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Article

Biomechanical study of weightlifting behavior in L5 lumbar spondylolysis using finite element simulation

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Abstract: Lumbar spondylolysis is related to weightlifting. The biomechanics of lumbar spondylolysis in weightlifting and the connection between lumbar spondylolysis and muscles are still unclear. Therefore, this study clarified the influence of decreased muscle strength on lumbar spondylolysis through finite element (FE) analysis. We used computed tomography to scan the L1-S1 segment of the patient and constructed a three-dimensional FE model. Apply a moment of 7.5 N·m and a weight of 280 N at the top of L1 after fixing the sacroiliac joint. The dumbbell weight was set to 15 kg. Apply muscle strength and follower loads representing the muscles of the back and abdomen in the FE model. The back muscle strength was reduced to 50%. The results showed that L4 with incomplete lumbar spondylolysis under decreased muscle strength and L5 with incomplete lumbar spondylolysis under normal muscle strength had the higher range of motion (ROM) in the flexion stage (45°). The ROM of L4 was affected by the decreased muscle strength, and the ROM of L5 was affected by the lumbar movement. L4 and L5 of incomplete lumbar spondylolysis showed the greatest stress range changes in the lifting and the final stage, respectively. Stresses at L4 and L5 are affected by the defect and decreased muscle strength. This study shows that the ROM of incomplete spondylolysis is vulnerable to flexion during weightlifting. Decreased muscle strength leads to increased stress on the defect and adjacent segments in the lifting and final stages, which might aggravate the fracture.

Keywords: lumbar spondylolysis; weightlifting; biomechanics; finite element analysis

1. Introduction

Weightlifting as a high-load and high-intensity sport can cause great damage to the lumbar. The effects on the lumbar include lower back pain, lumbar muscle strain, lumbar disc herniation, and even lumbar spondylolysis [1]. The lifting action causes the lumbar to endure huge axial loads and shear stresses [2]. These forces are particularly pronounced during exercises such as deadlifts, squats, and clean-and-jerk movements, where the lumbar spine is subjected to both compressive and shearing forces. Over time, these forces can lead to microtrauma in the lumbar structures, increasing the risk of chronic injuries. The load generated by the repeated bending of the lumbar in weightlifting will increase the pressure on the pedicles. This can lead to lumbar spondylolysis [3]. Spondylolysis, a stress fracture in the pars interarticularis of the vertebra, is a common injury among weightlifters due to the repetitive hyperextension and loading of the lumbar spine. This condition can progress to spondylolisthesis if not properly managed, further complicating the athlete's spinal health. Especially, young athletes who perform weightlifting training might affect the lumbar because of incorrect posture and high-intensity training, which might lead to lumbar spondylolysis [4]. Young athletes are particularly susceptible to lumbar injuries due to their still-developing musculoskeletal systems. Incorrect lifting techniques, such as rounding the back or failing to engage the core muscles, can place undue stress on the lumbar spine, increasing the risk of injury. Studies have shown that repeated stress on the lumbar in exercise can cause multi-segmental spondylolysis [5]. In addition, long-term training will affect the intrinsic compensation. It might influence the curvature of the lumbar sagittal profile [6]. The combination of poor technique and excessive training loads can create a perfect storm for lumbar injuries in young athletes. Coaches and trainers must emphasize proper form and technique, as well as monitor training intensity to prevent overuse injuries. The damage to the lumbar structure will bring other symptoms, including disc degeneration, cartilage tissue, and ligament damage [7].

. Manual material handling is common in daily life, such as among movers or people engaged in activities like relocation. This bending to lift objects can cause lumbar injury [8]. The lumbar spine’s natural curvature, or lordosis, can be altered by prolonged weightlifting training. This change in spinal alignment can lead to biomechanical imbalances, affecting the distribution of forces across the spine and increasing the risk of injury. This condition can lead to biomechanical lumbar deficiencies, abnormal body posture, and decreased balance [9]. To achieve better performance and increase the intensity of the competition, athletes will stretch and flex their lumbar to the limits of the physiological threshold. This increases the risk of lumbar spondylolysis. The pursuit of peak performance often drives athletes to push their bodies beyond safe limits, leading to extreme lumbar flexion and extension. These movements, when performed repeatedly and under heavy loads, can exceed the spine’s capacity to absorb stress, resulting in injuries such as spondylolysis. The competitive nature of weightlifting, where athletes are constantly striving to lift heavier weights, further compounds this risk. Although weightlifting has been identified as a major factor in lumbar spondylolysis, studies have failed to establish the connection between spondylolysis, lifting posture, and lumbar.

Lumbar spondylolysis is common in weightlifting. Lumbar spondylolysis can influence not only training but also daily life [10]. To investigate the biomechanics of the lumbar spine during weightlifting, we used finite element (FE) analysis to simulate the bending to lift of the lumbar spine. Previous studies have established FE lumbar spine models for weightlifters [11,12]. Previous studies have created models under different conditions and analyzed the stress changes in the models to predict risks and provide guidance for treating lumbar injuries [13]. Ramakrishna used FE models to analyze the function of the sacrum in lumbar spondylolysis and the relationship between disc degeneration and bilateral lumbar spondylolysis [14]. Although previous studies have provided various opinions on lifting movements and spondylolysis by FE analysis, it is unclear how weightlifting could cause or worsen spondylolysis.

