Stress Analysis of the Lower Lumbar Spine Three-joint Complex According to Different Pelvic Incidences

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Research article

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Abstract

Background: Pelvic incidence is closely related to degeneration of the facet joint and intervertebral disc and is related to the orientation of the facet joints. Currently, very few studies have been conducted on the force analysis of the three-joint complex in patients with different pelvic incidence measurements under different sports postures. We designed this study to better assess the influence of pelvic incidence on the stress of the lumbar three-joint complex. Finite element analysis can provide a biomechanical basis for the relationship between different pelvic incidences and degenerative diseases of the lower lumbar spine.

Methods: We developed three nonlinear finite element models of the lumbar spine (L1-S1) with different pelvic incidences (27.44°, 47.05°, and 62.28°) and validated them to study the biomechanical response of facet joints and intervertebral discs with a follower preload of 400 N, under different torques (5 Nm, 10 Nm, and 15 Nm), and compared the stress of the three-joint complex of the lower lumbar spine (L3-S1) in different positions (flexion-extension, left-right bending, and left-right torsion).

Results: In the flexion position, the stress of the disc in the low pelvic incidence model was the largest among the three models; the stress of the facet joint in the high pelvic incidence model was the largest among the three groups during the extension position. During torsion, the intradiscal pressure of the high pelvic incidence model was higher than that of the other two models in the L3/4 segment, and the maximum von Mises stress of the annulus fibrosus in the L5/S1 segment with a large pelvic incidence was greater than that of the other two models.

Conclusions: Pelvic incidence is related to the occurrence and development of degenerative lumbar diseases. The stress of the lower lumbar facet joints and fibrous annulus of individuals with a high pelvic incidence is greater than that of individuals with a low pelvic incidence or a normal pelvic incidence. Although this condition only occurs in individual segments, to a certain extent, it can also reflect the influence of pelvic incidence on the force of the three-joint complex of the lower lumbar spine.

Background

Pelvic incidence (PI) is an important component of the spine-pelvic sagittal parameters; it refers to the angle between the line drawn from the midpoint of the upper endplate of S1 to the midpoint of the femoral head and the line perpendicular to the upper endplate of S1. The concept of PI was first proposed by Duval-Beaupère et al in 1992 [1]. Since then, the orthopaedic academic community has conducted an increasing amount of research on PI. In the early stages, the study of PI mainly focused on two-dimensional imaging research. A study by Weinberg et al [2] involving 880 cadaver specimens showed that the normal range of PI was 46.0° ± 11.0°. The PI is of great significance in maintaining the sagittal balance of the human body. Additionally, Diebo et al [3] also affirmed the importance of PI in maintaining sagittal balance in the human body via two-dimensional imaging.

With the advancement of scientific research and technology, the study of PI has overcome the limitations of two-dimensional imaging. Currently, the study of PI is mainly conducted through in vitro studies of specimens. The research objectives are mainly focused on PI and sagittal spine position, its relationship with balance [4], and the size of the PI and its influence in in vivo studies. A study by Imagama et al[5] showed that a low PI is a risk factor for intervertebral disc degeneration and lumbar osteophyte formation. Low PI (PI < 51°) is a risk factor for lumbar osteophyte formation and intervertebral disc degeneration. Jentzsch et al[6] suggested that PI is linearly related to the longitudinal ligament (LL). A greater PI indicates a greater LL. At the same time, the study also showed that the PI is related to lumbar facet osteoarthritis (FJOA) and the lumbar facet joints. A greater PI indicates a greater probability of lower lumbar facet joint arthritis and a smaller sagittal angle of the facet joints at the L5/S1 segment. Many studies have shown that the PI correlates with spinal degeneration [4–8].

However, we questioned the mechanism by which PI affects spinal degeneration. Some authors hold that different PIs affect the movement of the three-joint complex and finally lead to spinal degeneration. The three-joint complex is the basic unit of spine activity, which is composed of facet joints and lumbar intervertebral discs of the same segment. Therefore, the facet joints, together with the intervertebral discs and spinal ligaments, bear most of the compressive load of the spine [9]. The 'three-joint complex' of the lumbar spine can maintain the normal movement of the spine. The latest biomechanical studies on the 'three-joint complex' have shown that the load sharing between facet joints and intervertebral discs is complicated by the influence of spinal posture and motion, and the intervertebral discs usually have the main weight-bearing role. Under normal circumstances, owing to different postures, the facet joint surface carries 6–30% of the axial compressive load [10], and in the case of severe intervertebral disc space narrowing, the facet joints can withstand up to 70% of the axial load [11–12]. However, very few studies have been conducted on the force analysis of the three-joint complex in patients with different PI measurements under different sports postures.

To further explore the mechanism by which PI affects the stress of the three-joint complex of the lumbar spine under different sports postures, we designed an experiment to simulate the stress of the lower lumbar spine three-joint complex, which has different PIs under different sports postures.

Methods

Development of the model

Three healthy volunteers with different PIs were selected for this study. Exclusion criteria included the presence of other spinal diseases, including structural spinal deformity and rheumatic immune system diseases, spondylolisthesis of the isthmus, severe osteoporosis, pregnancy, and previous spine or hip surgery. All volunteers underwent lumbar spine radiography and computed tomography (CT) examinations when admitted to the hospital. The PI of the three volunteers was 27.44°, 47.05°, and 62.28°. The age of the volunteers ranged from 33 to 38 years (Table 1). The experimental design was approved by the appropriate institutional review board prior to the initiation of the study. Written consent was obtained from each participant prior to the study.

Table 1
Material properties of the finite element model

Anatomic structure	Modulus of elasticity(MPa)	Poisson's ratio
Osseous cortex	12000	0.3
Cancellous bone	100	0.2
Annulus fibrosus	500	0.3
End plate	24	0.4
Nucleus pulposus	1666.7	-
Ligamenta longitudinale anteriust	20	0.3
Ligamenta longitudinale posterius	70	0.3
Ligamentum flavum	50	0.3
Ligamenta interspinalia	28	0.3
Ligamenta supraspinale	28	0.3
Articular capsule ligament	50	0.3
ligamenta intertransversaria	20	0.3

Three nonlinear finite elements models of complete lumbar spinal segments (L1-S1) were generated on the basis of geometrical reconstruction from the CT scans (Fig. 1). The image segmentation and reconstruction of the geometrical model were completed using medical image processing software (Mimics 19.0; Materialise Technologies, Leuven, Belgium) and reverse engineering and scanning software (Geomagic Studio 10.0; Geomagic Inc., North Carolina, USA), respectively. The mesh was generated using CAE pre-processing software (Hypermesh 11.0; Altair Engineering Corp, Michigan, USA). The intervertebral disc was made up of the nucleus and annulus, and the annulus was assumed to be composed of a ground substance reinforced by a collagen fibre network (Fig. 2). Seven ligaments were included: the anterior (ALL) and posterior (PLL) longitudinal ligaments and the intertransverse (ITL), flavum (FL), supraspinous (SSL), interspinous (ISL), and capsular (CL) ligaments. The facet cartilage was represented by three layers of hex elements with inhomogeneous thickness, and the gap between the facet articular surfaces was approximately 0.1 mm. The distribution of cartilage thickness in each facet was in accordance with the findings of Woldtvedt et al [13, 14].

