

# Does TLIF Aggravate Adjacent Segmental Degeneration More Adversely than ALIF? A Finite Element Study

## *TLIF Komşu Segmental Dejenerasyonu ALIF'den Daha Olumsuz Etkiler mi? Bir Sonlu Eleman Çalışması*

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### ABSTRACT

**AIM:** To determine whether transforaminal lumbar interbody fusion aggravate adjacent segmental degeneration more adversely than anterior lumbar interbody fusion.

**MATERIAL and METHODS:** A normal finite element model, an anterior lumbar interbody fusion model and a transforaminal lumbar interbody fusion model of L3-5 were developed. 800 N compressive loading plus 10 Nm moments simulating flexion, extension, lateral bending and axial rotation were imposed on L3 superior endplate. The intradiscal pressure, intersegmental rotation and tresca stress in L3-4 were investigated.

**RESULTS:** Under all loading directions, the values of three parameters in transforaminal lumbar interbody fusion model were the highest and those in normal model were the lowest, and the values in the ALIF model were higher than the normal model while lower than the TLIF model. In the ALIF and normal models, the values of the three parameters on left lateral bending and axial rotation were equal to those on right lateral bending and axial rotation. However, in the TLIF model, the values on right lateral bending and axial rotation were larger than those on left lateral bending and axial rotation.

**CONCLUSION:** Transforaminal lumbar interbody fusion aggravates adjacent segmental degeneration more adversely than anterior lumbar interbody fusion.

**KEYWORDS:** Anterior lumbar interbody fusion (ALIF), Transforaminal lumbar interbody fusion (TLIF), Adjacent segmental degeneration (ASD)

### ÖZ

**AMAÇ:** Transforaminal lomber interbody füzyonun komşu segment dejenerasyonunu anterior lomber interbody füzyondan daha olumsuz etkileyip etkilemediğini belirlemek.

**YÖNTEM ve GEREÇLER:** Bir normal sonlu eleman modelini kullanarak L3-5 için bir anterior lomber interbody füzyon modeli ve bir transforaminal lomber interbody füzyon modeli geliştirildi. L3 superior uç plakasına fleksiyon, lateral bükülme ve aksiyal rotasyon simülasyonu yapan 10 Nm momentler artı 800 N kompresif yükleme uygulandı. İntradisk basıncı, intersegmental rotasyon ve tresca stresi L3-4'te incelendi.

**BULGULAR:** Tüm yükleme yönlerinde üç parametrenin değeri transforaminal lomber interbody füzyon modelinde en yüksek olurken normal modelde en düşüktü ve ALIF modelindeki değerler normal modelden yüksekken TLIF modelinden düşüktü. ALIF ve normal modellerde üç parametrenin sol lateral bükülme ve aksiyal rotasyon değerleri sağ lateral bükülme ve aksiyal rotasyon için olanlara eşitti. Ancak, TLIF modelinde sağ lateral bükülme ve aksiyal rotasyon değerleri sol lateral bükülme ve aksiyal rotasyon için olandan daha büyüktü.

**SONUÇ:** Transforaminal lomber interbody füzyon komşu segmental dejenerasyonu anterior lomber interbody füzyondan daha olumsuz etkilemektedir.

**ANAHTAR SÖZCÜKLER:** Anterior lomber interbody fusion (ALIF), Transforaminal lomber interbody fusion (TLIF), Komşu segmental dejenerasyonu (ASD)

### INTRODUCTION

Transforaminal lumbar interbody fusion (TLIF) and anterior lumbar interbody fusion (ALIF) have similar indications, and both modalities have been a standard technique of lumbar interbody fusion used to treat degenerative disc disease, spondylolisthesis, spinal deformity, and pseudarthrosis. Many reports have been published about the surgical results of

the two modalities, most of which suggested that the two techniques had no significant difference in terms of morbidity or mortality rates, blood loss, surgical time and functional outcomes (5, 6). However, no reports, comparing the influence of ALIF and TLIF on adjacent segmental degeneration have been published.