This study established the FE model of lumbar spondylolysis and divided weightlifting into four main stages. FE analysis was used to clarify the changes in the range of motion (ROM) and stress of the lumbar and lumbar spondylolysis during weightlifting. In addition, we also analyzed and clarified the effects of weightlifting on lumbar spondylolysis under conditions of decreased muscle strength. This study revealed the main movements that affect the lumbar and lumbar spondylolysis during weightlifting and provided a biomechanical foundation for standardizing weightlifting movements.

2. Materials and methods

2.1. Patients and study design

The subject of this study is a 25-year-old healthy female from Wannan Hospital in China (Weight 56 kg, height 162 cm). Lumbar CT data were obtained in the Digital Imaging and Communications in Medicine file format. The participant signed an informed consent and this study was reviewed and approved under the Declaration of Helsinki statement (protocol code: RAGH 20240118). The weightlifting exercise was divided into 4 stages, as shown in **Figure 1**. The 4 stages include the starting (0°), flexion (45°), lifting (45° with load), and final stage (0° with load). This load represents the weight of dumbbells about 15 kg. The moment direction of the starting and flexion stages is flexion, and the lifting and the final stage is extension. The CT data were loaded into Mimics software (version 21.0; Materialise, Leuven, Belgium) to reconstruct the basic model from the first lumbar vertebra to the sacrum. The incomplete defect of the L5 vertebra was created by deleting a part of the L5 isthmus using SolidWorks software (version 2017, Dassault Systèmes, Versailles, France). The FE model and the CAD (geometry) model are shown in **Figure 2**. Muscle loading was added to the FE models, and the ROM and von Mises stress of the model were analyzed and compared in ANSYS (Ansys Inc., Version 17.0, Cannonsburg, PA, USA).

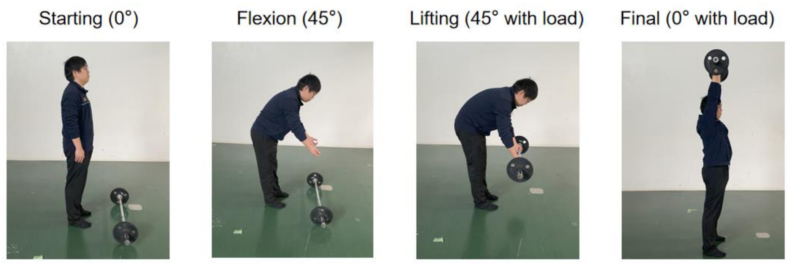

Figure 1. The weightlifting exercise was divided into 4 stages.

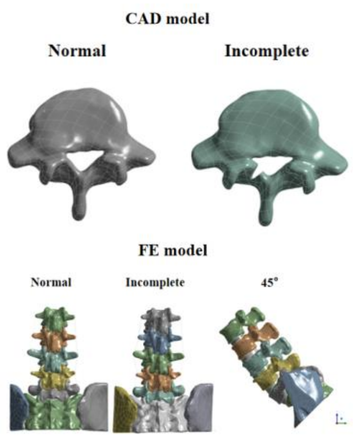

Figure 2. The FE model and the CAD (geometry) model.

2.2. Construct three-dimensional model

The patient's lumbar CT data (DICOM format) was loaded into Mimics software (Version 21.0; Materialise, Leuven, Belgium) to construct the basic model of the L1-L5 lumbar vertebrae. Then, the basic model was smoothed in Geomagic Studio (3D Systems, Rock Hill, SC, USA), including eliminating redundant triangular facets and edge smoothing. In addition, SolidWorks (version 2017, Dassault Systèmes, Versailles, France) was used to solidify the model and a complete lumbar vertebrae model (nodes: 620220, meshes: 381506) was constructed according to previous literature and anatomy. C3D4 tetrahedron elements were used to construct the cortical bone and cancellous bone. The model was reduced in Geomagic software to construct cancellous bone and a 1.5 mm cortical bone shell was constructed on the outer layer [15]. The endplates on the upper and lower surfaces of the intervertebral disc were constructed, about 0.5 mm in thickness [16]. The nucleus pulposus, annulus fibrosus, and articular cartilage models were constructed on the endplate surface. The nucleus pulposus makes up 43% of the entire intervertebral disc, and the cartilage between the facet joints is 1.5 mm [17]. The ligaments were constructed according to human anatomy and were set in ANSYS software (Ansys Inc., version 17.0, Canonsburg, PA, USA) to bear tension (**Figure 3**) [18]. The lumbar vertebrae model was meshed in ANSYS. The nucleus pulposus, endplate, and annulus fibrosus are tetrahedral elements, and the ligaments are hexahedral elements. To reduce the deviation between the lumbar spondylolysis and normal lumbar models, we modeled them as homogeneous. The material standards are shown in **Table 1**. The incomplete spondylolysis model (nodes: 620220, mesh: 381506) was created in SolidWorks by deleting part of the L5 isthmus (2 mm). The other conditions were consistent with the normal lumbar model.

