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An Approach to Lumbar Vertebra Biomechanical Analysis Using the Finite Element Modeling Based on CT Images

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1. Introduction

Lumbar disc herniation is one of the most common causes of low back pain. Relevant research indicates that it is generally induced by the degenerated deformation of a disc due to too much labor or spine abnormality. The number and variety of clinical interventions for lumbar disc herniation continue to grow. The lumbar intervertebral disc is a viscoelastic tissue located in the middle of the two intervertebral bodies, which is the largest cartilaginous structure in human body that contributes to flexibility and load support in the spine. They can transfer labor loads, balance body, stabilize spine and absorb vibration[1]. All the functions depend on the intact disc. In pathological cases such as disc herniation caused by excessive load on the spine. The spine will undergo a series of changes in its biomechanical properties.

The intervertebral disc consists of endplate, annulus fiber, and nucleus pulposus. The penetration between endplate and the cancellous bone will change feebly if the disc bears an abnormal force and thus can affect the nutrient supply of annulus fiber, which results in the degeneration of the disc. Annulus fiber with poor nutrient supply is likely to rupture if there exists a stress concentration on this region. The nucleus pulposus will dissociate from the disc along the ruptured region. At last, the disc extrudes out[2,3].

In order to biomechanically evaluate the treatment options for spine related diseases and gain a better understanding of spinal biomechanical behaviour, there are three types of methodologies available. *In vitro* biomechanical experimental measurements on spinal components or segments tested in the laboratory have been used extensively to compare different treatment scenarios, but there are inherent limitations in using the specimens[2,4,5]. The limitations include the availability, the range of samples and the representativity of specimens. *In vivo* measurements can provide true subject-specific information on the spine in its physiological state but are confined by the requirement for limited invasiveness. More recently, *in silico* testing method has become more prevalent in the biomechanics assessment of the spine. In particular, there has been a rapid rise in the use of finite element analysis for this purpose over the last decade. The finite element analysis is a standard engineering technique generally used in the design of airplanes, machines and bridges. Using a special software, it allows the modeling of complex structures by splitting the structure into numerous simple finite elements, each of which is easy to characterize and

model mathematically. It is a numerical method for solving problems of engineering and mathematical physics[6]. Finite element modelling and calculating can be applied to show detailed biomechanical characteristics of lumbar and provide intrinsic parameters (stress, strain, strain energy, etc). Finite element models of the spine have been developed to evaluate surgical interventions, and also to investigate the risk of fracture in the vertebrae and the progression of degeneration in the intervertebral discs.

Our study is to analyze the biomechanical characteristics of lumbar in the compression using the finite element method based on medical image. We are sure that better understanding on the biomechanical characteristics of surgical procedures will ultimately get to better diagnosis and treatment on intervertebral disk herniation.

2. Methodology

In this chapter, a CT image based computational method has been presented which is applied to perform biomechanical analysis for lumbar vertebrae *in vivo*. The method involves establishing a subject-specific geometric model and a finite element model for vertebra. The geometric model derived from CT images represents three-dimensional anatomy characteristics of the vertebra. the finite element model based on geometric model provides a useful tool to perform biomechanical calculation for the vertebra. These numerical models are created from CT image data, in a process, which draws from an array of segmentation and mesh generation tool to define the geometry model. The geometry model is then augmented into finite element model with material properties, boundary conditions and interactions between multi-element models. We can get the stain and stress distribution of the whole structure by studying the relation between the displacement of particle and force for every element[6,7]. It is a feasible method for resolving biomechanical characteristic problem of complex structure.

2.1 Geometric model

The subject-specific geometric model is established by segmenting vertebrae in CT slice images and performing three-dimensional reconstruction slice-by-slice using extracted vertebrae morphology information.

