Relationship between biomechanical changes at adjacent segments and number of fused bone grafts in multilevel cervical fusions: a finite element investigation

Technical note

MOZAMMIL HUSSAIN, PH.D.,1 AHMAD NASSR, M.D.,2 RAGHU N. NATARajan, PH.D.,3,4 HOWARD S. AN, M.D.,3 AND GUNNAR B. J. ANDERSSON, M.D., PH.D.3

1Division of Research, Logan University, Chesterfield, Missouri; 2Department of Orthopedic Surgery, Mayo Clinic, Rochester, Minnesota; 3Department of Orthopedic Surgery, Rush University Medical Center, Chicago; and 4Department of Bioengineering, University of Illinois, Chicago, Illinois

Object. Biomechanical studies have shown that anterior cervical fusion construct stiffness and arthrodesis rates vary with different reconstruction techniques; however, the behavior of the adjacent segments in the setting of different procedures is poorly understood. This study was designed to investigate the adjacent-segment biomechanics after 3 different anterior cervical decompression and fusion techniques, including 3-level discectomy and fusion, 2-level corpectomy and fusion, and a corpectomy-discectomy hybrid technique. The authors hypothesized that biomechanical changes at the segments immediately superior and inferior to the multilevel fusion would be inversely proportional to the number of fused bone grafts and that these changes would be related to the type of fusion technique.

Methods. A previously validated 3D finite element model of an intact C3–T1 segment was used. Three C4–7 fusion models were built from this intact model by varying the number of bone grafts used to span the decompression: a 1-graft model (2-level corpectomy), a 2-graft model (C5 corpectomy and C6–7 discectomy), and a 3-graft model (3-level discectomy). The corpectomy and discectomy models were also previously validated and compared well with the literature findings. Range of motion, disc stresses, and posterior facet loads at the segments superior (C3–4) and inferior (C7–T1) to the fusion construct were assessed.

Results. Motion, disc stresses, and posterior facet loads generally increased at both of the adjacent segments in relation to the intact model. Greater biomechanical changes were noted in the superior C3–4 segment than in the inferior C7–T1 segment. Increasing the number of bone grafts from 1 to 2 and from 2 to 3 was associated with a lower magnitude of biomechanical changes at the adjacent segments.

Conclusions. At segments adjacent to the fusion level, biomechanical changes are not limited solely to the discs, but also propagate to the posterior facets. These changes in discs and posterior facets were found to be lower for discectomy than for corpectomy, thereby supporting the current study hypothesis of inverse relationship between the adjacent-segment variations and the number of fused bone grafts. Such changes may go on to influence the likelihood of adjacent-segment degeneration accordingly. Further studies are warranted to identify the causes and true impact of these observed changes.

Key words • anterior cervical decompression • corpectomy • discectomy • finite element • adjacent-segment biomechanics

Abbreviations used in this paper: ASD = adjacent-segment degeneration; FE = finite element.
Adjacent-segment biomechanics

disc stresses\(^1\) and facet forces,\(^4\) but the specimens in those experiments generally have some degree of degenerative changes. When using older cadaveric specimens, it still remains unclear whether biomechanical changes at the adjacent segments are the result of the fusion or of the altered mechanics of the degenerative segments. Finite element (FE) modeling can help address these issues, by allowing simulation with healthy properties at adjacent segments immediately superior and inferior to the fusion construct.

The aim of the current study is to investigate the motion patterns, disc stresses, and facet loads at the segments adjacent to a C4–7 fusion when the number of bone grafts used for the reconstruction is varied. We hypothesize that the changes in these biomechanical parameters will be inversely proportional to the number of fused bone grafts used for reconstruction and that these changes will be related to the type of fusion technique.

Methods

A CT scan obtained in a healthy 38-year-old woman was used to develop a 3D FE model of an intact C3–T1 segment (Fig. 1). Validation data of the intact model have been previously reported.\(^{18}\) The following spinal structures were included in the model: cortical bone, cancellous bone, posterior elements, annulus fibrosus, nucleus pulposus, and articulating posterior facets. Ligamentous structures were also incorporated, including the anterior longitudinal ligament, the posterior longitudinal ligament, the intransipinous ligament, the ligamentum flavum, and the capsular ligaments. The insertion points and areas of the ligaments were closely matched with available published data.\(^{23,51}\) Anisotropy imparted by the thick interwoven anterior alar fibers and thin vertically oriented posterior fibers was also modeled in the disc annulus.\(^{22}\) The elastic moduli of the anterior and posterior annulus (one-tenth of the annulus volume each) were chosen as 10% higher and 10% lower than that of the lateral annulus, respectively.

