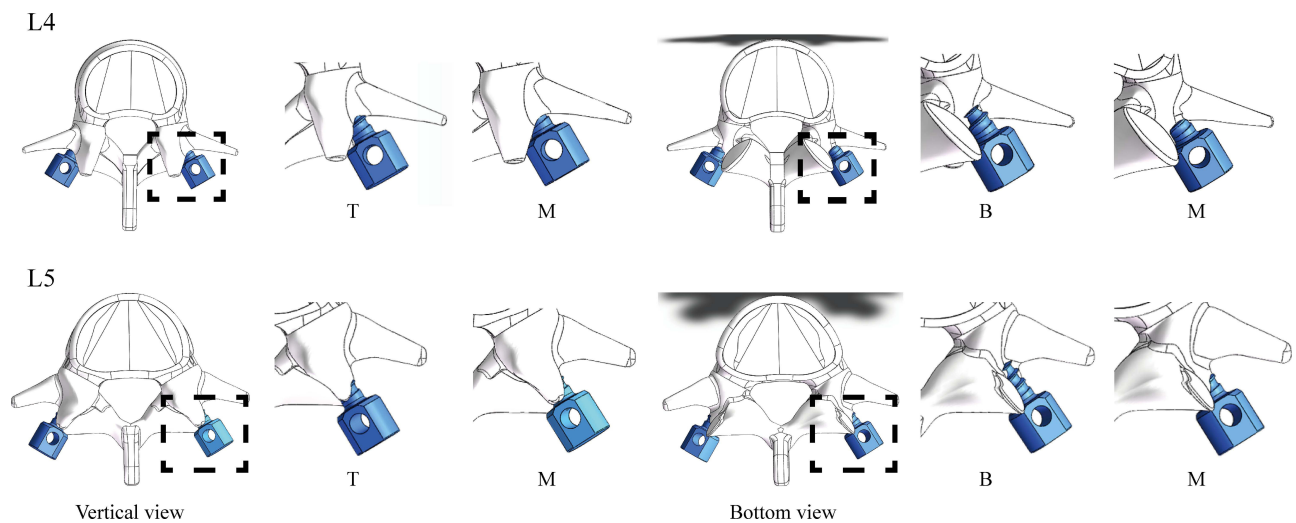


Figure 5 The maximum von-Mises stress on SCB.

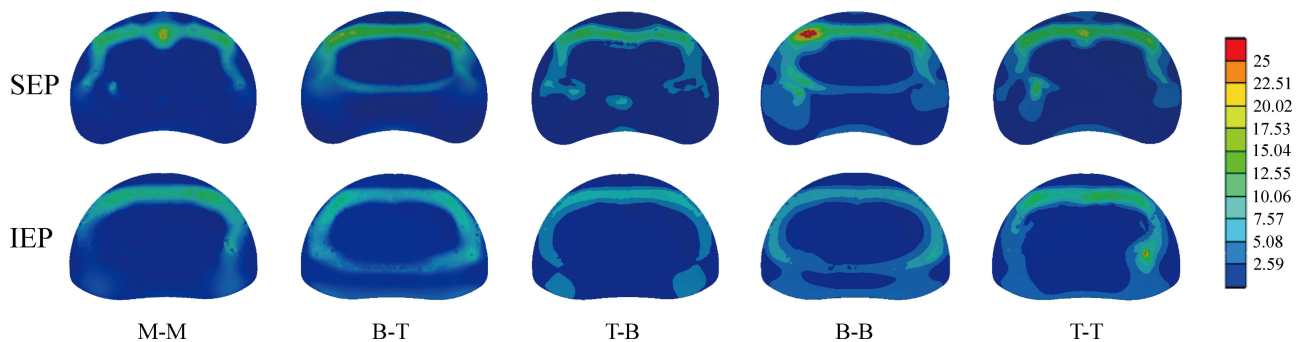
yield strength of medical titanium alloy (758 Mpa),<sup>20,49</sup> we believe that the change of insertional screw positions and BPS’s fixation length will not significantly affect the risk of non-fusion and BPS failure.

