Biomechanical evaluation of four surgical scenarios of lumbar fusion with hyperlordotic interbody cage: A finite element study

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Abstract.
BACKGROUND: Lumbar spinal fusion in the interbody space is augmented with interbody fusion cages to provide structural support while arthrodesis occurs. Subsidence is a serious complication of interbody fusion. However, the biomechanical influence of anterior longitudinal ligament (ALL) and pedicle screws on subsidence has not been fully understood.
OBJECTIVE: To investigate biomechanical effects of the hyperlordotic cages in different surgical conditions using finite element analysis.
METHODS: Four surgical finite element (FE) models were constructed by inserting 15 degree lordosis cage at the L3-L4 disc space. The four surgical conditions were ALL intact (M1), ALL resected (M2), ALL intact and bilateral pedicle screws (M3), and ALL resected and bilateral pedicle screws (M4). Follow loads were applied at the L2 vertebral body while the inferior surface of L5 was fixed. FEA was implemented to simulate the four motion modes and biomechanical properties of four fusion scenarios with hyperlordotic interbody cage were compared.
RESULTS: The range of motion (ROM) and facet joint force (FJF) at L3-L4 decreased significantly after fusion during all the motion modes. The cage stress and endplate stress at L3-L4 increased significantly after fusion during all the motion modes. The cage stress and endplate stress at L3-L4 for M3 and M4 were smaller than that for M1 and M2 during all the motion modes. The FJF at L3-L4 for M3 and M4 were smaller than that for M1 and M2 during extension, bending, and rotation.
CONCLUSIONS: ALL has little effect on the biomechanics after lumbar fusion with hyperlordotic interbody cage. The bilateral pedicle screws significantly decreased the stress in cage, stress in endplate at L3-L4, and lowered facet contact force except for flexion mode. The implication is that the supplemental bilateral pedicle screws are recommended whether or not the ALL is resected.

Keywords: Lumbar spine, biomechanics, lumbar fusion, hyperlordotic interbody cage, finite element analysis (FEA), bilateral pedicle screws, anterior longitudinal ligament (ALL), range of motion (ROM), subsidence, facet contact force

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1. Introduction

The interbody fusion has been gradually increasing in the treatment of degenerative lumbar disease [1,2]. It has been indicated that the cage with higher lordotic angle seemed to overcome some limitations of the standard cage [3,4]. Lumbar spinal fusion in the interbody space is augmented with structural interbody cage instrumentation to provide structural support while fusion occurs. Restoration of the sagittal alignment until recently involved posterior osteotomies of facets or pedicle. Recent investigations involve addition of hyperlordotic cages to the disc space from a lateral minimally invasive approach with selective release of the anterior longitudinal ligament (ALL) [4–9]. Interest regarding the influence of the ALL with hyperlordotic cages has primarily been directed at the degree of lordosis achieved with resection of the ALL. Uribe et al. [4] in 2012 and later Uribe [9] in an elegant finite element analysis study of the ALL and lordosis in 2015 showed lordosis of 20–30 degrees may be achieved with release of the ALL and increased with additional facetectomy. However subsidence is the serious complication of interbody fusion increased with the use of lordotic implants. Subsidence is defined as endplate fracture with displacement of more than 2 mm into the vertebral body. Excessive subsidence may be associated with loss of disc height, pseudarthrosis, collapse of intervertebral foramen, and progressive spinal deformity.

Although several factors such as device design and placement, vertebral trabecular and endplate bone quality, and endplate preparation are known to contribute to subsidence risk, the etiology of subsidence is not understood. The current knowledge comes from retrospective clinical and in vitro studies. The in vitro studies include range of motion, axial compression, endplate indentation, cyclical motion, and FEA. Spinal biomechanics needs a subsidence model to study the effects of various surgical conditions with lordotic interbody cage devices. The effect of sectioning the ALL, presence of bilateral pedicle screws, and damage to the adjacent vertebra and disc space has not been determined. The purpose of this study was to investigate the biomechanical properties of these surgical conditions with a hyperlordotic interbody cage using finite element method.