A reliable segmentation of vertebrae is essential for subject-specific geometric model. Although bony structures show high contrast in CT images, a precise segmentation of vertebrae still remains challenging. Many factors could influence the accuracy and robustness of the vertebrae segmentation, e.g., complex anatomical structure, degenerative deformations, unclear object boundaries and similar structures in close vicinity, and these factors also could complicate the application of fully automated segmentation. Several approaches to precise segmentation of lumbar vertebrae in CT images were proposed so far. Admittedly, exact vertebrae segmentation has not been addressed. It has to be noted that these methods were developed for specific vertebrae regions. Several methods have been presented aiming at automated segmentation and labelling of the vertebrae, combining respective characteristics of MRI and CT image. In MRI images, bony tissue structures show as dark regions with sparse details and low contrast whereas the intervertebral disks often appear brighter. In comparison with MRI, CT offers higher contrast for bony tissue structures while it is somewhat difficult to distinguish the intervertebral disks from surrounding soft tissue in CT images.

We developed a method to establish three-dimensional subject-specific geometric model of vertebrae based on the CT and MRI data-based anatomical structure of spine by using reconstruction software VTK. A GE Lightspeed 16 CT scanner (General Electric, Fairfield, CT) was used to scan the human subject. A young man with no history of present and past disc disease was selected as normal subject. Initially L1-L2 motion segment data were taken in the axial direction, from which we could get 30 contiguous slices images from the CT scans. The CT slice images had a slice thickness of 0.8mm, and each pixel size is 0.33×0.33mm. The 3D modeling procedure was in the VTK software. We obtained three groups of 3D data from CT scans, the disc, L1, and L2 vertebral bones. The original 2D image slices from CT scans has 512×512 resolution. Considering the cost of the calculation time, in order to get the balance between the accuracy and time cost, we reconstructed the 3D image by using the 2D subsampled image slices which had 64×64 resolution substituted of the original 2D image slices with a 512×512 resolution. The subsample ratio is 8:1. Then, we have these slice images segmented. In order to precisely extract the morphology of vertebrae in CT images, we apply both automated and manual segmentation methods for slice images. First we adopted gray threshold segmentation algorithms for 2D image slices. As we know, the gray value of bone is greater than those of other tissues. We can separate the bone tissue from other tissue by setting a proper gray value. However, the automated gray threshold segmentation may result in artificial isolated section or discontinuous boundary as the gray value of cancellous bone is lower than that of cortical bone and similar to those of other tissues such as soft tissue. That's why we employ the manual segmentation to modify the contour and boundary of the 2D slice images of the vertebrae. The manual segmentation is a live-wire-based semi-automatic segmentation method. MRI image is used as a reference priori knowledge for radiologist to identify the corresponding anatomical structure in CT image. The radiologist could interactively modify the segmented result using the manual method after the automated gray threshold segmentation. The 3D L1-L2 segmentation modeling was fundamentally performed based on a set of segmented image data of axial slices in VTK software using Maching cubes algorithm, a surface reconstruction method. The segmentation result shows in Fig. 1, and the reconstructed 3D model of the segment shows in Fig. 2. All reconstruction models were saved in a VTK file format. Its content included space coordinates of keypoints as well as topologic structure on the surface of the model. We



