Comparison of the biomechanical effects of lumbar disc degeneration on normal patients and osteoporotic patients: a finite element analysis

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Research Article

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Abstract

Background: With an increasing life expectancy, more and more people suffered from age-related diseases and sought more medical care. Disc degeneration and osteoporosis were typical orthopaedic diseases in this situation. And there was a lack of FE studies that predicted combined effects of the disc degeneration and osteoporosis.

Methods: A normal lumbar spine finite element model (FEM) was developed based on the geometric information of a normal male subject (age 35 years; height 178 cm; weight 65 kg). This normal lumbar spine FEM was modified to build three lumbar spine degeneration models simulating mild, moderate, and severe grades of disc degeneration at L4-L5 segment. Then the degenerative lumbar spine models for osteoporotic patients were constructed on the basis of the above mentioned degeneration models. Under a 400N follower compressive load, the 7.5 Nm moment was applied on these models to simulate different motion postures. Finally, the range of motion (ROM), Mises stress in cortical (MSC1), Mises stress in endplate (MSE), Mises stress in cancellous (MSC2) and Mises stress in post (MSP) were solved and analyzed.

Results: The results indicated that compared with disc degeneration patients without osteoporosis, the ROM, MSC1 and MSE of osteoporosis patients suffering from various disc degeneration decreased in all postures, while the MSC2 and MSP of disc degeneration patients with osteoporosis increased. With the increase in degree of disc degeneration, the reduction proportions of ROM and MSE in osteoporotic patients almost gradually increased, while the reduction percentages in MSC1 of osteoporotic patients almost gradually decreased. And the increase percentages of MSC2 in osteoporotic patients gradually increased. It's worth noting that, with the progressive changes of disc degeneration, the changes of MSP in osteoporosis patients were uneven.

Conclusions: In summary, the effect of disc degeneration on flexibility in the two kinds of patients (osteoporosis and non-osteoporosis patients) was nearly identical. By comparing the remaining biomechanical parameters (MSC1, MSE, MSC2 and MSP), we found that osteoporosis may accelerate disc degeneration and the osteoporotic patients who suffer from disc degeneration had a higher risk of vertebral fracture.

Highlights

1. Osteoporosis and lumbar disc degenerations are common orthopedic diseases.
2. The combined study of osteoporosis and lumbar disc degeneration is lacking.
3. The interaction of osteoporosis and lumbar disc degeneration is explored by finite element methods.

Introduction

Lumbar degenerative disease is a fairly common disease in orthopedics caused by disc degeneration. And disc degeneration is a progressive condition, which will change the geometrical morphologies and material properties of some important spinal components, and ultimately affects their abilities to transmit and bear loads [1]. That is because, the disc is a connecting medium between two adjacent vertebral bodies, which can transmit external forces and maintain spinal motion in the physiological environment. The deterioration of disc biomechanical environment usually leads to occurrence of radiculopathy, which in turn causes lower back pain and other symptomatic diseases, seriously affecting the quality of life [2–3].

Worldwide, as life expectancy increases, the increasing people will face age-related diseases and seek more medical care. Osteoporosis is a skeleton disease in this situation and it is characterized by a decrease in bone mass, accompanied by the microstructures deterioration of bone tissue, which greatly increases the tendency of fragility fractures [4]. The occurrence and development of osteoporosis is a painless process and this makes its diagnosis and prevention become more difficult. Elderly patients with osteoporosis are often accompanied by more concerned complications, such as cardiovascular disease, facet joint osteoarthritis, spinal stenosis, especially the degenerative disc changes [5]. Because the loss of bone mass may accelerate the deterioration of these comorbidities [6]. Unfortunately, in current finite element (FE) studies on osteoporosis, these compications have not been considered. For example, Giuseppe et al. developed a FE model of the L3-sacrum and explored biomechanical effects of metastasis in the osteoporotic lumbar spine. They found that osteoporotic lumbar spine metastasis lead to a higher risk of vertebral fractures [7]. Wang et al. analyzed the effect of osteoporosis on internal fixation after spinal osteotomy by FE method and they drew a conclusion that osteoporosis may increase the risk of fracture and internal fixation failure [8]. Anne et al. used FE analysis way to study the effect of cement reinforcement on the functional spinal unit of the lumbar spine in osteoporosis and proved that cement reinforcement can restore the strength of the vertebral body [9]. There has been scholar used clinical statistical methods to explore the relationship between osteoporosis and spinal fusion surgery [10]. That is to say, the current scholars mainly focus on the studies of the relationship between osteoporosis and spinal fusion surgery, and there is a lack of combination research on lumbar disc degeneration and osteoporosis. And the disc degeneration may affect the biomechanics of osteoporosis patients, causing changes in the stress and strain states of spinal components (such as vertebra and endplate).

Disc degeneration is related to the gradual changes in the compositions and morphologies of intervertebral disc, such as disc height loss, anterior osteophytes formation, water loss and endplate sclerosis [11–13]. These changes can be present individually, or more frequently in different combinations [14–21]. This study aims to conduct a biomechanical study on a wide range of clinical conditions of disc degeneration, where the most common degeneration exists in various combinations. In other words, the disc degeneration will cause many morphologic combination changes. In the previous FE studies of lumbar spine degeneration, those scholars only simulated few lumbar disc degeneration geometrical characteristics, and most of these studies mainly changed the disc height to simulate the disc degeneration [14–17]. Only some investigations have modified 2–3 disc degeneration geometric changes to build degenerative lumbar spine models [18–21]. Therefore, there is a lack of lumbar degeneration models with different degeneration degrees that can be more clinically representative. This study combined the aforementioned geometrical characteristics of disc degeneration, and changed the material properties of some important spinal components to simulate a more realistic physiological state of disc degeneration. Elderly patients who need to carry out the lumbar spine fusion surgery due to disc degeneration are often accompanied by osteoporosis. In turn, osteoporosis is likely to cause progressive spinal deformities and potential nerve damage in elderly patients, which is the main concern before spinal surgery. And the degeneration of intervertebral disc is basically a prerequisite for the above diseases. So in this study, the combined effects of intervertebral disc degeneration and osteoporosis have been considered.

As far as we know, due to moral and ethical restrictions, it is difficult to simulate and test some physiological behaviors of the lumbar spine in vivo. At the same time, in order to determine some biomechanical effects caused by the degeneration of spinal components (such as intervertebral discs and facet cartilages), sometimes the human cadaver specimens is not particularly useful. That's because in vitro experiments cannot control the degeneration degrees of cadaver specimens and highly depends on the limited availability of cadaver

specimens. FE analysis can provide an easier and more reliable method to study the internal spinal biomechanical changes caused by disc degeneration or osteoporosis, and has been widely used in previous FE studies [7–9, 14–21]. In the FE analysis, the degrees of the lumbar spine degeneration and osteoporosis can be controlled, which is very difficult to achieve in cadaver specimens experiment studies. That is to say, FE analysis method makes up for the insufficiency of in vitro experimental research to a certain extent.

The aim of this study was to investigate differences of the effects of lumbar disc degeneration on normal patients and osteoporotic patients by using three progressive lumbar spine degeneration finite element models (FEMs) with different morphological changes and higher clinical representativeness. To do this, the range of motion (ROM), Mises stress in cortical (MSC1), Mises stress in endplate (MSE), Mises stress in cancellous (MSC2), Mises stress in post (MSP) were calculated and analyzed.

Methods

The FEM of one normal lumbar spine

Establishing a FE model that similar to the real geometries, internal structures and tissue material properties of the lumbar spine was an important step in the spinal biomechanics studies. This investigation used the most common and effective modeling method in the current spinal FE analysis [14–21, 22–23]. Based on the computed tomography (CT) image of a healthy male subject (Age: 35 years old; Height: 178 cm; Weight: 65 kg), the normal lumbar spine model (L3-S1) was developed, as shown in Fig. 1. In the first place, the CT image was imported into Mimics 14.0 software (Materialise Technologies, Leuven, Belgium) to create a binary STL file. Then, the STL file was imported into a medical reverse engineering software (Geomagic Studio 11.0; Geomagic Inc., NC, USA) for further smoothing geometric surface. Thirdly, the processed geometric model was imported into the FE pre-processing software (Hypermesh 14.0; Altair Engineering Corp, Michigan, USA) for constructing meshes. Finally, the normal lumbar spine FEM was biomechanically simulated in a commercial FE software (Abaqus 6.14; Dassault Systemes Simulia Corporation, Pennsylvania, USA). The detailed steps of building a normal lumbar spine FEM were described from previous studies [14–21, 22–23].