2. Materials and methods

CT images of lumbar spine (L1-L5) were obtained from a healthy woman (age 36 yr, height 158 cm, weight 52 kg). A total of 492 CT images with an interval of 0.7 mm were imported into the software Mimics (Materialise Inc, Leuven, Belgium) to construct the 3D geometry structure. The geometry structure consists of intact lumbar vertebrae, intervertebral disc and cartilage endplate. The meshing of geometric structure was prepared using the software Hypermesh (Altair Technologies Inc, Fremont, CA, USA). The ligaments were modeled with truss elements (T3D2). Then the mesh structure was imported into the software Abaqus (Simulia Inc, Providence, RI, USA) to simulate the completed FE model. The computer used in the simulation is ThinkStation (Lenovo, China) which is configured with 24 processors and 64 GB memory.

The FE model of intact lumbar spine L1-L5 is shown in Fig. 1. The vertebrae were divided into three parts: cortical bone, cancellous bone, and posterior bone. The intervertebral discs included nucleus pulposus and annulus fibrosus. The ligaments included anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamenta flava (LF), interspinous ligament (ISL), supraspinal ligament (SSL), intertransverse ligament (ITL), and capsular ligament (CL). The thickness of cortical bone was 1.0 mm, and the thickness of endplate was 0.5 mm. The ligaments were modeled with tension-only truss
elements (T3D2). The 3D tetrahedral elements were used to mesh the total model except for the ligaments. The intact model contained 195,533 nodes and 841,038 elements, which could effectively eliminate the influence of meshing on the accuracy of calculation.

The 15 degree lordosis cage was modeled based on Nuvasive cage (Nuvasive, Inc, San Diego, CA, USA). The material of cage was polyetheretherketone (PEEK). The pedicle screws were modeled based on EXPEDIUM 5.5 System (DePuy Synthes Spine, Inc, Raynham, MA). The pedicle screws were made from titanium alloy (Ti6Al4V). The material properties of each component are shown in Table 1 [10–18].

To validate the intact spine model, the lumbar spine (L1-L5) was chosen to calculate the ROM and the motion segment L4-L5 was chosen to calculate the compression displacement and intervertebral disc pressure (IDP). Loading methods of pure moment and pure compression were simulated. For all of the models, the interfaces of vertebrae and discs were assigned to tie constraints. The contact between the facet joints was simulated as frictionless surfaces [1,12–14]. The simulation results were compared with the experimental data from the previous literature. Firstly, three different moments (2.5 Nm, 5.0 Nm, and 7.5 Nm) were applied to the superior surface of L1 while the inferior surface of L5 was fixed. The ROM of the lumbar spine was compared with previous in vitro results [13,19]. Then, the superior surface of L4 was loaded with four preload values (100 N, 200 N, 300 N, and 400 N) as described by Berkson et al. [20]. The compression displacement and IDP of L4-L5 were compared with previous results [19–21].

To simulate the fusion surgery, the segment L2-L5 was chosen to evaluate the biomechanics changes of surgical level and adjacent levels. The 15 degree lordosis cage was inserted into the segment L3-L4 laterally. The four surgical conditions were ALL intact (M1), ALL resected (M2), ALL intact and bilateral pedicle screws (M3), and ALL resected and bilateral pedicle screws (M4). The FE models of the four surgical conditions are shown in Fig. 2. For all models, the interfaces of vertebrae and discs were assigned to tie constraints. The interfaces of vertebrae and cages were also assigned to tie constraints. The contact between the facet joints was simulated as frictionless surfaces [1,13,14]. The inferior surface
Table 1
Material properties used in the finite element models

