SCIENTIFIC ARTICLE

Kinematic Characteristics and Biomechanical Changes of Lower Lumbar Facet Joints Under Different Loads

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Objective: To explore the kinematic biomechanical changes and symmetry in the left and right sides of the facet joints of lumbar spine segments under different functional loads.

Methods: Participants (n = 10) performing standing flexion and extension movements were scanned using computed tomography (CT) and dual fluoroscopy imagine system. Instantaneous images of the L_3–S_1 vertebrae were captured, and by matching a three-dimensional CT model with contours from dual fluoroscopy images, in vivo facet joint movements were reproduced and analyzed. Translations and rotations of lumbar vertebral (L_3 and L_4) facet joints of data were compared for different loads (0, 5, 10 kg). The participants performed flexion and extension movements in different weight-bearing states, the translations and angles changes were calculated respectively.

Results: From standing to extension, there were no statistical differences in rotation angles for the facet joint processes of different vertebral segment levels under different weight loads ($P > 0.05$). Mediolateral axis and cranio-caudal translations under different weight loads were not statistically different for vertebral segment levels ($P > 0.05$). Anteroposterior translations for L_3 (1.4 ± 0.1 mm) were greater than those for L_4 (1.0 ± 0.1 mm) under the different load conditions ($P = 0.04$). Bilaterally, mediolateral, anteroposterior, and cranio-caudal translations of the facet joints under different weights (0, 10 kg) for each segment level (L_3 and L_4) were symmetric ($P > 0.05$). From flexion to standing, there were no statistical differences in rotation angles for different weights (0, 5, 10 kg) for each level (L_3 and L_4) ($P > 0.05$). There were no statistical differences between mediolateral, anteroposterior, and cranio-caudal translations at each segment level (L_3 and L_4) under different loads ($P > 0.05$). Under the condition of no weight (0 kg), L_3 mediolateral translations on the left side (1.7 ± 1.6 mm) were significantly greater ($P = 0.03$) than those on the right side (1.6 ± 1.6 mm). Left side (1.0 ± 0.7 mm) L_4 mediolateral translations were significantly smaller ($P = 0.03$) than those on the right side (1.1 ± 0.7 mm). There were no statistical differences between different weights for either anteroposterior and cranio-caudal translations ($P > 0.05$). There were no statistical differences for mediolateral, anteroposterior, and cranio-caudal translations for 10 kg ($P > 0.05$).

Conclusion: Lumbar spine facet joint kinematics did not change significantly with increased loads. Anteroposterior translations for L_3 were greater than those for L_4 of the vertebral segments are related to the coronal facet joint surface. Changes in facet surface symmetry indicates that the biomechanical pattern between facet joints may change.

Key words: Lumbar facet joints; Kinematics; Range of motion; Weight bearing; Symmetry

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Introduction

The bilateral facet joints of the lumbar spine are an important component of the vertebral arch. The lumbar intervertebral disk and bilateral facet joints, as a three-joint complex, are essential for the stability of lumbar spinal movement. It has previously been shown that one of the causes of low back pain is abnormal facet joint movement and that weight-bearing can induce biomechanical changes in the pattern of motion of the facet joints, resulting in degeneration and osteoarthritis of these joints[1]. Degenerative disease of the lumbar spine leads to changes in intervertebral spacing (affecting the disks) and increases the load on the facet joints. These changes induce low back pain. It is also believed that the normal kinematic patterns of bilateral facet joints change[2], which results in morphological deformities, poor long-term posture, and the secretion of humoral factors[3,4].

At present, many studies on facet joint movement patterns are based on the analysis of morphological data collected by imaging methods such as computed tomography (CT) and magnetic resonance imaging (MRI)[5–7]. In several studies, the movement characteristics of facet joints have also been verified using cadaver specimens and animal models[8,9]. There is also the viewpoint that facet joint degeneration can be analyzed from the perspective of the influence of cytokine factors in the spinal biochemical environment; however, because low back pain from facet joint degeneration is related to the altered biomechanical patterns of spinal segment movement, even if cytokine injection or stem cell transplantation can be used to improve the spinal internal environment, this improvement would be meaningless without addressing abnormal lumbar motion. Understanding the movement patterns of the lumbar facet joints under weight-bearing states, in vivo rather than using cadaver specimens or animal models, can clarify the pathogenesis of lumbar spinal disease; in vivo, internal biochemical (the environment of body fluid) and physiological (forces at the attachment point of the muscles and ligaments) conditions are very different from those in cadaver specimens. Current literature concentrates on comparative analysis of the disease-related characteristics of the facet joints but not on analysis of different weight-bearing conditions[10,11]. It is clear that loads on facet joints increase under weight-bearing, accelerating degeneration of the facet joints and altering associated motion patterns. The bilateral processes of the facet joints are very important for the stability of the entire spine as a unit[12]. Lumbar facet joint tropism can be used to objectively evaluate biomechanical movement patterns related to the facet joints.

