Biomechanical Evaluation of strategies for Adjacent Segment Disease after Lateral lumbar interbody fusion: is extension of pedicle screw necessary?

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Abstract

Background: Adjacent segment disease (ASD) is a well-known complication after interbody fusion. Pedicle screw-rod revision possessed sufficient strength and rigidity. However, is a surgical segment with rigid fixation necessary for ASD reoperation? This study aimed to investigate the biomechanical effect on LLIF with different instrumentation for ASD treatment.Methods: A validated L2~5 finite element (FE) model was modified to simulate. ASD was considered the level cranial to the upper-instrumented segment(L3/4). Bonegraft fusion in LLIF with bilateral pedicle screw fixation (BPS) has occurred at the L4/5. The ASD segment for each group was underwent a) LLIF + posterior extension of BPS, b) PLIF + posterior extension of BPS, c) LLIF + lateral screw, d) Stand-alone LLIF. L3/4 Range of motion (ROM), interbody cage stress and strain, screw-boneinterface stress, cage-endplate interface stress, and L2/3 nucleus pulposus of intradiscal pressure (NP-IDP) analysis were calculated for the comparisons among fourmodels.Results: All reconstructive models displayed decreased motion at L3/4. In each loading condition, difference was not significant between model a and b, which providedthe maximum ROM reduction (73.8% to 97.7%, 68.3% to 98.4%, respectively). Model c also provided a significant ROM reduction (64.9% to 77.5%). Model d provided a minimal restriction of ROM (18.3% to 90.1%), which exceeded that of model a by 13.1 times in flexion-extension, 10.3 times in lateral bending and 4.8 times in rotation. Model b generated greater cage stress than other models, particularly in flexion. The maximum displacement of the cage and the peak stress of cage-endplate interface were found to be the highest in the model d in all loading conditions. For the screw bone interface, the stress was significantly greater in lateral instrumentation than that of posterior instrumentation.Conclusions: Stand-alone LLIF is likely to have limited stability, particularly in lateral bending and axial rotation. Posterior extension of BPS can provide the reliablystability and excellently protective effect on instrumentation and endplate. However, LLIF with in situ screw may be an alternative for ASD reoperation.

Background

Lumbar degenerative disease (LDD) is one of the most common reasons of dysfunction and decline in quality of life in elderly people[1]. Interbody fusion surgery for unstable spinal segments of LDD is currently the gold standard operative treatment. To achieve effective fusion with an interbody cage, supplemented internal fixation is often used. As pedicle screw-rod instrumentation becomes more widespread, spine surgeons are inevitably faced with a growing number of patients presenting with symptomatic adjacent segment degeneration (ASD)[2-5]. The symptomatic ASD ranged from 5.2% to 18.5% as reported by Park et al[6]. Ghiselli et al. reported the rate of symptomatic ASD following either decompression or fusion was predicted to be 16.5% at 5 years and 36.1% at 10 years[7]. Although the predisposing factors for developing adjacent segment problems after spinal fusion are largely unknown, altered biomechanics of the adjacent segments has been emphasized. In 2014, Kyaw et al. utilizing 10 cadaveric boars' spines at the L2–L5 levels, evaluated the biomechanical impact of pedicle screws on ASD in the lumbar spine[8]. The loss of ROM of the fusion segments led to greater torque applied to adjacent levels, and this then contributed to further degenerative changes in the disc. In current ASD

treatment strategies, traditional approach is to extend the previous screws-rods structure through the posterior approach[9, 10].

To this day, few biomechanical studies examine the ASD occurrence after lateral lumbar interbody fusion (LLIF), which has been developed more than a decade[11]. When ASD occurs in upper segment while the bone graft has successful spinal fusion in the lower segment that using the LLIF with bilateral pedicle screw fixation, how to select the surgical choice? The reoperation choice is often quite diverse and now we lack some high-quality clinical proofs for superiority of any surgical treatment. Choi et al. reported LLIF supplemented with lateral screw fixation is an alternative surgical option for ASD[12]. Louie et al. selected stand-alone LLIF to treat symptomatic ASD[13]. They considered segmental and regional lordosis, as well as intervertebral disc height were improved and remained stable by the surgery. Because the posterior spinal structure and pedicles were preserved, these two surgical techniques may not hamper further surgery. However, using a lateral surgical approach required reopening the previous scar, leading to a prolonged operative time. Hence, a posterior lumbar interbody fusion (PLIF) procedure will also be selected for ASD treatment after LLIF surgery.