<table>
<thead>
<tr>
<th>Components</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson ratio</th>
<th>Cross section area (mm²)</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>12,000</td>
<td>0.3</td>
<td>—</td>
<td>[10,11]</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>100</td>
<td>0.2</td>
<td>—</td>
<td>[10,11]</td>
</tr>
<tr>
<td>Posterior bone</td>
<td>3500</td>
<td>0.25</td>
<td>—</td>
<td>[10,11]</td>
</tr>
<tr>
<td>Endplate</td>
<td>4000</td>
<td>0.3</td>
<td>—</td>
<td>[12]</td>
</tr>
<tr>
<td>Annulus fibrous</td>
<td>4.2</td>
<td>0.45</td>
<td>—</td>
<td>[10,11]</td>
</tr>
<tr>
<td>Nucleus pulposus</td>
<td>1</td>
<td>0.49</td>
<td>—</td>
<td>[13,14]</td>
</tr>
<tr>
<td>ALL</td>
<td>20</td>
<td>0.3</td>
<td>63.7</td>
<td>[11,15]</td>
</tr>
<tr>
<td>PLL</td>
<td>20</td>
<td>0.3</td>
<td>20</td>
<td>[11,15]</td>
</tr>
<tr>
<td>LF</td>
<td>19.5</td>
<td>0.3</td>
<td>40</td>
<td>[11,15]</td>
</tr>
<tr>
<td>ISL</td>
<td>11.6</td>
<td>0.3</td>
<td>40</td>
<td>[11,15]</td>
</tr>
<tr>
<td>SSL</td>
<td>15</td>
<td>0.3</td>
<td>30</td>
<td>[11,15]</td>
</tr>
<tr>
<td>TL</td>
<td>58.7</td>
<td>0.3</td>
<td>3.6</td>
<td>[11,15]</td>
</tr>
<tr>
<td>CL</td>
<td>32.9</td>
<td>0.3</td>
<td>60</td>
<td>[11,15]</td>
</tr>
<tr>
<td>Cage (PEEK)</td>
<td>3500</td>
<td>0.3</td>
<td>—</td>
<td>[16,17]</td>
</tr>
<tr>
<td>Pedicle screws (TiAl)</td>
<td>110,000</td>
<td>0.3</td>
<td>—</td>
<td>[16,18]</td>
</tr>
</tbody>
</table>

(ALL, anterior longitudinal ligament; PLL, posterior longitudinal ligament; LF, ligamentum flavum; ISL, interspinous ligament; SSL, supraspinous ligament; TL, transverse ligament; CL, capsular ligament.)

of L5 was fixed in all directions. A follower load of 280 N and the moment of 7.5 Nm were applied to the superior surface of L2 as in previous study [19,23]. The follower load of 280 N corresponded to the partial body weight of a person, and the moment of 7.5 Nm simulated the movement occurred in different conditions (flexion, extension, bending, and axial rotation). Considering the symmetry of the sagittal plane, this study simulated the biomechanics of the fusion surgeries under four conditions: flexion, extension, bending-left, and rotation-left. The ROM, IDP (cage stress), endplate stress, and FJF were analyzed and exported. The intact L2-L5 model under combined loading was recalculated, so a total of 20 simulation calculations for five models and four motion modes were performed. Analysis results were in accordance with the requirements of visualization, mechanics data was expressed using Von Mises stress contours. The maximum Von Mises stresses were exported as cage stress, IDP, and endplate stress in this study.

3. Results

3.1. Model validation

Under the pure moment 7.5 Nm, the total L1-L5 ROM was consistent with the in vitro median values in previous experimental studies [13,19]. The whole movement angles were 24.75° during flexion-extension, 26.66° during lateral bending, and 18.43° during axial rotation. As is shown in Fig. 3(a), all values in different conditions were within the in vitro range. The load-deflection curves are displayed in Fig. 3(b), which was consistent with the existing results of previous studies [13,19]. The rotation angles were 15.04° during flexion, 9.71° during extension, 13.45° during lateral bending-left, and 10.33° during axial rotation-left. As is demonstrated in Fig. 3(b), there was a stiffness effect with increasing moment.
The compression-displacement curves are displayed in Fig. 3(c), which indicate that the axial displacement of L4-L5 segment increased almost linearly with the applied axial compressive loading. Compared to the previous in vitro experiment study [20], the results were reliable and reasonable. The compression-IDP curves are displayed in Fig. 3(d), which were also compared with the previous FE and in vitro experimental studies [19,21]. The predicted results demonstrated that IDP of L4-L5 segment increased almost linearly with the applied axial compressive loading.