More significantly, aberrant stress concentration in the BEP and SCB will damage these bony structures and increase the risk of cage subsidence; evident stress concentration in bony structures can be observed when the screw position is close to the surgical segment, especially for the inferior BEP and SCB.<sup>5,49</sup> According to the principle of stress concentration, a structure with high stiffness will bear a greater load. Therefore, BPS and its surrounding bony structures will be subjected to higher stress than other structures. Distances between the pedicle screw to BEP and SCB were larger in the cranial side, and the tendency of stress concentration of these bony structures was relatively slight. By contrast, when shifting up the insertional screw position of the caudal pedicle screw, the screw trajectory will be placed in the SCB of the inferior vertebral body. Thus, the apparent stress concentration of SCB and BEP can be recorded. The risk of SCB’s micro-damage may be increased dramatically under this mechanical circumstance, especially when the maximum value is higher than its yield strength,<sup>19,38</sup> and the damage to SCB will lead to the loss of the subendplate mechanical support.<sup>45,50</sup> This pathological change may be an even more important cause of cage subsidence in contrast to the damage to BEPs.<sup>36,37</sup> By contrast, although stress concentration can also be observed in BEP, considering the maximum stress was still less than the yield stress of BEP (110 Mpa),<sup>19,38</sup> the risk of BPE fracture may be increased slightly not (Figure 7).

The current study still faces some limitations. Firstly, we set ligaments as cable elements in current FE models. The mechanical effect of cables can only be acted on artificially selected positions rather than their original surfaces. Additionally, fibrosis scar tissues around the excision of the annulus and its mechanical effect were omitted in this study. Although the resulting potential risk of mechanical indicator distortions should be considered, we still believe that the computational results in this study are reliable for the following reasons. Firstly, the model construction strategy with these limitations was widely accepted.<sup>18,28,51</sup> Meanwhile, no attach positions of cable elements are defined on structures whose mechanical indicators were recorded in this study. Thus, even if there is computational distortion, it can be



**Figure 6** The relation between insertional screw positions and threads-bone integration. No matter which direction the insertional screw position shifts from the middle position of the pedicle, the length of screws that do not insert into the vertebral body will be increased.



**Figure 7** Nephograms of stress distribution on BEP under flexion loading condition.

excluded from the indicator's computation. More significantly, the intact model used in this study was validated and accepted in the studies mentioned above. These limitations should be optimized in our future in-silico studies.

Based on these computational results, we believe that the reduction of fixation length by changing insertional positions of BPS was not recommended for the insertional screw strategy and may increase the risk of segmental instability, non-union, and cage subsidence. Conclusions based on the biomechanical change should be revalidated in our further clinical studies.

## Conclusions

Adjusting the insertional position of BPS close to the surgical segment in OLIF models will lead to stress concentration of bony structures and surgical segmental instability. Therefore, reducing BPS's fixation length was not recommended, which may increase the risk of segmental instability, non-union, and cage subsidence. However, these conclusions should be validated in our future clinical studies.

## Abbreviations

BEP, bony endplate; BPS, bilateral pedicle screw; CEP, cartilage endplate; DC, disc compression; IDP, intradiscal pressure; IVD, intervertebral disc; LDD, lumbar degenerative diseases; LIF, lumbar interbody fusion; OLIF, oblique lumbar interbody fusion; ROM, range of motions; SCB, subendplate cancellous bone; SED, strain energy density.

## Data Sharing Statement

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

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## Author Contributions

All authors made a significant contribution to the work reported, whether that is in the conception, study design, execution, acquisition of data, analysis and interpretation, or in all these areas; took part in drafting, revising or critically reviewing the article; gave final approval of the version to be published; have agreed on the journal to which the article has been submitted; and agree to be accountable for all aspects of the work:

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Model construction and finite element analysis: Jingchi Li, Chen Xu and Zhongxin Fang;

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## Disclosure

The authors declare that they have no competing interests.

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