Therefore, the aim of this work is to explore renovation strategy in LLIF surgery and determine the mechanical parameters of several lateral-based constructs and posterior construct for ASD. To our knowledge, no study has analyzed the biomechanics of ASD following LLIF using finite-element analysis (FEA), which is well suited to physical parameter studies and allows the determination of many more values than an experimental study. It was hypothesized that a stand-alone LLIF would not provide adequate stability in the upper segment, but when adding the supplementary instrumentation would have comparable stability. Moreover, the supplementary instrumentation can reduce the stress loads on the cage device and endplate structure.

Methods

A three-dimensional FE model of the L2-5 lumbar spine was constructed in this study (Figure 1). The image data was obtained from 1-mm-thick computerized tomography (CT) scans of a male volunteer. The 3D geometry structure was constructed by using Mimics (version 19.0; Materialise Inc., Leuven, Belgium), which transformed the dicom format image into a digital model. The model was smoothed, amended and spherized by Geomagic Studio (version 2015; Geomagic, SC, U.S.A.). The cortical bone, cancellous bone, bony endplate, zygapophyseal cartilage and intervertebral disc were used to generate the solid model in the Solidworks CAD software (version 2017; SolidWorks Corp, Dassault Systèmes, Concord, MA). Bony endplate was simulated on superior and inferior surfaces of each vertebra. The gap of zygapophyseal joints was approximately simulated by CT images. The intervertebral disc was partitioned into annulus fibrosis and nucleus pulposus, which was defined to be composed of 43% of the total disc volume and located slightly posterior to the center of the disc[14]. All seven ligaments, including the anterior/posterior longitudinal ligament (ALL/PLL), ligamentum flavum (LF), interspinous ligament (ISL), supraspinous ligament (SSL), intertransverse ligament (ITL) and facet capsular ligament (FCL), were constructed in the FE model.

Then ABAQUS software (version 2016, Simulia Inc., USA) was used to set properties of the lumbar spine components. The material properties were referred to the previous literature as specified in Table 1[15-17]. The nucleus pulposus and ground substance of annulus fibrosis were modeled as a homogeneous, hyper-elastic material using the Mooney–Rivlin model[18]. Two nodes truss elements (T3D2) with non-compressible properties were assigned to fibers of the annulus fibrosis. Four reticular fiber layers were added to the ground substance at an angle between 24° and 45° [18]. The contact between adjacent facet joint surfaces was defined as coefficient of friction and set at 0.1[19]. Seven ligaments were defined as non-compressible T3D2 and different cross-sectional area (CSA). Each lumbar spine component was created mesh in ABAQUS. Mesh was conducted quality inspection and revised by using topological combination for mesh optimization. Element types and element numbers of each lumbar spine component were placed in Table 2.

Boundary and loading condition

The inferior surface of the L5 vertebra was totally constrained in all directions, and the loading condition was applied on the superior surface of the L2 vertebra. Utilizing a similar approach to Chen, et al[20] and Zhong, et al[21], a 150 N axial compressive pre-load was set and pure moment of 10 N-m was applied to simulate the model in six directions: (1) flexion (Flx); (2) extension (Ext); (3) left bending (LB); (4) right bending (RB); (5) left rotation (LR); (6) right rotation (RR). Applied load in this study was deemed to be sufficient to generate maximum physiological motion, but small enough not to harm the specimens according to previous researches[16, 20, 22]. ABAQUS 2016 software was used for these analyses.

FE model validation

The intact L2–L5 FE model was compared to the ROM among previously published studies[20, 21]. Kinematic behavior of the FE model was verified under the conditions of flexion, extension, lateral bending, and axial rotation.