Three C4–7 fusion models were built from the intact model by varying the number of bone grafts used for the fusion purpose. The 1-graft model, a 2-level corpectomy procedure, was developed by using a single long-strut bone graft between the C-4 inferior endplate and the C-7 superior endplate. The 2-graft model, a combined corpectomy-discectomy procedure, was created by placing one bone graft between the C-4 inferior endplate and the C-6 superior endplate (a corpectomy procedure) and another bone graft at the C6–7 level, by removing the respective intervertebral disc (a discectomy procedure). The 3-graft model, a 3-level discectomy procedure, was built by replacing the discs at the C4–5, C5–6, and C6–7 levels with 3 bone grafts. The bone grafts covered up to 50% of the area of the opposing endplates with rigid bone-screw interfaces.\(^{5,47}\) The anterior and posterior longitudinal ligaments for the C4–5, C5–6, and C6–7 motion segments were removed. The fusion construct was stabilized with a locked C4–7 anterior titanium alloy plate (height 52.12 mm, width 13.12 mm, thickness 2.30 mm). Unicortical screws of 16 mm length with outer and inner diameters of 3.5 mm and 2.5 mm, respectively (mean diameter 3.00 mm) were placed parallel to the corresponding vertebral endplates with rigid bone-screw interfaces. Terminal screws at the cephalad and caudad construct ends were used in the corpectomy model, while segmental screws were used in both the hybrid and multilevel discectomy models. The material properties for the FE models were adopted from the literature (Table 1). Both corpectomy and discectomy models were previously validated and compared well with the literature findings.\(^{37}\)

The moments were chosen within the in vivo physiological range: flexion (45°)/extension (35°),\(^{7,9,14,25,26,37,40}\) axial rotation on each side (20°),\(^{20,28,38}\) and lateral bending on each side (25°).\(^{21,37}\) Next, these moment loads in 3 planes were applied to the C-3 vertebra under a compressive preload of 73.6 N.\(^{11}\) Two isotropic truss elements connecting the midlateral sides of the vertebral bodies were used to apply the compressive load, a follower-load technique.\(^{36}\) The inferior surface of the T-1 vertebra was fixed. For the 3 instrumented fusion models, range of motion, von Mises disc stresses, and facet loads were recorded at the segments immediately superior and inferior to the fused construct, at C3–4 and C7–T1, respectively. Range of motion was also calculated for the instrumented C4–7 fused segment. Facet loads were computed by averaging the resultant loads on the right and left articulating facets. The analysis was performed using commercially available FE code ADINA software (ADINA R & D, Inc.).

Results

Increasing the number of bone grafts for reconstruction was associated with a decrease in the range of motion at the instrumented level (Fig. 2) and a corresponding increase at the adjacent segments (Fig. 3). Relative to the intact model, the increase in motion at the adjacent segments was as follows. In the 1-graft model, flexion increased by 144% at C3–4 and 131% at C7–T1, extension increased by 109% at C3–4 and 65% at C7–T1, axial rotation increased by 124% at C3–4 and 83% at C7–T1, and lateral bending increased by 189% at C3–4 and 67% at C7–T1. In the 2-graft model, flexion increase was 142% at C3–4 and 127% at C7–T1, extension increase by 107% at C3–4 and 63% at C7–T1, axial rotation increase by 181% at C3–4 and 14% at C7–T1. In the 3-graft model, flexion increased by 140% at C3–4 and 125% at C7–T1, extension increased by 102% at C3–4 and 63% at C7–T1, axial rotation increase was 103% at C3–4 and 36% at C7–T1, and lateral bending increased by 179% at C3–4 and 5% at C7–T1. Superior C3–4 motion was greatest during lateral bending, while inferior C7–T1 motion was most affected during flexion.