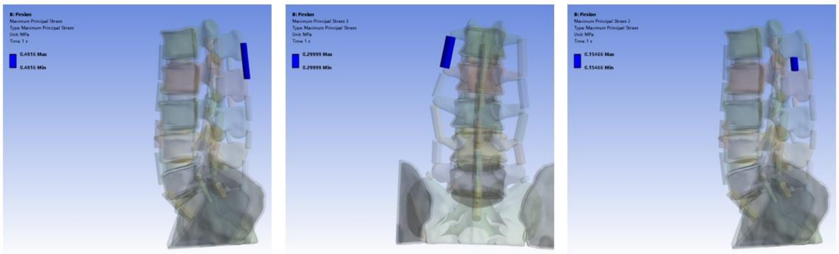

Figure 3. The ligament is under tension only in the model.

Table 1. Material properties used by FE analysis.

Components	**Young's modulus (MPa)**	**Poisson's ratio**	**Element type**	**Cross sectional area (mm2）**	**References**
Vertebra					
Cortical bone	12000	0.3	Tetra		Eric Wagnac et al., 2012
Cancellous bone	100	0.2	Tetra		Eric Wagnac et al., 2012
Endplate	24	0.4	Tetra		Rohlmann A et al.,2006b
Sacrum	5000	0.2	Tetra		Huang et al.,2016
Facet cartilage	11	0.2	Tetra		V K Goel et al.,1988
Annulus	110	0.3	Tetra		Rohlmann A et al.,2006b
Nucleus pulposus	1	0.49	Tetra		Rohlmann A et al.,2006b

Table 1. (*Continued*).

Components	Young's modulus (MPa)	Poisson's ratio	Element type	Cross sectional area (mm2）	References
Ligament					
ALL	10	0.3	Hex	63.7	Cheng-Cheng Yu et al.,2016
PLL	10	0.3	Hex	20	Cheng-Cheng Yu et al.,2016
LF	1.5	0.3	Hex	40	Cheng-Cheng Yu et al.,2016
CL	10	0.4	Hex	30	Cheng-Cheng Yu et al.,2016
ISL	1.5	0.3	Hex	40	Cheng-Cheng Yu et al.,2016
SSL	10	0.3	Hex	30	C S Chen et al.,2001
ITL	10	0.3	Hex	63.7	C S Chen et al.,2001

2.2.1. Loading conditions

The 280 N load was applied at 30 mm in front of the disc and the 7.5 N·m moment was applied at the top of L1 to simulate motion along the spine axis. The sacroiliac joints were fixed on both sides to prepare for the model validation. Next, static loads representing lifting movements (flexion and extension) were simulated. The model's ROM was compared with previous literature to ensure that the model is consistent with the previous literature model. In addition, we used the muscle-loading conditions from Zhu et al. [9] to simulate the real body structure. Three groups of muscle loads were applied to the FE model. All muscle loads were applied to the whole spine model beginning from L1. 170 N was applied 40 mm posterior to the center of the disc to simulate the back muscle load about the erector spinae. 20 N was applied 153 mm anterior to the center of the disc to simulate the abdominal muscle load about the rectus abdominis. A 200 N follow-up load was added to the model to simulate the compressive effect of local muscle forces [19]. The FE lumbar model under muscle load is shown in **Figure 4**. This study analyzed and compared the ROM, the L4-L5 von Mises stress, and intradiscal pressure in healthy and unilateral incomplete spondylolysis models under lifting movement [20,21].

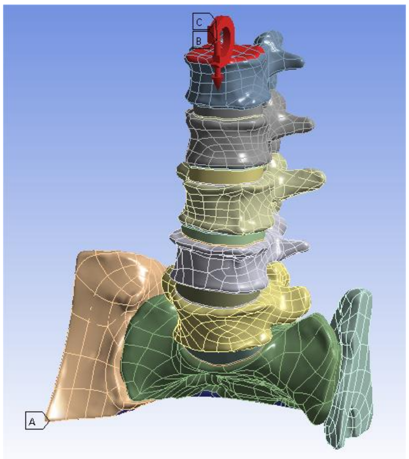

Figure 4. The FE lumbar model under muscle load.

2.2.2. Mesh convergence and material sensitivity test

This three-dimensional model is a linear model. Mesh convergence and material sensitivity tests can be used to judge that the model can be used [22]. This model generates five different unit sizes including automatic system division, 5, 4, 3, and 2.5 mm. The von Mises stresses in vertebrae were calculated and compared for different mesh resolutions. The mesh division was considered to be converged when the deviation between the results for two consecutive mesh resolutions was less than 5% [23]. According to previous research, material property parameters have been linearized [24]. The model was constructed into a linear model, a nonlinear model, a low-value model (a 25% decrease in material properties), and a high-value model (a 25% increase in material properties). The ROM and intradiscal pressure of L1-L2 in the linear, nonlinear, low-value, and high-low models were compared. Jebaseelan et al. showed that the annulus fibrosus was insensitive to material changes, while the material properties of the ligament were highly nonlinear [25]. Therefore, this study only tested the sensitivity of the vertebrae and nucleus pulposus.

