Effects of laminectomy and facetectomy on the stability of the lumbar motion segment

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Abstract

A ligamentous, nonlinear, sliding contact, three-dimensional finite element (FE) model of L2–L3 complex was developed to investigate the biomechanical effect of laminectomy with and without facetectomy. The L2–L3 FE model was validated against experimental study under various physiological loadings and found to match well with the experimental data. Four iatrogenic models (unilateral laminectomy, unilateral laminectomy with unilateral facetectomy, unilateral laminectomy with bilateral facetectomy and total bilateral laminectomy) were evaluated under flexion, extension, torsion, lateral bending, anterior and posterior shear load vectors to determine alterations in kinematics and annulus stress. Results show that total laminectomy with facetectomy induces considerable increase in motion and annulus stress, except for lateral bending, whereas unilateral laminectomy shows the least increases.

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Keywords: Finite element method; Lumbar spine; Laminectomy; Facetectomy; Stability

1. Introduction

Spinal stenosis is a degenerative process, caused by progressive narrowing of the lumbar spinal canal and neural foramen, leading to a constriction of the nerve roots of the cauda equine. Non-operative therapy with analgesics, activity modification, exercises, bracing, and epidural steroid injection may be helpful, but seldom results in sustained improvement [1]. Consequently, patients with persistent symptoms due to degenerative lumbar spinal stenosis, particularly lower extremity discomfort, are generally undergone surgery. Currently, facetectomy and laminectomy are the standard methods of decompression for the degenerative lumbar spinal stenosis with resultant alteration in established inter-relationships between various vertebral column components, even though good results have been reported as 85–90% [2–4]. As a result, the change in functional associations of these components may change both the load-bearing and motion characteristics of the spine with a possible increased propensity for back pain or acceleration of segmental degeneration. Therefore, in order to reduce risk for the post-operative complications, resection of more than 50% of facet joint bilaterally or a complete unilateral facetectomy and posterior ligaments transection will not recommend for symptomatic lumbar stenosis in surgery. Sometime, the uses of bone grafts (arthrodesis) of fixation devices (instrumentation) to fuse the affected levels (fusion) are used to treat spinal instability resulting from extensive decompression.

Hitherto, a number of in vitro experiments [5–7] conducted using fresh human cadaveric specimens have been performed to investigate three-dimensional stability of the lumbar spine, wherein significant anatomic variation may play a role in the resulting data. However, there are inherent limitations in using the cadaveric specimens. On the other hand, the finite element (FE) modeling technique mitigates these problems because of its reproducibility and repeatability characteristics, and is an excellent analysis tool in performing parametric studies (whereby one variable at a time is altered). It could also show detailed degree of intervertebral motion.

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in terms of translational/angular motion and provide intrinsic parameters (stress, strain, strain energy, etc.) that experiments cannot. Accordingly, there are few studies conducted using FE method to investigate segmental stability due to graded facetectomy and transection of ligaments. In 1995, Sharma et al. [9] studied the issue of segmental stability on L3–L4 motion segment with and without posterior elements (ligaments and facets) and concluded that the rotational stability in flexion is unlikely without prior damage of ligaments, whereas instability in extension rotation is unlikely before facet degeneration or removal. Natarajan et al. [10] reported a substantial sudden change in rotation motion of the motion segment, due to applied torsion moment, after 75 % removal of any one of the facet joints. The information from the results of these studies is primarily limited to the effect of facetectomy and ligaments transection on biomechanical responses of the motion segment. In recent year, our group, Teo et al., studied the effect of graded facetectomy on the L2–L3 motion segment under anterior shear force and concluded that complete unilateral facetectomy of greater than 75% and bilateral facetectomy of 75% resection markedly alters the translational displacement and flexibilities, especially the motion segment with complete bilateral facetectomy.

From the above review, although the biomechanical responses of the motion subjected to graded facetectomy/transection of ligaments have been investigated experimentally and analytically by researchers, analytical studies about biomechanical effect of laminectomy with and without facetectomy on the motion segment are sparse. Therefore, the current study proposes to analyze the biomechanical alterations in terms of translational/angular motion and annulus stress of the iatrogenic-altered motion segments under physiological flexion, extension, torsion, lateral bending, anterior and posterior shear load vectors. Investigation of these surgical procedures could be useful to both the orthopaedic surgeons and neurosurgeons, who often have slightly different ideas regarding treatment strategies [12]. From this, it may be possible to identify clear evidence-based superiority of procedure performed. We believe that better understanding on the effect of surgical procedures will ultimately translate into better decisions and treatment with low back pain.