The purpose of our research was to obtain movement data (translations and rotations) of the facet joints of the lumbar spine under different weights to reveal the etiology of lumbar degenerative diseases. The results may be relevant to limiting poor posture and delaying the progression of lumbar degenerative disease; these data are typically used to guide treatment in clinic, especially in planning surgical approaches, developing relevant prostheses, in positioning implants, for a better understanding of specific patient anatomy, and to improve postoperative rehabilitation[10]. However, exploring the characteristics of in vivo motion of the facet joints under weight-bearing conditions can theoretically reveal whether asymmetric motion aggravates the deterioration of lumbar intervertebral disks because the moment of the three-joint load body is reduced, causing lumbar segmental instability and related lumbosacral degenerative disease.

Therefore, we hypothesized that six-degree-of-freedom kinematic data of the lumbar facet joints of asymptomatic participants, captured in vivo under weight-bearing conditions using a combination dual fluoroscopic imaging system and CT imaging system to reproduce with modeling facet joint kinematics and analyze the motion characteristics of the different segments of the lower lumbar facet joints, which could obtain the translations and angles of the facet joints under different loads, and explore the relationship between the facet joints under different loads and disease progression. The purposes of study are: (i) to reveal the etiology of lumbar facet joint under the different loading conditions, such as low back pain; (ii) to guide surgical operation of the spine according the translations of lumbar facet joint; and (iii) to trace the data of kinematics at postoperation to conduct a plan for rehabilitation.

Methods

Participant Recruitment

We recruited 10 healthy young volunteers (four males, six females, 24.8 ± 1.8 years old, body mass index [BMI] 20.7 ± 2.1) between 20 and 30 years from the university campus. Using imaging and clinical examinations, the experimental plan was approved by the institutional review board, and an informed consent form was signed by each participant.

Inclusion Criteria

The inclusion criteria were: (i) volunteers aged 20–50 years; (ii) without chronic diseases such as cardiovascular, liver, or kidney diseases; (iii) without abnormal lumbar and physical radiology examinations; (iv) with BMI between 18.5 and 23.9; (v) with T value in bone density between − 1 and 1.

Exclusion Criteria

Participants were excluded based on the following criteria: (i) a history of spinal trauma ; (ii) history of lumbar disk herniation; (iii) lumbar spondylolisthesis; (iv) lumbar hyperplasia; (v) lumbar spine tuberculosis; (vi) history of spine tumor.

Modelling Technique

Thin-slice CT (Sensation 16, Siemens AG, Germany) scans were taken, with each participant lying supine, to obtain L_3–S_1 vertebral cross-sectional images (0.75-mm slice thickness and 512-pixel × 512-pixel resolution). The CT images were imported into Materialise Interactive Medical Image Control System software (version 17.0, Materialise NV, Belgium) and

Rhinoceros software (version 5.0, Robert McNeel & Associates, United States) to create network models based on outlines of the spine.

Image Acquisition Using Dual-Plane Fluoroscopy Image System

Using two identical forward intersecting (90° vertical) C-arm fluoroscopy machines, with each volunteer wearing a lead suit and collar to protect the thyroid gland, instantaneous images of the lumbar spine were obtained during movement for the maximum range of motion[13,14] from flexion to extension, with the pelvis and hips stationary and the duration of each end state of motion not less than 1 s. The images were filtered to remove interference and were stored in a specific format. The process was completed under the guidance of a professional orthopaedist and radiologists (Fig. 1).

Kinematics of the Facet Joint

The models based on the images obtained from the CT scans introduced into the virtual dual-screen fluoroscopy system (using Rhinoceros software) were aligned, through translation and rotation, until the bone contours of the three-dimensional CT image matched with those in the virtual fluoroscopic images (Fig. 2). The models were used to simulate the activities of the lumbar spine under different weights to obtain data in six degrees of freedom for analysis. With repeated verification, the accuracy of this technology can reach 0.3 mm and 0.7° for translations and rotation, respectively[11,13].

Coordinate System

A right-hand Cartesian coordinate system was defined by taking the upper and lower endplate surfaces and the front and rear edges of each vertebral body as the surfaces and edges, respectively, of a cylindrical structure. The geometric center of the bilateral upper articular process was defined as the origin of the coordinate system with horizontal, the direction parallel to the upper endplate of the vertebral body, defined as the x-axis (mediolateral) and pointing to the left defined as positive; the y-axis (anteroposterior) was defined as perpendicular to the x-axis pointing in the direction of the spinous process and with posterior as positive; and the z-axis was defined as cranio-caudal and was perpendicular to the x- and y-axes with the cranial direction as positive. Euler angles α, β, and γ were defined as rotations about the x-axis, y-axis, and z-axis, respectively (Fig. 3).