2.2.3. The ROM and the von Mises stress

The ROM is the possible deformation of the model under load and moment. In this study, the Sacroiliac joint was fixed, which means that this part is the origin of the coordinate system. Therefore, the possible deformation in this study is an angle (degree). The von Mises stress can evaluate the stresses and ranges to which a model is subjected under loading conditions. It is made of the normal (σx, σy, σz) and shear stress (τxy, τyz, τzx) on each mesh of the model. This formula follows Hooke's law.

$$\sigma vm = \sqrt{\frac{1}{2}[(\sigma x-\sigma y)^2+(\sigma y-\sigma z)^2+(\sigma z-\sigma x)^2+6(\tau^2 xy+\tau^2 yz+\tau^2 zx)]} \quad (1)$$

The direction of ROM and stress is mainly determined based on the vector. **Figure 5** is a vector of L4 and L5 during flexion. It can be seen that the normal group and weightlifting group move forward in the vector direction during flexion. The weightlifting group has greater displacement and stress due to the high load of dumbbells.

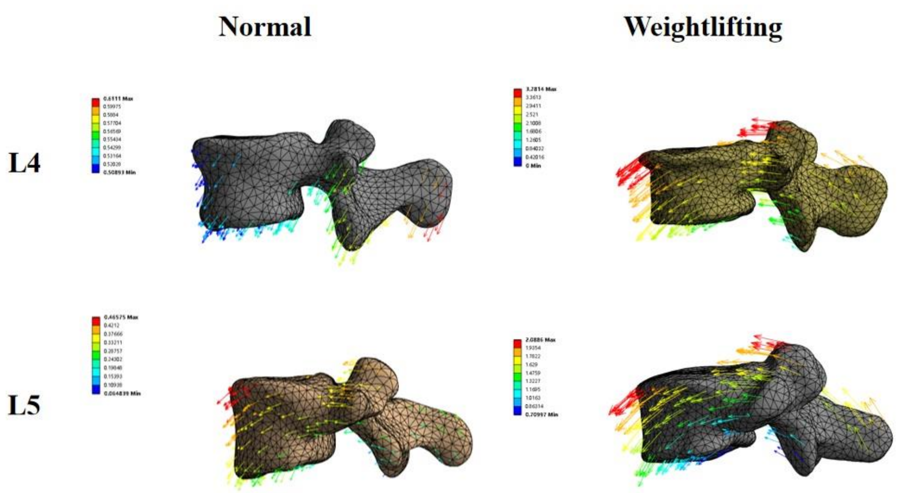

Figure 5. The vector of L4 and L5 during flexion.

3. Results

3.1. FE model validation

The consistency of the model in this study was verified based on the research results of Zhu et al. [9] In flexion-extension, bending, and torsion, the moment (7.5 N·m) was applied to the center of the upper surface of the L1 vertebra. L5-S1 segments were compared with the other studies. As shown in **Figure 6**, the model's ROM results match previous studies' data. Therefore, it can be considered that the FE model of this study is suitable for subsequent analysis. The mesh convergence test shows the number of elements and nodes for each mesh resolution in **Table 2**. Fix the sacroiliac joint and apply a force of 50 N from the L1 surface to test the mesh resolution. When the element size is 4 mm, the stress result is 1.9031 MPa; when the element size is 3 mm, the stress result is 1.8758 MPa. The difference between the two calculated results is 1.434%. The results of this study show that the model mesh remains convergent. **Table 3** shows the data of material sensitivity analysis. **Figure 7** shows the material property sensitivity analysis in IDP.

ROM: The linearized basic model was reduced by 2.3% compared to the nonlinear model; improved by 16.4% compared to the low-value model; and reduced by 14.3% compared to high-value models.

IDP: The linearized basic model was reduced by 1.9% compared to the nonlinear model; improved by 10.3% compared to the low-value model; and reduced by 13.4% compared to high-value models.

The comparison results in ROM and IDP are close.

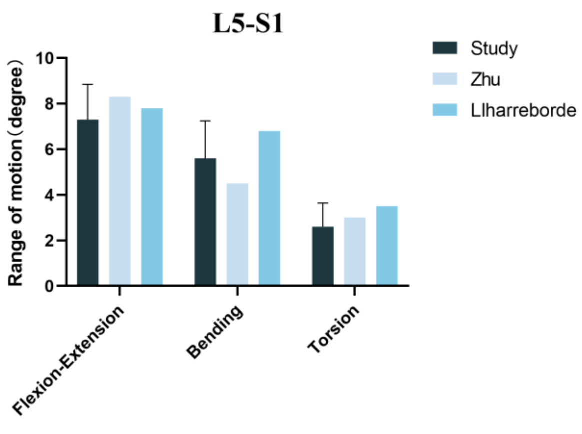

Figure 6. Comparison of the ROM of L4-L5 segment between the current and previous studies.

Table 2. Element and node numbers for six different mesh resolutions.

Unit size	Element number	Node number	von-Mises (MPa)
system	73730	37028	2.3948
5mm	104363	55422	2.3612
4mm	147879	81259	1.9031
3mm	258498	149645	1.8758
2.5mm	381788	228737	2.0512

Table 3. Material properties sensitivity analysis in ROM.