2. Materials and methods

An anatomically accurate three-dimensional FE model of L2–L3 complex was generated from geometrical data of an embalmed lumbar spine (59-year old male subject) obtained using a multi-axis digitizer (FaroArm, Bronze Series, Faro Technology Inc., FL, USA). The digitized geometrical data of the bony vertebrae were processed for subsequent volume and FE mesh construction in a downstream process. To obtain the vertebrae geometrical co-ordinates, prominent cross-sections (such as body, pedicle, lamina, facets, etc.) were firstly marked using pencil over the surface profile of each vertebra, dividing it into numerous sectional outlines. A small hole was drilled across the vertebral body, and a rod was tightened to it. This enabled the rod to be mounted in a vice to hold the vertebra upright and allowed the digitizer to capture the pre-marked cross-sections on the vertebra as point data. The point data were processed accordingly into planar cross-sections as shown in Fig. 1A. The data
were then input into a commercially available FE code, ANSYS version 6.0, for FE construction using a bottom-up approach. The systematic combination of the digitizing techniques and discretization process allows anatomical details of the spinal components to be preserved in the FE model (Fig. 1B), which consisted of 8581 eight-noded isoparametric solid element and 817 cable elements with 32,641 degrees of freedom. To take into account the effect of lordosis for the vertebral displacement, the anterior and posterior height of the intervertebral disc were taken as 8.9 and 6.7 mm, respectively, based on literature to complete the FE model of L2–L3 motion segment. Details of the model generation have been previously reported [13].

Table 1 summarizes the type and number of elements used to model the various components of the L2–L3 motion segment. The vertebral body was assumed to have a cancellous core with the periphery composed of cortical bone (1.0 mm thick). Cartilaginous endplates, 0.5 mm thick, were assumed at the superior and inferior surfaces of the intervertebral disc. The annulus was assumed as a composite material consisting of annular fibers embedded in a homogeneous matrix material. It was assumed to consist of three consecutive laminar layers. In each layer, annulus fibers were oriented on average at $\pm 30^\circ$ to the endplate. The fiber diameters or fiber cross-sectional areas, which are averaged in each annulus layer, were calculated based on the fiber volume fraction assumed, the number of fiber elements used, length of the fibers, and the volume of each annulus layer.

All solid components including cortical and cancellous bone, endplates, annulus ground substance and posterior elements were simulated by a total of 8581 eight-node brick elements. The annulus fibers were simulated by three-dimensional (3D) cable elements that sustained tension only. The nucleus pulposus was modeled using 8-noded brick element to simulate an incompressible behavior, in which has a Poisson’s ratio close to 0.5 and Young’s Modulus of 1 MPa. In order to appropriately model the changing areas of contact of facet articulating surface with loads, facet articulations were simulated by frictionless sliding surface contact elements. The pair of contact surface in each facet joint was assumed cover with cartilage material of 0.5 mm thick, which was assumed to behave linearly [9]. A total of 150 contact elements were used to idealize the facet joints and this number exceeded by more than half the number of contact elements used in the analysis by Natarajan et al. [10], who performed a convergence test for optimum mesh density for the facet joints. More details of modeling the facet joints have been reported in our earlier study [14].

Seven different ligaments (anterior and posterior longitudinal ligaments, ligamentum flavum, the intertransverse ligament, the capsular ligament, the interspinous ligaments, and the supraspinous ligament) approximated the ligamentous structure in the motion segment and their attachment points to the bony prominence were chosen from anatomy books to mimic anatomic observations as closely as possible. The ligaments were assumed to be nonlinear and defined based on data derived by experimental study of Chazal et al. [15]. The cross-sectional areas used were averages of the values reported previously. Also, three types of nonlinearities (i.e. geometrical, material and contact nonlinearities) exhibited by the motion segment were incorporated into the model.