3.2. Range of motion (ROM)

Figure 4 displays the ROM of the surgical models under the combined compression-moment. For all of the surgical models, the predicted ROM at surgical level L3-L4 decreased significantly during flexion, extension, bending, and rotation. The ROM at adjacent levels decreased during flexion while that changed very little during extension, bending, and rotation.

3.3. Intervertebral disc pressure (IDP) and cage stress

The maximum stresses at surgical level L3-L4 and IDP at adjacent levels are shown in Fig. 5. The maximum stresses in cages for M3 and M4 were smaller than that for M1 and M2. The maximum stresses in cages for M1 and M2 were similar, the maximum stresses in cages for M3 and M4 were also very similar. Compared to M1 and M2, the median stresses for M3 and M4 reduced by 12.74% during flexion,
25.52% during extension, 18.20% during bending, and 11.48% during rotation, respectively. The IDP at adjacent level L2-L3 increased significantly in extension, while that changed very little during the other modes.

3.4. Endplate stress

Figure 6 depicts the maximum stresses in endplates of all surgical models. The maximum stress in endplates at surgical level in four surgical conditions increased significantly during all motion modes. The maximum stresses in endplates for M3 and M4 were smaller than that for M1 and M2. The maximum stresses in endplates for M1 and M2 were almost equal, the maximum stresses in endplates for M3 and M4 were also very similar. Compared with M1 and M2, the median stresses for M3 and M4 reduced by 14.35% during flexion, 24.66% during extension, 18.22% during bending, and 11.11% during rotation, respectively. The stress in endplates at adjacent level L2-L3 increased significantly in extension, while there was no obvious change during flexion, bending, and rotation.
3.5. Facet joint force (FJF)

The FJF of the surgical models is displayed in Fig. 7. The FJF at surgical level in four surgical conditions decreased during all motion modes. The FJF for M3 and M4 at surgical level were smaller than that for M1 and M2 during all motion modes except for flexion. The FJF for M1 and M2 were almost the same, the FJF for M3 and M4 were also almost the same. Compared with the forces in facet joint for M1 and M2, the forces for M3 and M4 reduced by 22.51% during extension, 12.08% during bending, and 3.46% during rotation, respectively; while increased by 41.93% during flexion. In addition, the FJF of the four models at adjacent levels decreased during bending. The location of maximum facet joint force at L4 posterior bone of different surgical models is shown in Fig. 8. With changes to kinematic motion of lumbar spine in various scenarios, the location of maximum facet joint force was changed as well.

4. Discussion

In this finite element analysis with hyperlordotic interbody cage results were comparable to previous studies [24–26]. In Figs 4 and 7, the ROM and FJF decreased significantly at surgical level L3-L4 after fusion during flexion, extension, bending and rotation. In Figs 5 and 6, the cage stress and endplate stress at L3-L4 increased significantly after fusion during all the motion modes. The ROM, FJF, cage stress, and endplate stress were directly affected at L3-L4 during all the motion modes, and changed with each condition (Fig. 2). The biomechanical effect of sectioning the ALL in M1 and M2 was not significantly
different, and addition of pedicle screws in M3 and M4 did not significantly change with sectioning of the ALL. Compared to M1 and M2, M3 and M4 showed obvious advantages, such as the decreases in cage stress, endplate stress and FJF at L3-L4. Pedicle screws decreased the max stress in cage and endplate during all the motion modes, and lowered FJF except for the flexion mode, which may decrease subsidence. Bilateral pedicle screws produced greater stiffness with the 15 degree interbody cage, but the effect of ALL sectioning was very small. It can be concluded that supplemental bilateral pedicle screws were recommended in lumbar fusion with hyperlordotic interbody cage. Sectioning the ALL was not statistically important.