	The linearized basic model	The nonlinear model	The low-value model	The high-value model
L1	13.997	15.004	16.0	12.5
L2	5.5004	6.3002	6.0	5.143
Difference	8.4966	8.7038	10.0	7.357

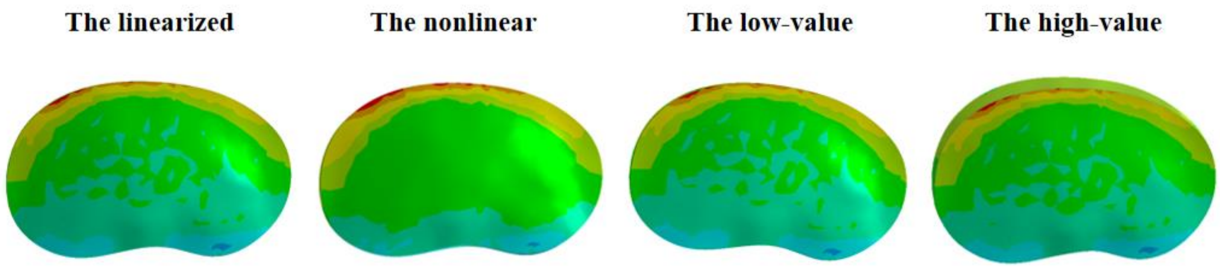

Figure 7. The material property sensitivity analysis in IDP.

3.2. ROM

Figure 8 shows the influence of decreased muscle on the ROM of normal and incomplete lumbar spondylolysis in L4 and L5 during weightlifting. Detailed data are shown in **Table 4**. The ROM in incomplete spondylolysis under the decreased muscle conditions is changed significantly in L4 and is highest when lumbar with 45° flexion, and decreases when lifting dumbbells with 45° and 0° extension. Then is the control group under the decreased muscle conditions. Although the control group under the decreased muscle conditions is lower than the incomplete spondylolysis group in general, it also reaches the highest when lumbar is in 45° flexion. The control group and incomplete spondylolysis group under the normal muscle conditions have similar ROM, and the trend is consistent with the model under the decreased muscle conditions. The ROM in incomplete spondylolysis is also higher than that of the normal in L5. The ROM in incomplete spondylolysis under normal muscle conditions is the highest, and all models are highest when the lumbar is in 45° flexion, and decrease when the lumbar is in 45° and 0° extension. From the results of the muscle conditions, the incomplete spondylolysis group under the decreased muscle conditions was lower than the incomplete spondylolysis group with normal muscle conditions. The normal group under the decreased muscle conditions was also lower than the normal group under normal muscle conditions.

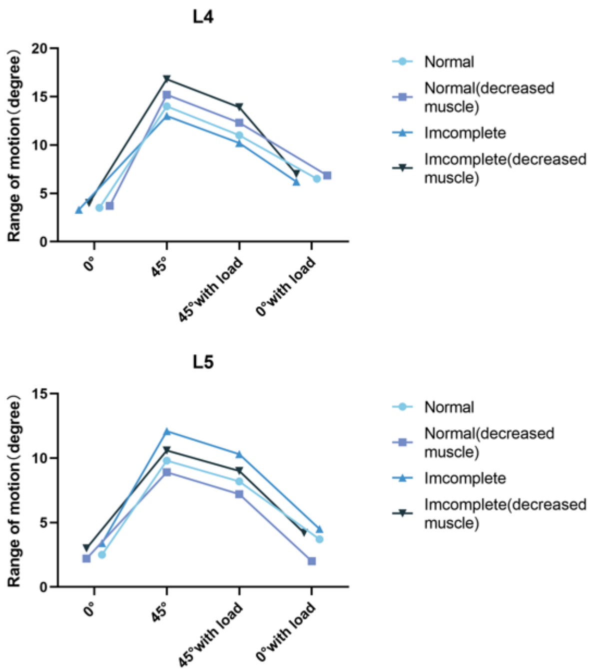

Figure 8. ROM of normal and lumbar spondylolysis with and without decreased muscle conditions.

Table 4. ROM of normal and incomplete lumbar spondylolysis in L4 and L5 during weightlifting.

L4	**0°**	**45°**	**45°with load**	**0°with load**
Normal	3.5	14	11	6.5
Normal (decreased muscle)	3.7	15.2	12.3	6.85
Incomplete	3.3	13	10.2	6.2
Incomplete (decreased muscle)	4	16.8	13.9	7
L5	**0°**	**45°**	**45°with load**	**0°with load**
Normal	2.5	9.8	8.2	3.7
Normal (decreased muscle)	2.2	8.9	7.2	2
Incomplete	3.4	12.1	10.3	4.5
Incomplete (decreased muscle)	3	10.6	9	4.2