The material properties, element types and references of normal lumbar spine model were displayed in Table 1 [18, 21, 24–30]. The vertebral body, consisting of cartilage endplate, cortical bone, cancellous bone and post element, which were defined as linear elastic materials [18, 21, 24–25]. And the sacrum was also modeled with linear elastic materials [21, 26]. The facet cartilage was modeled by Neo-Hookean hyper-elastic elements [18, 21]. The initial gap of two facet cartilage surfaces was set as 0.1–0.2 mm, and the interaction was set as frictionless surface-to-surface contact [27–28]. Intervertebral disc was composed of nucleus pulposus and annulus ground, and they were defined as hyper-elastic materials (Mooney–Rivlin) [21, 26, 29–30]. The annulus fibers were embedded in annulus ground at an angle of approximately ± 30° to endplate surface, were simulated as nonlinear spring elements [11, 14, 26]. The seven major ligaments were built in corresponding place in the form of one-dimensional spring element with nonlinear characteristics [18, 21, 26].

The FEMs of different lumbar spine disc degeneration for normal patients

According to the normal lumbar spine FEM, three FE models of degenerative lumbar spine with different degeneration degrees (mild degeneration, moderate degeneration, severe degeneration) were built by modifying

the geometric morphologies (disc height, anterior osteophytes, disc volume, the thickness and curvatur of endplate) of intervertebral disc and the material properties of some important spinal components [14, 18–21, 26]. The simulation of disc degeneration is only in the L4-L5 segment, which has the highest incidence of degeneration [1, 15–18].

Table 2 showed the lumbar disc degeneration grading system and the implementation of FE model. In mild, moderate and severe degeneration models, the proportions of disc height loss were 1/5, 2/5 and 3/5 of normal disc height, respectively [18, 21]. In the process of disc degeneration, the shape of anterior osteophyte was very complicated, so we defined its height and width as equal to simplify the process of model building. In different degenerative models, the height and width of anterior osteophyte were set as 1/10, 2/10 and 3/10 of sagittal diameter of normalvertebra (L4 vertebra and L5 vertebra), as shown in Fig. 2 [19–21]. With the increase of the disc degeneration degrees, the volume of the nucleus pulposus and the annulus ground are constantly decreasing. In this study, their volume changes were adjusted by reducing the disc height and the area of nucleus pulposus [18–19, 21]. Simulating the volume change of intervertebral disc is actually simulating the change in the hydration degree during the disc degeneration. Endplate sclerosis exhibited different characteristics in different stages of disc degeneration. In the early stage of disc degeneration (mild degeneration), endplate sclerosis was only reflected in the reduction of endplate thickness, and its curvature did not change much. In moderately and severely degenerated models, endplate sclerosis will cause the changes in thickness and curvature of endplate. As the degree of disc degeneration increased, the curvature of the endplate gradually flattened and the thickness of endplate decreased from 0.6 mm to 0.3 mm, as shown in Fig. 2. The sizes of the cartilage endplate thickness are shown in Table 2 [20–21, 26]. Figure 2 displayed the schematic diagrams of L45 segment of the three lumbar spine degeneration models.

During the process of disc degeneration, the material properties of some important spinal components in degenerative segment have also changed. In general, the hardness of these spinal components gradually increases. The material properties of cartilage endplate are gradually the same as that of cancellous bone [19–21, 26]. Frozen sections showed that the soft tissues between anterior osteophytes and anterior osteophytes were similar in structure to annulus ground and cancellous bone, respectively [20–21]. According to previous literatures [18–21], the material properties of nucleus pulposus and annulus ground were determined in mild and severe degeneration models. Their material properties in the moderate degeneration model were obtained by the linear interpolation method. Table 3 showed the material properties, element types and references of some important structures in degenerative lumbar spine models.

The FEMs of different lumbar spine disc degeneration for osteoporotic patients

Based on these degenerative lumbar spine models, the FEMs of different lumbar spine degeneration for osteoporotic patients were built by changing the material properties of cartilage endplate, cortical bone, cancellous bone, post element and sacrum. Specifically, osteoporosis was defined as a decrease in elastic modulus of all bony structures (cartilage endplate, cortical bone, cancellous bone, post structure and sacrum) by 33%, 33%, 66%, 33% and 33%, respectively [9, 31–33]. The soft tissue structures remained unchanged, such as intervertebral discs, facet cartilage and ligaments. Table 4 showed the material properties, element types and references of bone structures in degenerative lumbar spine models for osteoporotic patients.

Boundary and loading conditions

The six degrees of freedom in sacrum were all constrained. A node at the middle position of upper endplate of L4 segment was created, then this node was coupled with the L4 segment upper endplate. In the same way, other nodes were built at the center of corresponding endplates and the nodes coupled with these endplate surfaces, a total of 6 nodes were established. Using these nodes to create a connector element, and a 400 N follower load was added on the connector element to simulate the physiological compression condition (the synergy of upper body weight and muscles) of lumbar spine [24–25]. A 7.5 Nm moment load was applied on the upper surface of L4 upper endplate to simulate different motion postures (Flexion, Extension, Lateral bending and Torsion).

Calibration and Validation

The calibration factors of annulus fibers and ligaments were adjusted to complete the calibration process, and the detailed calibration process was introduced in previous literatures [34–36]. Then, the range of motion was predicted in various loading conditions (Flexion: 8 Nm; Extension: 6 Nm; Lateral Bending: 6 Nm; Torsion: 4 Nm), and compared with the data from in vitro experiments for validating the developed normal lumbar spine FE model [36]. Under 10 Nm pure moment, the ROM of L4-L5 segment in different degeneration models were also compared with in vitro experimental data [37]. These simulation works were implemented in a commercial finite element software (Abaqus 6.11; Dassault Systemes Simulia, Pennsylvania, USA).

Results

Calibration and Validation

Figure 3 showed the comparison between the FE data (ROM) in this study and in vitro cadaver experiment data of Renner et al. [36]. The results illustrated that all ROM of this study was within the standard deviation of in vitro experimental data, which meant that these FE data closely match the in vitro experimental data of Renner et al. [36]. In addition, the comparison of FE data and experimental data (L4-L5 degeneration segment's ROM) for degenerative lumbar spine models under 10 Nm pure moment load was implemented. We found that the most of FE data were within the standard deviation of the in vitro experimental results of Mimura et al [37], as shown in Table 5. Therefore, the normal and degenerative lumbar spine FEMs of this study were deemed to complete the process of calibration and validation. These models can be further used to construct other FE models and analyze the biomechanical responses of lumbar spine under different boundary and loading conditions.

Range of motion, Mises stress in cortical and Mises stress in endplate

Figure 4 **-** Fig. 6 showed the range of motion, Mises stress in cortical and Mises stress in endplate in different postures for osteoporosis patients with disc degeneration and normal patients with disc degeneration, respectively. Compared with normal patients that experienced different disc degeneration, the ROM, MSC1 and MSE of osteoporosis patients suffering from various disc degeneration decreased in all motion postures. Compared with normal patients that accomplished by disc degeneration, the decline percentages of ROM, MSC1 and MSE in osteoporosis patients with disc degeneration were 0.53–17.64%, 2.77–12.88% and 2.9–23.10%, respectively. At the same time, the most of ROM and MSC1 decrease percentages were within 10%, as shown in Fig. 4 **(d)** and Fig. 5 **(d)**. The reduction percentages of MES mainly were within 20%, as displayed in Fig. 6 **(d)**. It is worth noting that, compared with non-osteoporotic patients, the decrease percentages of ROM and MSE of

osteoporotic patients gradually increases as the degree of disc degeneration increases in the most postures. With disc degeneration progressively changing, the reduction percentages of MSC1 were slightly reduced in osteoporotic patients compared with normal patients.

Mises stress in cancellous and Mises stress in post

Compared with disc degeneration patients without osteoporosis, the Mises stress in cancellous and Mises stress in post of disc degeneration patients with osteoporosis increased, as displayed in Fig. 7 - Fig. 8. Compared with non-osteoporosis patients that experienced by disc degeneration, the increase percentages of MSC2 and MSP were 43.01–73.17% and 1.46–24.09%, respectively. The increase percentages of MSC2 were basically above 45%, as displayed in Fig. 8 **(d)**. Figure 9 **(d)** showed that the growth rate of MSP almost all were just below 20%. Compared with normal patients, the increase percentages of MSC2 in osteoporotic patients gradually increased with increasing disc degeneration. However, with an increasing degree of disc degeneration the growth percentage of MSP was uneven.

