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Research Article

Biomechanical Effects of Spinal Flexibility and Rigidity on Lumbar Spine Loading: A Finite Element Analysis Study

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Abstract

Objective: The effects of spinal rigidity correlate positively with low back pain. Although spine flexibility has also been considered important for preventing stress-related lumbar disorders, the effects of spinal rigidity on mechanical stress during lumbar motion have rarely been reported.

Methods: A biomechanical investigation and finite element analysis were conducted to elucidate the effects of flexibility of the lumbar spine. We created two kinds of rigid and flexible spine models by changing the material properties of the ligaments and discs. A hybrid loading condition was applied and the maximum stresses on the discs and pars inter articularis were computed.

Results: The FE analysis, in the rigid spine models, the range of motion decreased to about 80%, while in the flexible spine models it increased to about 110% that of the intact model. The pars and disc stress increased in both rigid models.

Conclusions: Biomechanically, both pars stress and disc stress increased in the rigid spine; a flexible spine is important for preventing stress-related disorders such as lumbar pars stress fracture and degenerative disc disease. Thus, flexibility of the spine is beneficial for preventing low back pain by reducing the lumbar loading.

Keywords: *Biomechanics; Finite element analysis; Lumbar spine; Flexibility*

Introduction

Mechanical loading of the lumbar spine is a common cause of lumbar disorders in both adults and children [1-7]. Pars stress fracture [4,6] and apophyseal ring fracture [5,7] are the most common disorders in children and are prevalent in boys and girls who are very active in sports. Pars stress fracture (i.e., lumbar spondylolysis) is caused by repetitive mechanical stress on the pars during sports activity [4,6]. Apophyseal ring fracture is often caused by mechanical loading around a posterior apophyseal ring with a weak growth plate [5].

Mechanical loading of the intervertebral disc is also a notable causative factor of disc degeneration disease, including herniated nucleus pulposus [1-3]. Furthermore, mechanical loading of facet joints and the ligamentum flavum has been shown to play an important role in the development of degenerative lumbar spinal canal stenosis. Thus, it is reasonable to assume that decreased lumbar loading would decrease the incidence of such mechanical stress-induced lumbar disorders.

Understanding the effects of the rigidness of the spine itself on mechanical stress in the lumbar spine during motion is very important for understanding the path mechanism of stress-induced lumbar disorders. However, few studies have investigated this area. Therefore, the purpose of this study was to determine the effects of spinal rigidity on mechanical stress in the lumbar spine by three-dimensional finite element analysis (FEA). Our hypothesis is that improving the flexibility of the spine would be beneficial for preventing low back pain by reducing the lumbar loading during motion.

Methods

An experimentally validated model of the L3-S1 spine segment was used in this investigation (Figure 1). We used this model extensively in our previous research on biomechanical simulation of a variety of surgical procedures and instrumentation systems, and the model details and validation data are documented in these reports [4,8-11]. Bony structures including the cortical and cancellous bone layers and the intervertebral discs were constructed using three-dimensional eight-noded hexagonal elements. All seven major ligament bundles-interspinous, supraspinous, inter transverse, posterior longitudinal, capsular, anterior longitudinal, and ligamentum flavum-were represented using truss elements, and an appropriate theoretical cross-sectional area was assigned to each ligament group. The intra-articular facet joints were simulated using unidirectional gap elements, which transferred compression forces between contact nodes along a single direction as the gap closed. Contact element stiffness was increased exponentially as the gap closed to simulate the physiologic nature of the cartilaginous layer lining the contact surfaces. The intervertebral discs were simulated as composite structures that included a solid matrix embedded with fibers in concentric rings. This structure simulated the annulus fibrosis surrounding the pseudo-fluid nucleus. The three-dimensional hexagonal elements were used to define the ground substance, and the REBAR option with no-compression behavior was used to define the reinforced fibers oriented at alternating angles 30° to the horizontal. Fiber thickness and stiffness were increased in the radial direction. The material properties of the FEA model components are detailed in (Table 1). To simulate naturally changing physiological ligament stiffness, lower stiffness at lower strains and higher stiffness at higher strains, we used a hypo elastic material that provided axial stiffness as a function of axial strain.