In addition, inherent limitations in clinical research and conventional experiments make discrete characterization of the effects of lumbar interbody fusion on the adjacent segment significantly difficult. Specimens for cases and controls are difficult to obtain and standardize, and subsequent repetitive testing under various loads can result in significant sample variability (10). In contrast, the finite element modeling technique mitigates these problems. It is highly reproducible and repeatable. Adjustments can be made to the models to affect the material properties, simulate different situations of spine either in normal or fusion conditions, show stress distribution under different loading modes, or reflect any structural change. Compared with other experimental methods, the finite element method has many advantages, facilitating a comparative study among different models (10).

We therefore developed a three-dimensional finite element model of L3-5 for the normal, healthy spine, along with an ALIF model and a TLIF model, and our objectives were: 1) to compare the biomechanical effect of ALIF and TLIF on the adjacent segment; 2) to determine if TLIF aggravates adjacent segmental degeneration more adversely than ALIF.

## MATERIAL and METHODS

A three-dimensional numerical model of a two-level ligamentous lumbar segment (3 vertebrae and 2 discs) was built and implemented with the finite element model (FEM) software ANSYS 10.0. Three different configurations of the model were considered: (1) a normal healthy model of L3-5 (Figure 1); (2) an ALIF model and (3) a TLIF model. The normal model was well validated in the past studies (9). The ALIF and TLIF models were modified from the normal model by changing the geometry or material properties of the L4-5 segment. We assume the elements of the L3-4 segments including intervertebral disc, facet joints, endplates, and vertebral bodies to be normal in all finite element models.

### 1. Normal model

A non-linear, three-dimensional FEM of L3-5 was created with geometry based on the high resolution computed tomography scan (slice thickness of 0.75mm) of a 26-year-old healthy male volunteer. The FEM of the ligamentous lumbar spine consisted of vertebrae, intervertebral discs, endplates, superior and inferior facet articulating surfaces, ligaments and capsules. Spinal structures that could not be recognized using computer tomography were determined by magnetic resonance imaging and histological observations (8).

The modeled vertebrae and intervertebral discs were meshed using eight-node solid 186 elements. The vertebrae and intervertebral discs model consists of 156806 elements and the seven spinal ligaments were represented by 2-node link 10 elements, which do not offer resistance in compression. The surfaces of facet joints were simulated by a cartilaginous layer, which was assumed to be multi-linear elastic in compression. The contact between the facet joints was simulated by surface to surface contact elements without



Figure 1: Normal finite element model of L3-5.

friction. The facet joints have a gap of 0.5 mm and can only transmit compressive forces.

The intervertebral discs consisted of the nucleus pulposus and the surrounding annulus fibrosus. The nucleus pulposus, modeled as an incompressible material, was 43% of the total disc volume and located slightly posterior to the center of the disc (12). The annulus fibrosus was assumed to be a composite of a homogenous ground substance reinforced by collagen fibers. Seven crisscross fiber layers were defined in the radial direction and the fibers were mounted in two times seven layers with orientation of  $\pm 30^\circ$  to the midcross-sectional plane. The fiber content of the modeled annulus fibrosus was approximately 19%, appropriate for the natural collagen content of the annulus. The material constants for the nucleus and annulus ground substance were defined according to previously accepted values (4) (Table I).

### 2. The ALIF and TLIF models

To mimic ALIF and TLIF, the L4-5 disc of the normal model was removed and replaced by the interbody bone graft, which was assumed to have the same dimensions as healthy disc and the same material properties as cancellous bone. In addition, bone grafts were placed between the transverse processes of L4-5 segment of the TLIF model bilaterally. To facilitate the study, all the grafts were assumed to have gained solid fusion and internal fixation instrumentations were removed. TLIF was assumed to be performed on the left so the total facetectomy was performed at the L4-5 segment and the left facet joint and ligamentum flavum were removed, and the left superior and inferior lamina were partially removed, but the posterior elements, contralateral facet joint, supraspinous ligaments, and interspinous ligaments were preserved.