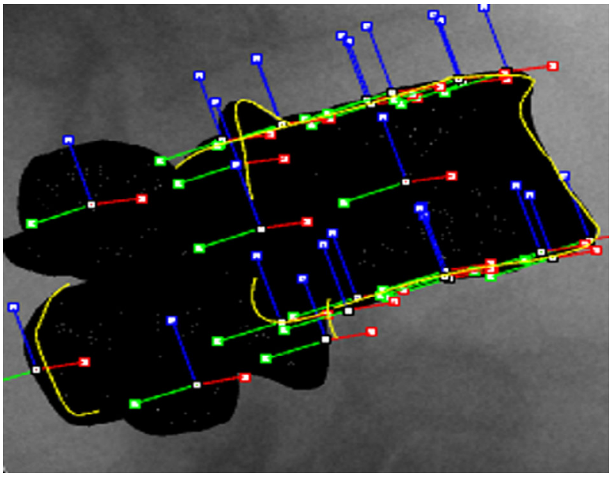

Fig. 2 Three-dimensional reconstruction of the bone contours from stereo CT and virtual fluoroscopic images to reproduce the kinematics in vivo of each segment under different loads.

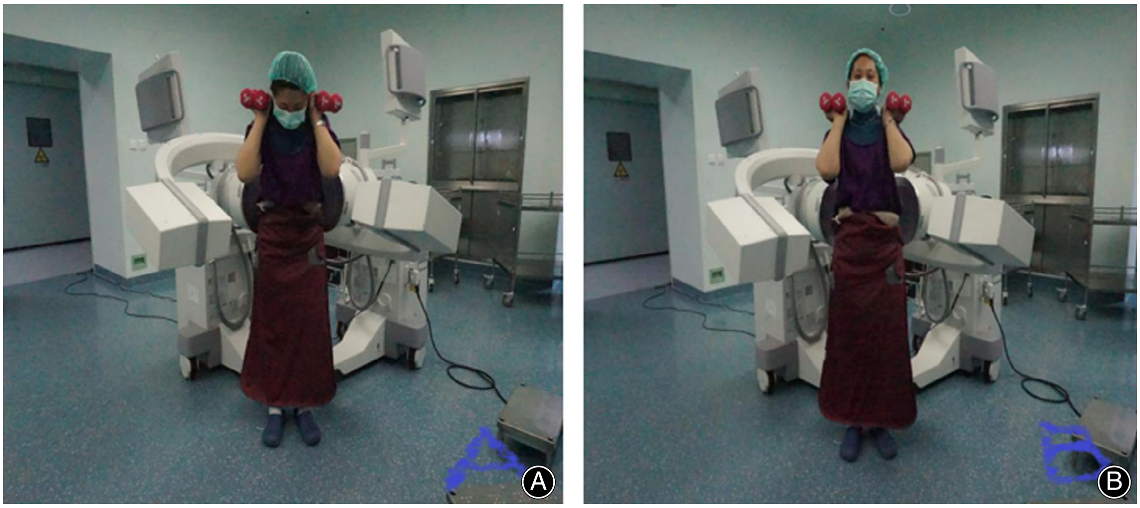

Fig. 1 (A) Dual-plane fluoroscopy used to collect flexion images of the spine under 10 kg load. (B) Fluoroscopy machines, with each volunteer wearing lead suit and collar while standing.

Calculation of the Range of Motion of Facet Joint

Facet joint translations and rotations were measured using the range of motion difference method from flexion to standing. The apex of the spinous process of the lower spine segment to the upper center joint of the adjacent upper facet joint were used, and the absolute value of the difference between the translation (or rotation) of the origin of the process were regarded as the translations (or rotations) (Fig. 4). The translations (or rotations) of the bilateral facet joints were averaged when calculating symmetry of bilateral facet joint movement, and measurements were standardized to positive values for statistical comparison.

Statistical Analysis

Continuous variables were expressed as mean ± standard deviation. In each group, L_3 and L_4 absolute displacements and angles under different load conditions (0, 5, and 10 kg) were compared using two-way analysis of variance. The absolute translations (angles) were the dependent variables, and the vertebral body level and weight were the independent variables. The differences in bilateral facet joint displacements between non-weight bearing (0 kg) and weight bearing (10 kg) were compared post hoc using paired *t* tests and were statistically significant if $P < 0.05$. The data were analyzed using SPSS software (version 26.0, IBM Corp., Armonk, NY, USA).