Facet joint degeneration grading, vertebral body height and intervertebral space height, facet sagittal angles, and disc degeneration degree were obtained from CT and magnetic resonance imaging. The degree of degeneration in the L3–S1 facet joints and discs was graded according to the Weishaupt scales [15] and the Pfirmann classification [16].

Biomechanical study of three-joint complex

An additional follower preload of 400 N was applied to each segment and for each loading case to determine the effect of the preload on the three-joint complex. Then, pure moments of 5, 10, and 15 Nm in different postures (flexion, extension, left bending, right bending, left torsion, and right torsion) were imposed on a node that was coupled with the upper endplate of L1. The stress of the three-joint complex was measured under three torques of 5, 10, and 15 Nm, and the average stress on the facet joints of each segment of L3-S1, the internal pressure of the intervertebral disc, and the fibrous annulus of each segment maximum von Mises stress were recorded (Fig 3). Finally, we compared the stress of the three-joint complex of each segment of L3-S1 in different groups.

Results

These models were validated before analysing the stress of the three-joint complex (Table 2). The stiffness results measured from three models of different PI were similar to the results of previous studies from cadavers [17–19] (Fig. 4). We believe that the difference is due to the differences in the model details.

Table 2
Stiffness comparison with the results of the list literature

	Moment(Nm)	Anteflexion(N·m/°)	Postextension(N·m/°)	Left rotation(N·m/°)	Left bending(N·m/°)
Heth et al. [17]	10	1.1	2.35	1.33	2.61
Liu et al. [18]	10	2.35	3.58	2.86	8.98
Yamamoto et al.[19]	10	1.75	3.22	2.44	5.66
Low PI	10	2.02	3.05	2.53	6.18
Normal PI	10	1.83	3.18	2.58	6.43
High PI	10	1.76	3.31	2.46	5.74

The age and body mass index (BMI) of the three volunteers were similar, and all were women (Table 2). We measured the pelvic parameters (Table 3), the height of the vertebral body, and the height of the intervertebral space (Table 4) and graded the degree of degeneration of the facet joints (Table 5). In general, the three models were comparable.

Table 3
Demographics for the three groups of different PI

Group	Age	Gender	SS(°)	PT(°)	PI(°)	BMI(kg/㎡)
Low PI	36	F	18.03	9.95	27.44	21.48
Normal PI	38	F	32.56	16.27	47.05	20.54
High PI	33	F	46.26	15.38	62.28	20.87

Table 4
Weishaupt classification of facet joint degeneration at L3-S1 of different groups

Group	L3/4		L4/5		L5/S1	
	Left	Right	Left	Right	Left	Right
Low PI	0	0	1	1	1	1
Normal PI	1	1	1	1	1	1
High PI	1	1	1	1	1	1

Table 5
The orientation of facet joint at L3-S1 of different groups

Group	L3/4		L4/5		L5/S1	
	Left	Right	Left	Right	Left	Right
Low PI	51.88	49.32	50.37	49.54	44.88	44.76
Normal PI	48.93	52.3	49.32	46.24	46.15	47.32
High PI	46.64	52.54	47.42	44.78	44.23	42.56

Table 6
The vertebral height and intervertebral space height at L3-S1 of different groups

Group	Vertebral height (mm)			Intervertebral space height (mm)		
	L3	L4	L5	L3/4	L4/5	L5/S1
Low PI	21.50	21.08	20.93	9.51	9.41	10.19
Normal PI	21.12	21.70	21.33	11.12	10.31	9.76
High PI	23.09	24.94	25.24	11.06	9.25	9.87

In the flexion position, the average von Mises stress of the facet joint decreased with the increase in torque. In the L3/4 segment, the average stress of the right facet joint of the low PI group was the smallest when the torque was 15 Nm (Fig. 5), which was 0.57 MPa. In the L4/5 segment, the average stress of the facet joints in the low PI group was lower than that in the L3/4 segment, and the average stress of the facet joints in the normal PI group and in the high PI group was smaller than that of the low PI group. In the L5/S1 segment, the differences between the three groups were irregular, and the average stress of the facet joints in each group did not change significantly with the increase in torque. As the torque increased during flexion, the average stress of the facet joint gradually decreased, but this trend was obvious in the upper two segments of the lower lumbar spine (L3/4 and L4/5). In the flexion position, the average stress of the facet joints of the L5/S1 segment was less affected by the torque. The intradiscal pressure and maximum von Mises stress of the annulus fibrosus gradually increased with the increase in torque (Figs. 6 and 7). The intradiscal pressure in the small PI group was the highest among the three groups.

In the extension position, with the increase in torque, the average stress of the facet joints of each subject in the three segments typically showed an increasing trend (Fig. 8). The average stress of the facet joints in the high PI group was greater than that in the low PI group or the normal PI group. This phenomenon was significant in the L3-S1 segments. In the extension position, the intradiscal pressure of the low PI group at the L3/4 and L5/S1 segments was greater than that of the other two groups (Fig. 9), and the maximum von Mises stress of the annulus fibrosus in the low PI group was the largest in the three groups under the same loading conditions (Fig. 10).

In the left bending position, the average von Mises stress of the left facet joint was greater than that of the right side. This phenomenon existed in the three different PI groups (Fig. 11), and as the torque increased, the average stress of the left facet joint also increased. On the contrary, the average stress of the contralateral facet joint decreased with the increase in torque. In the three groups, the left facet joint was still significantly affected by torque change. The increase in torque resulted in an increase in the average stress of the concave facet joint, while the average stress on the convex side decreased. In the L5/S1 segment, this trend was no longer significant, especially when the average stress of the right facet joint was only minimally affected by the torque. This feature is more pronounced in the L3/4 and L4/5 segments but less obvious in the L5/S1 segment. The intradiscal pressure and the maximum von Mises stress of the annulus fibrosus in the left bending position changed with torque, but there was no significant difference between the groups (Figs. 12 and 13).