FE model with implants

The intact lumbar spine model was modified to simulate instrumented LLIF with different types of internal fixation. L2-5 lumbar spine was adjusted the loading conditions to the surface of the L2 vertebra. In each group, ASD was assumed to occur at the segment cranial to the upper-instrumentation (L3/4). Successful bone graft fusion with LLIF + bilateral pedicle screw fixation (LLIF+BPS) was simulated at L4/5. The ASD segment for each group was underwent a) LLIF + posterior extension of bilateral pedicle screw, b) PLIF + posterior extension of bilateral pedicle screw, c) LLIF + lateral screw, d) Stand-alone LLIF. In the ASD model, nuclear pulposus and lateral annulus fibrosis resection were performed at L4/5 segment and subsequent insertion of a lateral cage with bilateral pedicle screw fixation. At L3/4 segment, model a, c, and d were undergone the typical L3/4 LLIF surgery with or without additional fixation. In PLIF model b, laminectomy, nuclear pulposus and posterior annulus fibrosis resection were performed at L3/4, with a posterior cage and bilateral pedicle screw fixation (Figure 2-4). The rest of L2-5 element components were preserved.

Lateral-inserted cage (48 mm length, 22mm width, 9 mm height) was box-shaped, with 8-degree incline between superior and inferior surfaces (DePuy Synthes Spine, Inc, Raynham, MA). Posterior-inserted cage (23 mm length, 10mm width, 9 mm height) was placed in the PLIF model (DePuy Synthes Spine, Inc, Raynham, MA). Two kinds of cages were centered on the middle sagittal plane in the disc space. Three simulated constructs were adopted for internal fixation except the model d (Figure 4). The internal fixation and cage implants were reconstructed in Solidworks CAD software and fitted closely to the vertebral and endplate structure. In these ASD models, the diameter of the pedicle screws was 6.0 mm, and the lengths of the screws were set to reach the anterior or lateral cortex of the vertebral body. All screws were fixed to the vertebral bodies without allowing relative motion, which were assigned the contact surfaces to be tied in ABAQUS software. The rods connecting the screws were selected for lofting and reconstruction to ensure the exact fit. Pedicle screws and rods were defined as using a “Tie” constraint at the interfaces. A finite sliding algorithm with a coefficient of friction of 0.2 was defined between the cage and end plate to allow for any small relative displacements between the two contacting surfaces. Titanium (E=110 GPa) and polyetheretherketone (E=3.6 GPa) material properties were defined for the posterior/lateral configuration and interbody cages[23].

Analysis

L3/4 Range of motion (ROM), interbody cage stress (von Mises stress) and strain (mm), screw-bone interface stress, cage-endplate interface stress, and L2/3 nucleus pulposus of intradiscal pressure (NP-IDP) analysis were tracked and calculated for the comparisons among four models.

Results

Model validation

ROM data of the intact lumbar spine were compared to the results of the previous studies, which were under the act of the same load as listed in Figure 5. The ROM tendency of each segment was closely correlated with the results of Chen, et al[20]and Zhong, et al[21]. In flexion, the maximum ROM took place at L4-5 and the maximum in extension and bending were respectively seen at L3-4 and L4-L5. The mean values in torsion were under 3°. The ROM of the L2-L5 segments were 11.2°, 10.9°, 12.0°, and 7.1°in flexion, extension, bending, and torsion, respectively. Overall, the ROM discrepancy was within the acceptable range of error. The results of our study confirm the rationality of the model and can be further analyzed.

Range of motion

In Fig 6, there was a significant reduction in ROM at L3/4 for model a, b and c when compared with the intact model for all loading conditions. Model d slightly decreased ROM in axial rotation and lateral bending. The supplemented fixation device provided the additional fixed effect on the fusion segment.

Differences in the ROM between model a and b were not significant at less than 1 degree for all loading conditions. The ROM of each instrumented model was shown in more detail in Fig 7.