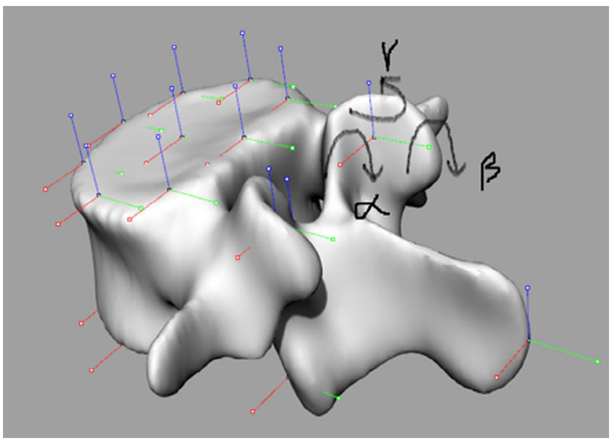

Fig. 3 Right-hand Cartesian coordinate system. Red is the x-axis, green is the y-axis, and blue is the z-axis with Euler rotation angles α, β, and γ, respectively.

Result

Movement of the Facet Joints from Flexion to Standing

Under loading of L_3 for 0, 5, 10 kg, respective mediolateral translations were 1.7 ± 1.6 mm, 1.3 ± 1.0 mm, 1.5 ± 0.8 mm; anteroposterior translations were 1.6 ± 1.2 mm, 1.8 ± 1.1 mm, 1.6 ± 1.5 mm; cranio-caudal translations were 1.9 ± 1.4 mm, 1.8 ± 1.4 mm, 1.6 ± 1.8 mm; α angles were 2.4° ± 2.1°, 2.8° ± 2.1°, 2.7° ± 2.9°; β angles were 3.1° ± 3.1°, 1.9° ± 1.9°, 1.9° ± 1.1°; and r angles were 2.6° ± 1.4°, 1.8° ± 1.6°, 1.6° ± 0.9°. Under loading of L_4 for 0, 5, 10 kg, respective mediolateral translations were 1.1 ± 0.7 mm, 1.1 ± 0.8 mm, 1.1 ± 1.0 mm; anteroposterior translations were 1.9 ± 1.9 mm, 1.6 ± 1.2 mm, 2.2 ± 1.4 mm; cranio-caudal translations were 2.3 ± 1.7 mm, 1.1 ± 0.9 mm, 2.3 ± 1.6 mm; α angles were 3.4° ± 2.6°, 2.1° ± 1.4°, 4.0° ± 2.4°; β angles were 2.2° ± 1.6°, 1.5° ± 1.4°, 2.4° ± 1.6°; and r angles were 1.6° ± 1.3°, 1.5° ± 1.3°, 3.4° ± 2.0°.

From flexion to standing, there were no statistical differences in rotation angles (α, β, r) at the level (L_3, L_4) of the facet joints under different weights (0, 5, 10 kg) ($P > 0.05$). For different weights, there were also no statistical differences between the mediolateral, anteroposterior, and cranio-caudal translations for each segment ($P > 0.05$) (Figs 5 and 6). The translations of each vertebra (L_3 or L_4) of the lumbar spine under different weights are shown in Table 1. Under the condition of no weight (0 kg), there were no statistical

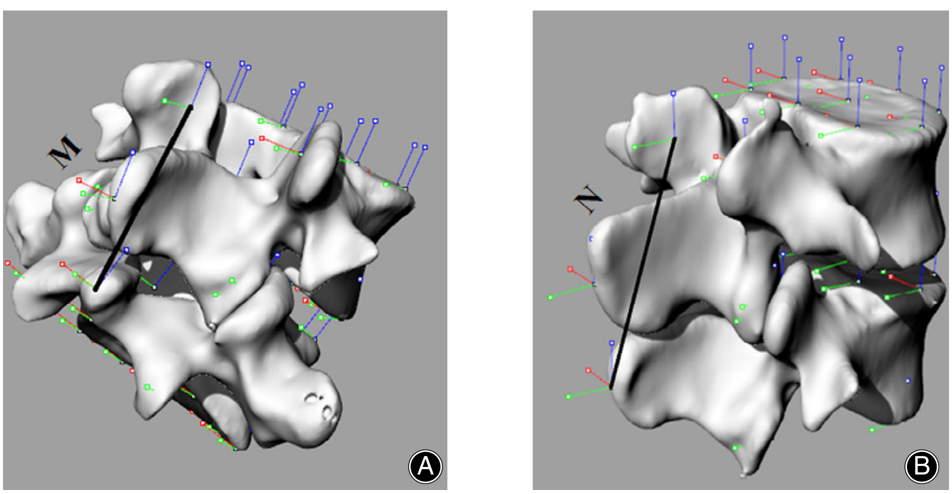

Fig. 4 (A) Shows translation A from the apex of the spinous process of the lower vertebra to the origin of the upper articular process of the adjacent vertebra during flexion. (B) Shows translation B from the apex of the spinous process of the lower vertebra to the origin of the upper articular process of the adjacent vertebra while standing. The absolute value of the difference between M and N is the translation of the absolute motion.