Discussions

In this study, a normal lumbar spine model and three degenerative lumbar spine models with different degeneration degrees were developed and validated. Then the degenerative lumbar spine models for osteoporotic patients were constructed on the basis of the above mentioned degeneration models. Finally, the range of motion, Mises stress in cortical, Mises stress in endplate, Mises stress in cancellous and Mises stress in post were solved and analyzed, to explore the differences of biomechanical effects of disc degeneration on osteoporotic patients and non-osteoporotic patients.

Various biomechanical studies using FE models had been published in literatures to clarify the spinal biomechanics, including the biomechanical studies of lumbar disc degeneration [14–22] and the investigations of osteoporotic vertebral compression fractures [8, 38–39]. These studies individually evaluated the effect of lumbar disc degeneration on the stress, strain and flexibility of the lumbar spine [14–22], or analyzed the load or stress of osteoporotic vertebral compression fractures and the outcomes after vertebral augmentation [6–9, 38–39]. However, there is a lack of researches that combines disc degeneration and osteoporosis to explore their combined effects.

Whether it is for osteoporosis patients or non-osteoporosis patients, when they encounter lumbar disc degeneration and need to carry out spinal fusion surgery, the stability of surgical segment after surgery is a recovery indicator that patients and clinicians are very concerned about. That is because the spinal instability often leads to appearances of many complications, such as disc height decrease, cage subsidence, endplate damage, and vertebral body non-fusion. This greatly increases the risk of patients undergoing secondary operations, which in turn brings severer pain and greater financial burden to patients. Therefore, it is of great significance to evaluate the comprehensive influence of disc degeneration and osteoporosis on spinal flexibility. Currently, many scholars have confirmed that the disc degeneration will decrease the spinal flexibility and correspondingly increase the spinal rigidity, which is inevitable result of the changes in geometries and material properties of intervertebral disc due to the disc degeneration [14–15, 18, 21]. The results of this paper showed that the effect of intervertebral disc degeneration on the flexibility of normal patients and osteoporotic patients was almost the same. Although the ROM of osteoporotic patients suffering from disc degeneration is slightly smaller than that of normal patients, the change rate is almost all within 10%, as shown in Fig. 4. This may be

because the intervertebral disc is responsible for bearing main motion of the lumbar spine, and as long as the disc degeneration degree of osteoporotic patients and non-osteoporotic patients is approximately similar, their ROM will not show a big difference. At the same time, a previous study has found that compared with patients with normal bone quality, patients with osteoporosis have almost no changes in ROM [40]. Yang et al. used an electromagnetic tracking device to investigate the ROM of whole lumbar spine of 90 elderly subjects with different bone densities. They showed that there was no significant change in the ROM of lumbar spine in patients with or without osteoporosis [41]. In addition, the kinematic characteristics of the lumbar spine were found to be largely dependent on the disc morphologies, such as disc height, cross-sectional area and disc degeneration degress [18–19, 21, 41]. These investigations all proved from the side that the intervertebral disc played a major role in maintaining spinal motion. Therefore, it is understandable that osteoporotic patients and non-osteoporotic patients did not show a significant difference in ROM, which is because their disc geometries and disc degeneration degrees are almost same.

However, it is an indisputable fact that the respective effect of disc degeneration or osteoporosis will both cause changes in stress and strain of spinal important components [7–9, 18–19, 21–23, 39–40]. The results of this study showed that, compared with non-osteoporotic patients suffering from disc degeneration, the MSC1 and MSE of osteoporosis patients suffering from disc degeneration reduced, as shown in Fig. 5 - Fig. 6. Their reduction ranges were 2.77–12.88% and 2.9–23.10%, respectively. The structure of a vertebral body is composed of porous trabecular bone and dense cortex. Osteoporosis reduced the bone density of a patient and weakened the structural strength of the bones [42]. The cortical bone and endplate are on the outside of the vertebrae, and osteoporosis decreases their elasticity modulus, which leads to a reduction in their ability to bear loads. This may be the reason for the decline of MSC1 and MES in osteoporotic patients. In daily life, the spine often bears more than this degree of load and the vertebral fractures are related to the loss of bone density and changes in load patterns, which may increase the risk of fractures in osteoporotic patients with disc degeneration. However, it is worth noting that different spinal components exhibit different load-bearing capabilities (for example, cortical bone is much harder than other bony structures). Therefore, it is difficult to explain the stress distribution in the model when considering the general edge strips, because the upper limit of the edge strips may indicate the critical area of one tissue type, while the fracture strength is significantly lower in another tissue type [43]. This explains that MSC1 is far greater than the stress on other spinal structures. In addition, after the reduction of endplate's ability to bear the load, excessive peak stress may cause the endplate to damage and rupture, and even have the risk of accelerating endplate degeneration. Once endplate suffers from degeneration, it may accelerate the degeneration of the intervertebral disc [44]. Since the endplate is a bridge for the transport of nutrients, degeneratie endplate can interrupt the transport of nutrients, resulting in impaired viability of the intervertebral disc cells [45]. In general, for osteoporotic patients suffering from disc degeneration, the reduced abilities of cortical bone and endplate to bear loads may increase the risk of fracture and accelerate the disc degeneration.

Cancellous bone in the middle of the vertebral body is surrounded by cortical bone and endplate, and the bone density loss rate of cancellous bone is greater than the latter two, which leads to a further decrease in the bearing capacity of the vertebral body [42]. At the same time, because cancellous bone is a brittle material, a large loss of cancellous bone density will further increase the brittleness of cancellous bone, which will increase the probability of vertebral fractures. Previous literatures have also shown that patients with osteoporosis are much more likely to suffer from vertebral fractures than normal patients [7, 9, 42–43]. Disc degeneration changed the geometries and tissue material properties of the interertebral disc. The main manifestation was

that disc height decreased and the nucleus pulposus gradually changed from liquid behavior to solid behavior, which reduced the bearing capacity of the intervertebral disc. The total load of lumbar spine is generally constant, which may lead to the changes in the load distribution of spinal components. This means that excess loads may be transferred to other spinal components, and a significant portion of the load may be transferred to them due to endplates and cancellous bone lie below the disc. The slight reduction of endplate stress in osteoporotic patients with disc degeneration in this study may be due to the cartilage endplate being simulated in our model. At the same time, the results of this study showed that compared with patients without osteoporosis, the MSC2 and MSP increased of osteoporosis patients with disc degeneration, as shown in Fig. 7-Fig. 8. Wang et al. explored the effect of osteoporosis on internal fixation after spinal osteotomy, and they found a substantial increase in fixation instrument stress and cancellous bone stress in patients with osteoporosis, which corroborated our analysis above [8]. The increase of MSP in osteoporotic patients with disc degeneration may be due to the increased force between posterior facet joints. Anagnostidis et al. performed an assessment of stress patterns in spinal motion segments in normal and osteoporotic models with or without disc degeneration. They found that osteoporosis caused an uneven increase in facet joint loading, which was more pronounced in scenarios that simulating disc degeneration. This suggest that the abnormally increased facet joint force in osteoporotic patients results in increasing more load on the posterior elements [43]. In sum, the increase in the loss rate of bone density in cancellous bone leads to an increase in its fragility, which may be one of the reasons why cancellous bone of osteoporosis patients is prone to fracture. The increased stress of posterior elements in osteoporosis patients with disc degeneration may be a consequence of the increased force of the posterior facet joints.