Stresses in adjacent-segment discs were found to be greatest in the 1-graft model, followed by the 2-graft model, and least in the 3-graft model (Fig. 4). Compared with the disc stresses in the intact model, the disc stresses in the instrumentation models generally increased at the segments adjacent to the fusion construct. In the 1-graft model, the disc stress during flexion, relative to the intact model, increased by 150% at C3–4 and 142% at C7–T1; during extension, it increased by 163% at C3–4 and 135%
at C7–T1; during axial rotation, it increased by 109% at C3–4 and 92% at C7–T1; and during lateral bending, it increased by 115% at C3–4 and 89% at C7–T1. In the 2-graft model, during flexion, the disc stress increased by 136% at C3–4 and 121% at C7–T1; during extension, it increased by 142% at C3–4 and 116% at C7–T1; during axial rotation, it increased by 87% at C3–4 and 53% at C7–T1; and during lateral bending, it increased by 113% at C3–4 and 76% at C7–T1. In the 3-graft model, during flexion, the disc stress increased by 94% at C3–4 and 88% at C7–T1; during extension, it increased by 108% at C3–4 and 106% at C7–T1; during axial rotation, it increased by 75% at C3–4 and 53% at C7–T1; and during lateral bending, it increased by 96% at C3–4 and 76% at C7–T1. Stress changes were greater in the superior C3–4 disc than in the inferior C7–T1 disc. The highest disc stress was observed during extension.

Facet loads were found to be greatest in the 1-graft model, followed by the 2-graft model, and least in the 3-graft model (Fig. 5). In flexion, since the facets slide against each other and resist mainly anterior shear, negligible facet loads were observed in the axial direction. Compared with the intact model, the facet loads in the instrumentation models increased at the segments adjacent to the fusion construct. In the 1-graft model, the facet load during extension, compared with the intact model, increased by 556% at C3–4 and 296% at C7–T1; during axial rotation, it increased by 239% at C3–4 and 185% at C7–T1; and during lateral bending, it increased by 199% at C3–4 and 135% at C7–T1. In the 2-graft model, the facet load during extension increased by 524% at C3–4 and 280% at C7–T1; during axial rotation, it increased by 203% at C3–4 and 87% at C7–T1; and during lateral bending, it increased by 197% at C3–4 and 51% at C7–T1. In the 3-graft model, the facet load during extension increased by 503% at C3–4 and 268% at C7–T1; during axial rotation, it increased by 161% at C3–4 and 25% at C7–T1; and during lateral bending, it increased by 147% at C3–4 and 40% at C7–T1. Facet load changes were greater in the superior C3–4 segment than in the inferior C7–T1 segment. As predicted, facet loads were greatest during extension.
Adjacent-segment biomechanics

Discussion

The present FE study addressed the load-sharing mechanics between the disc and the posterior facets at normal segments adjacent to an anterior instrumented C4–7 fusion using 3 different reconstruction techniques—corpectomy, discectomy, and combined corpectomy-discectomy.

At adjacent segments, biomechanical changes in discs and posterior facets were reduced in the multilevel discectomy (3 fused bone grafts) compared with the corpectomy (1 fused bone graft), thereby supporting the current study hypothesis of inverse biomechanical relationship between the adjacent-segment variations and the number of fused bone grafts. These biomechanical changes may go on to