In the right bending position, the average stress of the right facet joints in the three groups was greater than that of the left side (Fig. 14), and the change in the average stress of the left and right facet joints was exactly opposite to that of the left side bending. There were no regular differences in the internal disc pressure and the maximum von Mises stress of the annulus fibrosus in each segment (Figs. 15 and 16). In the lateral bending position, the difference in the average stress of the facet joints between the different PI groups under the same load conditions and segments was not large, and there was no obvious rule among the three groups.

In the left torsion position, the right facet joint is blocked, causing the average stress of the right facet joint to be significantly greater than the average stress of the left facet joint, and as the torque increases, the average stress of the bilateral facet joints also gradually increases. In the L3/4 segment, compared with the other two groups, the average stress of the facet joints in the high PI group was the largest, reaching a maximum in the right facet joint of the high PI group under a torque of 15 Nm, which was 4.21 MPa (Fig. 17). The increase in torque leads to an increase in the average stress of the facet joints, and PI has less influence on the average stress of the facet joints in the left-torsion position; in addition, the intradiscal pressure of the high PI group was the largest among the three groups (Fig. 18), and the maximum von Mises stress of the annulus fibrosus had no obvious regularity (Fig. 19). In the right-torsion position, the average stress of the left facet joints in the three groups was greater than that on the right side (Fig. 20). The maximum intradiscal pressure of the L3/4 segment (Fig. 21) and the maximum von Mises stress of the L5/S1 segment occurred in the high PI group (Fig. 22). We found that PI had little effect on the average stress of the facet joints, and torque had a different effect on the average stress of the facet joints between the upper (L3/4) and lower segments (L5/S1). Torque had a greater effect on the average stress of the facet joints in the L3/4 segment than in the L5/S1 segment.

Discussion

Three healthy volunteers with different PIs (27.44°, 47.05°, and 62.28°) were selected to perform biomechanics experiments. The age, sex, and BMI of the three volunteers were not significantly different. Additionally, the height of the vertebral body, the height of the intervertebral space, and the degree of degeneration of the facet joints of the three models also showed no obvious divergence. Therefore, the authors hold that the three models were comparable, and the choice of experimenter was reasonable. The force of the three-joint complex of the low lumbar spine under different motion states is realistically simulated using the latest biomechanics software, namely Mimics, Geomagic studio, and CAE. Although the current research is limited to imaging and two-dimensional research [20–23], research reports similar to ours are rare. However, combined with our experimental design, object selection, and experimental equipment considerations, the authenticity and rationality of our experimental results are beyond doubt.

Some scholars have found that PI may cause degeneration of the intervertebral discs or facet joints. However, they did not clarify the specific mechanism of PI and the three-joint complex. Through the experiments we designed, we confirmed that the stress of the disc in the low PI model was the largest among the three models in the flexion position, the stress of the facet joint in the high PI model was the largest among the three groups in the extension position, and the intradiscal pressure of the high PI model was higher than that of the other two models in the L3/4 segment in the torsion position. Obviously, a low PI will cause intervertebral disc degeneration in hyperflexion, and a high PI will cause facet joint degeneration in hyperextension. This result is consistent with the research findings of other scholars. Ferrero et al [24] reported a close relationship between

PI and degenerative lumbar spondylolisthesis (DLS), and related studies [25–27] found that the PI of patients with lumbar spondylolisthesis was higher than that of healthy people, and the greater the PI, the more severe the spondylolisthesis. Interestingly, degeneration of the lumbar facet joints is more serious in patients with a high PI and lumbar spondylolisthesis [28, 29, 30]. However, the performance of intervertebral disc degeneration is more prominent in people with low PI [31]. Therefore, we believe that our research results are consistent with the conclusions of spinal degeneration reached by other scholars, and they better confirm the mechanism of PI leading to spinal degeneration.

In terms of the clinical application of our findings, we believe that individuals with too small a PI bear most of the load of the lumbar spine through the anterior and middle columns of the lumbar spine, which may lead to premature lumbar disc degeneration and lumbar disc herniation, especially in long-term flexion. Therefore, we recommend that people with a small PI reduce the time spent in flexion to reduce the probability of lumbar disc herniation. In contrast, an individual with a high PI has a greater axial load on the posterior structure of the lumbar spine and is more likely to experience facet joint degeneration. Severe degeneration of the facet joints of the lower lumbar spine may cause diseases such as lumbar spondylolisthesis and osteoarthritis (OS) [32, 33]. Therefore, we recommend that people with a high PI reduce the time spent in extension to reduce the probability of DLS and OS. Based on this experimental study, we believe that PI is valuable for guiding the prevention of clinical spine diseases.

However, this study has some limitations. First, only women were included in the study through screening, and the age range was 33–38 years. Further studies could increase the sample size and study both sexes. Second, the three torques of 5 Nm, 10 Nm, and 15 Nm do not have a significant impact on the stress of the model, and more can be done on the axial load in the later period. Grouping should be performed, simulating the human body's activities in various functional positions under different weight-bearing conditions and further analysing the force of the important load-bearing parts of the lumbar spine under the weight-bearing state to obtain more comprehensive biomechanical data of the lumbar spine three-joint complex.

Conclusions

PI affects the force of the three-joint complex of the lumbar spine. The stress on the small joints and annulus of the lumbar spine of individuals with a large PI is greater than that of individuals with a small or normal PI, although this condition only occurs in some segments. However, it can also reflect the influence of PI size on the force of the lower lumbar spine three-joint complex to a certain extent. This study analyses the stress of the three-joint complex of the lower lumbar spine from different PIs and provides a reference for the connection between spine-pelvic sagittal parameters and spinal degenerative diseases.

Abbreviations

ALL: anterior longitudinal ligament; CL: capsular ligament; CT: computed tomography; DLS: degenerative lumbar spondylolisthesis; FJOA: facet osteoarthritis; FL: flavum ligaments; ISL: interspinous ligament; ITL: intertransverse ligament; OS: osteoarthritis; PI: pelvic incidence; PLL: posterior longitudinal ligament; SSL: supraspinous ligament

Declarations

Ethics approval and consent to participate:

The experimental design was approved by the appropriate institutional review board prior to the initiation of the study. Written consent was obtained from each participant prior to the study.

Consent for publication:

Not applicable

Availability of data and materials:

Not

Competing interests:

The authors declare that they have no competing interests.

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Authors' contributions:

QL and JY conceived and designed the study. JY performed the experiments. ZZZ and JY analyzed the data. QL wrote the paper. ZMW reviewed and edited the manuscript. All authors read and approved the manuscript.