The results of this study showed that the effects of different degrees of disc degeneration on the flexibility of osteoporotic and non-osteoporotic patients were not significantly different. Although the differences between the ROM values of osteoporotic patients and normal patients were very small. With the increase in degree of disc degeneration the reduction proportion of ROM in osteoporotic patients almost gradually increased. This may be because patients suffering from more severe disc degeneration had greater reductions in ROM. Therefore, osteoporotic patients suffering from moderate and severe disc degeneration will show a greater percentage of ROM decline. Compared with non-osteoporosis patients that experienced disc degeneration, the MSC1 in osteoporotic patients slightly reduced. From the perspective of the numerical size, the MSC1 in both normal and osteoporotic patients was much larger than the stresses on other spinal components. This demonstrated that the cortical bone's ability to bear loads was much greater than that of other spinal components. In addition, the decrease percentages of MSE in osteoporotic patients gradually increased with the progressive changes in disc degeneration. Both disc degeneration and osteoporosis caused changes of endplate’s geometries and material characteristics, which in turn lead to the stresses changes in endplate. The former stiffened the endplate and reduced the thickness of endplate, which result in an increase in endplate stress, which had been confirmed by previous studies [26]. The latter will greatly decrease the elasticity modulus of the endplate, which reduces the endplate stress from the perspective of this study. Therefore, we can think that the impact of osteoporosis on the endplate is greater than that of intervertebral disc degeneration. With the increase of disc degeneration degree, the MSC2 in osteoporosis patients gradually increased compared with normal patients. Most likely because the cancellous bone is located just below the disc, and disc degeneration and osteoporosis alter the load distribution of the lumbar spine, making it bear a larger load. It's worth noting that, with the progressive changes of disc degenerations, the increase percentages of the MSP in osteoporosis patients has no obvious

regularity, which may be due to the increased unevenness of posterior facet joint force caused by disc degeneration. That is to say, this phenomenon caused the changes of MSP uneven.

This study has some limitations. Firstly, the degeneration model and osteoporosis model in this study were established based on CT images of a single subject, and morphological differences between individuals may lead to differences in movement patterns and stress distributions. Secondly, we only validated the range of motion in the normal and degenerated models, which is an inherent limitation of current spine finite element analyses. Finally, the follower loads was used to simulate the synergy of the spinal muscles, but we did not build muscles in our model, which played an important role in regulating lumbar spine movement and maintaining stability. Although this study had the above limitations, the established lumbar spine model in this paper showed good agreement with the published experimental data, which will help to study the synergistic effect of disc degeneration and osteoporosis.

Conclusions

Our FE biomechanical study compared the flexibilities and stresses of lumbar spine in non-osteoporotic and osteoporotic patients suffering from disc degeneration. The results showed that both disc degeneration and osteoporosis altered the flexibility of lumbar spine and the stresses of important spinal components. Overall, the impact of disc degeneration on flexibility in the two kinds of patients (non-osteoporosis patients and osteoporosis patients) was nearly identical. By analyzing the stresses of important spinal components in the two kinds of patients, we found that osteoporosis may accelerate disc degeneration and the osteoporotic patients who suffer from disc degeneration had a higher risk of vertebral fracture.

Declarations

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Ethics approval and consent to participate: Our study was approved by the institutional ethics committee of the Affiliated Hospital of Hebei University. Written informed consent was obtained from all participants included in this study.

Consent for publication: Not Applicable.

Availability of data and materials: The datasets used and/or analysed during the current study available from the corresponding author on reasonable request.

Author's contributions: Xin-Ying Zhang and Ye Han carried out the model development and simulation, data analysis and drafted the manuscript. Xin-Ying Zhang and Ye Han participated in revising the manuscript.

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Tables

Table 1
Material properties, element types and references of normal lumbar spine model [18, 21, 24–26, 29].

Components	Young's Modulus (MPa)	Poisson's ratio	Element types	References
Cartilage endplate	24	0.4	Hexahedron	[18, 21, 24]
Cortical bone	12000	0.3	Hexahedron	[18, 21, 24]
Cancellous bone	100	0.2	Hexahedron	[18, 25]
Post	3500	0.25	Tetrahedron	[18]
Sacrum	5000	0.2	Tetrahedron	[21, 26]
Facet cartilage	Neo-Hookean, C10 = 2.0		Hexahedron	[18, 21]
Nucleus pulposus	Hyper-elastic, Mooney–Rivlin, C1 = 0.12, C2 = 0.03		Hexahedron	[21, 26, 29]
Annulus ground	Hyper-elastic, Mooney–Rivlin, C1 = 0.18, C2 = 0.045		Hexahedron	[21, 26, 29]
Annulus fibers	Nonlinear curves (stress–strain)		SPRINGA	[18, 21, 26]
Seven ligaments	Nonlinear curves (deflection–force)		SPRINGA	[18, 21, 26]

Seven ligaments：Anterior longitudinal ligaments, Posterior longitudinal ligaments, Capsular ligaments, Flavum ligaments, Interspinous ligaments, Supraspinal ligaments, Intertransverse ligamentss.

Table 2
Grading system and FE implementation of lumbar spine disc degeneration [14, 18–21, 26].

Variables of lumbar disc morphology	Normal	Mild	Moderate	Severe
Disc height loss[14, 21]				
Height loss of anterior disc (mm)	0	2.95 (1/5)	5.91 (2/5)	8.86 (3/5)
Height loss of posterior disc (mm)	0	2.08 (1/5)	4.16 (2/5)	6.24 (3/5)
Anterior osteophytes formation [19–21]				
L4-L5 segment upper osteophytes (mm)	0	3.21 (1/10)	6.42 (2/10)	9.63 (3/10)
L4-L5 segment lower osteophytes (mm)	0	3.09 (1/10)	6.18 (2/10)	9.27 (3/10)
The volume of lumbar spine disc [18–19, 21]				
Nucleus pulposus volume (mm3)	6953	4623	2755	1556
Annulus ground volume (mm3)	7382	6794	5902	5312
Endplate sclerosis [20–21, 26]				
Degree of sclerosis	No sclerosis	Mild sclerosis	Moderate sclerosis	Severe sclerosis
Changes of curvature	No changes	No changes	Flatten slightly	Flatten slightly
Thickness reduction of endplate (mm)	0	0.1 (1/6)	0.2 (1/3)	0.3 (1/2)

Disc height loss: As shown in Table 2, anterior disc height loss and posterior disc height loss were different. But in this study, the disc height was decreased in the same proportion. In mild, moderate and severe degeneration models, their disc heights decreased by 1/5, 2/5 and 3/5 of the normal lumbar disc height, respectively.

Anterior osteophytes formation: The shapes of anterior osteophytes in degenerative lumbar spine models were complicated. So the height and length of the anterior osteophytes in the sagittal plane were defined as equal to simplify formation of the osteophytes, as shown in Fig. 2. In mild degeneration model, moderate degeneration model and severe degeneration model, the height and length of L4-L5 segment upper osteophytes was defined as 1/10, 2/10 and 3/10 of sagittal diameter of the L4 vertebrae. The related dimension proportions of L4-L5 segment lower osteophytes were defined the same as L4-L5 segment upper osteophytes, as shown in Table 2.

The volume of lumbar spine disc: As the degree of disc degeneration increased, the volume of the nucleus pulposus and annulus ground both declined. In present study, the decrease in the volume of nucleus pulposus and annulus ground were simulated by a combination of their areas and disc heights. The changes of intervertebral disc volume were shown in Table 2.

Endplate sclerosis: Endplate sclerosis exhibited different characteristics in different stages of disc degeneration. In the early stage of disc degeneration (mild degeneration), endplate sclerosis was only reflected in the reduction of endplate thickness, and its curvature did not change much. In moderately and severely degenerated models, endplate sclerosis will cause the changes in thickness and curvature of endplate. As the degree of disc

degeneration increased, the curvature of the endplate gradually flattened and the thickness of endplate decreased from 0.6 mm to 0.3 mm. The detailed sizes of the cartilage endplate thickness were shown in Table 2.

Table 3
Material properties, element types and references of some important tissues in degenerative lumbar spine models [18–21, 26].

Different degenerative models	Young's modulus (MPa)	Poisson's ratio	Element types
Mild degeneration [18–21, 26]			
Cartilage endplate	24	0.4	Hexahedron
Osteophytes	100	0.2	Hexahedron
Soft tissue	Hyper-elastic material, C1 = 0.4, C2 = 0.1		Hexahedron
Annulus ground	Hyper-elastic material, C1 = 0.4, C2 = 0.1		Hexahedron
Nucleus pulposus	Hyper-elastic material, C1 = 0.14, C2 = 0.035		Hexahedron
Moderate degeneration [18–21, 26]			
Cartilage endplate	50	0.4	Hexahedron
Osteophytes	100	0.2	Hexahedron
Soft tissue	Hyper-elastic material, C1 = 0.6, C2 = 0.15		Hexahedron
Annulus ground	Hyper-elastic material, C1 = 0.6, C2 = 0.15		Hexahedron
Nucleus pulposus	Hyper-elastic material, C1 = 0.17, C2 = 0.041		Hexahedron
Severe degeneration [18–21, 26]			
Cartilage endplate	100	0.4	Hexahedron
Osteophytes	100	0.2	Hexahedron
Soft tissue	Hyper-elastic material, C1 = 0.9, C2 = 0.23		Hexahedron
Annulus ground	Hyper-elastic material, C1 = 0.9, C2 = 0.23		Hexahedron
Nucleus pulposus	Hyper-elastic material, C1 = 0.19, C2 = 0.045		Hexahedron

Table 4
Material properties, element types and references of some important tissues in degenerative lumbar spine models for osteoporotic patients [31–33].