TABLE 1: Material properties of the spinal components*

<table>
<thead>
<tr>
<th>Description</th>
<th>Element Type</th>
<th>Young’s Modulus (MPa) E</th>
<th>Poisson’s Ratio ν</th>
<th>Authors &amp; Year</th>
</tr>
</thead>
<tbody>
<tr>
<td>cortical bone</td>
<td>3D solid (4 node)</td>
<td>10,000.00</td>
<td>0.29</td>
<td>Argoubi &amp; Shirazi-Adl, 1996; Lee et al., 2000; Teo &amp; Ng, 2001</td>
</tr>
<tr>
<td>cancellous bone</td>
<td>3D solid (4 node)</td>
<td>100.00</td>
<td>0.29</td>
<td>—</td>
</tr>
<tr>
<td>posterior elements</td>
<td>3D solid (4 node)</td>
<td>3,500.00</td>
<td>0.29</td>
<td>—</td>
</tr>
<tr>
<td>endplate</td>
<td>3D solid (4 node)</td>
<td>500.00</td>
<td>0.40</td>
<td>—</td>
</tr>
<tr>
<td>annulus fibrosus</td>
<td>3D solid (4 node)</td>
<td>2.50</td>
<td>0.40</td>
<td>—</td>
</tr>
<tr>
<td>nucleus pulposus</td>
<td>3D solid (4 node)</td>
<td>1.50</td>
<td>0.49</td>
<td>—</td>
</tr>
<tr>
<td>facet cartilage</td>
<td>3D solid (4 node)</td>
<td>10.40</td>
<td>0.40</td>
<td>Kumaresan et al., 1998</td>
</tr>
<tr>
<td>ALL</td>
<td>3D tension truss (2 node)</td>
<td>15 (ε&lt;12%), 30 (ε&gt;12%)</td>
<td>—</td>
<td>Goel &amp; Clausen, 1998</td>
</tr>
<tr>
<td>PLL</td>
<td>3D tension truss (2 node)</td>
<td>10 (ε&lt;12%), 20 (ε&gt;12%)</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>interspinous ligament</td>
<td>3D tension truss (2 node)</td>
<td>2 (ε&lt;40%), 8 (ε&gt;40%)</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>ligamentum flavum</td>
<td>3D tension truss (2 node)</td>
<td>5 (ε&lt;25%), 10 (ε&gt;25%)</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>capsular ligaments</td>
<td>3D tension truss (2 node)</td>
<td>7 (ε&lt;12%), 30 (ε&gt;12%)</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>bone graft</td>
<td>3D solid (4 node)</td>
<td>3,500.00</td>
<td>0.30</td>
<td>Natarajan et al., 2000; Rohlmann et al., 2006; Zander et al., 2002</td>
</tr>
<tr>
<td>anterior plate</td>
<td>3D solid (4 node)</td>
<td>110,000.00</td>
<td>0.30</td>
<td>Ratner et al., 1996</td>
</tr>
<tr>
<td>anterior screws</td>
<td>3D solid (4 node)</td>
<td>110,000.00</td>
<td>0.30</td>
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</table>


Fig. 2. Range of motion of the C4–7 corpectomy construct.
influence the likelihood of ASD, and our findings support
the hypothesis that the adjacent-segment biomechanics is
dependent on different types of anterior decompression
and fusion constructs. The differences in biomechanical
changes observed with corpectomy and discectomy may
be due to inherent differences in these constructs, with
different bone graft heights used in corpectomy and dis-
cectomy causing changes in spinal lordosis, which may
further affect adjacent disc behavior.

The current findings show an increase in the stresses

![Graph 1: Range of motion](image1)

**Fig. 3.** Range of motion of the segments immediately superior (C3–4) and inferior (C7–T1) to the C4–7 corpectomy con-
struct.

![Graph 2: Disc stresses](image2)

**Fig. 4.** Stresses inside the superior C3–4 and inferior C7–T1 discs following a C4–7 corpectomy fusion with 3 instrumentation
techniques.
Adjacent-segment biomechanics

![Facet Load Graph](image)

**Fig. 5.** Loads on posterior facets at the superior C3–4 and inferior C7–T1 segments following a C4–7 corpectomy fusion with 3 instrumentation techniques.

inside the discs above and below the level of corpectomy and/or discectomy and fusion. Lopez-Espina et al.26 investigated 1- and 2-level discectomy and fusions with stand-alone bone grafts using a FE model, and demonstrated an increase in adjacent-level disc stress in flexion, axial rotation, and lateral bending, but not in extension. We have shown a step further that the biomechanical changes at adjacent segments are not limited solely to the discs, but also propagate to the posterior facets. Chang et al.4 measured both intradisc pressure and facet forces after C6–7 anterior cervical discectomy and fusion at C6–7. They noted a significant increase in both intradisc pressure and facet forces at the adjacent segments. The intradisc pressure was highest in flexion at the superior adjacent segment. The facet loads were minimal in flexion and highest in extension, as seen in our models. Negligible contact between the posterior facets in flexion may be the reason why the model demonstrated the highest disc stress during this motion.30 It is still unclear whether high facet forces at adjacent segments are the result of intermediate fused segments and/or increased adjacent disc stresses. This remains to be ascertained. Consistent with other results reported in the literature,3,4,6,10,50 the current study noted higher biomechanical changes above the fused segments. A few biomechanical investigations have also shown greater disc stress (or pressure) and motion changes at the inferior levels than at the superior levels in 1- and 2-level cervical fusions,26,30 whereas no significant22,41,43 or even decreased45 biomechanical changes have been recorded at the adjacent segments after 1-level cervical arthrodesis.