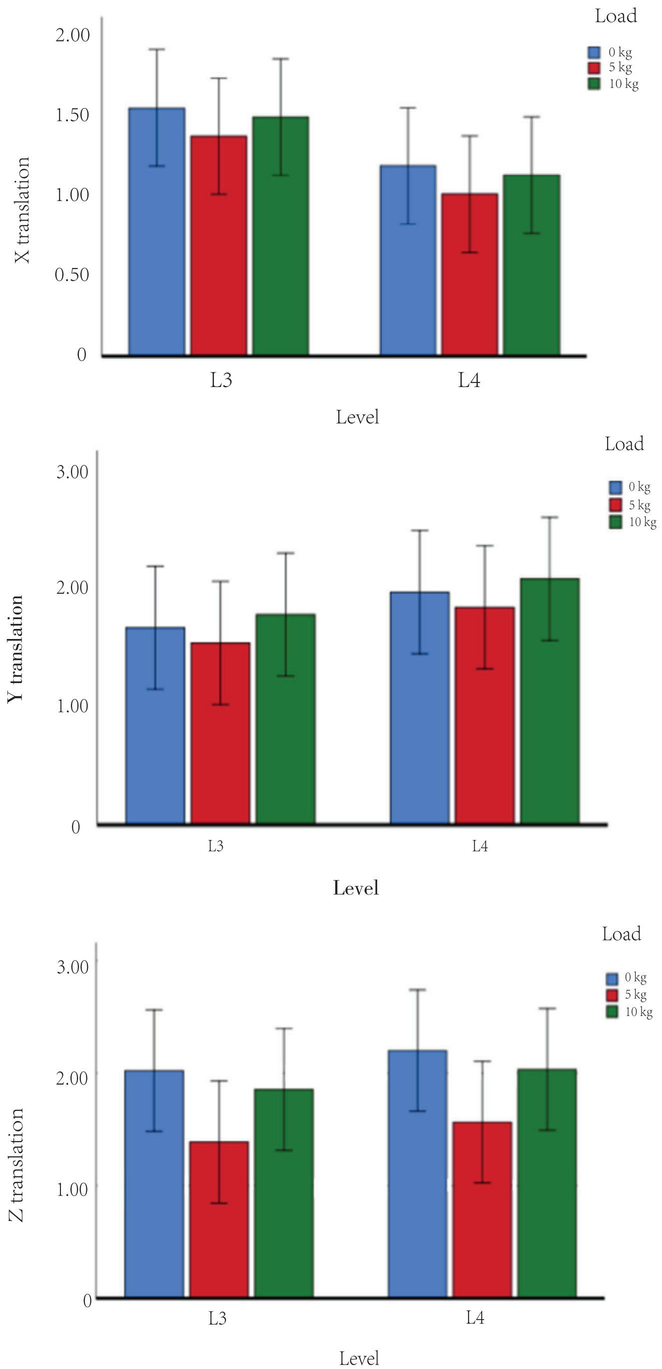

Fig. 5 Each lumbar segment level X, Y, and Z translation under different load from flexion to standing.