Components	Young's modulus (MPa)	Poisson's ratio	Element types
Cartilage endplate (normal)	16 (24 decreased 33%)	0.4	Hexahedron
Cartilage endplate (mild L4-L5)	16 (24 decreased 33%)	0.4	Hexahedron
Cartilage endplate (moderate L4-L5)	33 (50 decreased 33%)	0.4	Hexahedron
Cartilage endplate (severe L4-L5)	66 (100 decreased 33%)	0.4	Hexahedron
Cortical	7920 (12000 decreased 33%)	0.3	Hexahedron
Cancellous	33 (100 decreased 66%)	0.2	Hexahedron
Post	2310 (3500 decreased 33%)	0.25	Tetrahedron
Sacrum	3333 (5000 decreased 33%)	0.2	Tetrahedron

The lumbar disc degeneration was accompanied by endplate sclerosis and it will cause the material properties of endplate to harden, so the material properties of endplate in the L45 degeneration segment were different. But the percentage of decrease in material properties of endplate was same when the lumbar disc degeneration occured for osteoporotic patients.

Table 5
Comparison of FE data and experimental data (L4-L5 segment's ROM) for lumbar spine degeneration models under 10 Nm pure moment load [37].

ROM (°)	This study (Finite element results)			Experimental data by Mimura et al. [37]		
	Mild degeneration	Moderate degeneration	Severe degeneration	Mild degeneration	Moderate degeneration	Severe degeneration
Flexion-Extension	11.02°	8.53°	5.39°	12.5 ± 3.5°	12.0 ± 2.8° 10.8 ± 2.5°	8.7 ± 2.3°
Lateral-Bending	9.07°	5.58°	2.53°	11.3 ± 2.2°	9.8 ± 3.0° 7.9 ± 2.7°	2.5 ± 1.9°
Torsion	3.79°	1.93°	1.04°	2.5 ± 1.9°	3.0 ± 1.9° 4.0 ± 2.2°	2.6 ± 2.3°

Figures

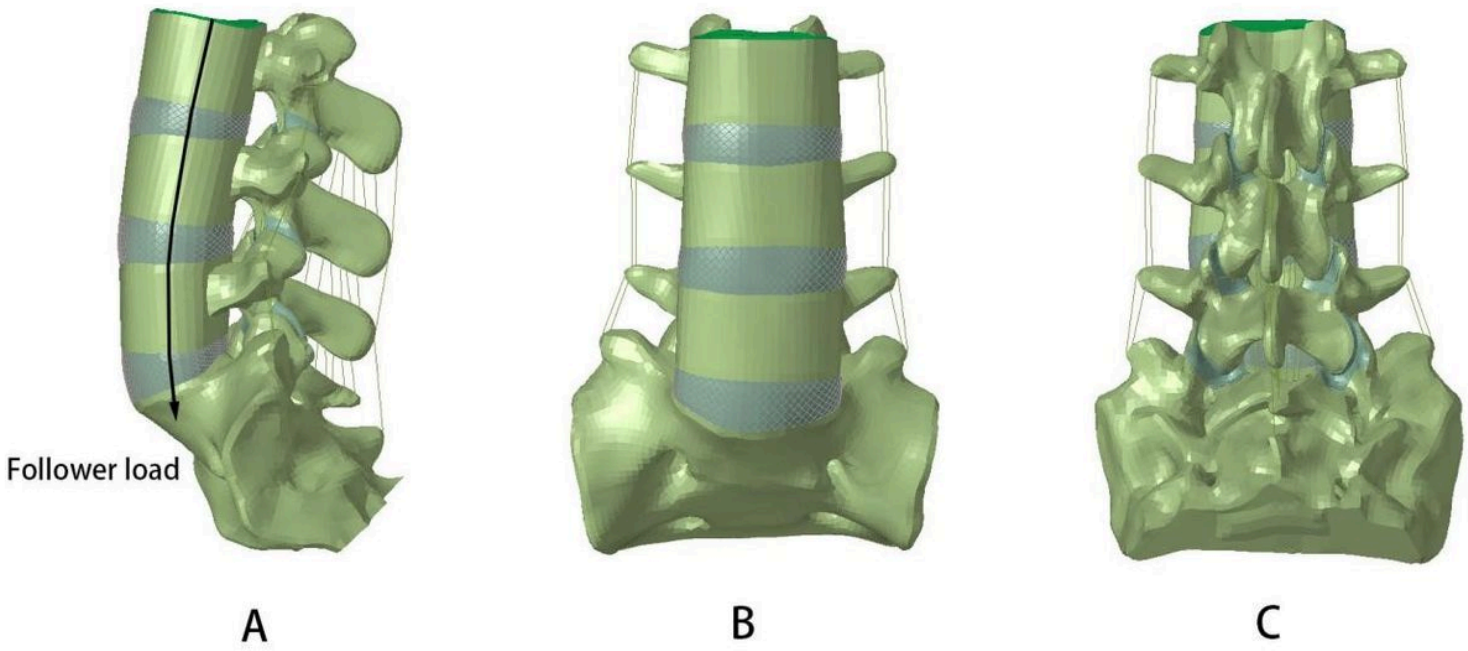

Figure 1

Non-linear three-dimensional FE model of L3-S1 normal lumbar spine.

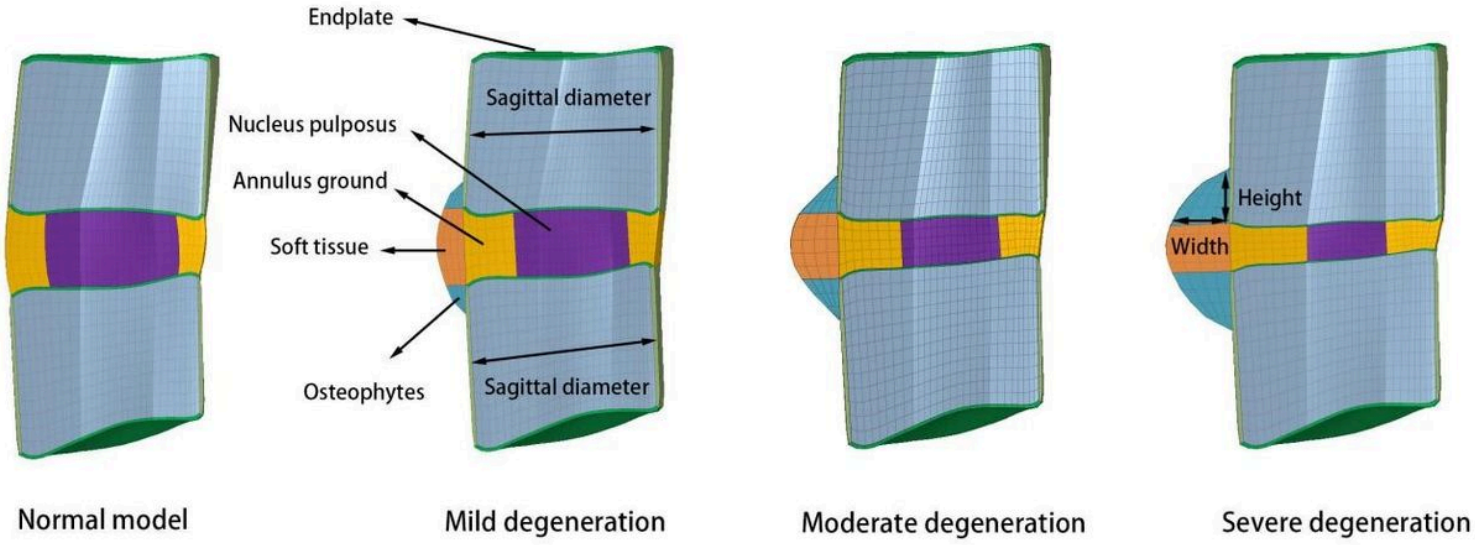

Figure 2

Non-linear three-dimensional L4-L5 segment FE models of a normal lumbar spine and three degenerative lumbar spine.