Given these mixed findings, what causes ASD progression and why higher changes are observed at the superior level and not at the inferior level (or at the inferior level and not at the superior level) remains poorly understood. Of course, one of the probable reasons is the type of fixation technique—anterior and/or posterior. In our previous investigations,15,16 we supplemented the anterior fixation with the posterior segmental fixation (lateral mass screws with rods). We found that the use of posterior fixation, whether alone or in combination with anterior fixation, confers comparable stability and decreases the chance of hardware failure at the expense of likely affecting the kinematics and biomechanics of the adjacent segments accordingly. Which segment(s) of the cervical spine is fused may also play a critical role in superior or inferior progression of ASD. A higher risk of developing ASD is reported when fusion ends at the C-5 or C-6 level.12

Results and conclusions drawn from the current FE study are not without limitations. As FE model development from a CT scan is very time consuming, only one CT scan (obtained in a healthy 38-year-old woman) was used to build the intact C3–T1 model. Furthermore, instrumentation models are developed by modifying intact models and not by analyzing the CT scans of corresponding surgical patients. Given that spinal fusion procedures are also performed in older patients and that the current FE models are based on a healthy individual, generalizing our data to the population of males and females in different age groups is not recommended. Further, the fact that a single CT scan of a normal spine was used does not allow for examination of varying degrees of lordosis. The multilevel fusion constructs examined also limit the choices of levels to model. Future studies should examine 1-, 2-, and 3-level fusions at different segment levels (and...
hence different points along the lordotic curve) that might help explain the observed findings in light of different degrees of lordosis.

Simple geometries of the bone grafts, screws, and plates are adopted. Also, the bone grafts are rigidly fixed to the endplates in this study, representing a complete bony fusion as seen during the late postoperative stage; therefore, the current study conclusions should not be applied for the immediate postoperative stage. Furthermore, the surgical FE models are not simulated with the clinical steps performed during the surgery, such as endplate preparation and slight vertebral distraction applied to insert and accommodate the grafts, and they are subjected to the same loads as the intact healthy FE model.

The range of motion of the cervical spine is affected by the neck muscles, which are not incorporated into the current FE models. Disc mechanics is further affected by annular fibers, which are absent in the current analysis. Rather, fiber anisotropy in the disc is modeled by varying the regional tissue elasticity. Moreover, the facet loads in this study may not completely represent the behavior of the facets on any given side, especially in axial rotation and lateral bending, due to the averaging of left and right facet loads. Last but not the least, FE modeling, in large part, depends on multiple parameters that are entered into the model. These can include tissue material properties of the bone, disc, and ligaments that have to be simplified to allow for reasonable numerical calculations to be performed. We used known tissue material properties (homogeneous and isotropic) from the literature, which have been shown to be closely correlated with biomechanical results. This simplification often does not mimic the complex behavior of the biological tissues involved. Additionally, FE models may not well mimic the extremes of motion or stresses, as these conditions may generate a different set of material properties from the same tissues, especially those conditions that are closer to the failure loads of these tissues.

Conclusions

While there are several limitations in the current FE models, they still provide thoughtful and reasonable preliminary ways to study questions about the biomechanical effects of surgical techniques that are not easily answered with other tools. The current FE models have clearly allowed us to investigate the theoretical changes in both adjacent-level disc pressures and posterior facet loads, which are otherwise quite challenging to measure. While it was not possible to validate all current surgical FE models due to unavailable literature data, we have previously shown an appropriate level of fidelity of these surgical FE models that predicted the biomechanical results of the intact cervical spine, as well as the cervical spine subjected to anterior and/or posterior fixation(s) after corpectomy and/or discectomy fusion(s). Considering these study limitations, further studies are warranted to identify the cause and true impact of the current biomechanical outcomes.

Disclosure

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Author contributions to the study and manuscript preparation include the following. Conception and design: all authors. Acquisition of data: Hussain. Analysis and interpretation of data: Hussain, Nassr, Natarajan. Drafting the article: Hussain, Nassr. Critically revising the article: all authors. Reviewed submitted version of manuscript: all authors. Approved the final version of the manuscript on behalf of all authors: Hussain.

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Address correspondence to: Mozammil Hussain, Ph.D., Logan University, 1851 Schoettler Rd., Chesterfield, MO 63017. email: mozammil.hussain@logan.edu.