Table 1

<table>
<thead>
<tr>
<th>Material</th>
<th>Element type</th>
<th>Number of elements</th>
<th>Young’s modulus (MPa)</th>
<th>Possion’s ratio</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>8-node brick</td>
<td>528</td>
<td>12000</td>
<td>0.3</td>
<td>Natarajan et al. [10]</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>8-node brick</td>
<td>2448</td>
<td>344</td>
<td>0.2</td>
<td>Teo et al. [11]</td>
</tr>
<tr>
<td>End-plate</td>
<td>8-node brick</td>
<td>260</td>
<td>25</td>
<td>0.25</td>
<td>Sharma et al. [9]</td>
</tr>
<tr>
<td>Posterior elements</td>
<td>8-node brick</td>
<td>4373</td>
<td>3500</td>
<td>0.25</td>
<td>Natarajan et al. [10]</td>
</tr>
<tr>
<td>Annulus matrix</td>
<td>8-node brick</td>
<td>288</td>
<td>4.2</td>
<td>0.45</td>
<td>Sharma et al. [9]</td>
</tr>
<tr>
<td>Nucleus</td>
<td>8-node brick</td>
<td>252</td>
<td>1.0</td>
<td>0.499</td>
<td>Sharma et al. [9]</td>
</tr>
<tr>
<td>Annulus fiber</td>
<td>3D-cable (tension only)</td>
<td>766</td>
<td>500</td>
<td></td>
<td>Teo et al. [11]</td>
</tr>
<tr>
<td>Facet contact</td>
<td>Sliding surface contact</td>
<td>150</td>
<td></td>
<td></td>
<td>Lee et al. [13,14]</td>
</tr>
<tr>
<td>Ligaments</td>
<td>3D-cable (tension only)</td>
<td></td>
<td></td>
<td></td>
<td>Chazal et al. [15]</td>
</tr>
</tbody>
</table>

ALL, anterior longitudinal ligament; PLL, posterior longitudinal ligament; LF, ligamentum flavum; TL, transverse ligament; CL, capsular ligament; IS, interspinous; SS, supraspinous ligament.
2.1. Model validation

As validation studies form the vital link between the development of the FE model and its final intended use, the present model was validated under physiological loading modes (compression, tension, anterior and posterior shear, flexion, extension, lateral bending and torsion), and the predicted responses were compared against results by Berkson et al. [16], Schultz et al. [17] and Panjabi et al. [18]. Accordingly, the inferior surface of L3 vertebral body and its spinous process were fixed in all directions. Two loadcases were performed for the loading conditions. A compressive preload of 400 N, achieved in five steps, was applied uniformly to all nodes of the top surface of the L2 vertebral body (force control) in the first load step so as to mimic the loading conditions during physiological movement in the upright posture by simulating the effect of in vivo body weight. At the second load step, for validation under axial force and rotational moment configurations, an axial force of 150/400 N and pure moment of 7.5 Nm was applied on top of L2 vertebral body from zero to predetermined value, respectively. As for anterior and posterior shear load, a shear load of magnitude 150 N applied incrementally at the center of the L2 vertebra in a horizontal direction parallel to the base of the inferior vertebra. All these loadings for the second load step were applied in such a way to mimic the loading conditions in the mechanical testing of the cadaveric motion segments [16–18].

The predicted results for compression and shear loading were based on the displacement of the center of the superior vertebral body and compared with the aforementioned in vitro experimental results. As for the rotational moments, two nodal points on the superior surface of L2 were selected to determine the rotation of the L2 with respect to L3 (fixed) in the plane of moment application. Other nodal points were also chosen for the determination of coupled rotation in other planes. This procedure assumes that the nodal displacement caused by bony vertebral deformation is negligible, which is also the basic assumption in almost all the vertebral kinematics measurement during in vitro biomechanical lumbar spine experiments.

2.2. Iatrogenically altered analysis

The validated motion segment model was then used to predict the changes in motion responses and annulus stress due to four iatrogenically altered conditions under flexion, extension, left lateral bending, left torsion, anterior and posterior shear loadings. Accordingly, the posterior parts of the FE models were modified accordingly to simulate four iatrogenic-altered conditions designed to replicate the common methods of decompression for the degenerative lumbar spinal stenosis [19,20] as follows:

- Unilateral total laminectomy (Unilam): removal of the left lamina of L3 and left part of flaval ligament between L2 and L3 (Fig. 2A).
- Unilateral total laminectomy with unilateral facetectomy (Unilam+Unifac): same as (a) plus removal of left articular process of L3 with the corresponding facet capsular ligament (Fig. 2B).
- Unilateral total laminectomy with bilateral facetectomy (Unilam+Bifac): same as (c) plus removal of right articular process of L3 with the corresponding facet capsular ligament (Fig. 2C).
- Bilateral laminectomy (Bilam+Bifac): same as (d) plus removal of spinous process and right lamina of L3. The supraspinous, flaval and interspinous ligaments were removed between L2 and L3 (Fig. 2D).

