Comparison of the load-sharing characteristics between pedicle-based dynamic and rigid rod devices

Yoon-Ho Ahn¹, Wen-Ming Chen², Kwon-Yong Lee³, Kyung-Woo Park⁴ and Sung-Jae Lee⁵

¹ Korea Orthopedics and Rehabilitation Engineering Center, Incheon 403712, Korea
² Division of Bioengineering, National University of Singapore, Singapore 117576, Singapore
³ Department of Mechanical Engineering, Sejong University, Seoul 143747, Korea
⁴ Department of Neurosurgery, Kwang-Hye Spine Hospital, Seoul 135280, Korea
⁵ Department of Biomedical Engineering, Inje University, Gimhae 621749, Korea

E-mail: sjl@bme.inje.ac.kr

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Abstract
Recently, numerous types of posterior dynamic stabilization (PDS) devices have been introduced as an alternative to the fusion devices for the surgical treatment of degenerative lumbar spine. It is hypothesized that the use of ‘compliant’ materials such as Nitinol (Ni–Ti alloy, elastic modulus = 75 GPa) or polyether-etherketone (PEEK, elastic modulus = 3.2 GPa) in PDS can restore stability of the lumbar spine without adverse stress-shielding effects that have been often found with ‘rigid’ fusion devices made of ‘rigid’ Ti alloys (elastic modulus = 114 GPa). Previous studies have shown that suitably designed PDS devices made of more compliant material may be able to help retain kinematic behavior of the normal spine with optimal load sharing between the anterior and posterior spinal elements. However, only a few studies on its biomechanical efficacies are available. In this study, we conducted a finite-element (FE) study to investigate changes in load-sharing characteristics of PDS devices. The implanted models were constructed after modifying the previously validated intact model of L3-4 spine. Posterior lumbar fusion with three different types of pedicle screw systems was simulated: a conventional rigid fixation system (Ti6Al4V, Φ = 6.0 mm) and two kinds of PDS devices (one with Nitinol rod with a three-coiled turn manner, Φ = 4.0 mm; the other with PEEK rod with a uniform cylindrical shape, Φ = 6.0). To simulate the load on the lumbar spine in a neutral posture, the axial compressive load (400 N) was applied. Subsequently, the changes in load-sharing characteristics and stresses were investigated. When the compressive load was applied on the implanted models (Nitinol rod, PEEK rod, Ti-alloy rod), the predicted axial compressive loads transmitted through the devices were 141.8 N, 109.8 N and 266.8 N, respectively. Axial forces across the PDS devices (Nitinol rod, PEEK rod) and rigid system (Ti-alloy rod) with facet joints predicted to take over 41%, 33% and 71% of the applied compression load, respectively. Our results confirmed the hypothesis on the PDS devices by showing the substantial reduction in stress-shielding characteristics. Higher axial load was noted across the anterior structure with the PDS devices, which could slow the degeneration process of bony structures and lower the possibility of implant failure.

(Some figures in this article are in colour only in the electronic version)
1. Introduction

‘Rigid’ metal alloys such as titanium alloy and stainless steels have been used widely as biomaterials for spinal fixation systems (e.g. pedicle screw system) partly due to their superior mechanical strength and fixation capabilities [1–4]. However, several reports in the literature have shown that rigid spinal systems may cause abnormal changes in the load transfer and stresses that may lead to degeneration of the intervertebral disc and bony structures [5–8].

Recently, more ‘compliant’ spinal stabilization systems have been introduced for the lumbar spine to avoid these adverse effects. They are clinically called ‘dynamic’ stabilization systems (PDS) [9–11]. Studies have shown that devices made of more compliant materials were able to provide flexibility of a motion segment and induce more uniform load distribution across the spine than conventional rigid stabilization devices [9, 10]. They are designed to maintain or restore the intersegmental motions and load transfer of the intact spine so that they have less negative effects on the segments adjacent to the stabilized one. Recent studies have suggested that dynamic stabilization may be able to provide better surgical alternatives to the conventional fusion for the patients with chronic low back pain [9, 10]. As a result, various types of PDS devices have been developed for the lumbar spine. They employ more ‘compliant’ materials (such as Nitinol, polymers or PEEK (polyether-etherketone)) or design changes (coiled or twisted rods instead of straight rods). It is believed that they can maintain or restore the load-sharing characteristics to the level of the intact spine with less stress shielding [10, 12]. In this study, we created a finite-element (FE) model of the lumbar spine to investigate changes in load-sharing characteristics of PDS devices.

2. Materials and methods

2.1. Intact finite-element model

A three-dimensional, intact osteoligamentous lumbar spinal segment L3-L4 finite-element (FE) model was used as the baseline case (figure 1); its geometry was reconstructed from computer tomography (CT) scans of a 44-year-old male with no pathologies, while the material properties were selected from the published literatures (table 1) [13–17]. The detailed FE model included vertebral bodies, bony posterior elements, intervertebral discs and seven major groups of ligaments: anterior longitudinal (ALL), posterior longitudinal (PLL), ligament flavum (LF), facet capsular (CL), intertransverse (ITL), interspinous (ISL) and supraspinous ligaments (SSL). The facet joint articulations were modeled by contact gap elements which are restricted to transmit loads normal to the surface. Axial stiffness of spinal segment (L3-L4) and the predicted intradiscal pressure under compressive load were compared with those published results of measurements [18, 19]. The results appear relatively linear and good agreements were found with previous studies. The general-purpose FEA package ABAQUS (Hibbitt, Karlsson & Sorensen, Inc., RI, USA) was used in this study.