Flexion-extension

In Fig 7, there were no ROM differences in flexion among four models (90.1% to 98.8% restriction). In extension, model a and model b provided similar stability (97.7% and 98.4% restriction, respectively) when compared with the intact model. Model c reduced 77.5% ROM of the intact model, which was 9.8 times greater than that of model a. Model d reduced the lowest ROM (65.3% restriction), which was less restrictive than that of the model a (15.1 times).

Lateral bending

Model a and model b provided the largest reduction of ROM, by 95.7% and 94.5% restriction in lateral bending, compared with the intact model. Model c presented less than 30% of intact ROM (76.3% restriction). Like flexion-extension, model d reduced the lowest ROM (55.9% restriction), which was 10.3 times greater than that of model a.

Axial rotation

Compared with the intact model, the largest reduction of axial rotation ROM was found in model a. But there was no significant difference in the ROM observed within model a, b and c (73.8%, 68.3%, 64.9% restriction, respectively). Significant differences were found in model d, which merely provided 18.3% ROM restriction when compared with the intact model. In addition, axial rotation ROM was the least restricted mode of kinematic behavior.

The magnitudes of the maximum Von Mises stress in interbody cage

The maximum Von Mises stress in the interbody cage was displayed in Figure 8. In all loading conditions, the stress of the cage was found to be largest in model b. In flexion, the maximum stress of the cage reached 172.6 MPa in the model b, which was significantly increased maximum stress compared with other models. The cage stress in the model b was 13.2, 6.1, and 6.7 times greater than that of model a, c and d in flexion, respectively. Similarly, the peak stress in the model b was 4.8 and 2.3 times greater than that of model a and c in the lateral bending and 2.0 and 1.5 times greater than that of model a and c in the axial rotation. Difference was not significant between model b and d in lateral bending and axial rotation.

The magnitudes of the maximum Von Mises stress on interbody cage-L4 superior endplate interface

In all loading conditions, model d generated the largest endplate stress among implanted models (Fig 9). However, in flexion, the maximum stress caused by the model b exceeded model a, c and d by 3.9, 2.3 and 1.6 times, respectively. The stress in model a and c was 40.9% and 68.9% of that compared with model d. In lateral bending, the maximum endplate stresses caused by the model d exceeded the model a, b, and c

by 2.6, 3.0, 5.4 times in left bending and 2.7, 2.5, 1.7 times in right bending. In axial rotation, the largest stress on the pedicle screw was found in the model b, which exceed model a, c and d by 1.7, 1.8 and 1.3 times, respectively.

The maximum displacement(mm) in interbody cage

For interbody cages without supplementary fixation, the maximum displacement of the cage was found to be high in the model d in all loading conditions (Fig 10). In flexion, the displacement caused by the model d exceeded the model a, b and c by 121.3%, 116.8%, and 116.8%, respectively. Greater differences could be seen in lateral bending, the displacement caused by the model d exceeded the model a, b and c by 173.8%, 225.8%, and 166.3%. In extension and axial rotation, model d was slightly higher than that of other models, but the difference was not significant.

The magnitudes of the maximum Von Mises stress on screw-bone interface

The stress peak of the screw-bone interface is investigated to show the load distribution between the vertebrae and the spinal implants. It is important to evaluate the risk of screw loosening and migration[24]. Fig. 11 summarizes maximum Von Mises stress of the screw-bone interface for implanted models. At L3/4 segment, the stress was greater in lateral instrumentation than that of posterior instrumentation in all loading conditions. In flexion-extension, the stress in the model c was 5.7 and 5.1 times greater than that of model a and model b. The largest stress of screw-bone interface was found to be 617.5 MPa in the model c in axial rotation, which exceeded the model a and model b by 4.1 and 3.4 times. Greater differences could be seen in lateral bending, the stress caused by the model c was 7.0 and 6.1 times greater than that of model a and model b. Besides, the stress caused by model b was slightly higher than that of model a in all loading conditions. Particularly in axial rotation, the difference was more than 30 MPa.