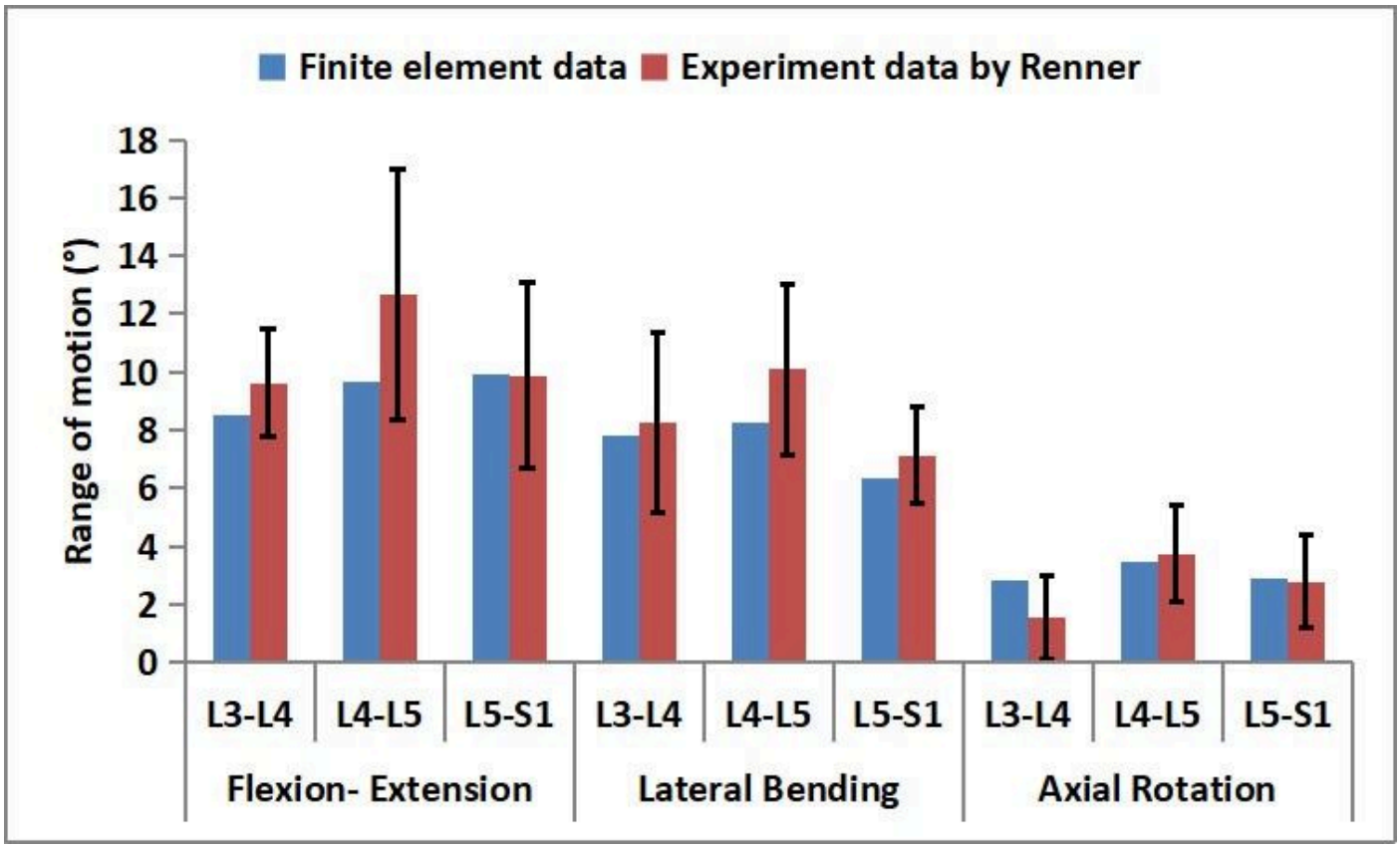

Figure 3

Comparison of finite element data (ROM) in this study and experimental data by Renner [36].

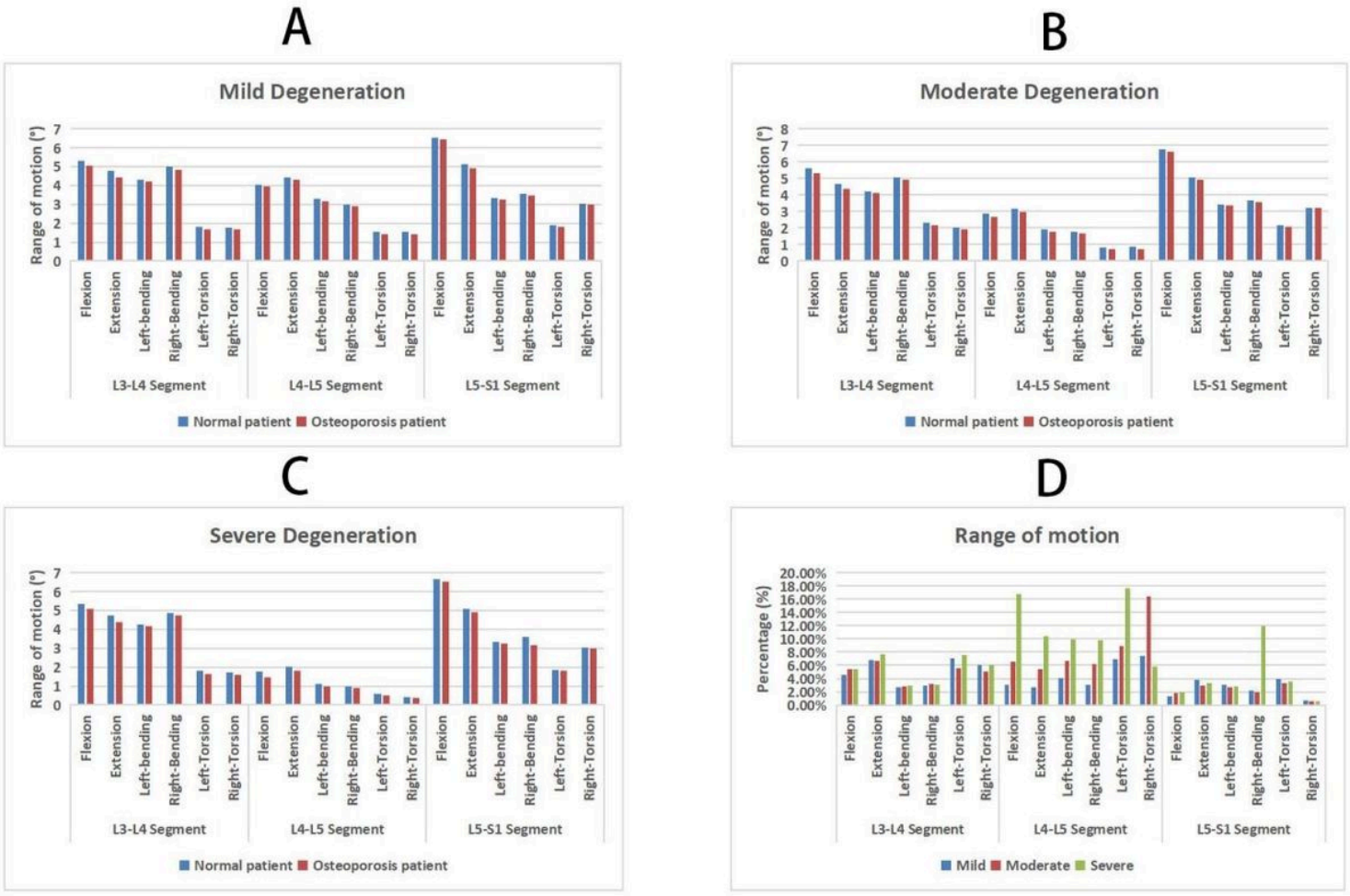

Figure 4

The ROM in different postures for osteoporosis patients and normal patients , which both experienced disc degeneration. (a): L34 segment (b): L45 segment (c): L51 segment (d): The decrease percentages (Range of motion) of osteoporosis patients compared with normal patients.