**Table I:** The Material Properties in the Finite Element Models

Material	Young's modulus (Mpa)	Poisson's ratio	Element number	Cross-section (mm <sup>2</sup> )
Cortical bone	12000	0.3		
Cancellous bone	100	0.2		
Posterior elements	3500	0.30		
Nucleus pulposus	1	0.499		
Annulus ground substance	4.2	0.45		
Fiber	92	0.45		
Endplate	500	0.25		
Facet cartilage	3500	0.25		
Capsular ligament	7.7	0.39	24	102.5
Anterior longitudinal ligament	11.9	0.39	20	75.9
Posterior longitudinal ligament	12.5	0.39	10	51.8
Ligamentum flavum	2.4	0.39	6	78.7
Interspinous ligament	3.4	0.39	6	36.3
Supraspinous ligament	3.4	0.39	2	75.7
Transverse ligament	3.4	0.39	10	2

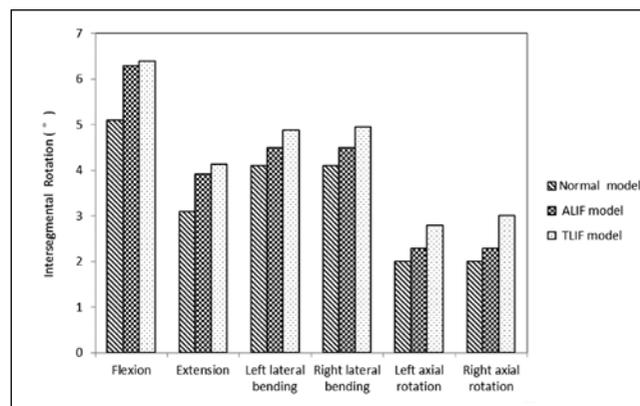
**3. Boundary and loading conditions**

The degrees of freedom of inferior surfaces of L5 were completely fixed in all directions, and 10Nm flexion, 10Nm extension, 10Nm lateral bending, and 10Nm axial rotation moment under 800N compressive loading were imposed on the L3 superior endplate of all models respectively. The maximum load was achieved in five load steps in each model and the main parameters including intradiscal pressure, intersegmental rotation and maximum tresca stress of annulus fibrosus at the L3-4 segment were investigated.

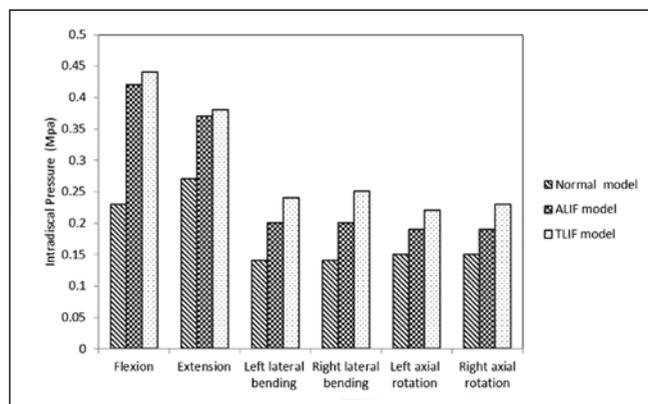
**RESULTS**

The results of the three parameters of intradiscal pressure, intersegmental rotation and tresca stress at the L3-4 segment are represented in Figure 2, 3 and 4. In ALIF and normal models, the values of the three parameters on left lateral bending and axial rotation were equal to those on right lateral bending and axial rotation. While in the TLIF model, the values on right