All resected models were analyzed under similar loading/boundary configurations according to the validation study explained earlier. For each iatrogenic-altered condition, a pure moment of 7.5 Nm and a shear force of 150 N were applied to the L2–L3 model in flexion, extension, left torsion, left lateral bending, anterior and posterior shear, respectively. In order to simulate loss of compressibility for nucleus, disc degeneration was simulated at L2–L3 intervertebral disc by assuming the nucleus to have an elastic modulus twice that of the annulus and Possion’s ratio to be 0.48 [8,21]. Thus, a total of 54 simulations were completed for the analyses, which were performed in commercially available finite element code, ANSYS 6.0.

The following biomechanical quantities were analyzed in every iatrogenic-altered condition relative to the intact model. Under every loading conditions: translational/angular motion of L2 with respect to L3 in the direction of loading and maximum von Mises stress in the L2–L3 disc annulus. Percentage changes (PC) from intact displacement/rotation were normalized to the intact case by use of the following equations:

\[ PC = \frac{R_{\text{variant}} - R_{\text{intact}}}{R_{\text{intact}}} \times 100 \]

where \( R_{\text{variant}} \) is the rotation for a given model variant and \( R_{\text{intact}} \) represents the intact motion. A positive value in the normalized rotation/displacement indicates an increase in mobility of the iatrogenic-altered model. The maximum von Mises stress was the peak value in all the elements representing the disc annulus. The stresses were normalized similarly with respect to the corresponding values of the intact model. A positive value in the normalized stress indicated an increase in stress magnitude. In this way, all quantities were
expressed as percentage change from those of the intact model.

3. Results

The predicted maximum principal motions of the motion segment under various loading configurations compared against experimental results [16–18] are shown in Fig. 3. As seen for load vectors, the FE model predicted response falls within the range of the aforementioned experimental data. Fig. 3 also shows that the experimental results exhibit considerable divergence due to biological variations among specimens and the use of different experimental techniques.

For the iatrogenic-altered (unilateral laminectomy with bilateral facetectomy and total bilateral laminectomy) models, the percentage increase in angular rotation (Fig. 4) was the highest under torsion compared with other loading modes and the lowest under lateral bending. In general, increasing levels of iatrogenic conditions resulted in increased changes in the angular rotation and displacements under all the physiologic load vectors, except for lateral bending. It is observed that, for degenerated disc at L2–L3 level, the angular and displacement responses were generally predicted lower than that of the normal disc.

Under flexion, bilateral laminectomy with bilateral facetectomy resulted in a considerable increase in the angular motion due to the removal of supraspinous and interspinous ligaments. It is known that these ligaments play a restrictive role in maintaining the motion segment’s integrity under flexion loading. This concurs with the finding by Pintar et al. [7], who evaluated spinal components under various surgical alterations and noted a significant increase in overall deflection of the function unit (bilateral facetectomy with posterior ligaments transection) under flexion–compression load. Under extension, there was an increase in angular motion predicted for the model with unilateral
laminectomy with unilateral facetectomy, but the higher increase was seen when bilateral facetectomy is done. This underscores the importance of facet joints in resisting extension moment. Sharma et al. [9], in their study of segmental stability, found that lumbar motion segments were susceptible to rotational instability in extension after facetectomy. This finding is in agreement with their analysis.

As for torsion, a considerable increase in angular motion was observed when compression facet joint was removed, indicating that the facet joint restricts torsion movement. This is in agreement with Natarajan et al. [10] finding which concluded that a sudden change in rotation motion of the motion segment due to applied torsion moment after 75% of the compression facet joint was removed.