2.2. Implanted models

The implanted models were constructed after modifying the intact model to simulate post-operative changes with these kinds of pedicle screw systems: two kinds of PDS devices and a conventional rigid fixation system (table 2 and figure 2). Since our models were designed to simulate the biomechanical

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Figure 1. Three-dimensional FE model of the human intact L3-L4 multi-motion segment model.

Figure 2. FE models of (a) stabilized motion segment with Nitinol rods with a three-coiled turn manner; and stabilized motion segment with (b) PEEK and (c) Ti-alloy rods with uniform cylindrical shape.
Table 1. Material properties used in finite-element model of lumbar spine.

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s modulus $E$ (MPa)</th>
<th>Poisson’s ratio</th>
<th>Cross-sectional area (mm$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>12 000</td>
<td>0.3</td>
<td>–</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>100</td>
<td>0.2</td>
<td>–</td>
</tr>
<tr>
<td>Bony</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Posterior element</td>
<td>3500</td>
<td>0.25</td>
<td>–</td>
</tr>
<tr>
<td>End plate</td>
<td>25</td>
<td>0.25</td>
<td>–</td>
</tr>
<tr>
<td>Annulus ground</td>
<td>4.2</td>
<td>0.45</td>
<td>–</td>
</tr>
<tr>
<td>Nucleus pulposus</td>
<td>1.0</td>
<td>0.499 (incompressible)</td>
<td>–</td>
</tr>
<tr>
<td>Annulus fibers</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Layer $\frac{1}{2}$</td>
<td>550</td>
<td>–</td>
<td>0.50</td>
</tr>
<tr>
<td>Layer</td>
<td>495</td>
<td>–</td>
<td>0.39</td>
</tr>
<tr>
<td>Layer 5/6</td>
<td>413</td>
<td>–</td>
<td>0.31</td>
</tr>
<tr>
<td>Layer 7/8</td>
<td>358</td>
<td>–</td>
<td>0.24</td>
</tr>
<tr>
<td>Ligaments</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ALL</td>
<td>7.8 (&lt;12%)</td>
<td>20 (&gt;12%)</td>
<td>63.7</td>
</tr>
<tr>
<td>PLL</td>
<td>10 (&lt;11%)</td>
<td>20 (&gt;11%)</td>
<td>20</td>
</tr>
<tr>
<td>LF</td>
<td>15 (&lt;6.2%)</td>
<td>19 (&gt;6.2%)</td>
<td>40</td>
</tr>
<tr>
<td>CL</td>
<td>7.5 (&lt;25%)</td>
<td>33 (&gt;25%)</td>
<td>30</td>
</tr>
<tr>
<td>ITL</td>
<td>10 (&lt;18%)</td>
<td>59 (&gt;18%)</td>
<td>1.8</td>
</tr>
<tr>
<td>ISL</td>
<td>10 (&lt;14%)</td>
<td>12 (&gt;14%)</td>
<td>40</td>
</tr>
<tr>
<td>SSL</td>
<td>8 (&lt;20%)</td>
<td>15 (&gt;20%)</td>
<td>30</td>
</tr>
</tbody>
</table>

All: anterior longitudinal ligament; PLL: posterior longitudinal ligament; LF: ligament flavum; CL: capsule ligament; ITL: intertansverse ligament; ISL: interspinous ligament; SSL: supraspinous ligament.

Table 2. Material properties and design specification of longitudinal rod.

<table>
<thead>
<tr>
<th>Implant type</th>
<th>Diameter (mm)</th>
<th>Design</th>
<th>Elastic modulus (GPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nitinol rod$^a$</td>
<td>4.0</td>
<td>Three-coiled turn</td>
<td>75</td>
</tr>
<tr>
<td>PEEK rod$^a$</td>
<td>6.0</td>
<td>Cylindrical</td>
<td>3.2</td>
</tr>
<tr>
<td>Ti-alloy rod$^b$</td>
<td>6.0</td>
<td>Cylindrical</td>
<td>114</td>
</tr>
</tbody>
</table>

$^a$ PDS system.
$^b$ Conventional rigid fixation.

2.3. Loading and boundary conditions

To simulate the load on the lumbar spine in a neutral posture, the axial compressive load ($F_T$) was applied as a uniform pressure at the superior endplate of the L3 vertebral body and the inferior endplate of L4 vertebral body was totally constrained [21]. Subsequently, the changes in load-sharing characteristics were investigated. Axial force across the facet joints ($F_{FJ}$) was the axial component of the facet contact force predicted from the gap elements. Axial force transmitted through the implants ($F_I$) was computed from the reaction forces ($F_R$) in the case of the total discectomy model under same deformation under an assumption that the ‘residual’ forces ($F_D$) are carried by the anterior column (figure 3):

$$F_T = F_D + F_{FJ} + F_I$$

($F_T = 400$ N $F_I = F_R$ of discectomy model).