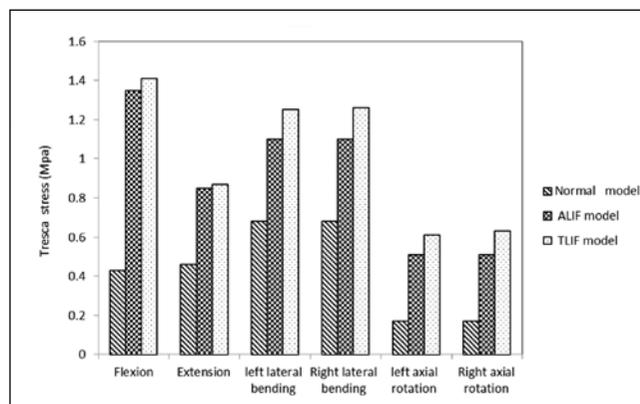
lateral bending and axial rotation were larger than those on left lateral bending and axial rotation.



**Figure 3:** Intersegmental rotations in different loading directions and models.



**Figure 2:** Intradiscal pressure in different loading directions and models.



**Figure 4:** Tresca stress on annulus fibrosus in different loading directions and models.

The intradiscal pressure of L3-4 in different loading directions and models are displayed in Figure 2. Under all loading directions, the intradiscal pressure in the TLIF model is the highest and presented with 0.44Mpa, 0.38Mpa, 0.24Mpa, 0.25Mpa, 0.22Mpa and 0.23Mpa on flexion, extension, left lateral bending, right lateral bending, left axial rotation and right axial rotation respectively. The intradiscal pressure in the normal model is the lowest and presented with 0.23Mpa, 0.27Mpa, 0.14Mpa and 0.15Mpa on flexion, extension, left-right lateral bending and left-right axial rotation. In all loading directions, the intradiscal pressures in ALIF model were higher than normal model, while lower than the TLIF model.

The intersegmental rotations of L3-4 in different models and loading directions are displayed in Figure 3. Under all loading directions, the intersegmental rotation of the TLIF model is the largest with 6.4° on flexion, 4.1° on extension, 4.8° on left lateral bending, 4.9° on right lateral bending, 2.8° on left axial rotation and 3.0° on right axial rotation, and the values in the normal model are the lowest with 5.1° on flexion, 3.1° on extension, 4.1° on left-right lateral bending and 2° on left-right axial rotation. The ALIF model presented with 6.3° on flexion, 3.9° on extension, 4.5° on left-right lateral bending and 2.3° on left-right axial rotation, which are larger than the normal model, but lower than the TLIF model.

The tresca stresses on the L3-4 annulus fibrosus in different models and loading directions are displayed in Figure 4. These are higher in the TLIF model than in any of the other models in all loading planes with 1.41MPa on flexion, 0.87MPa on extension, 1.25MPa on left lateral bending, 1.26MPa on right lateral bending, 0.61MPa on left axial rotation and 0.63MPa on right axial rotation. The normal model has the lowest values of tresca stress with 0.43MPa on flexion, 0.46MPa on extension, 0.68MPa on left-right lateral bending, and 0.17MPa on left-right axial rotation. Similar to intradiscal pressure and intersegmental rotation, the tresca stresses in the ALIF model are larger than the normal model, but lower than the TLIF model in all loading directions.

## DISCUSSION

In the present study, we analyzed and compared the biomechanical influence of TLIF and ALIF on the superior adjacent intervertebral disc using finite element techniques, and we attempted to determine if TLIF aggravates adjacent segment degeneration more adversely than ALIF by comparing the parameters of intradiscal pressure, intersegmental rotation and tresca stress in L3-4.

Adjacent segment degeneration is a well-recognized and frequent long-term consequence of lumbar interbody fusion. Lumbar fusion may lead to decreased elasticity and increased stiffness, resulting in stress concentration at the adjacent segments. This may adversely affect the adjacent disc and aggravate its degeneration. Many studies have reported the effect of lumbar fusion on the adjacent segments. These findings are confirmed through in vitro experiments by Rao (7), Weinhofer (11), Chen (1) and Chow (2) all suggesting

that hypermobility, tresca stress and intradiscal pressures are increased within adjacent segments. In the present study, we found at the adjacent superior segments in flexion, extension, axial rotation and lateral bending, the values of the range of motion, the intradiscal pressure and tresca stress at L3-4 segment are significantly larger in ALIF and TLIF models than normal spine model, which is consistent with the aforementioned studies, demonstrating that both ALIF and TLIF can promote adjacent segmental degeneration.