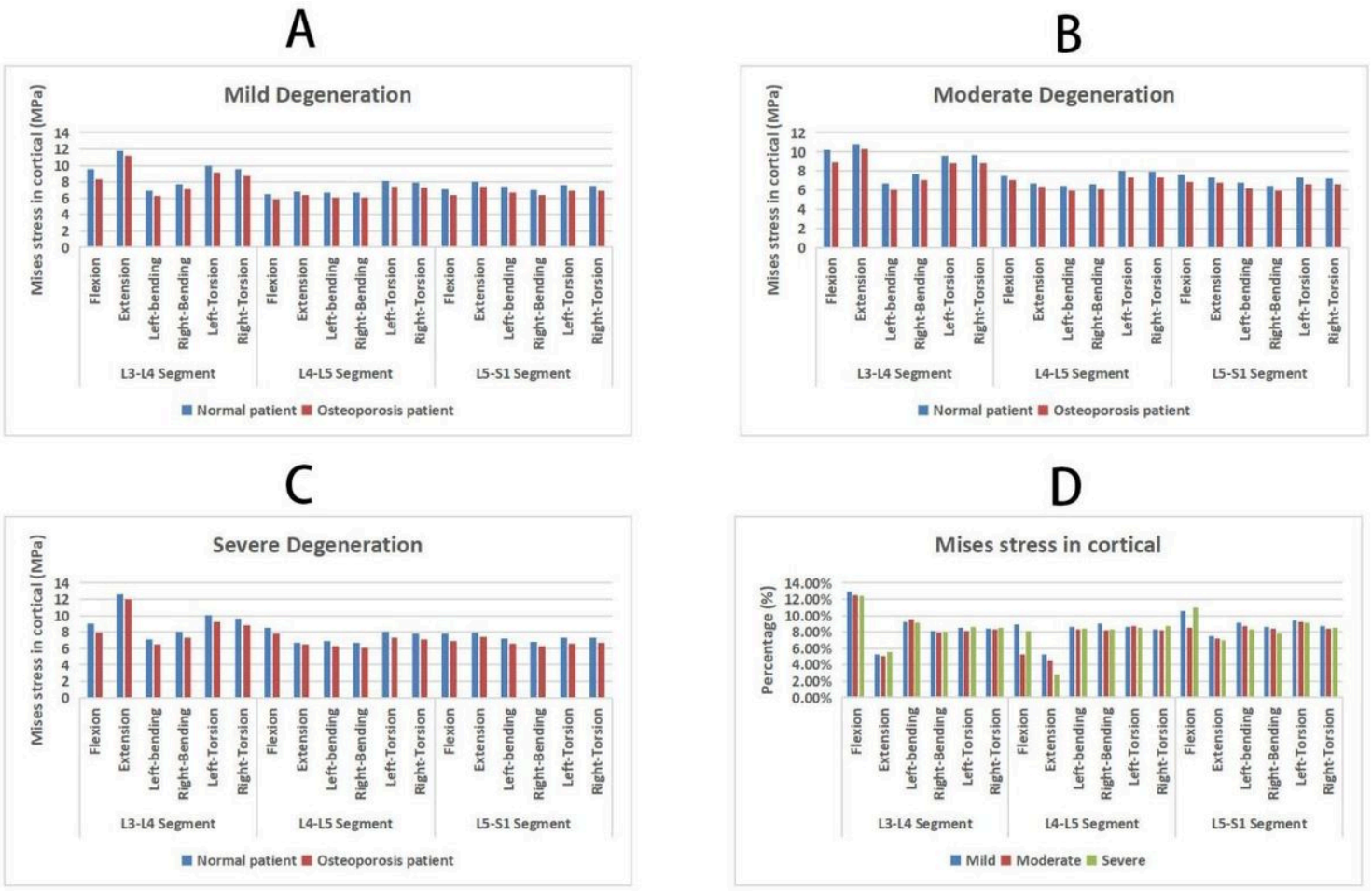

Figure 5

The Mises stress in cortical in different postures for osteoporosis patients and normal patients, which both experienced disc degeneration. (a): L34 segment (b): L45 segment (c): L51 segment (d): The decrease percentages (Mises stress in cortical) of osteoporosis patients compared with normal patients.

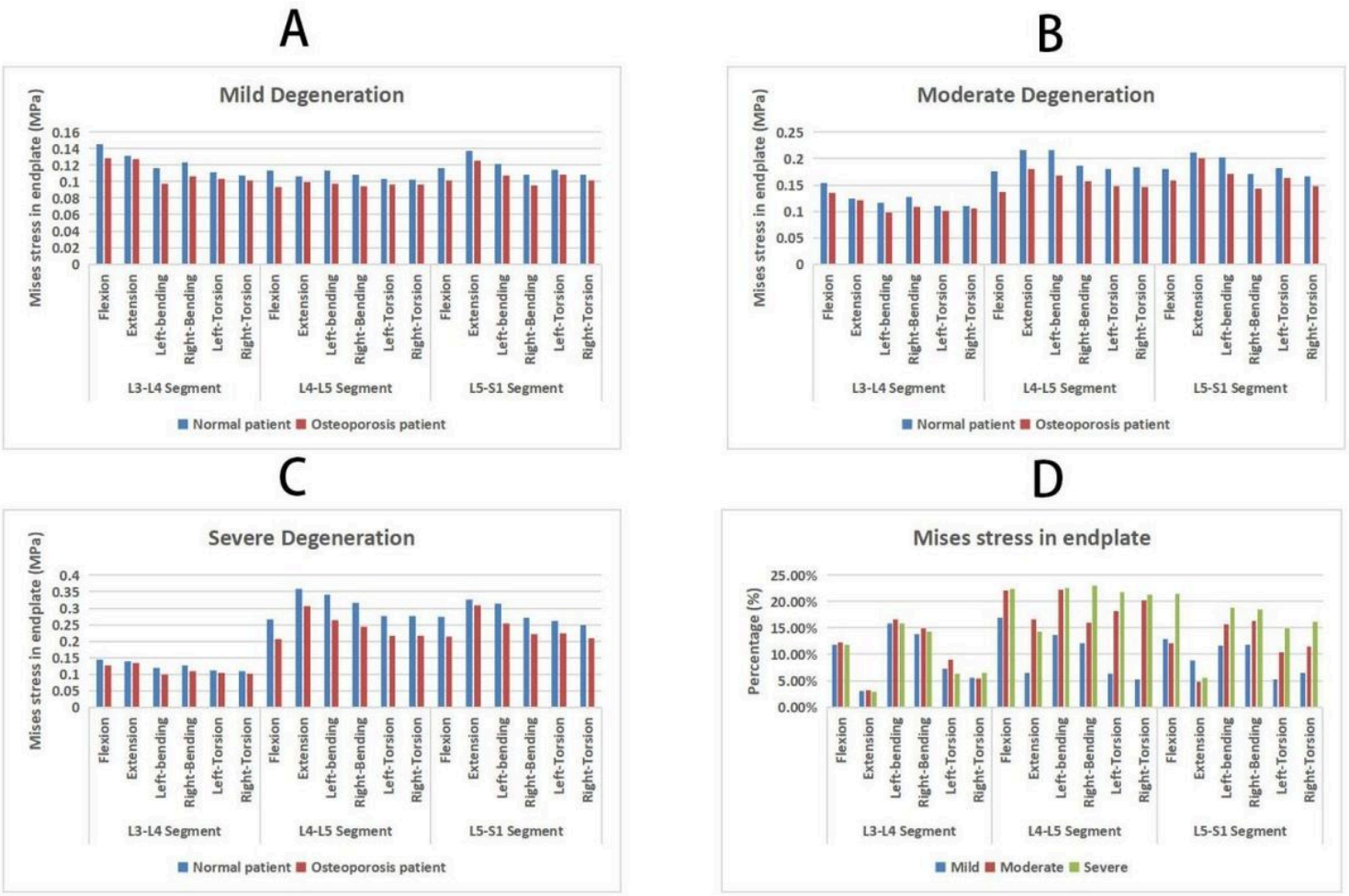

Figure 6

The Mises stress in endplate in different postures for osteoporosis patients and normal patients, which both experienced disc degeneration. (a): L34 segment (b): L45 segment (c): L51 segment (d): The decrease percentages (Mises stress in endplate) of osteoporosis patients compared with normal patients.