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QL, JY, ZZZ and JY are masters' graduate students. ZMW is Ph D

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Figures

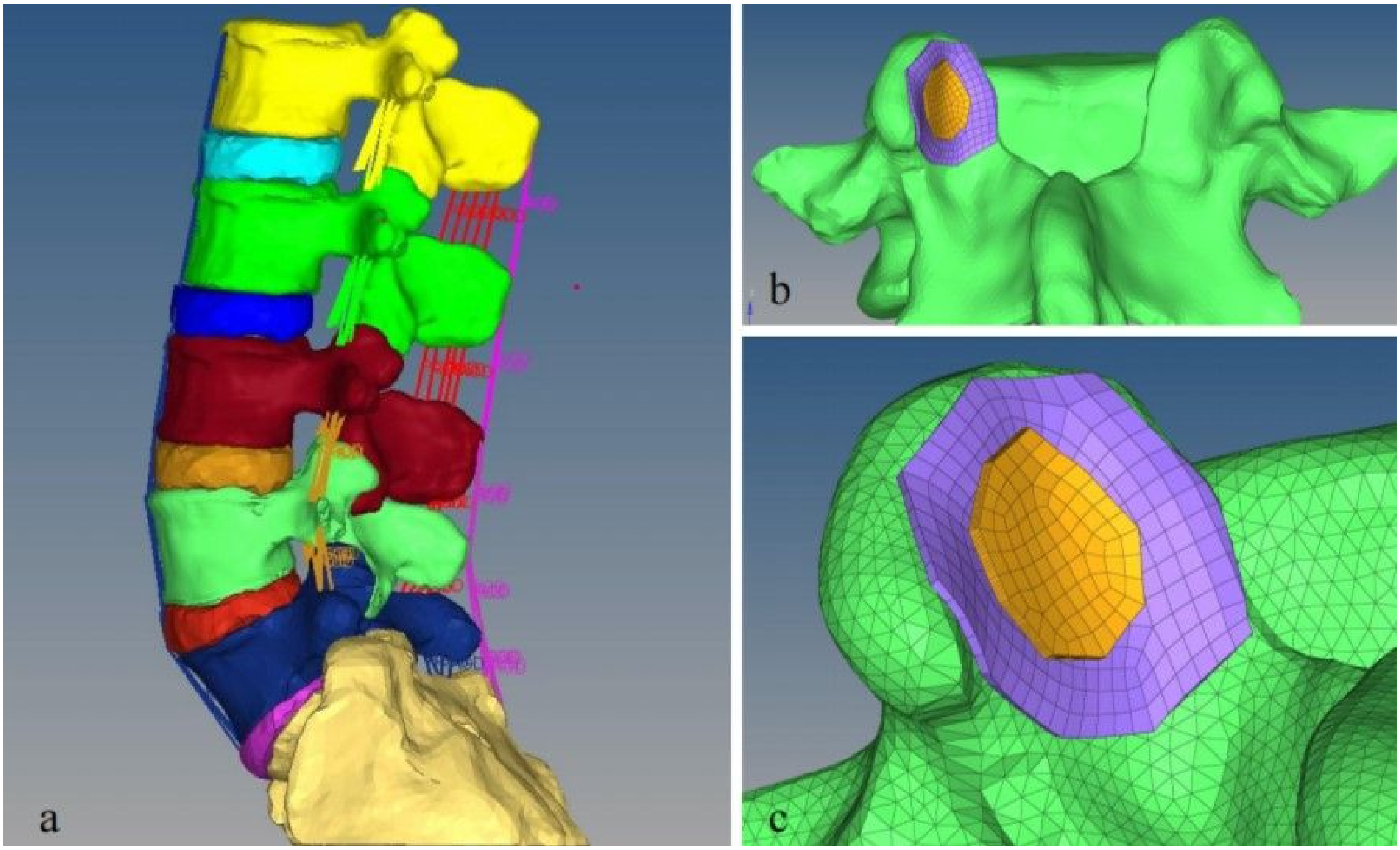

Figure 1

Lumbar model and details of the facet joint.

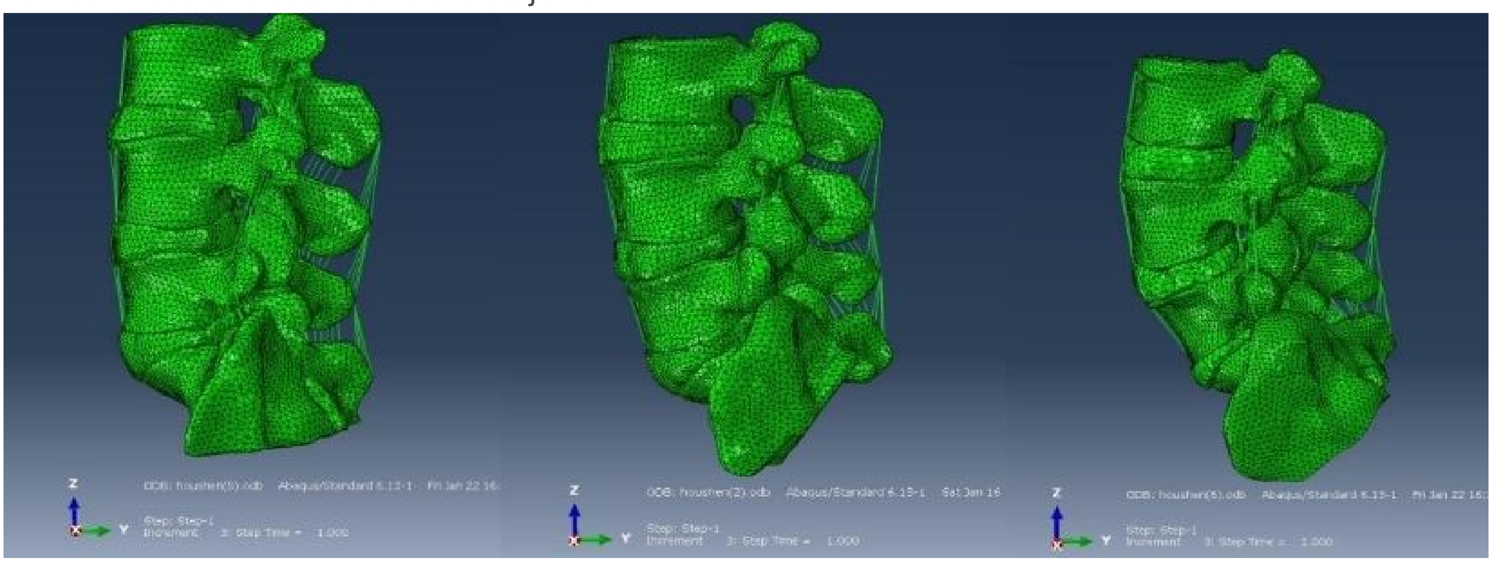

Figure 2

Finite element models with different PI (low PI, normal PI and high PI from left to right).