Fig. 1. The segmented image

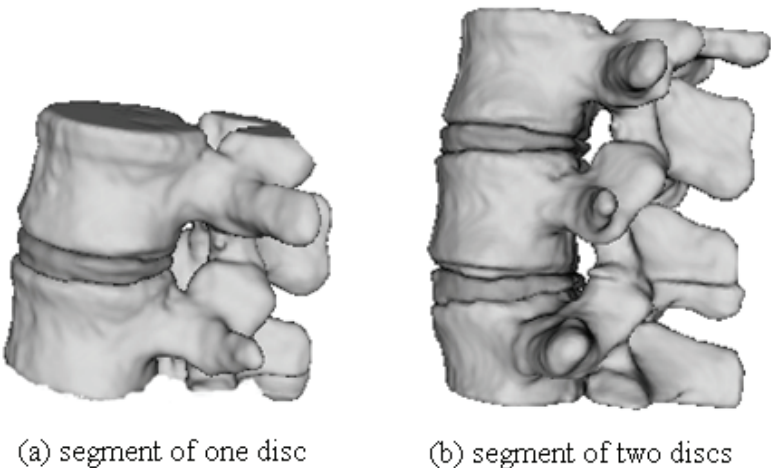


Fig. 2. The model of reconstruction in VTK

translate the data from the VTK file format to the macro file format in order to import the data to the finite element method software Ansys6.0. We finally create the geometric model of the L1-L2 segment in Ansys6.0. The space coordinates of the keypoints in the VTK software correspond to the keypoints in Ansys6.0. It is convenient to transfer data between the geometric model in VTK and the finite element model in Ansys6.0. The geometric model which was imported into Ansys6.0 is an entity model. We divided it into a grid of elements by applying the finite element meshing on it to form the finite element model.

2.2 The finite element model

The generation of a subject-specific finite element model consists of the determination of the geometrical parameters, the creation of a finite element mesh and the definition of the material properties, and all the procedures should be associated to the given subject. The common method for creating a subject-specific mesh of the vertebrae is the direct conversion from image voxel to hexahedral mesh elements. In this method, the segmented region of the image is extracted and each image voxel is directly converted to a brick element in the mesh. Materials properties are defined on an element-by-element basis using densities derived from the image data. The density value is generally derived for each element based on the image intensity in that region, then the density is applied to calculate the Young's modulus value for that element. Several formulae used for this calculation conversion have been published, as yet there is no consensus as to which of these formulae is most appropriate. How to get more accurate material properties of vertebrae tissue is still challenging for most image-based models.

We applied the Ansys to create an isotropic, 3D, nonlinear finite element model of an intact human L1-L2 motion segment. Details of the model developing have been given and are briefly summarized here: the shape of the lumbar segment is reconstructed from data obtained from CT scans of a human L1-L2 segment. The finite model created is shown in Fig. 3.

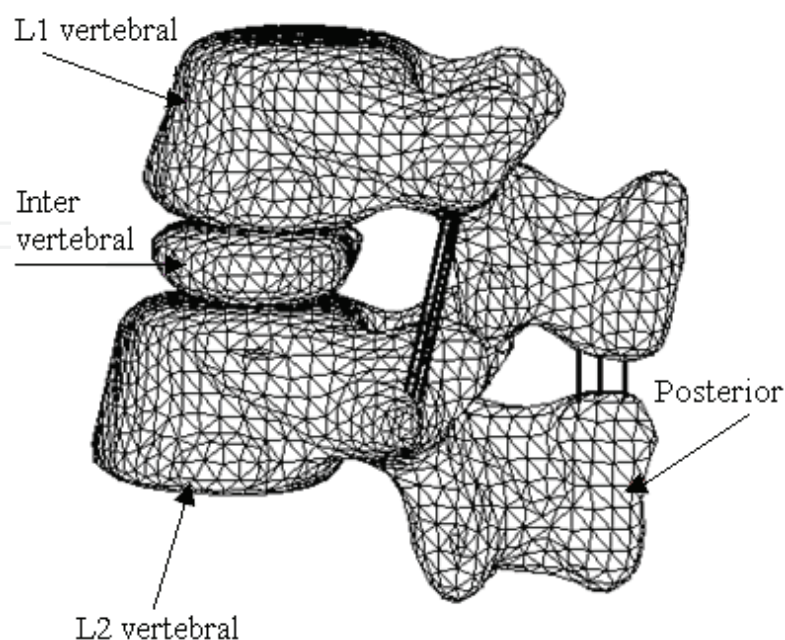


Fig. 3. The finite element model of L1-L2 segment

Each vertebral bone was modeled as 20-node isoparametric material elements using homogeneous and isotropic material properties[8]. The intervertebral disc was modeled using solid elements to simulate an incompressible behavior with a low Young modulus and a Poisson ratio close to 0.4999[8]. In order to appropriately model the changes of contacting areas of facet articulating surface which were applied upon with the load, facet articulations were modeled using contact elements. The ligaments were modeled as two-node axial cable elements that sustained tension only, oriented along the respective ligament fiber alignment. Their attachment points to the bony prominence were determined by referring to anatomy books in order to mimic anatomic observations as close as possible. The cross-sectional areas used were averages of the values reported previously. The material properties used in the study were derived from the literature[8-10]. The behavior of material properties in the model response better reflected those of published experimental lumbar response. Here, we hypothesize that the strain of spine is small. Table 1 lists the type, number and material properties of element used to model the various components of the L1-L2 motion segment and the complete model consists of 37449 elements.