3.2.1. von Mises stress

Figure 9 shows the von Mises stress of L4 in normal and incomplete lumbar spondylolysis under muscle conditions during weightlifting. The trend is similar to the results of ROM, and the stress change of incomplete spondylolysis under the decreased muscle conditions is the largest. The overall stress increase is most obvious when lifting dumbbells with 0° extension, especially at the part of the upper vertebrae surface. Under 45° with dumbbell loading, the normal model and incomplete lumbar spondylolysis under the decreased muscle conditions have a high stress on the inner side of the vertebrae pedicle. Under normal muscle conditions, although the normal and incomplete spondylolysis also increases the most when lifting dumbbells with 45° extension, the overall pressure is not obvious compared with the decreased muscle conditions. **Figure 10** shows the von Mises stress of L5 in normal and incomplete lumbar spondylolysis under muscle conditions during weightlifting. However, what is interesting is that the von Mises stress of L5 shows a different trend. Compared with the L4 segment, the incomplete spondylolysis group with normal muscle conditions has the most obvious change in L5. The overall stress is the largest when lifting dumbbells with 45° extension. Compared with the control group, the stress in the unilateral incomplete group was concentrated on the opposite side of the fracture and vertebrae surface. The stress in the incomplete spondylolysis group under decreased muscle strength was also higher. It increased by 19.6% and 9.8% when lifting dumbbells with 45° and 0° extension respectively compared with muscle conditions. The stress distribution at the defect site is shown in the **Figure 11**.

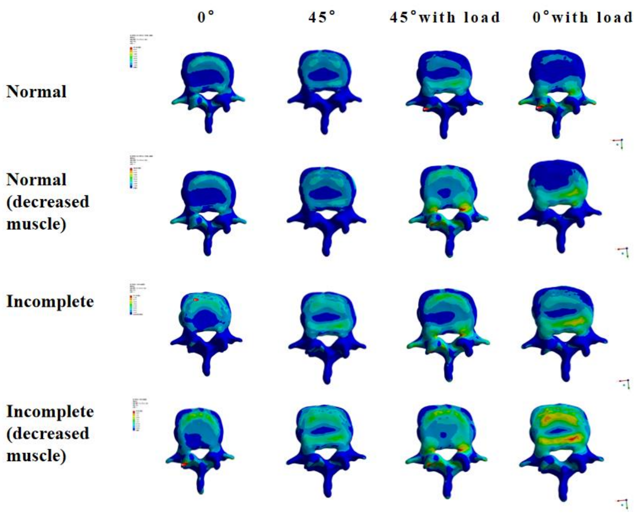

Figure 9. L4 stress distribution of normal and spondylolysis with and without decreased muscle conditions.

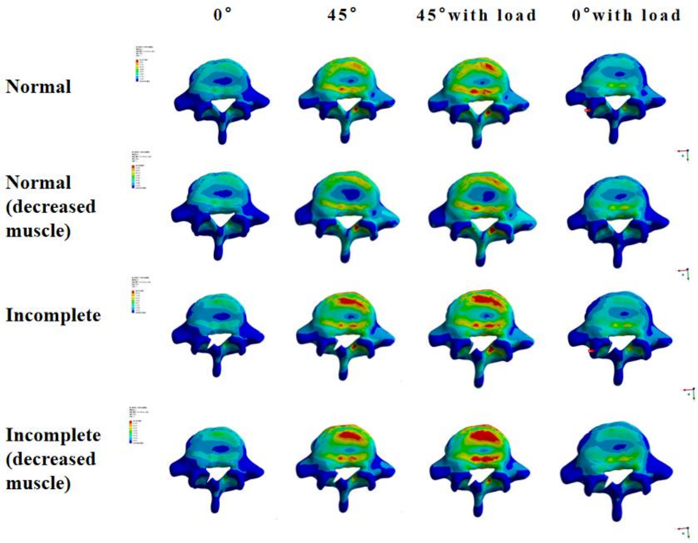

Figure 10. L5 stress distribution of normal and spondylolysis with and without decreased muscle conditions.

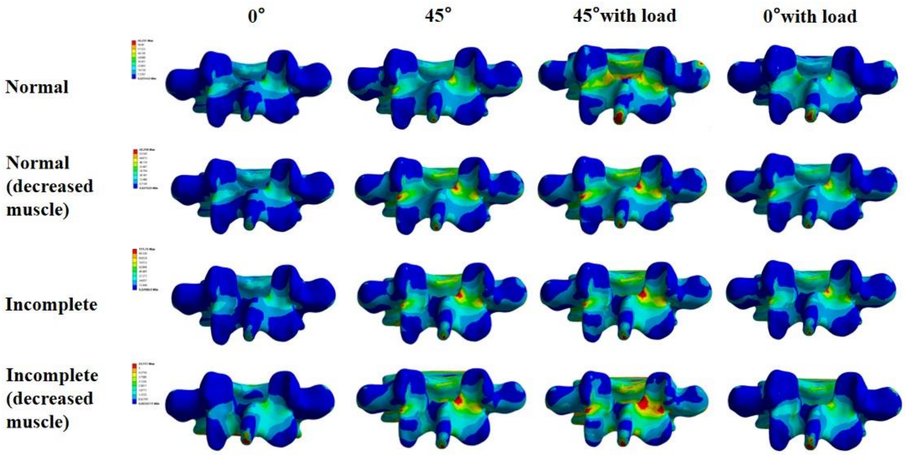

Figure 11. The stress distribution at the defect site.