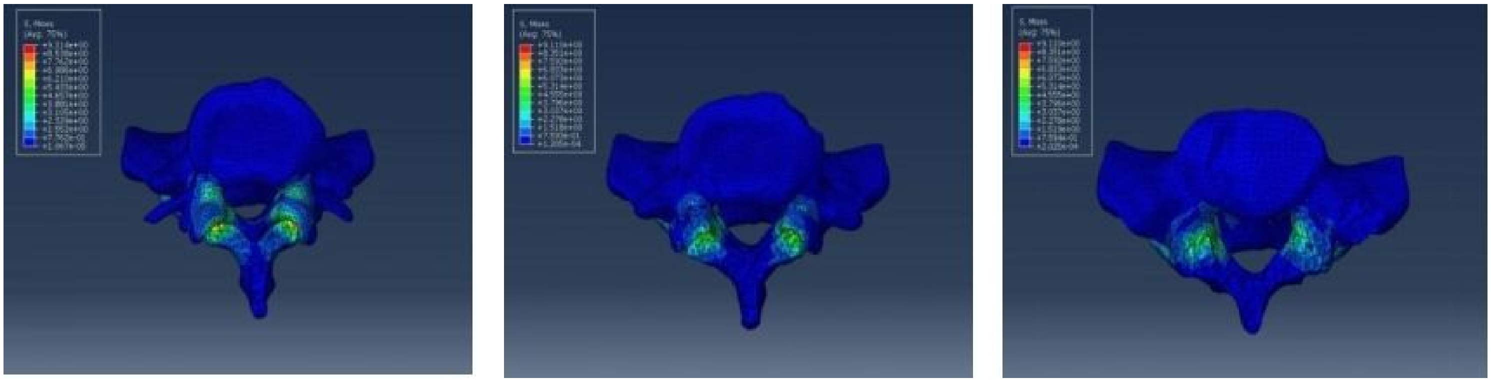

Figure 3

Facet joint stress of low PI model (L3/4, L4/5, L5/S1 facet joints from left to right).

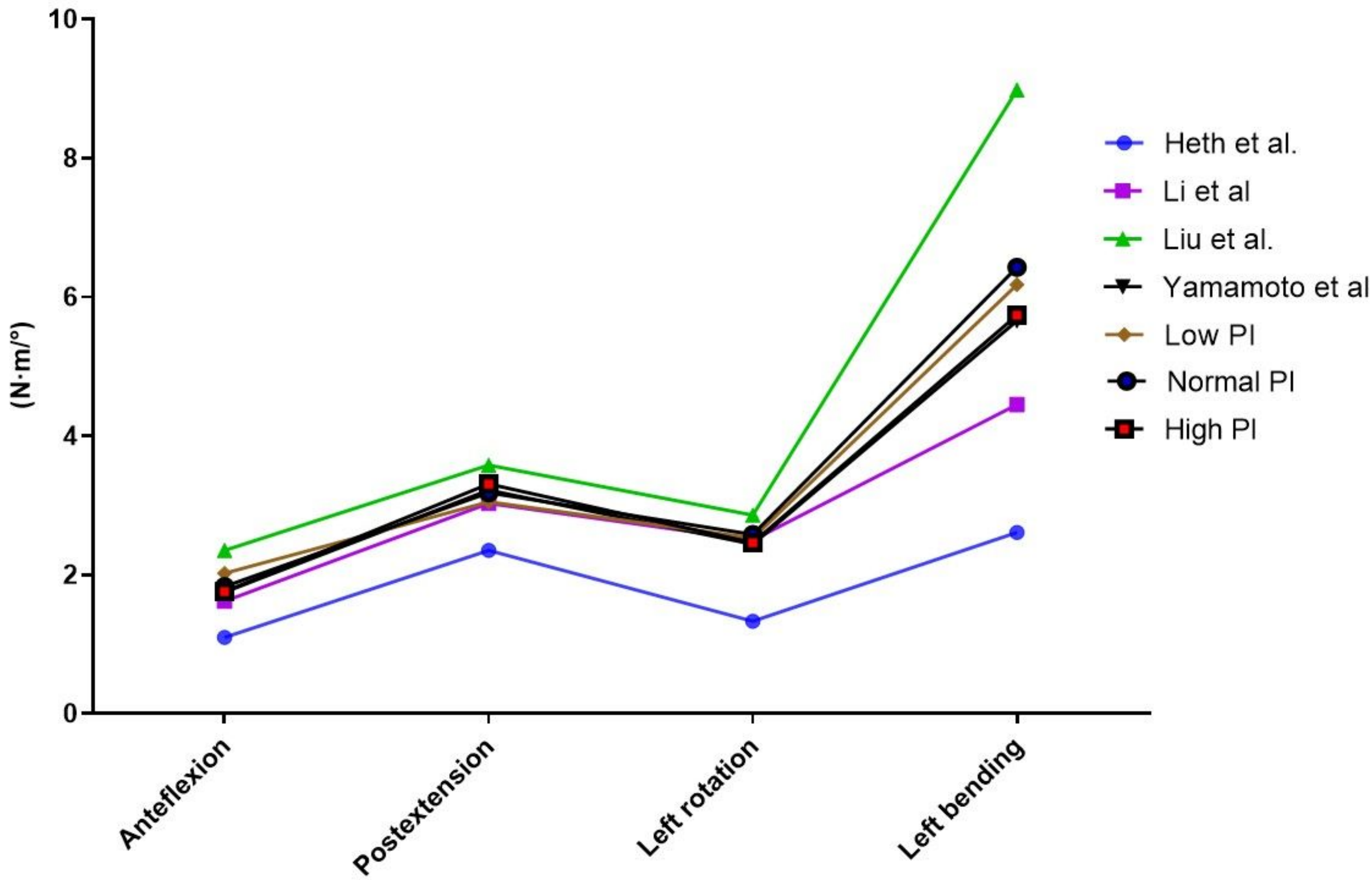

Figure 4

Comparison of results by the current lumbar models and previous studies.