The magnitudes of the maximum Von Mises stress in NP-IDP of adjacent intervertebral disc

Fig. 12 included the maximum Von Mises stress in NP-IDP of the superior adjacent level(L2/3) for each instrumented construct. In all loading conditions, the L2/3 NP-IDP caused by four models was slightly higher than that of the intact model, but the differences were not significant.

Discussion

This study represents the first FEA to explore an existing fusion strategy in treating ASD patients with previous LLIF+BPS from a biomechanical standpoint. Compared with the intact model, four instrumented constructs in all loading conditions provided immediate postoperative stability. Although these findings are only fit in describing the static effect on mechanical behavior, they did reflect an overall trend.

When a LLIF cage was placed into the ASD segment that previous successful spinal fusion in the lower segment, stand-alone LLIF reduced ROM when compared with the un-instrumented disc of the intact model, particularly during flexion-extension. This study confirmed movement in flexion and extension did

not destabilize the cage if the anteroposterior annulus fibrosus, anterior longitudinal ligament and posterior longitudinal ligament remain intact. However, the facet joint movement remains, stand-alone LLIF is unable to effectively limit axial rotation activity. These results are supported by Laws et al. in vitro study. He reported when compared to the intact disc, the stand-alone LLIF cage provided a significant decrease in flexion-extension, lateral bending except axial rotation[25]. Some studies considered the use of a stand-alone LLIF was associated with a high risk of subsidence in up to 30% of patients[26, 27]. The lack of accessorial instrumentation leads to stress directly distributes to the surface of cage and endplate, which increase the chance of bony endplate damage and cage subsidence. Meanwhile, the movement of interbody cage was reduced when adding the supplementary instrumentation. In this study, adding the lateral instrumentation effectively reduced ROM in the lateral bending and axial rotation condition, appeared to be the effective minimally-invasive technique for the clinical application. However, due to the one column fixed, more stress was shifted to the screw. In this study, stress concentration of lateral screw fixation can be found at the screw-bone interface, which the peak stress reached 331.9 MPa in flexion-extension, 366.1 MPa in lateral bending and 617.5 MPa in axial rotation. The risk of screw loosening and breakage were potentially increased.

Previous biomechanical studies investigating lateral instrumentation further strengthened these findings. Shasti, et al. found the reduction of bending ROM would be more pronounced when supplemented with the lateral instrumentation in LLIF[28]. Zhang et al. reported the lateral plate increased stiffness in bending and axial rotation and reduced cage stress and endplate stress in all motion modes[29]. Fogel et al. also showed the lateral stabilization added in the vertebra and spinous process could achieve stiffness in all loading conditions similar to pedicle screws[30]. It was indicated that in performing LLIF, the combination of lateral instrumentation may offer an alternative. Clinically, Choi, et al. proposed LLIF combined with lateral instrumentation could be applicable for the ASD treatment. Those authors utilized LLIF and lateral screw fixation for adjacent segment stenosis of the lumbar spine. This revision method can shorten the operation time and decrease the bleeding. The radiological findings showed that the segmental angle and anterior disc height were significantly improved[12].

At present, posterior extension surgery remains the most regular strategy for ASD treatment. In this study, the posterior supplementary instrumentation provided the most biomechanically stable construct and less peak stress distribution. However, the prior surgery was LLIF with BPS. Sometimes reopening surgical scar tissue increased risk of complication, so that using a classic PLIF should be discussed in the ASD treatment. Our data showed LLIF or PLIF combined with posterior extension of bilateral pedicle screw provided the maximum reduction in ROM among all constructs at every plane of motion, ranging from 66.8 to 98.8% of the intact spine. Moreover, the results from this study demonstrate that the posterior extension of bilateral pedicle screw generated screw-bone interface stress, which reduced the risk of screw loosening. These findings reinforce previous studies' findings that bilateral rod fixation provides better structural stability in all loading modes[31, 32].