3.2.2. Intervertebral disc pressure (IDP)

Figure 12 shows the changes in L5/S IDP in normal and incomplete lumbar spondylolysis under muscle conditions during weightlifting. The L5/S IDP of incomplete lumbar spondylolysis under decreased muscle conditions has the largest change, especially when lifting dumbbells at 45° and 0°. The IDP of the control group under decreased muscle conditions when lifting dumbbells at 45° and 0° is consistent with trend of the incomplete lumbar spondylolysis. The intradiscal pressure at 45° with load is concentrated on the front side in all groups. There is no obvious change in the healthy lumbar spine in a normal state.

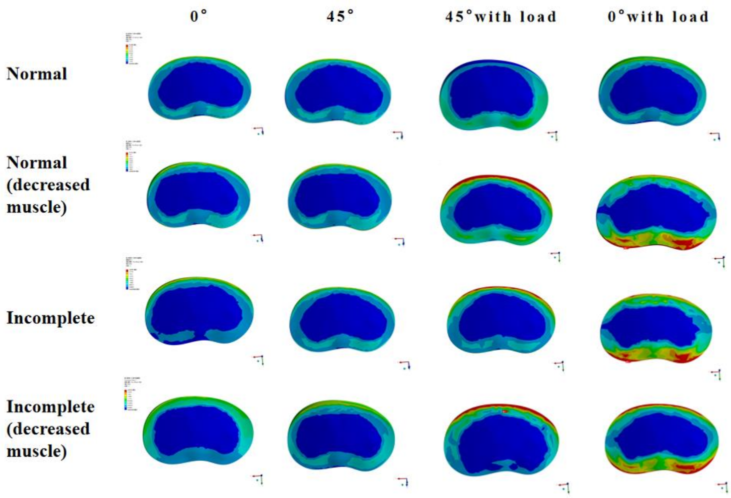

Figure 12. Stress distribution of IDP of normal and spondylolysis with and without decreased muscle conditions.

4. Discussion

Weightlifting and bending to lift objects in daily life are important risk factors for lumbar spondylolysis. The connection between the muscles around the lumbar and the movement of the lumbar is complex and remains unclear. Mohamad Y Fares' study showed that using too much weight and performing incorrect states can lead back to injury [26]. Therefore, our study fully simulated the biomechanical changes of the lumbar spine and spondylolysis during weightlifting through FE analysis with and without decreased muscle conditions. These results allow weightlifters to change their posture to reduce the risk of spondylolysis. ROM increased during the flexion stage of the weightlifting. Adjusting the flexion to the front squat can reduce the risk of lumbar spondylolysis. The stress results show that the weightlifters increased their load on the opposite side of the defect under the dumbbell load. It means that excessive pressure affects load distribution. Maintaining lumbar balance during lifting and final stages might reduce this effect. Regarding rehabilitation strategies, athletes can train in specific movements. The stress increased significantly during the lifting and final stages of the weightlifting in the lumbar. For normal athletes, spinal extension exercises can be performed more often in daily training to improve spinal mobility. For patients with early spondylolysis, flexion-extension and torsion can be reduced. The lumbar spine should be exercised with slight bending to the left and right to avoid symptom aggravation. In addition, Christopher S Patterson showed that hip and lumbar extensor weakness during weightlifting can change biomechanics to reduce lumbar load demands [27]. E Heidari's study also showed that unstable lifting activities caused significantly increased the L5-S1 and L4-L5 compressive loads in individuals with low back pain compared with stable lifting activities [28]. Our study further demonstrated that lumbar vertebrae defects (lumbar spondylolysis) under decreased muscle conditions can lead to an increase in stress range at the L5-S1 and L4-L5.

Our results showed that the ROM in the lumbar reached the highest during the weightlifting process when flexion and decreased during the 45° and 0° with load stages. This suggests that the movements during the flexion stage of weightlifting begin to affect the lumbar, especially spondylolysis. Athletes should avoid flexion motion to lift dumbbells in weightlifting [29]. Although many people commonly bend in weightlifting and lifting heavy objects, athletes who train in repetitive movements might be able to alleviate the effects of weightlifting by squatting to lift dumbbells. Because the static motion was simulated in this study, we compared our model and results with Caiting Zhang et al. research [30]. Under the condition of the 15 kg load, Caiting Zhang et al. found that during weightlifting, the lumbar flexion angle increased with increasing load and was accompanied by torsion and bending. Skeletal structural abnormalities can change the ROM and stress of specific areas by changing the mechanical transmission pathway [31]. We also confirmed that the L4 ROM in incomplete spondylolysis under decreased muscle conditions increased during weightlifting. The L5 ROM in incomplete spondylolysis also increased during weightlifting. It is related to muscle conditions. Decreased muscle strength will reduce the ROM in the fracture segment and increase in the adjacent segments. Especially the adjacent segments of the spondylolysis are more affected by decreased muscle strength. This is due to the adjacent segments compensated for the change in the ROM caused by the defective segment. This is consistent with the Rami Haj-Ali et al. Research has shown that the ROM in adjacent segments increased under the unilateral and bilateral spondylolysis models [32]. Hong Jin Kim et al. showed that the muscle volume around the fracture was lower than that around the non-fractured area [33]. The decreased muscle strength will increase the possibility of adjacent segments developing pedicle injury. The pars interarticularis plays a protective and connecting role in lumbar movement, but the decreased muscle strength can cause the stress of the posterior vertebral structure to increase. Incomplete spondylolysis increases stress on the opposite side in the pedicle and pars interarticularis. This increases the risk of bilateral multiple fractures. We consider that changes in L4 stress are also related to decreased muscle strength. Especially when lifting at 0°, the stress on L4 in incomplete spondylolysis increased significantly under the decreased muscle strength condition. This might be because the weight and dumbbell loads have a greater impact on the vertebral body after these directions are consistent.