Material	Element type	Element number	Young's modulus(MPa) /Possion rate
Cortical bone of L1	20-node brick (solid 95)	16596	200/0.3
Cortical bone of L2	20-node brick (solid 95)	18663	200/0.3
Disc bone	20-node brick (solid 95)	2082	4/0.4999
Facet joint Ligaments	Contact(CONTA 174) 3D-cable(tension only)	42	

Table 1. Element type and material properties

2.3 Boundary conditions

With regard to the validation and accurateness of model analysis, we applied the boundary conditions on the finite element model. The boundary conditions on the model use pressure and restraints assigned to surface areas of the model. The inferior surface of L2 vertebral body and its spinous process were fixed in all directions. The restraints were used to limit the models movement with six possible values at the node on the surface, three translations and three rotations. The value of freedom was zero. Then couple the inferior surface of vertebral L1 body and the upper surface of the intervertebral disc body, as well as the bottom intervertebral disc body and the upper surface of the L2 vertebral body in all directions of translation. The facet articulation was modeled as a 3D contact unit using interface elements [8].

2.4 Load cases

In this study we will analyze the stress and strain distribution of the spine. The evaluation was performed by following methods: (1) load-displacement behavior. We can observe the

displacement change of the vertebral and the strain distribution of L1-L2 segment under different loads. (2) A load of 1600N axial compression was applied to the superior surface of the model in the form of a uniformly concentrated load over all L1 superior surface nodes. We can observe the stress distribution of L1-L2 segment by applying the load and clue on the high stress concentration region as the most likely fracture areas. (3) Disc bulge: disc herniation is an important part in our study for the L1-L2 segment. It is of clinical significance to analyze the disc bulge degree under different directions at 400N axial compression, and such analysis can instruct the surgical treatment for disc herniation. (4) Nucleus populos resection: applying 500N loads on the L1-L2 segment with denucleated disc to observe stress and strain distribution on lumbar and the difference between normal disc and denucleated disc. From the load cases, we know that the finite element model can be used to predict the change of biomechanical behavior of the human lumbar spine under pressure.

3. Results

The stress and strain distributions of the vertebral column were obtained from biomechanical analysis by applying different axial compression loads using our proposed CT image based numerical modelling method. The results are presented in the following sections.

3.1 Load displacement

The loads we applied on the L1 superior surface were: 500, 1000, 1500, 2000, and 2500 N. The L1 superior surface bears a 500N load when a healthy person weighted 70 kg stands straight in a relaxation state. However, the number goes up to 2000N when he lifted 100N with two arms stretching straight. The heavier he lifts the more L1 superior surface bears. The results of load-displacement behavior in axial compression are shown in Table 2. From the table we can see an increase tendency of displacement of L1 superior surface with the increase of load. The tendency is approximately linear which also illustrates that the vertebral bone has flexible biomechanical characteristics.