Under lateral bending, various iatrogenic conditions did not show significant deviation in rotational response, as compared with the intact motion segment. This demonstrates minor role played by the posterior spinal elements and underscore intervertebral disc as main load bearing under lateral bending. This is in accordance to the findings by Shirazi-Adl [22] and Schultz et al. [17]. Both reported that the removal of facets had a negligible effect on the segmental lateral flexibility under lateral rotations of up to about 7°.

As for anterior shear, considerable increase in displacement was seen for the model with unilateral laminectomy with unilateral facetectomy. Increasing level of iatrogenic condition resulted an increased in the displacement. But, for posterior shear, the increase in the displacement was observed to be less, as compared with the anterior shear. In recent year, Teo et al. [11], in their study of facetectomy under anterior shear, also reported a considerable increase for their iatrogenically altered FE model with bilateral facetectomy.

Fig. 5 shows the increase in annulus stress at L2–L3 intervertebral disc for pure moment and shear load under different degrees of decompression. Similar trend was observed between the increase in annulus stress and flexibility of the motion segment. In addition, the predicted annulus stress for degenerated disc was generally lower than that of normal disc under various loadings, except for torsion. The maximum von Mises stress are found in the ventral, dorsal, right lateral, left lateral, ventral and dorsal region of the annulus during flexion, extension, torsion, lateral bending loadings, anterior and posterior shear, respectively.

4. Discussion

Results from previous studies depended mainly on postoperative radiographs, patient interviews, and examination were based on subjective opinions of clinical evaluation. Biomechanical investigation using FE method was not used previously to establish whether these surgical techniques could provide segmental stability. Surgically, posterior approach, involving the facetectomy and laminectomy, is commonly conducted for nerve root decompression that will inevitably affect the segmental stability. Therefore, in this study, a three-dimensional anatomically accurate L2–L3 FE model was modified accordingly to simulate laminectomy with and without facetectomy or combination to investigate the influence of these surgical procedures on segmental stability in terms of translational/rotational motion and annulus stress under various physiological loadings. The authors believe that this study reinforces some understandings into the role of posterior elements in preserving segmental stability under the loadings and would help orthopaedic surgeons and neurosurgeons make better decisions and treatment with low back pain.

The close agreement in the translational/angular responses predicted by the intact FE model and the
experimental data of Berkson et al. [16], Schultz et al. [17] and Panjabi et al. [18], under physiological compression, tension, flexion, extension, left lateral bending, left torsion, anterior and posterior shear load vectors, as shown in Fig. 3, provides the necessary confidence to further analyze the biomechanical effect of the iatrogenic changes due to the surgical procedure for decompression.

Clinically, total bilateral laminectomy is one of the common decompression methods for lumbar stenosis. Many studies have shown that total laminectomy increases and produces segmental instability unless fusion is performed [3,23,24]. In the study by Postacchini et al. [25], three of 32 laminectomy cases developed marked segmental instability. Because midline structures (i.e. supraspinous and interspinous ligaments) play a mechanical role in the movement of the lumbar spine, their ablation may contribute to the pathogenesis of postoperative instability, particularly when bilateral facetectomy has been performed simultaneously, thus altering the load-bearing and motion characteristics of the spine [6]. These postoperative alterations in intervened segment may produce further disc degeneration and change of alignment or slipping, which could induce low back pain. Other postoperative complications include severe atrophy of paraspinal muscles, and posterior scarring, which can lead to poor results [26,27]. However, clinical results from laminectomy do tend to deteriorate with time. This is partially a function of the progressive native of degenerative lumbar disease, and also of the mechanical disruption of the lumbar spine secondary to total laminectomy. It is important to preserve as much facet joint and pars interarticularis as possible at surgery, because extensive laminectomy leads to increased rate of spondylolisthesis postoperatively [3]. In recent years, unilateral lateral laminectomy has been proposed for the management of spinal stenosis due to the fact that it preserves the facet joints and neural arch of the contralateral side, limits postoperative destabilization and
protects the nervous structures against posterior scarring. Mariconda et al. [19] performed unilateral laminectomy to treat spinal stenosis in 44 patients and reported that unilateral laminectomy leads to better 4-year results than does conservative treatment of patients. This current study suggests that unilateral laminectomy does not lead to segmental instability in views of much lower translational/angular responses and annulus stress, compared to the other iatrogenic conditions, under various physiological loadings. Since the analyses have shown that total laminectomy increase or cause vertebral instability, this study underscores the need for arthorodesis if patients who undergo total laminectomy with complete bilateral facetectomy are at increased risk for postoperative instability and complications.