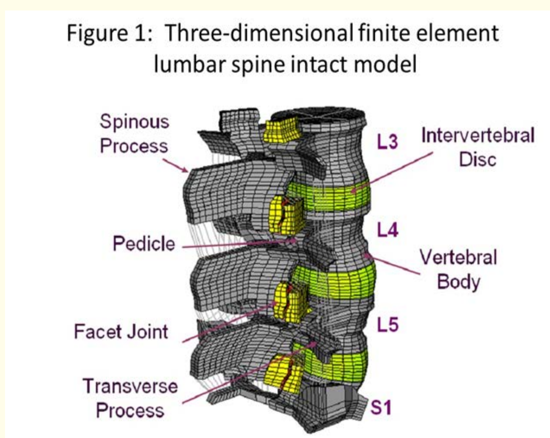

Figure 1: *The three-dimensional finite element lumbar spine model used in this study. This model has been experimentally validated and widely used in previous research.*

The FEA models were modified to simulate the following clinical cases: a) ligament stiffening (LS); b) ligament and disc stiffening (LDS); c) ligament relaxing (LR); and d) ligament and disc relaxing (LDR).

Component	Element Formulation	Modulus (MPa)	Poisson's ratio
Bony Structure			
Vertebral Cortical Bone	Isotropic, elastic hex elements	12,000	0.3
Vertebral Cancellous Bone	Isotropic, elastic hex elements	100	0.2
Posterior Cortical Bone	Isotropic, elastic hex elements	12,000	0.3
Posterior Cancellous Bone	Isotropic, elastic hex elements	100	0.2
Intervertebral disc			
Annulus (ground)	Neo Hookain, hex elements	C10 = 0.348, D1 = 0.3	
Annulus (Fiber)	Rebar	357-550	0.3
Nucleus pulposus	Incompressible fluid, cavity elements	1	0.499
Ligaments			
Anterior Longitudinal	Tension – only, Truss elements	7.8 (< 12%), 20.0 (> 12%)	0.3
Posterior Longitudinal	Tension – only, Truss elements	10.0 (< 11%), 20.0 (> 11%)	0.3
Ligamentum flavum	Tension – only, Truss elements	15.0 (< 6.2%), 19.5 (> 6.2%)	0.3
Intertransverse	Tension – only, Truss elements	10.0 (< 18%), 58.7 (> 18%)	0.3
Interspinous	Tension – only, Truss elements	10.0 (< 14%), 11.6 (> 14%)	0.3
Supraspinous	Tension – only, Truss elements	8.0 (< 20%), 15.0 (>20%)	
Capsular	Tension – only, Truss elements	7.5 (< 25%), 32.9 (>25%)	0.3
Joint			
Apophyseal joints	Non - linear soft contact, GAPPUNI elements	---	---

Table 1: *Material properties of finite element model used in this study.*

In the LS model, the stiffness of all ligaments was increased by 50%. Similarly, in the LDS model, the stiffness of the ground element of the annulus was also increased by 50%. In the LDS model, the stiffness of both the ligaments and discs was increased. In the LR model, the stiffness of the ligaments was reduced by 50%, and in the LDR model both the ligaments and the discs was reduced by 50%.

To simulate physiological loading, a follower load of 400 N was applied across the segment by attaching wire connectors to the left and right sides of each vertebral body following the curvature of the spine. A 10-Nm moment was applied to the superior surface of the L3 vertebra of the intact spine to simulate flexion, extension, left/ right bending, and left/ right axial rotation (Figure 2).

The hybrid loading protocol was simulated in the rigid and flexible models and included the 400-N follower load plus a gradually increasing bending moment applied to the top of L3. The moment increased until the overall motion of the L3-S1 segment matched that of the intact spine. The biomechanical data including the hybrid moment, segmental kinematics, intra discal pressure, and peak stress at the pars interarticularis across the segments were analyzed and compared between cases.