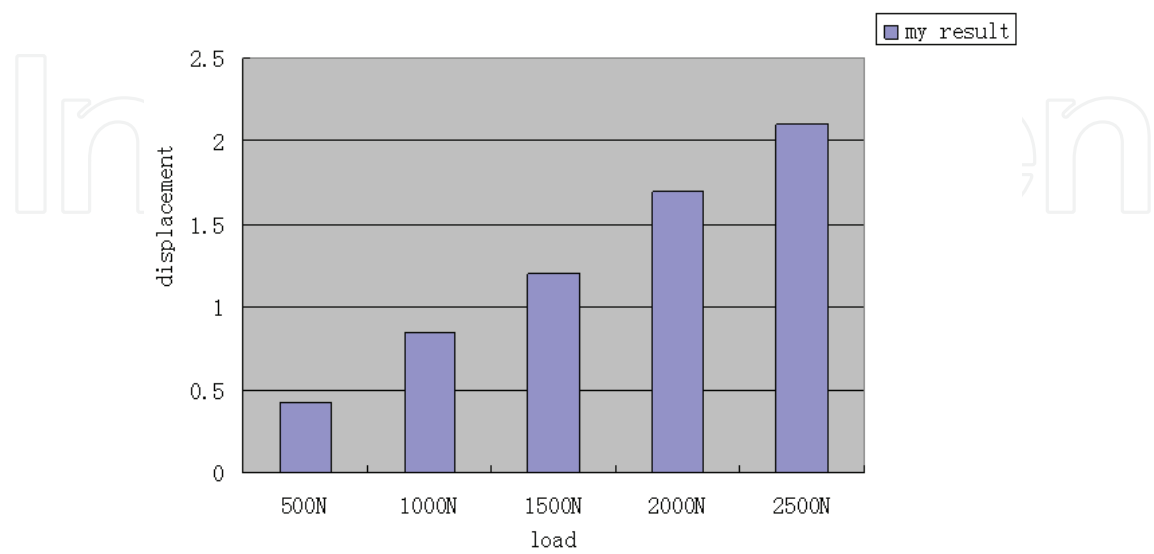


Table 2. The maximum displacement of the FE model in different loading

3.2 Stress distribution of model

Fig. 4 shows the stress distribution of the spine when an 1500N load is applied. It shows that the high stress concentrations are around the vertebral body and pedicle region due to the way the load applied. That is, they are mainly focusing on the upper body of vertebral. These areas show Von Mises stress that ranges gradually from blue, $1.7E-01\text{mN/mm}^2$ to the Maximum Von Mises stress of $1.57E-02\text{mN/mm}^2$ indicated in red. The stress on the vertebral body (the value ranges from 35 to 87) is higher than that on the pedicle (about 17), which makes it a common place for injuries due to loading. The stress concentration may be higher in pedicle region if the pedicle area is loaded with a greater proportion.

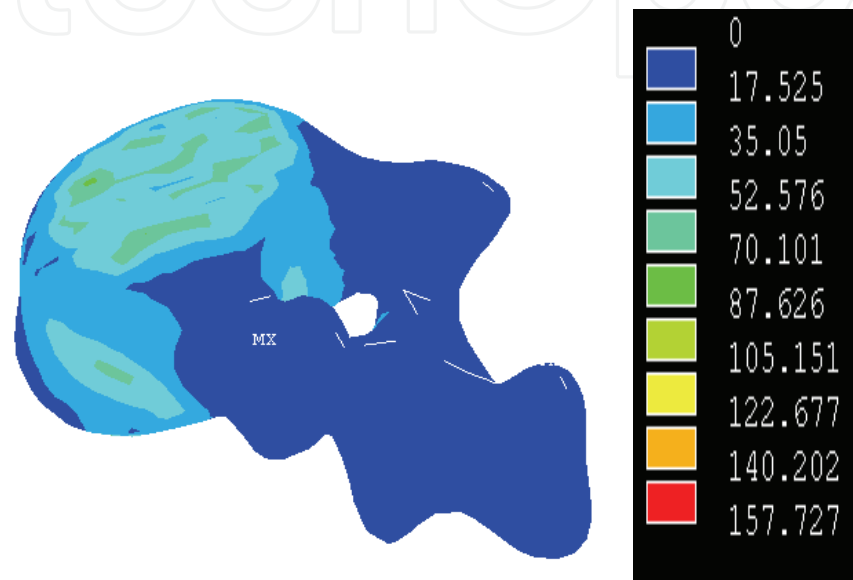


Fig. 4. The Vos Mises distribution of L1-L2 segment

As to the L1 vertebral displacement, the vertebral body and the superior articular processes are compressed downward. Consequently, the movement of vertebral due to the applied load and restraints placed on the model induces the high areas of stress in the pedicle.