Biomechanically, the excision of the supraspinous/interspinous ligaments and facetectomy greater than 50% affects the motion of the intervertebral joint and induces additional stresses to the remaining components of the spine [5–7]. Current study shows that an increase in flexibility of the motion segment resulted in an increase in annulus stress at L2–L3 intervertebral disc (Fig. 5). This is in consistent with Dai et al. [28], who showed that the stress level in all parts of the FE lumbar spine was elevated with posterior element resection.

The present iatrogenically altered models help to provide qualitative analysis of the biomechanical changes in percent increase in motion and annulus stress as compared to the intact one. The high increase in angular motion under flexion due to total bilateral laminectomy with facetectomy demonstrates the restrictive role of the ligaments of the spine. Under extension and torsion, the considerable increase in angular motion occurred only after laminectomy with bilateral facetectomy, indicating the importance role of the facet joints in resisting extension and torsion loading.

Fig. 5. Increase in annulus stress for pure moment and shear load under different degrees of decompression (see text for abbreviation).
As for anterior shear, the displacement increased progressively with an increase in level of iatrogenic conditions. Such increase in the angular/translational motion correlates with the increase in annulus stress in L2–L3 intervertebral disc (Fig. 5) and seems to emphasize the transfer of the internal resisting load vector from the posterior to the anterior region of the spine. The observed high increase in the motion/stress seems to confirm the clinical observation that patients undergoing laminectomy with facetectomy may be more susceptible to develop instability or change of alignment or slipping, which could induce low back pain [12,25].

Currently, there is significant interest in the development of minimally invasive spine surgery techniques. Various authors have discussed technical modifications of the stand laminectomy, applicable to the cervical, thoracic, and lumbar spine. These techniques have evolved in an attempt to adequately treat patients with spinal stenosis while the structural preservation of the spine is maximized. Wide variation in the use of spinal surgery indicates there is substantial room for discretion in clinical decisions about treatment for these conditions. But there is considerable controversy regarding which optimal surgical procedure should be performed. We believe that the remarkable degree of variation reflects clinical uncertainty about the relative efficacy of medical and surgical approaches and is largely attributable to the paucity of instability results available to guide decision-making. Hence, the use of FE analysis will lead to an improved understanding of these biomechanical characteristics, therefore assisting in defining clinical expectations for various forms of surgical intervention as well as offering insight into technical alterations of the operative procedures. The present study contributes to an improved understanding of the biomechanics through an analysis of the external motions and intrinsic parameter in disc annulus using the FE method.

4.1. Limitation and future works

Most previous studies (experimental and analytical) concerning spine stability were based mainly on the functional spinal unit [5–7,9–11]. Under multiple laminectomy or facetectomy, testing of the functional spinal unit cannot give the whole picture of the effects seen else where in the lumbar spine, such as redistribution of lumbar spinal motion and adjacent disc stress after multiple laminectomy and facetectomy. This motion pattern may well represent the in vivo situation, but will not be observed by function spinal unit testing. Also, the disc annulus stress predicted by the FE model is at L2–L3 level, more interesting would be the disc stress at affected L3–L4 level, rather than the adjacent level. Therefore, it should be emphasized that the stability of the lumbar spine as a whole should be examined under the physiological loading condition, which would be the subject for further study. Truck muscles, known to be the primary stabilizers for the spine, are not included in the analysis as this study explores only the mechanical factors that may cause segmental instability. The biological responses of the spine to these mechanical situations must be investigated to gain a clear understanding of segmental responses under iatrogenic conditions.

5. Conclusion

A three-dimensional nonlinear FE model of lumbar motion segment was established to simulate various methods of decompression (unilateral laminectomy, unilateral laminectomy with unilateral facetectomy, unilateral laminectomy with bilateral facetectomy and total bilateral laminectomy) used for the degenerative lumbar spinal stenosis. This study provides a qualitative analysis, which shows a considerable increase in kinematics and annulus stress under total bilateral laminectomy, and least increase under unilateral laminectomy. Results suggest that total bilateral laminectomy with facetectomy would lead to segmental instability under most of physiological loadings, in which might require arthrodesis on patients, if necessary, to minimize postoperative complications.

References