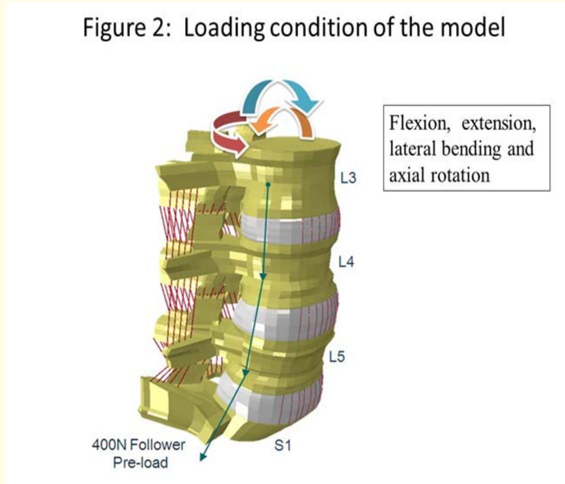

Figure 2: *Loading conditions of the spinal models. Follower loading was used to simulate the standing posture. A 10-Nm moment was applied to the superior surface of the L3 vertebra of the intact model to simulate flexion (Flex), extension (Ext), left/right bending (LB/RB), and left/right axial rotation (LR/RR) to elucidate the mechanical stresses in the lumbar spine during motion. Flex: flexion, Ext; extension, LB/RB; left bending/ right bending, LR/RR; left axial rotation/ right axial rotation. Finger to floor distance and straight leg raising before and after stretching were significantly (p < 0.01) improved after stretching.*

Results

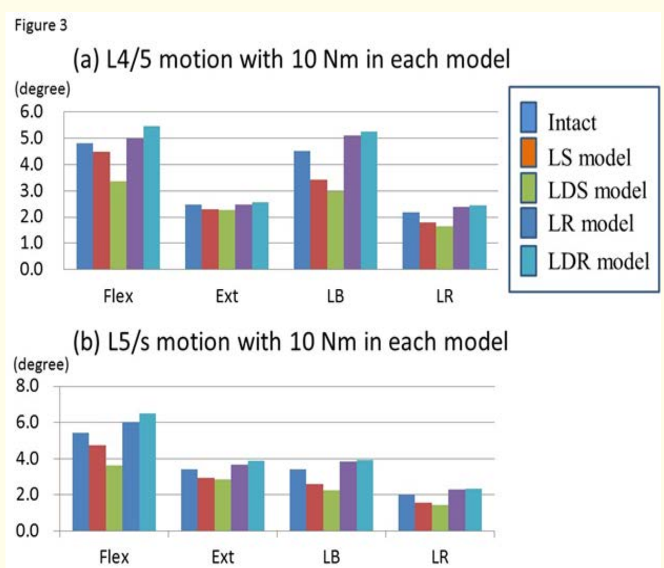

Figure 3: *L4/5(a) and L5/S1(b) motion with 10 Nm in each model. The angular motion decreased in the ligament stiffening (LS) model and ligament and disc stiffening (LDS) model compared with the intact model. On the other hand, the motion increased in the ligament relaxing (LR) model and ligament and disc relaxing (LDR) model compared with the intact model.*

At both the L4/5 and L5/S1 levels, the angular motion at 10 Nm decreased in the LS and LDS models and increased in the LR and LDR models, compared with the intact model (Figure 3). When the mean motion of the intact model was defined as 100%, the motion in the LS, LDS, LR, and LDR models was 86%, 73%, 107%, and 113%, respectively, at L4/5 and 83%, 71%, 112%, and 117%, respectively, at L5/S1. These data indicate that it was possible to create a rigid hypo mobile spine model by stiffening the ligaments and discs and a flexible hypermobile spine model by relaxing the ligaments and discs.

Next, for hybrid loading, we applied angular motion displacement of the intact model at 10 Nm to the other models. Increased torque was needed in the LS and LDS rigid spine models, while decreased torque was needed in the LR and LDR flexible spine models (Figure 4). Compared with the intact model of 10 Nm, the LS, LDS, LR, and LDR models required 14.6 Nm, 17.7 Nm, 7.2 Nm, and 5.9 Nm, respectively. Under such hybrid loading conditions, the highest stress on the disc (L4/5 and L5/S1) and pars (L4 and L5) was calculated for all models and compared.