Therefore, during the flexion of the whole weightlifting process, the pedicle stress of the L4 vertebra increased, while the stress on the anterior and superior surfaces of the L5 vertebra increased significantly. In the extension, due to the process of lifting dumbbells, the ROM decreases with the changed degree from 45° to 0°. From the whole stress in the lumbar, the stress distribution is uneven and dispersed. Under normal conditions, the stress is generally concentrated on the vertebrae [34]. In addition, the uneven stress distribution of L4 and L5 is also related to decreased muscle strength. The increase in the ROM and stress caused by decreased muscle strength is obvious in L4. Although there is only flexion and extension movement, the high-stress area is not completely concentrated above the vertebral body near the front and back. Because the lumbar segment structure will bear slight torsion under any movement [35]. This shows that even in weightlifting with only flexion and extension, the lumbar spine will relieve the high stress through slight axial torsion. Therefore, we speculate that this special change formed during the extension accompanied by axial torsion might be one of the reasons for the occurrence of lumbar spondylolysis caused by weightlifting. The stress of L4 changes the most when lifting dumbbells at 0°, while the stress of L5 changes the most when lifting dumbbells at 45°. Compared with L5, L4 bears weight and dumbbell load earlier when lifting dumbbells at 0°. When lifting dumbbells at 45°, the L5 fracture is more vulnerable to decreased muscle strength and load. This biomechanical change might progress to instability, degeneration, or even spondylolisthesis. Increased mechanical stress on the lumbar can accelerate disc degeneration, leading to an increased risk of conditions such as spondylolysis or facet joint osteoarthritis. Decreased muscle strength might lead to uneven load distribution. Without muscle support, the loads from weightlifting can lead to stress fractures and eventually spondylolysis. For rehabilitation and prevention, athletes can follow a specific strengthening program. Prioritize core stabilization exercises, such as plank exercises, to restore proper spinal control [36]. Strengthen the glutes, hamstrings, and lower back with exercises such as front squats and hip thrusts. This can reduce load changes at L5-S1 and L4-L5. After an injury, gradually increase training intensity to prevent cumulative stress. The results of the intradiscal pressure show that the stress is concentrated on the edge of the intervertebral disc during movement. This might be because the annulus fibrosus mainly bears these stresses [37]. Naserkhaki et al. also showed that the intervertebral disc changes caused by different lordosis showed similar load trends [38]. When lifting the dumbbell at 0°, the direction of the dumbbell load and weight remain consistent. This will bring huge stress to the annulus fibrosus that protects the nucleus pulposus.

As with previous studies, this study has limitations. The muscle loads and moments in this model are idealized. This is an ideal situation where the in vitro data from previous studies were simplified in a parametric way. Therefore, the results of this study are a comparative analysis to show a trend. Although these stresses do not require much data in vivo or in vitro, comparing biomechanics under the same assumptions can provide some relative trends to a certain extent. In addition, there are many different methods for modeling in FE studies, including CT grey, different moment verification, and heterogeneous materials for analysis [39,40]. These methods have to reduce the level of modeling sophistication due to the complexity of the calculation. This study analysis by the same moment and homogeneous material analysis, the main purpose was to maintain the universality of the results and reduce deviation. Furthermore, we consider that more clinical research and biomechanical analysis are needed on the mechanisms and relationships between weightlifting, muscle strength, and spondylolysis.

5. Conclusion

From the perspective of ROM, L4 is most affected by decreased muscle strength during weightlifting, while L5 as the fracture segment has the greatest change in flexion without dumbbell load. From a biomechanical perspective, decreased muscle strength increases the stress of normal and incomplete spondylolysis during weightlifting. The stress of L4 and L5 in incomplete lumbar spondylolysis increases under decreased muscle strength, and is greatest when lifting dumbbells at 0° and 45°. This study shows that the risk of spondylolysis was increased in the fracture segment during lifting dumbbells (45°). Decreased muscle strength further increases the risk and affects adjacent segments.

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Ethical approval: The study was conducted in accordance with the Declaration of Helsinki, and was approved by the Institutional Review Board of Huzhou Central Hospital, the Fifth School of Clinical Medicine of Zhejiang Chinese Medical University (project number RAGH 20240118). Informed consent was obtained from all subjects involved in the study.

Data availability: The data used to support the findings of this study are included within the article.

Conflict of interest: The authors declare no conflict of interest.

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