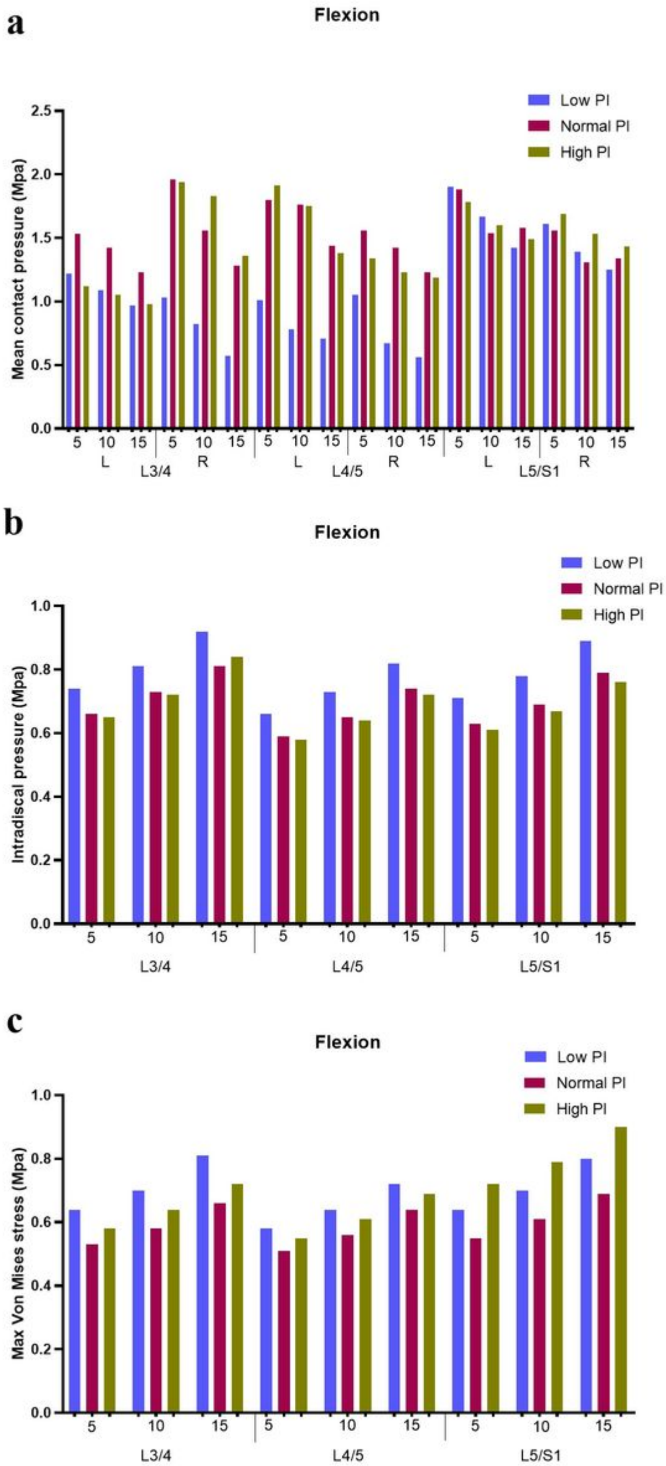

Figure 5

a:Mean contact pressures of facet joints at different levels in flexion posture under various torques. b:Intradiscal pressure at different levels in flexion posture under various torques. c:Max Von Mises stress of annulus at different levels in flexion posture under various torques.

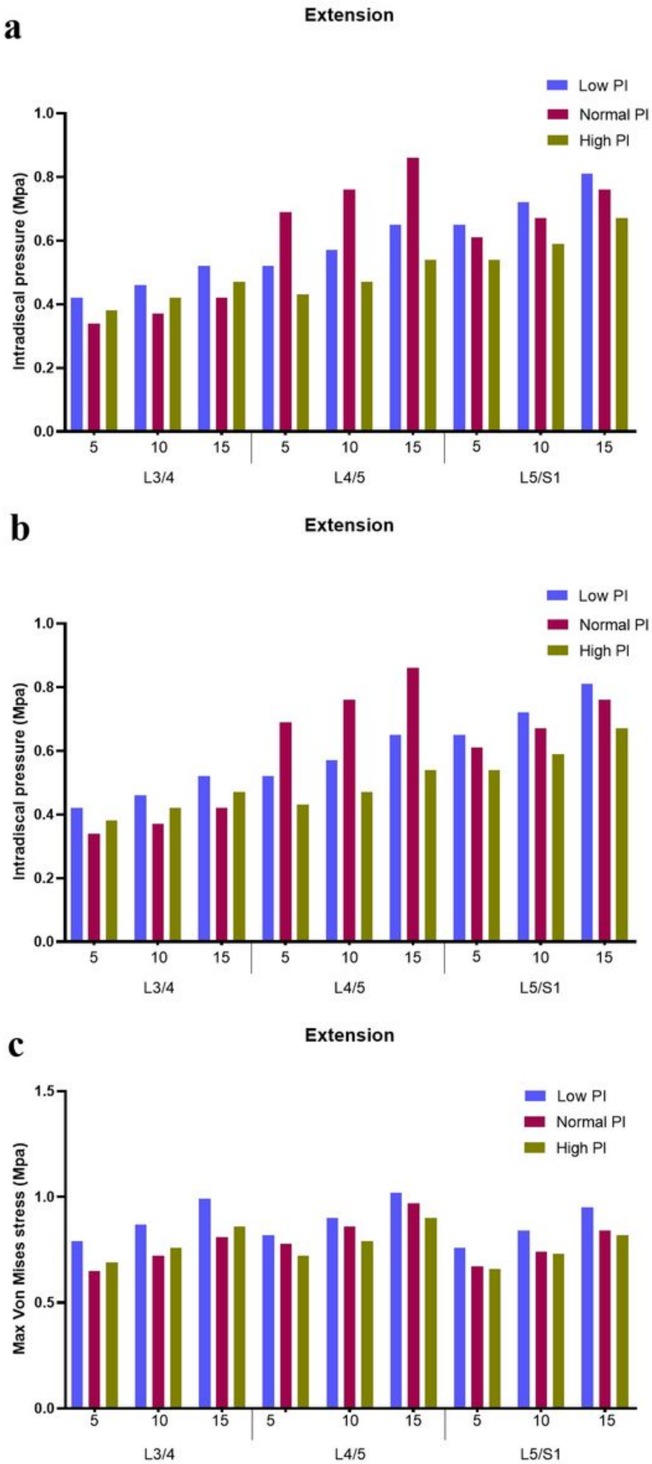

Figure 6

a:Mean contact pressures of facet joints at different levels in extension posture under various torques. b:Intradiscal pressure at different levels in extension posture under various torques. c:Max Von Mises stress of annulus at different levels in extension posture under various torques.