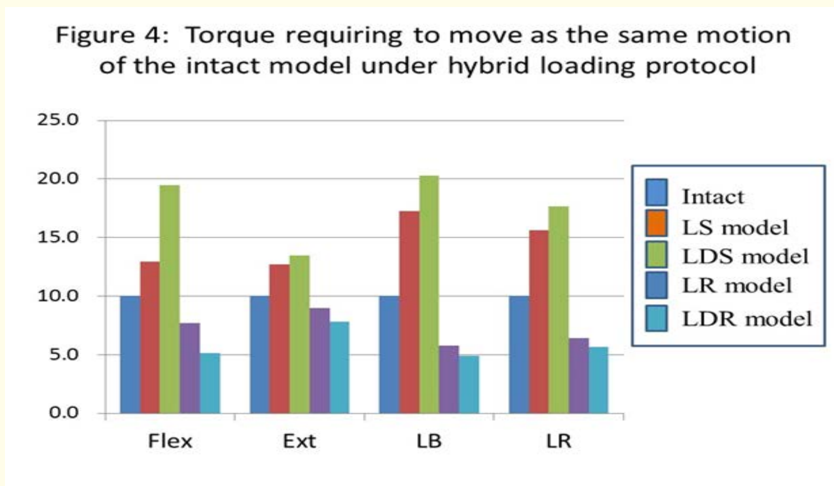

Figure 4: *Torque required to move in the same motion as that of the intact model under hybrid loading. Increased torque was needed by the rigid spine models, while decreased torque was needed by the flexible spine models.*

(Figure 5) show the maximum stress during lumbar motion under hybrid loading at the L4 and L5 pars interarticularis, respectively. The mean maximum stress of all flexion, extension, lateral bending, and axial rotation motions of the intact model at the L4 pars was 22.6 MPa. Likewise, in the LS and LDS models, the mean stress was 25.9 and 33.8 MPa, respectively. Stiffening the spine increased the pars stress to 114.8% and 149.6%, respectively. On the other hand, the stress in the flexible LR and LDR models decreased to 99.9% and 78.5%, respectively. At the L5 pars, the loading was 130.8 MPa, 151.7 MPa, 87.0 MPa, and77.3 MPa for the LS, LDS, LR, and LDR models, respectively. A notable trend was observed in L4 and L5 pars stress.

(Figure 6) show the maximum stress during lumbar motion during hybrid loading at L4/5 and L5/S1, respectively. In the LS and LDS rigid models, stress increased to 110.1% and 132.6% at L4/5 and 109.8% and 128.5% at L5/S1, respectively, compared with the intact model. In the LR flexible model, the stress decreased to 92.7% and 97.9% at the L4/5 and L5/S levels, respectively. On the other hand, in the LDR model, stress increased to 107.4% and 109.4% at the L4/5 and L5/S1 levels, respectively.

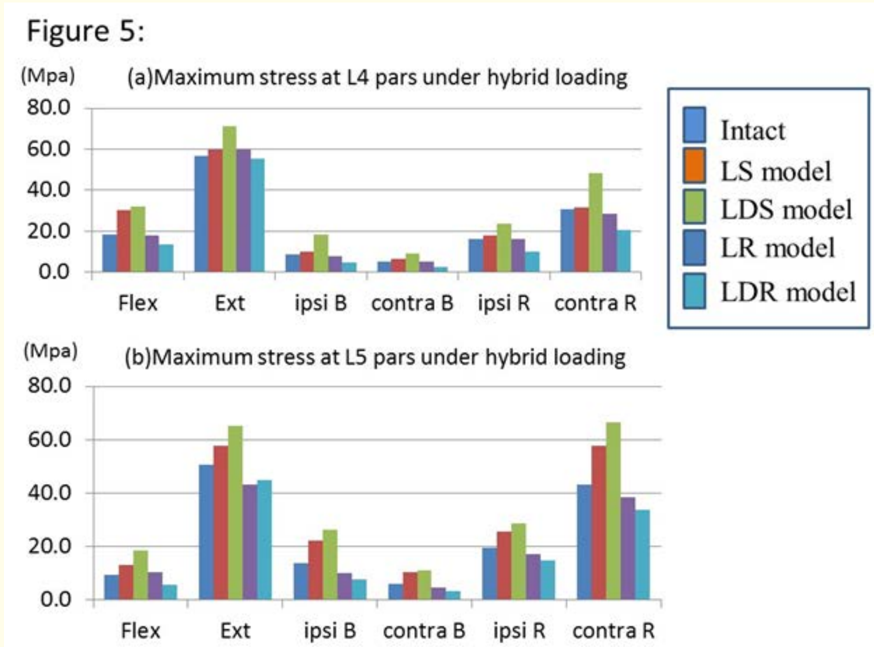

Figure 5: *Maximum stress (a) on the L4 pars and (b) on the L5 pars under hybrid loading. Stiffening the spine increased the pars stress in the rigid spine model but decreased it in the flexible model.*