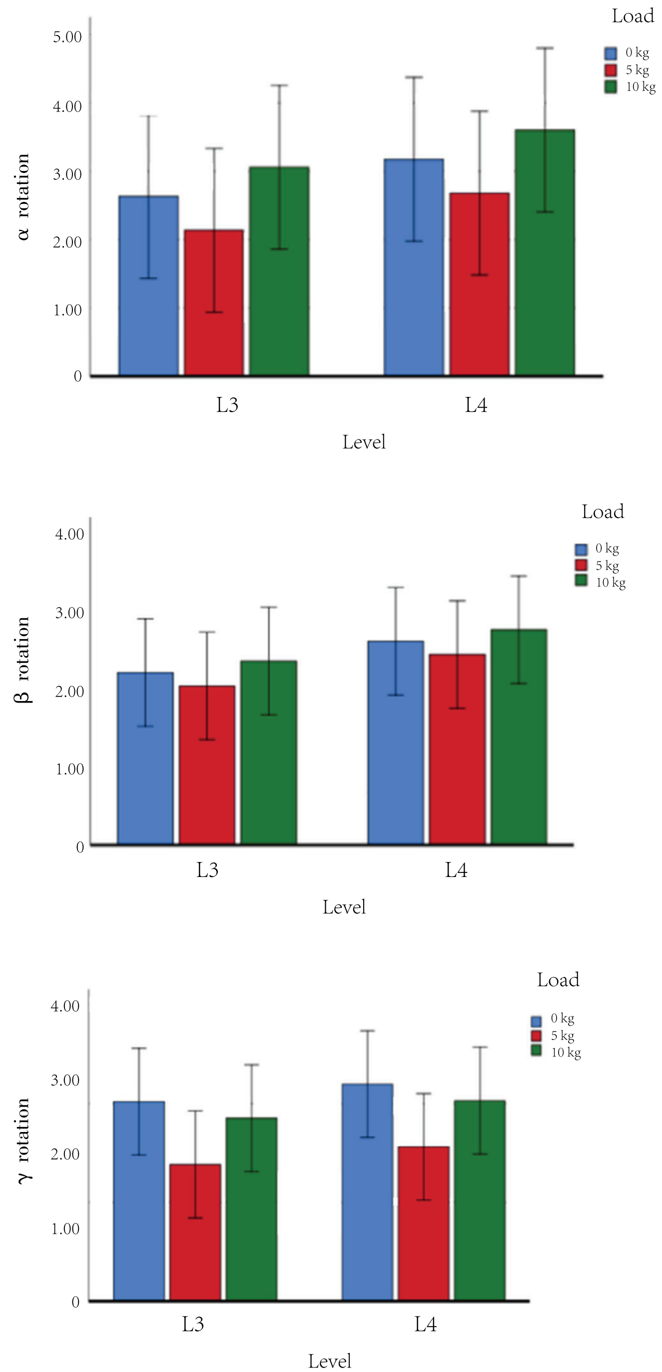

Fig. 6 Each lumbar segment level α, β and γ rotation under different loads from flexion to standing.

differences between L_3 and L_4 for anteroposterior and cranio-caudal translations ($P > 0.05$); however, mediolateral translations showed significant differences ($P < 0.05$). For 10 kg, there were no statistical differences between L_3 and L_4 in mediolateral, anteroposterior, or cranio-caudal translations for level ($P > 0.05$).

Movement of the Facet Joints from Standing to Extension

For L_3, under 0, 5, and 10 kg loading, respective mediolateral translations were 1.4 ± 1.5 mm, 1.4 ± 1.2 mm, 1.0 ± 0.9 mm; anteroposterior translations were 1.8 ± 1.2 mm, 1.3 ± 1.3 mm, 1.0 ± 1.0 mm; cranio-caudal translations were 1.2 ± 0.8

TABLE 1 The translations of the Bilateral facet joints under different weights (0 , 10 kg) for each segment level from flexion to standing (mm)

		0 Kg			10 Kg		
Translation		Left	Right	*P* value	Left	Right	*P* value
X	L_3	1.7 ± 1.6*	1.6 ± 1.6*	**0.03**	1.5 ± 0.8	1.5 ± 0.8	0.12
	L_4	1.0 ± 0.7*	1.1 ± 0.7*	**0.03**	1.1 ± 1.0	1.2 ± 1.0	0.05
Y	L_3	1.7 ± 1.3	1.5 ± 1.3	0.57	1.8 ± 1.6	1.8 ± 1.4	0.74
	L_4	1.8 ± 2.0	2.3 ± 1.9	0.20	2.6 ± 1.6	1.9 ± 1.2	0.08
Z	L_3	2.0 ± 1.2	2.0 ± 1.5	0.95	1.8 ± 2.1	1.4 ± 1.5	0.19
	L_4	2.1 ± 1.4	2.5 ± 1.7	0.26	2.0 ± 1.4	2.6 ± 1.7	0.14

Bold value represent statistical significance between left side and right side of the translation.
* Represent statistical significance between left side and right side of the translation ($P = 0.03$).

mm, 1.5 ± 1.2 mm, 1.1 ± 1.2 mm; α angles were 1.9° ± 1.4°, 1.9° ± 2.3°, 1.4° ± 1.7°; β angles were 2.3° ± 2.7°, 1.8° ± 1.9°, 1.4° ± 1.1°; and r angles were 3.2° ± 2.7°, 1.9° ± 2.3°, 1.9° ± 1.0°. For L_4, under 0, 5, and 10 kg loading, respective mediolateral translations were 0.8 ± 0.6 mm, 1.0 ± 1.2 mm, 1.1 ± 0.9 mm; anteroposterior translations were 0.7 ± 0.5 mm, 1.0 ± 0.8 mm, 1.2 ± 0.9 mm; cranio-caudal translations were 0.9 ± 0.7 mm, 1.0 ± 0.8 mm, 1.1 ± 0.7 mm; α angles 1.8° ± 1.1°, 1.3° ± 0.7°, 1.8° ± 1.2°; β angles were 2.3° ± 2.5°, 2.0° ± 1.8°, 2.7° ± 2.3°; and r angles were 1.7° ± 1.1°, 1.8° ± 1.1°, 3.7° ± 2.4°.

From standing to extension, there were no statistical differences in the rotation angles (α, β, r) of each segment level (L_3, L_4) under different weights (0, 5, 10 kg) ($P > 0.05$). There were no statistical differences between mediolateral translations and cranio-caudal translations for each segment level ($P > 0.05$). There were no statistical differences between the anteroposterior translations of L_3 and L_4 for different weights ($P > 0.05$). Anteroposterior translations were statistically different between L_3 and L_4 vertebrae ($P < 0.05$) (Fig. 7A-C). The translations of each segment level (L_3, L_4) under different weights are shown in Table 2. Movements of the bilateral facet joints at both levels (L_3, L_4) under different weights (0, 10 kg) were symmetric. There were no statistical differences in mediolateral, anteroposterior, and cranio-caudal translations.