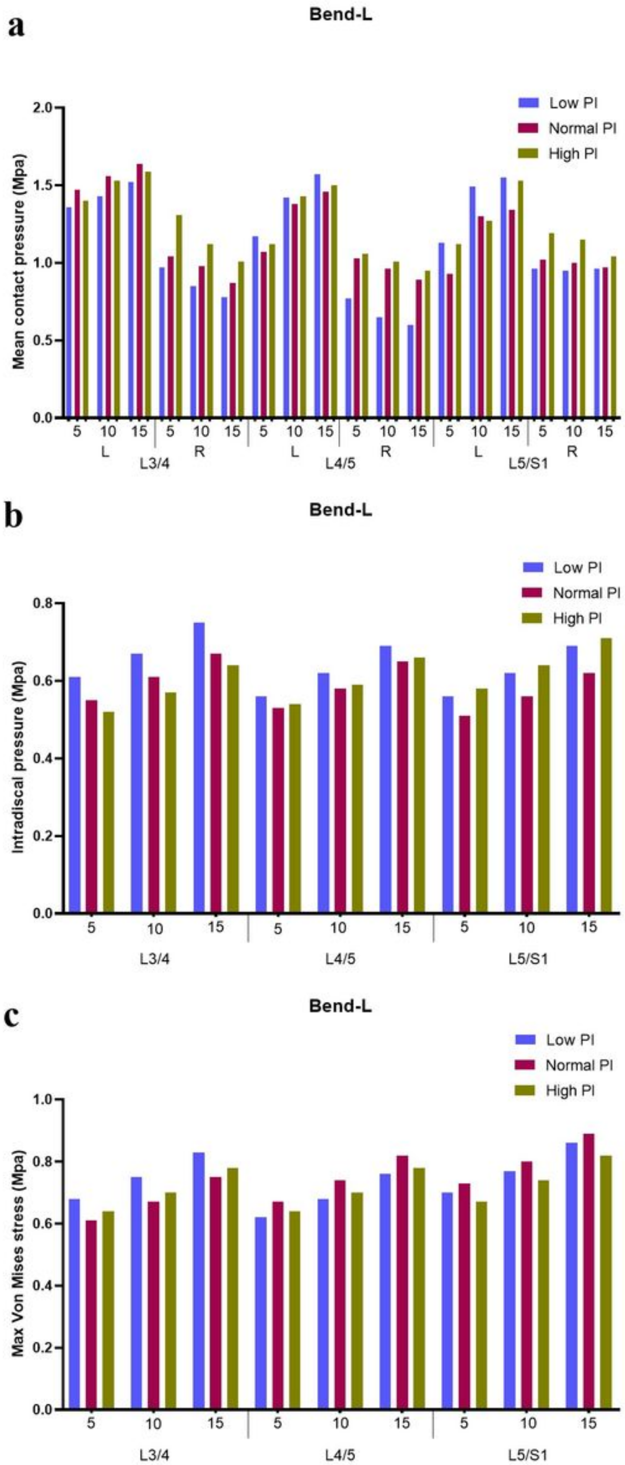

Figure 7

a:Mean contact pressures of facet joints at different levels in left-bending posture under various torques. b:Intradiscal pressure at different levels in left-bending posture under various torques. c:Max Von Mises stress of annulus at different levels in left-bending posture under various torques.

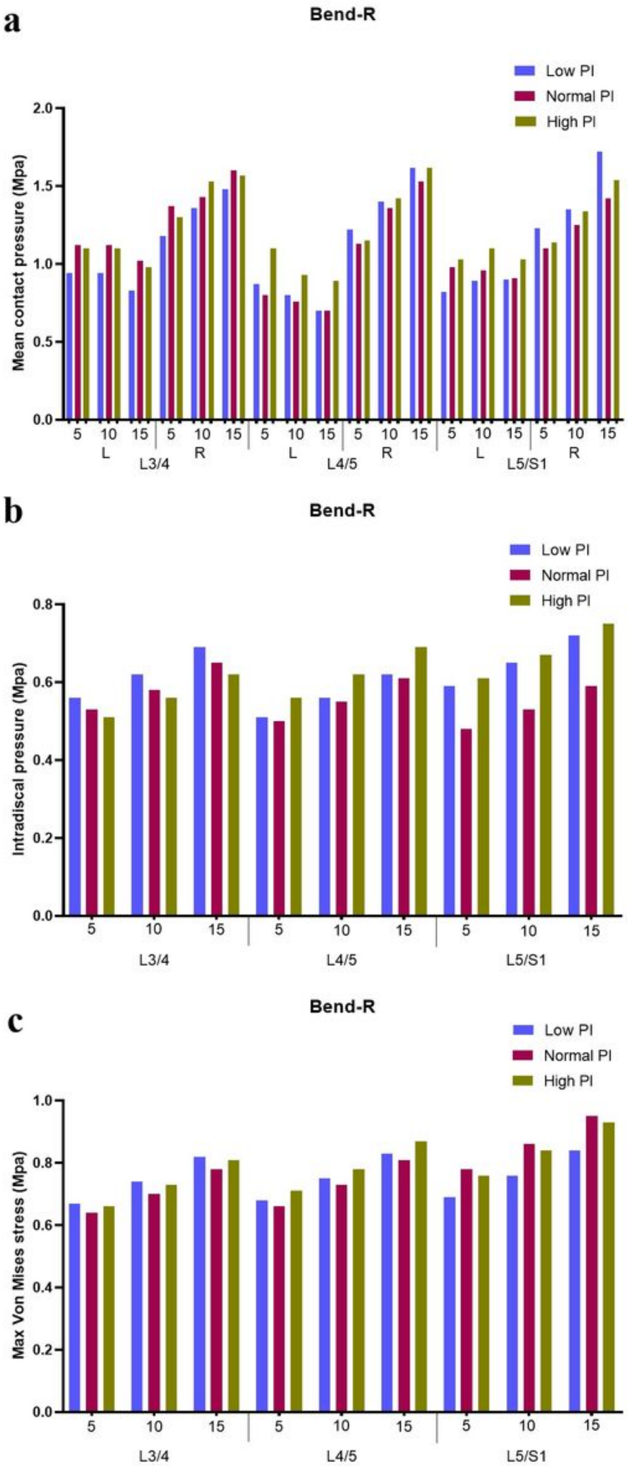

Figure 8

a:Mean contact pressures of facet joints at different levels in right-bending posture under various torques. b:Intradiscal pressure at different levels in right-bending posture under various torques. c:Max Von Mises stress of annulus at different levels in right-bending posture under various torques.