Despite the use of different cages did not affect the stability as assessed by ROM of the instrumented level based on this study, our findings demonstrated that the interbody cage stress and cage-endplate

interface stress vary with different cages. Significantly high peak stress was found in the traditional posterior cage. The peak stress in the traditional posterior cage ranged between 2.0 and 13.2 times greater than that of the lateral cage in all loading conditions. Similarly, the peak stresses of cage-L4 superior endplate interface for posterior cage was 3.9 times greater than that of lateral cage in flexion. Our results suggest that the LLIF interbody cage generates the least amount of cage-endplate interface stress. This finding is possibly because of the PLIF smaller cage surface area in contact with the endplate, in contrast to the larger area of a LLIF lateral cage. These findings are consistent with those previously published by Xu et al.[33], which used FEA to compare peak cage-endplate interface stresses for standard cage and crescent-shaped cages.

From a biomechanical standpoint, limitations inherent to excessive rigid fixation may contribute to acceleration of ASD[6, 34]. Compared with the intact model, four instrumented constructs in all loading conditions increased the adjacent segment NP-IDP, while the differences were not significant. Although these results are only capable of describing immediate effect, they did reflect an overall trend. At present, the term stability is misused. Reducing the ROM does not mean necessarily more stability. A stable system is one that does not undergo a large displacement under small perturbations. Clinically, Less than 5° ROM was considered to be the successful fusion in FDA definition[35]. Since biomechanical studies are unable to simulate the fusion process, ROM was chosen to compare. In this study, LLIF or PLIF with posterior extension of the bilateral pedicle screw were considered adequately stable, but reinforces previous studies' findings[28, 36]. However, it is worth noting that LLIF with lateral instrumentation investigated in this study could probably provide enough load sharing to allow the bone to fuse, more clinical studies are recommended.

Although previously mentioned findings in this study might be meaningful for the clinical practice, some limitations of this study need to be mentioned. Bone tissues, ligaments and implants were defined as linear-elastic material properties. Because the focus of this research is not to predict the post-yield mechanic behavior of implants, isotropic linear-elastic material models can be used to simulate the pre-yield mechanic behavior[24]. Many FEA on lumbar spine have assumed that the components of spine were linear in order to improve the calculation efficiency[23, 37-39]. The tendency of predicted results with various fixation options would not be substantially changed depending on the individual geometric model and simplified material properties. Further clinical studies evaluating the findings from this study also would be expected in the future.

Conclusions

This study indicates that stand-alone is likely to have limited stability, particularly in lateral bending and axial rotation. Posterior extension of bilateral pedicle screw can provide the reliably mechanical stability and excellently protective effect on interbody cage, screw-bone interface and cage-endplate interface. However, LLIF supplemented with lateral screw may be an alternative reoperation surgery option to treat ASD. Further clinical studies are necessary to evaluate the clinical effects of augmentation of LLIF with in situ screw fixation.

List Of Abbreviations

LDD: Lumbar degenerative disease

ASD: Adjacent segment degeneration

LLIF: Lateral lumbar interbody fusion

PLIF: Posterior lumbar interbody fusion

FEA: Finite-element analysis

CT: Computerized tomography

ALL: Anterior longitudinal ligament

PLL: Posterior longitudinal ligament

LF: Ligamentum flavum

ISL: Interspinous ligament

SSL: Supraspinous ligament

ITL: Intertransverse ligament

FCL: Facet capsular ligament

CSA: Different cross-sectional area

Flx: Flexion

Ext: Extension

LB: Left bending

RB: Right bending

LR: Left rotation

RR: Right rotation

BPS: Bilateral pedicle screw fixation

ROM: Range of motion

NP-IDP: Nucleus pulposus of intradiscal pressure

Declarations

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Availability of data and materials

The datasets generated and/or analyzed during the current study are available from the corresponding author by reasonable request.

Authors' contributions

ZYL, JCC and XBJ contributed to the conception and design of the study. JRZ, JHH, and ZYL performed the experiment. JJT and HR analyzed the data. DL and LQY revised the manuscript. ZYL and JCC played the main role in writing the manuscript. All authors read and approved the final manuscript.

Competing interests

The authors declare that they have no competing interests.

Consent for publication

Not applicable.

Ethics approval and consent to participate

No patient information is presented in this study.

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