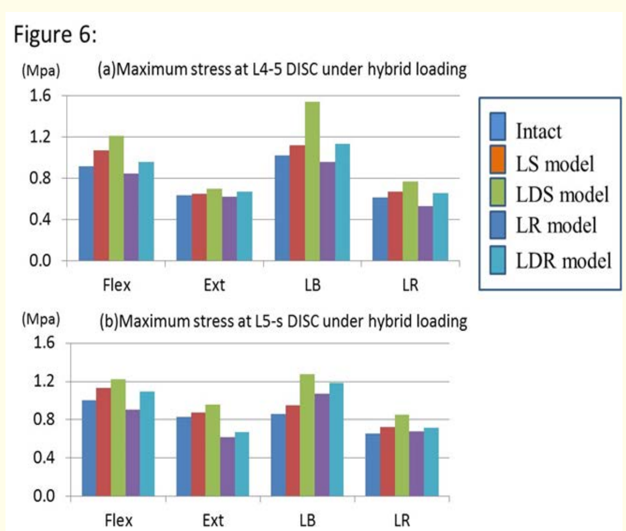

Figure 6: *Maximum stress (a) on the L4/5 disc and (b) on the L5/S1 disc under hybrid loading. Stress increased in the rigid LDR model and decreased in the flexible LR model compared with the intact model.*

Discussion

We evaluated the effects of spinal flexibility on lumbar loading as well as the effects of spinal rigidity on the mechanical stress in the spine during motion by FEA. Biomechanically, pars and disc stress during motion increased in the rigid spine and mostly decreased in the flexible spine. Thus, it can be concluded that tight hamstrings and a rigid spine are risk factors for stress-related spinal disorders. Conversely, improving the flexibility of spine will help to prevent such disorders.

Many mechanical stress-related spinal disorders exist. Spondylolysis [4,6] and apophyseal ring fracture [7] are reported to be the major two disorders in the pediatric population, while disc degenerative disorder is reported to be most common disorder in adults [1-3]. The importance of preventing stress-related disorders through improving flexibility is well known; however, only a few studies have investigated whether or not a rigid spine is harmful. To clarify this issue, we created four kinds of FEA models here, each with different kinds of stiffness, by changing the material properties of the spinal ligaments.

In the biomechanical investigation, we applied hybrid loading and calculated mechanical stresses at the pars by means of deformation control. We applied the same range of motion of the intact model to the other four models, and as we suspected, mechanical stress on the pars increased in the rigid models (LS and LDS) and decreased in the flexible models (LR and LDR) (Figures 5-6). These results strongly suggest that pediatric athletes with a rigid spine have a higher chance of developing pars stress fracture (i.e., spondylolysis) than those with a flexible spine. Based on these findings, we propose the importance of spine flexibility in young athletes for preventing pars stress fracture. A pars fracture can be healed osseously by conservative treatment with a brace and rest [12], but recurrence is still a potential problem after return to the sport. Thus, significant physical therapy including stretching of the spine and hamstrings is crucial for pediatric patients with spondylolysis before returning to sports activity.

Disc stress was calculated in the same manner in the biomechanical investigation. Disc pressure increased in the rigid LS and LDS models (Figures 5-6). Therefore, as with pars fracture, disc degeneration disease may be facilitated by a rigid spine. Although spine stretching is beneficial for preventing disc degeneration, simply having a flexible spine does not always decrease disc pressure. On the other hand, because the pressure increased unexpectedly in the LDR model, it is difficult to conclude that a flexible spine is beneficial for preventing degenerative disc disease. The differences between flexibility and laxity are well established, and many reports have shown that joint laxity and hypermobility are related to a higher incidence of sports-related injuries [13,14]. The LDR model, in which stiffness of the ligaments and discs was reduced to 50%, may in fact be a laxity model not a flexible model. This is a limitation of our biomechanical investigation.

Conclusion

Subjects with a rigid (hypo mobile) spine would have a higher chance of developing stress-related spinal disorders such as lumbar spondylolysis and degenerated disc disease. Taken together, improving the flexibility of the spine is beneficial for preventing low back pain.

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Conflict of Interest: All authors declare that we have no conflict of interest.

Ethical approval: This article does not contain any studies with human participants or animals performed by any of the authors.

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