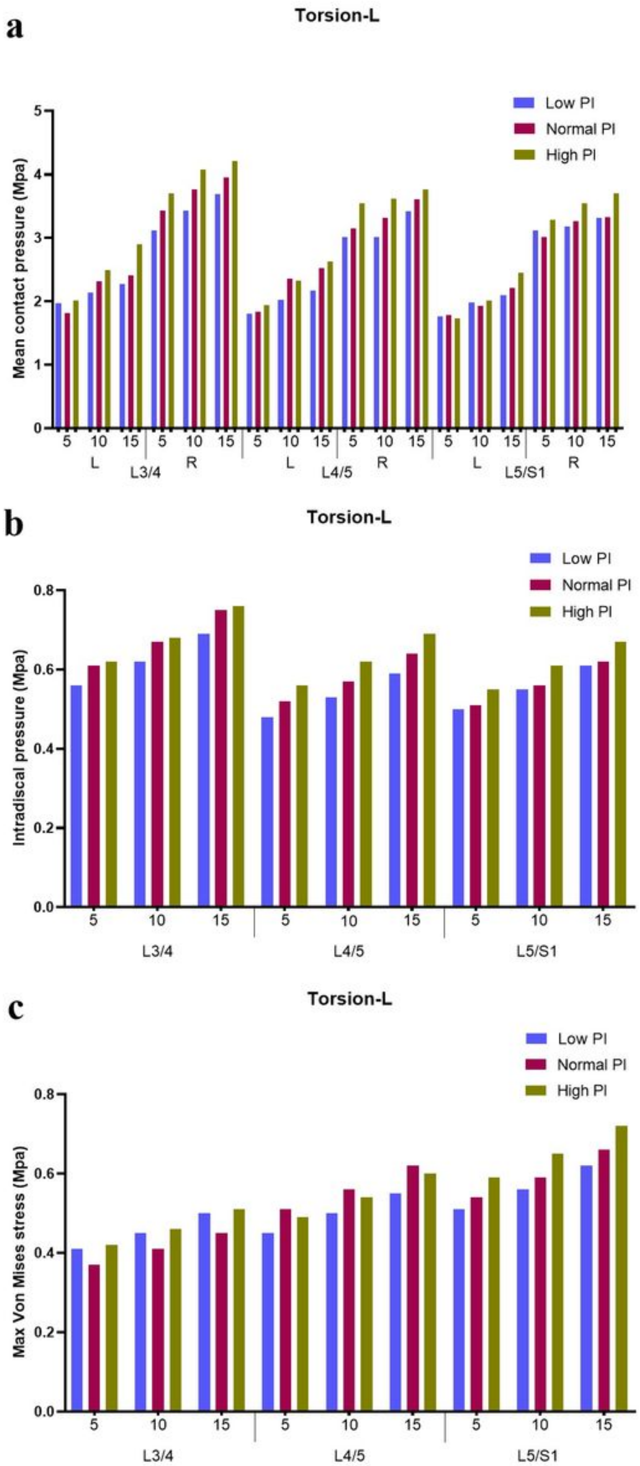

Figure 9

a:Mean contact pressures of facet joints at different levels in left-torsion posture under various torques. b:Intradiscal pressure at different levels in left-torsion posture under various torques. c:Max Von Mises stress of annulus at different levels in left-torsion posture under various torques.

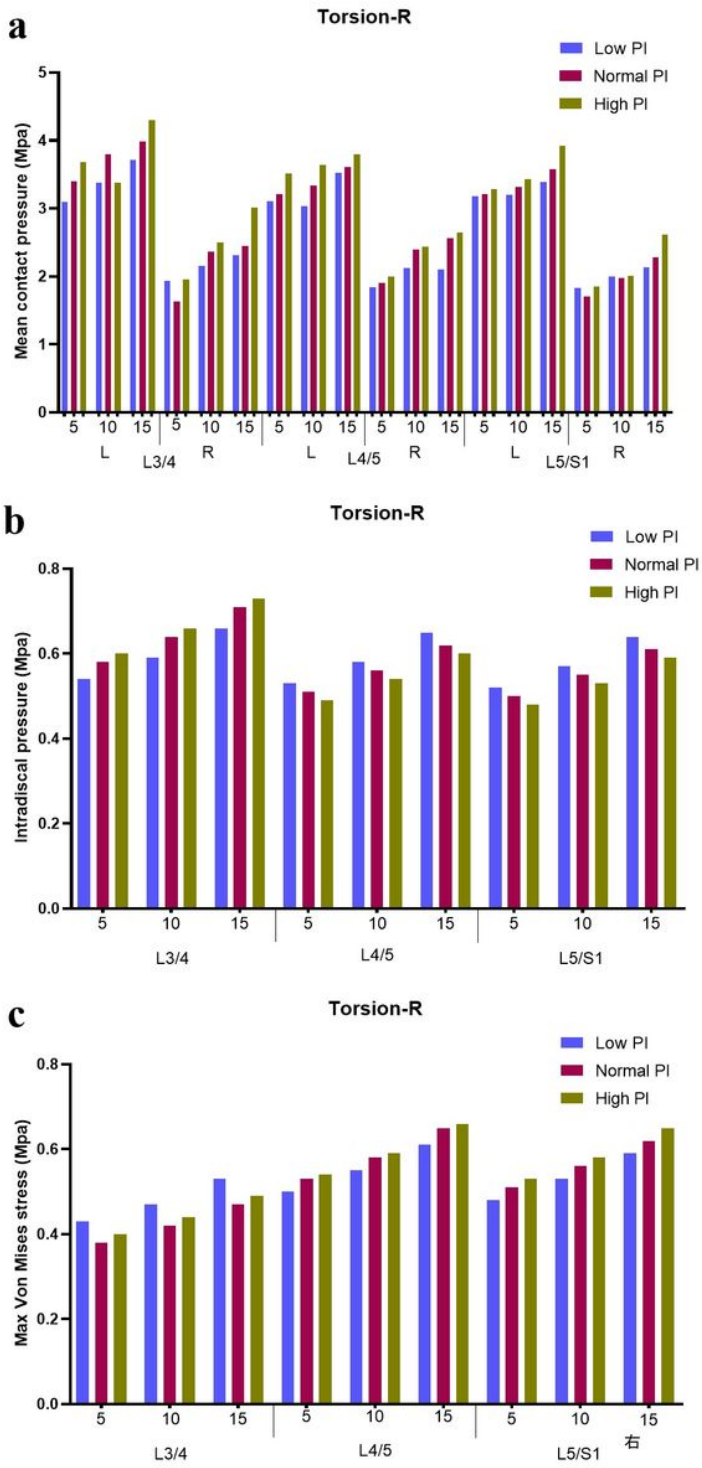

Figure 10

a:Mean contact pressures of facet joints at different levels in right-torsion posture under various torques. b:Intradiscal pressure at different levels in right-torsion posture under various torques. c:Max Von Mises stress of annulus at different levels in right-torsion posture under various torques.