The main idea of lumbar interbody fusion is to stabilize the lumbar spine by reducing the ROM at surgical level. However, the previous studies have shown that there is a stress shielding effect after fusion, which may increase the risk of degeneration at adjacent levels [27–29]. Figure 4 showed that the ROM at adjacent levels decreased after fusion during flexion, and the ROM at L4-L5 increased after fusion during extension and bending. It suggested the effect of fusion on ROM during flexion mode was greater than during other modes, and the effect of fusion on ROM at L4-L5 was greater than at L2-L3. Figure 5 indicated that the IDP at adjacent levels increased after fusion during extension, while that during other modes showed similar results with the case of intact model. Figure 6 showed that the endplate stress at adjacent levels L2-L3 increased after fusion during extension, while the endplate stress at adjacent levels L4-L5 changed very little. Figure 7 indicated that the FJF at adjacent levels decreased after fusion during bending, which may be related to the change in the contact area. According to the biomechanical comparison at the adjacent levels, it could be found that the fusion had a certain effect on the adjacent segments during flexion, extension and bending, but had the least effect during rotation. In addition, this
study was based on the same load condition, so the ROM was not significantly changed at adjacent levels. However, in order to achieve the desired ROM, the patient after fusion will naturally increase the driving force, which may further accelerate the degeneration of adjacent segments.

According to the recent clinical studies, direct lateral interbody fusion has been widely performed for degenerative lumbar disease, and hyperlordotic interbody cage (12 degree) seemed to result in more disc angle and less subsidence [3]. The FE study has shown that the increased segmental lumbar lordosis could be achievable with hyperlordotic interbody cages (20 degree) [9]. Our reasoning was that the hyperlordotic cages are widely available and increase the lordosis of the disc space compared to the standard cages. In this study, 15 degree cage was chosen according to the disc angle and foraminal height of the present lumbar model. As was shown in Figs 5 and 6, the use of bilateral pedicle screws could effectively decrease the cage stress and endplate stress, while the ALL had little influence to the stresses. This means that ALL has little effect on the subsidence, so it is recommended that ALL could be removed when the fusion with hyperlordotic interbody cage is performed. The results of this study may explain the ALL resection in clinical fusion with hyperlordotic interbody cages. However, the relationship between lordotic angle of interbody cage and the subsidence requires further investigation.

There are some limitations in this study, such as using a unique lumbar model, simplifying the material properties of some tissues, and ignoring the role of the muscle. Because of the differences in sex, age and height, the geometric model of lumbar spine varies from person to person such as the intervertebral disc space and the gaps between facet joints. However, only one model of lumbar spine was used in this study. The material properties were set to be linear elastic though the components of lumbar spine are nonlinear in reality. But many studies using FEM on lumbar spine have assumed that the components
of spine was linear in order to improve the computational efficiency [11,17,23,30,31]. In addition, the model of muscles was not been considered in this study although the muscles plays an important role in supporting the lumbar spine. However, the tendency with different surgical conditions would not be significantly changed depending on the individual geometric model, material properties, and model of the muscles.

5. Conclusions

In the current study, we comprehensively compared the biomechanical influences of ALL and bilateral pedicle screws on lumbar fusion, including ROM, cage stress and IDP, endplate stress, and FJS. We clarified the effects of sectioning the ALL, presence of bilateral pedicle screws, and the damage to the adjacent discs. According to the biomechanical evaluation, it was found that the supplemental bilateral pedicle screw can affect the motion and prevent subsidence of lumbar spine noticeably while ALL have little effect on that. Compared to previous studies on lumbar spine, the similar results were reported that bilateral pedicle screw fixation provides superior biomechanical stability in transforaminal lumbar interbody fusion although the ROM was not changed significantly with different surgery conditions [27,28]. The present study may confirm previous studies that the ALL resection may provide the possibility to achieve the lordosis of 20–30 degrees with higher lordotic cages [3,4]. However, we anticipate the higher lordosis cage designs will increase the subsidence risk. In future studies, a subsidence model will be established to determine the effect of the higher lordotic cages on subsidence.
Fig. 8. Location of maximum facet joint force at L4 posterior bone of different surgical models during flexion, extension, bending, and rotation. The red arrows indicate the locations where the maximum facet joint force occurred. (M1, ALL intact; M2, ALL resected; M3, ALL intact and bilateral pedicle screws; M4, ALL resected and bilateral pedicle screws.)

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Conflict of interest

None to report.

References