Discussion

The characteristics of the in vivo movements of the facet joints were shown by recruiting 10 asymptomatic participants and using three-dimensional CT and dual-plane fluoroscopy imaging system-based models to reproduce the movements of the facet joints under different weight-bearing conditions. We found that the rotation angles of the vertebrae did not significantly change between different weight-bearing conditions suggesting that low weight-bearing conditions are not an important factor affecting rotation around the main axis or coupled rotations.

From standing to extension, anteroposterior L_3 translation increased compared with that of L_4, which is related to the increased angle of the facet joint surface in the coronal plane, as the anteroposterior block force increases. It may also be related to the balanced traction of the lower waist muscles and ligaments on the lumbar facet joints. This can explain why degenerative disease and low back pain seem to affect L_4 more than L_3 vertebrae with increasing stress. Regarding the symmetry of facet joint movement, from flexion to standing, mediolateral displacements of L_3 and L_4 without weight bearing were not symmetric, which may cause long-term abrasion of the facet joints and aggravate[15] biomechanical changes that induce intervertebral disk degenerative disease[16] and lower back pain. It is important to restrict harmful intervertebral movements to delay degenerative disease of the lumbar segment.

Many previous studies are based on imaging that measures the morphology of facet joints to obtain data[3,4,17] and to analyze the relationship between changes in movement patterns and spinal disease. There are also applications of CT–MRI modeling and dual fluoroscopy technology to facet joints for experimental analysis[2,10,18]. However, data analysis of facet joint movement under weight-bearing conditions is still lacking. The innovation of this experiment was using CT and dual fluoroscopy technology to reproduce the in vivo motion trajectories of lumbar facet joints under loading to reflect true and accurate physiological data under weight-bearing conditions. These data are of great significance for studying the mechanisms of lower lumbar spine to delay disease progression.

The intervertebral disks and bilateral facet joints of the adjacent vertebral bodies of the spine are very important for the stability of the spine as a composite structure[19]. The facet joints carry tensile, compressive, and shear loads when the body performs flexion and extension movements, during which, as the area of the facet joint surface changes, torsional forces increase causing compression loads imbalance. When the motion of the bilateral facet joints is asymmetric, it may further aggravate joint degeneration causing low back pain

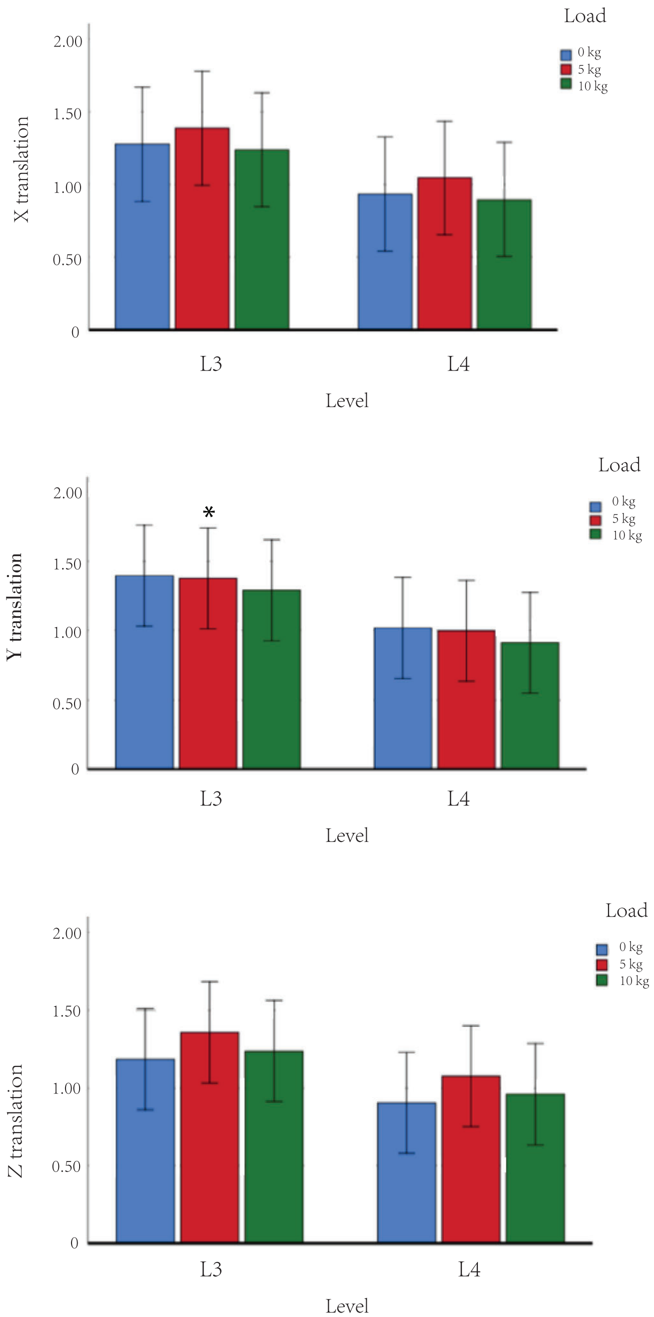

Fig. 7 Each lumbar segment level X, Y, and Z translation under different loads from standing to extension. There were statistical differences in anteroposterior translations for L_3 between L_4 under the different load conditions ($P = 0.04$).