The lumbar function unit, with interdependent components distributing loads and guiding movements, involves the intervertebral disc, the posterior facet joints and ligamentous system. The facet joints and posterior ligamentous system play an important role in loading distribution in addition to the intervertebral disc. The anatomical orientation of the lumbar facet joints indicates that their major function is the control and stabilisation of torsional forces, leading to some limitations on the motion of lumbar function units, especially on lateral bending and axial rotation, which may affect the intersegmental rotation and stress distribution, and subsequently affect the biomechanical characteristics of adjacent segments. The posterior ligamentous system also influences stress distribution, and Ekman (3) found a significantly higher incidence of ASD in patients with laminectomy compared to non-laminectomised patients in a prospective randomised study of 111 patients on the long-term effect of lumbar fusion on adjacent disc degeneration, suggesting laminectomy may be of pathogenetic importance for ASD development. It could be that the loss of posterior tension band function creates instability and, hence, accelerates disc degeneration (3). In the present study, we created the TLIF model according to the standard technique of TLIF surgery, in which the left facet joint was completely removed and left superior and inferior lamina were partially removed, and as a result, the biomechanical situation of the adjacent segment would be affected. Theoretically, the left facet joint removal may have more influence on the adjacent segment on lateral bending and axial rotation than on flexion or extension. In the present study, we found the intradiscal pressure, intersegmental rotation and tresca stress of TLIF model to be higher on right lateral bending and right axial rotation than left lateral bending and left axial rotation, indicating the effect of facetectomy and laminectomy on the loading conduction. There was just a minor difference between the left and right side, maybe because the effect of facetectomy and laminectomy was offset by the solid fusion obtained by interbody fusion and intertransverse fusion.

In addition, when compared to the ALIF model, the values of the intradiscal pressure, intersegmental rotation and tresca stress in the TLIF model on lateral bending and axial rotation were larger than those in ALIF, and our results demonstrate that the intradiscal pressure of the L3-4 segment increases 20%, 25%, 15.8% and 21.1%, the intersegmental rotation increases 8.4%, 10%, 21.7% and 30.9% , and the tresca stress increases 13.6%, 14.5%, 19.6% and 23.5% on left lateral bending, right lateral bending, left axial rotation and right

axial rotation respectively from the ALIF to the TLIF model. However, under flexion and extension directions, the values of the three parameters show a small increase, and the intradiscal pressure in the TLIF model increases 4.8% and 2.7%, the intersegmental rotation increases 1.6% and 5.4%, and the tresca stress increase 4.4% and 2.4% on flexion and extension respectively, compared with the ALIF model. We therefore suggest that the intertransverse fusion may have more influence on the loading conduction on lateral bending and axial rotation, while a relatively small influence on flexion and extension. Although the main plane of motion in the lumbar spine is in flexion and extension, we can find a clear trend that TLIF leads to higher intradiscal pressure, intersegmental rotation and tresca stress at L3-4 segment when compared to ALIF model, and we therefore conclude that TLIF influences adjacent segmental degeneration more adversely than ALIF.

However, our study does have inherent limitations. To facilitate the study, the structure of the vertebral body was assumed to be isotropic and homogenous and the bone graft was assumed to completely occupy the disc space and intertransverse space, and all the fusion was assumed to be solid fusion, which may not be truly representative of the in vitro or clinical situation. In addition, the present finite element model does not account for the mechanical effect of muscle contraction and the loading conditions were not truly physiological. More studies are needed to further clarify these effects.

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