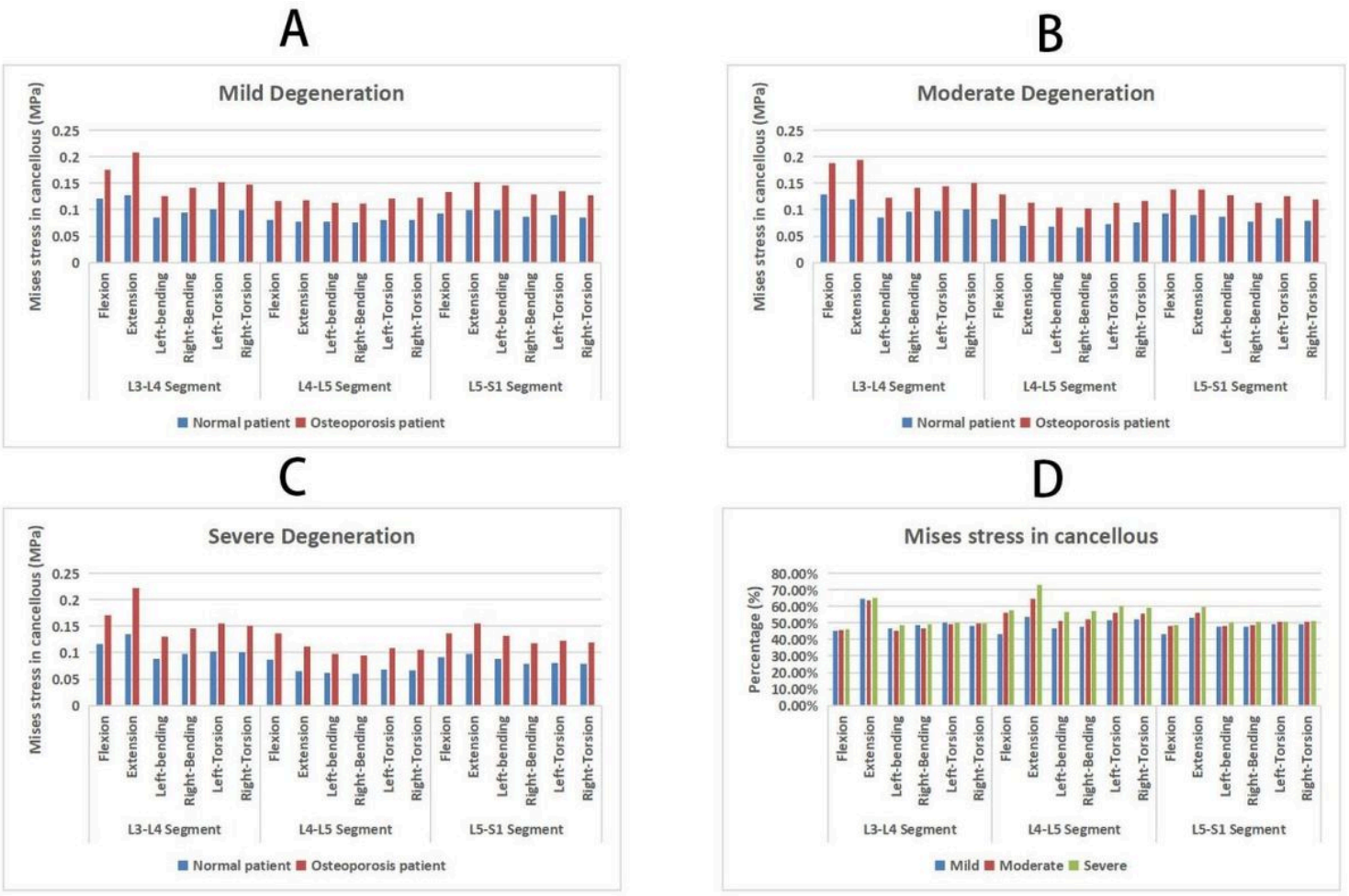

Figure 7

The Mises stress in cancellous in different postures for osteoporosis patients and normal patients, which both experienced disc degeneration. (a): L34 segment (b): L45 segment (c): L51 segment (d): The increase percentages (Mises stress in cancellous) of osteoporosis patients compared with normal patients.

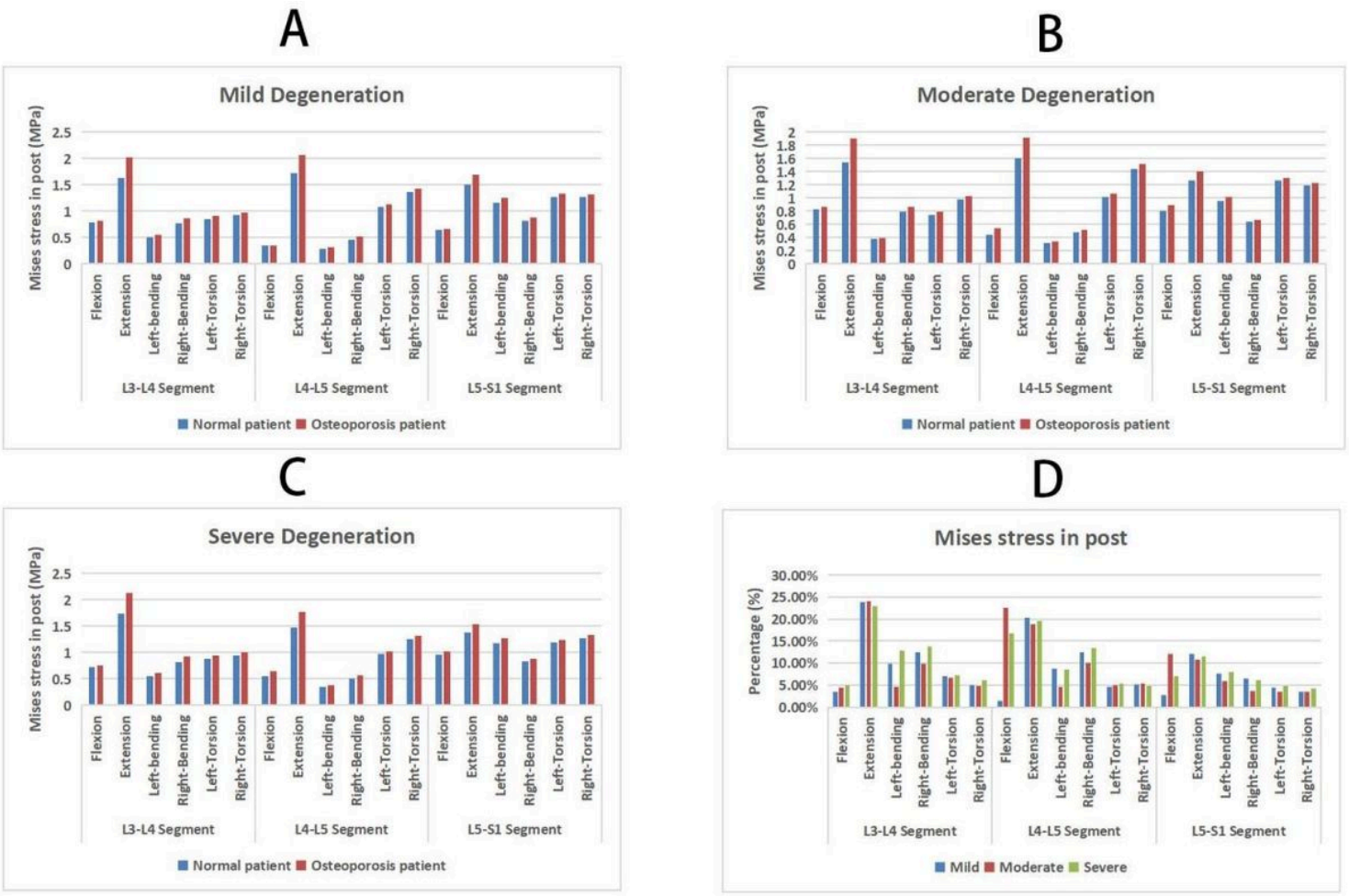

Figure 8

The Mises stress in post in different postures for osteoporosis patients and normal patients, which both experienced disc degeneration. (a): L34 segment (b): L45 segment (c): L51 segment (d): The increase percentages (Mises stress in post) of osteoporosis patients compared with normal patients.

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Figure 9

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