Figure 3. Analysis of load-sharing characteristics: (a) calculation of the facet load under 400 N compression and (b) prediction of axial force transmitted through the spinal fixator.
3. Results

3.1. Load-sharing characteristics in spinal column

The intact model showed that 12% of the compressive load (46.9 N) was shared by the facet joints. All fixation systems reduced the facet load regardless of rod types. No distinguishable differences were found between PDS systems (23.5–25.8 N). The facet loads for the rigid fixation system decreased to 37.9% (17.8 N) of that in the intact model.

When the compressive load was applied on the implanted models (Nitinol rod, PEEK rod, Ti-alloy rod), the predicted axial compressive loads transmitted through the devices were 141.8 N, 109.8 N and 266.8 N, respectively (figure 4). Axial forces across the PDS devices and rigid system with facet joints predicted to take over 41%, 33% and 71% of the applied compression load, respectively.

3.2. Peak implant stresses for instrumentation models

Peak von Mises stresses were investigated for all stabilization systems. Stress plots of all the implant are shown with the values of the peak von Mises stress occurring in each component of the fixation systems are also included (figure 5).

All of highest peak von Mises stresses of each component were seen in rigid fixation systems. PDS rods lower the stress values of pedicle screws by 75.5–90% as compared to those of the rigid fixation system. Moreover, stress on the PDS rod (Ni–Ti, PEEK) was decreased to 57.1%, 11.5% of that in the rigid rod, respectively.

4. Discussion

Posterior dynamic stabilization (PDS) is now a popular form of surgical intervention employing a highly diverse range of devices. These devices are designed to relieve several different pathological conditions, including spinal stenosis and discogenic pain, by fulfilling a range of biomechanical functions.

PDS devices have the potential for affecting the spine in several ways: ‘unloading’ and ‘modification of the distribution of loads’ within the intervertebral disc at the treated level. There is a hypothesis that discogenic pain results from an overloading or non-uniform loading of the disc that can be prevented through surgical intervention. PDS devices aim to share a proportion of the compressive load, hence control the magnitude of disc loading [22, 24].
The range of motion is the primary parameter reported in the biomechanical evaluation of spinal fusion devices. The aim of rigid fusion is to restrict the motion at the instrumented segment, based upon the hypothesis that instable motion segment is the cause for low back pain. On the other hand, PDS aims to preserve the motion of that of the intact segment. The concept of PDS devices is based upon the hypothesis that low back pain is caused due to abnormal changes in the pattern of load transfer rather than instable motion [12, 22].

Previous studies have shown that rigid fusions increase the risk of stress shielding, with a decrease in bone mineral density of the stabilized segment [20, 23]. Therefore, it is very important for the PDS devices to provide uniform load sharing. Implant failure or loosening following fusion surgery is common in the presence of pseudoarthrosis. PDS devices have to provide load-sharing throughout its life; implant failure is an important consideration.

PDS devices have a rich potential to modify the mechanical behavior of the implanted segment. They can change the neutral posture of the segment, control its range of motion and change both the loading and the deformation of individual regions of the segment. Many studies have shown the relationship between kinematic behavior and bending moment following the PDS implantation [9, 10, 25]. Therefore, we investigate the load-sharing characteristic and stress distributions after posterior fixation. A computational approach was used to evaluate the biomechanical efficacies of three types of pedicle screw systems.

In this study, we choose two commercial PDS devices which were made of representative materials (Nitinol, PEEK) in recent PDS systems. Our results show that PDS devices reduce the compressive load transmitted through the fixation systems. The lower elastic modulus of the posterior rod in PDS devices leads to less stress-shielding effects. Although we could not find which has ideal stiffness for the dynamic rods, we found that despite numerical difference between Nitinol and PEEK in elastic modulus, two systems have very similar load-sharing characteristics in spinal column.

Our results indicate that the PDS devices have the ability to allow for loading through the anterior bony structures and intervertebral disc. The magnitude of stress-shielding effect was reduced following fusion with PDS devices. All the implant stresses were well below the tensile yield strength of the material used. However, the materials of PDS longitudinal rod decrease the stresses of pedicle screws, and the stresses of PDS longitudinal rod were much below the stress of the rigid rod.

5. Conclusion

Our results confirmed the hypothesis on the PDS devices by showing the substantial reduction in stress-shielding characteristics. Higher axial load was noted across the anterior structure with the PDS devices, which could slow the degeneration process of bony structures and lower the possibility of implant failure.

Acknowledgment

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References


Queries

(1) Author: The meaning of the sentence ‘Stress plots of all the implant . . .’ is unclear. Please check.

(2) Author: In the sentence ‘Many studies have shown the relationship . . .’ reference citation [26] has been changed to [25]. Please check.

(3) Author: Please check whether the journal title in reference [19] is okay as set.

(4) Author: Please be aware that the colour figures in this article will only appear in colour in the Web version. If you require colour in the printed journal and have not previously arranged it, please contact the Production Editor now.

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