and lumbar degenerative disease. In patients with lumbar disk herniation, the compressive stresses near the surface of the facet joints increase, leading to degeneration of articular cartilage, which may also be the basis for lower lumbar spine disease[20]. When the body moves from flexion to standing, the center of gravity is displaced forward, resulting in downward forces in the facet joints anterior components, and intervertebral compressive stresses as the contact area between the joints changes during positional changes, counter resistance to mediolateral translation is weakened, and bilateral motion of facet joint is uncoordinated.

Changes in stress in the facet joint capsule during lumbar movement is an important measure for translations. Generally speaking, stress in the facet joint will increase with translation of the facet joint, and the subsequent change in joint capsule strain may be one of the reasons for cartilage degeneration. Little *et al.*[21] found that, during in vivo lumbar spine movement, strain in the facet joint capsule increases near the fixed segment. This finding of the changes in facet joint capsule stress during in vivo movement may be rare; it is a new perspective of the cause of degeneration of adjacent vertebral segments after lumbar internal fixation.

Changes in the rotational kinematics of the facet joints of the lumbar spine are also the cause of low back pain. Li *et al.*[11] found that lumbar facet joint rotation in patients with intervertebral disk degeneration are different and that this difference is mainly reflected as increased coupling rotation angles. This increase in coupling rotation angle can make the facet joints excessively mobile, which may be the cause of articular cartilage degeneration. Because there are pain receptors in the small joint capsule, changes in the strain of the facet joint capsule can explain the mechanism of low back pain. At present, the relationship between lumbar facet joint kinematics and diseases such as low back pain is not clear. The order of degeneration of the lumbar intervertebral disk and facet joint process is also unclear because reports are limited.

We have investigated in vivo kinematic data of lumbar facet joints to explore the relationship of lumbar facet joint movement with the pathogenesis of the lumbar degenerative disease. These data are difficult to obtain from animal experiments, cadaver specimens, and CT or MRI. Our research results may provide a theoretical basis for clinical surgery. For example, in the treatment of degenerative lumbar spondylolisthesis, either decompression and fusion or simple decompression can be selected based on the stability of facet joint process range of motion. Controversy over the efficacy of the two surgical methods has always existed[22]. Fusion treatment for unstable lumbar degenerative spondylolisthesis is considered to be mainstream. With our research, we can use experimental data to accurately assess the stability of the spinal segment to select surgical procedures and fixation devices according to the specific conditions of the disease.

Our study has several limitations. First, the sample size was small, and only young asymptomatic participants were included. Different age groups and different lumbar conditions should be investigated. Second, all individuals were healthy; individuals with pathologies should be added for comparison and analysis. Third, we only collected facet joint surface data from the lumbar spine from standing to flexion and extension. In the future, we should expand the analysis to lateral flexion and rotation.

TABLE 2 The translations of the Bilateral facet joints under different weights (0, 10 kg) for each segment level from standing to extension (mm)

		0 Kg			10 Kg		
Translation		Left	Right	*P* value	Left	Right	*P* value
X	L_3	2.0 ± 1.2	2.0 ± 1.5	0.28	1.8 ± 2.1	1.4 ± 1.5	0.35
	L_4	2.1 ± 1.4	2.5 ± 1.7	0.55	2.0 ± 1.4	2.6 ± 1.7	0.89
Y	L_3	1.8 ± 1.2	1.8 ± 1.2	0.93	1.2 ± 0.9	1.3 ± 1.1	0.83
	L_4	1.1 ± 1.3	0.8 ± 0.7	0.45	1.6 ± 0.8	1.4 ± 1.0	0.56
Z	L_3	1.0 ± 0.6	1.4 ± 0.9	0.38	1.0 ± 1.2	0.9 ± 0.8	0.86
	L_4	0.9 ± 0.7	0.9 ± 0.7	0.81	1.2 ± 0.8	1.3 ± 0.7	0.82

There was no statistical significance between left side and right side of the translation.

Nevertheless, this study provides data analysis of the in vivo movements of the facet joints of lumbar vertebrae under different weight loads based on a combination of CT modeling and a dual fluoroscopy image system. This can be used to associate lumbar degenerative disease with movement through the analysis of the lumbar spine's in vivo movement data and explain the pathogenesis of lumbar spine disease, providing theoretical information to guide clinical surgical procedures.

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