3.3 Disc bulge

3.3.1 Disc stress distribution

The disc consists of annulus fiber and nucleus pulposus. In a normal healthy disc, the hydrated nucleus pulposus exerts a hydrostatic pressure (intradiscal pressure, IDP) on the annulus fibrosus (annulus) fibers [11]. The fiber bears a higher IDP than the nucleus pulposus [12,13]. Applying 1000N axial compressive loads on the normal disc, we can see that the value of stress on fiber shows in green, and the range of the stress is from $1.77E-01$ to $2.67E-01\text{mN/mm}^2$. From the annulus fiber to the nucleus pulposus, the intensity of stress grows weaker and weaker (Fig. 5). It indicates that the nucleus pulposus can absorb some compressive loads. That's why the fiber shows higher stress than the nucleus pulposus. From the figure, we can also see that the stress and strain distribution of lumbar is mainly concentrated on the posterior and bilateral posterior of annulus fibers in the normal disc. We can also observe that the stress becomes higher correspondingly as the loads applied to the disc increases. The relationship between IDP and the external load is approximately linear as is shown in Table 3, indicating that intervertebral disc has flexible properties in a certain extent.

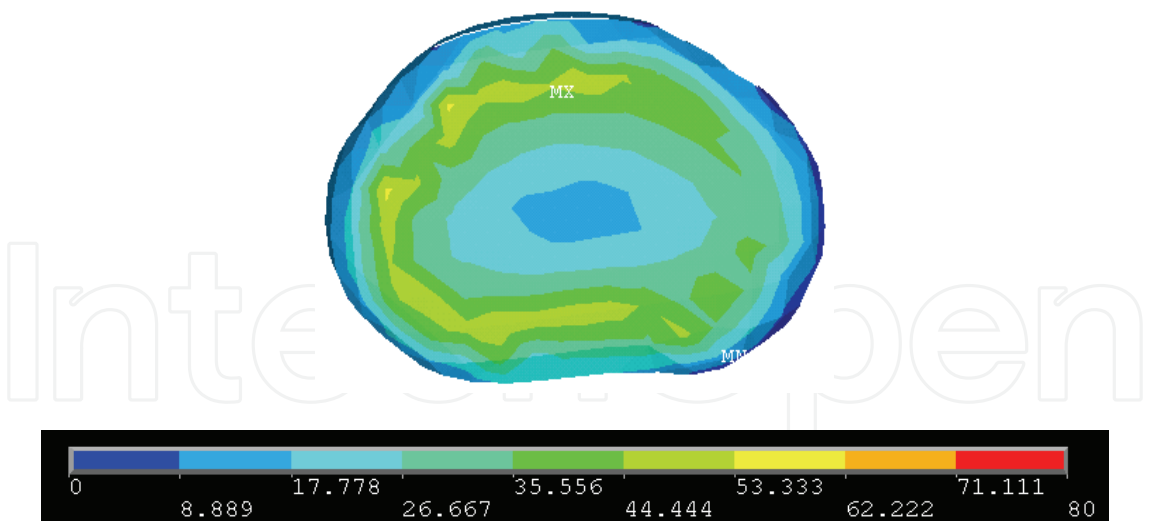


Fig. 5. Vos Mises distribution of disc at 1000N axial compressed load

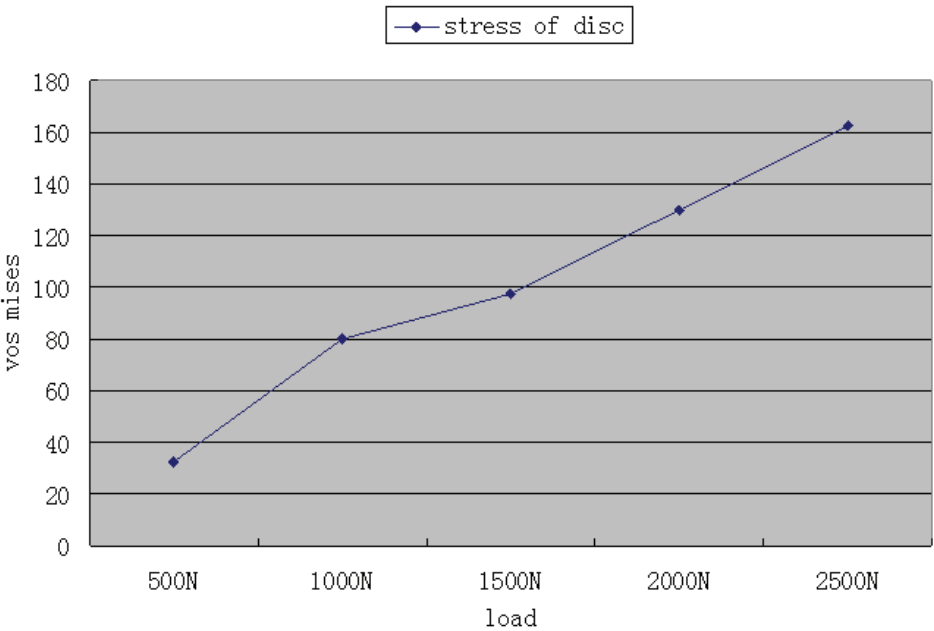


Table 3. The maximum stress of disc in different loading

3.3.2 Bulge direction and magnitude

Four nodal points on the intervertebral disc were taken to represent the directions of bulge, left lateral, and right lateral, left posterior and right posterior. They were marked to determine the bulge displacement of the disc at the 500N axial compression (Fig. 6) [9]. We observed the degree of the bulge at four directions. The result indicates that the displacement of posterior disc extrusion is more obvious than that of bilateral.

Table 4 displays the disc bulge magnitude at different directions on the 500N load. Table 5 lists the disc displacement magnitude at different axial compressive loads from 500 to 2500 N. The values of displacement at all directions increase with the increase of loads. The posterior direction has a more obvious tendency to extrude than others; at the same time, the bilateral direction has a greater tendency to extrude out than the anterior. This agrees

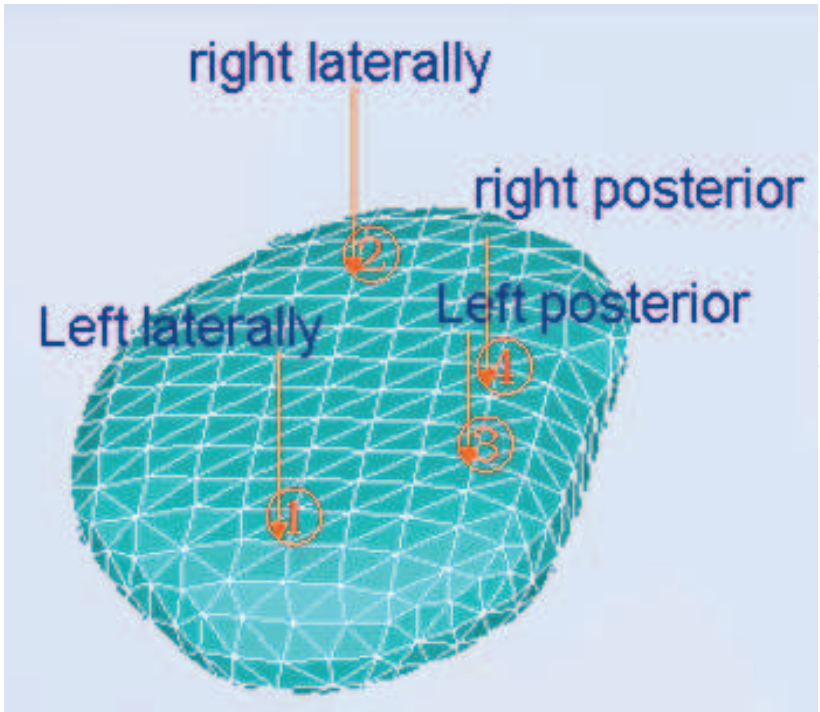


Fig. 6. The selection of four directional node

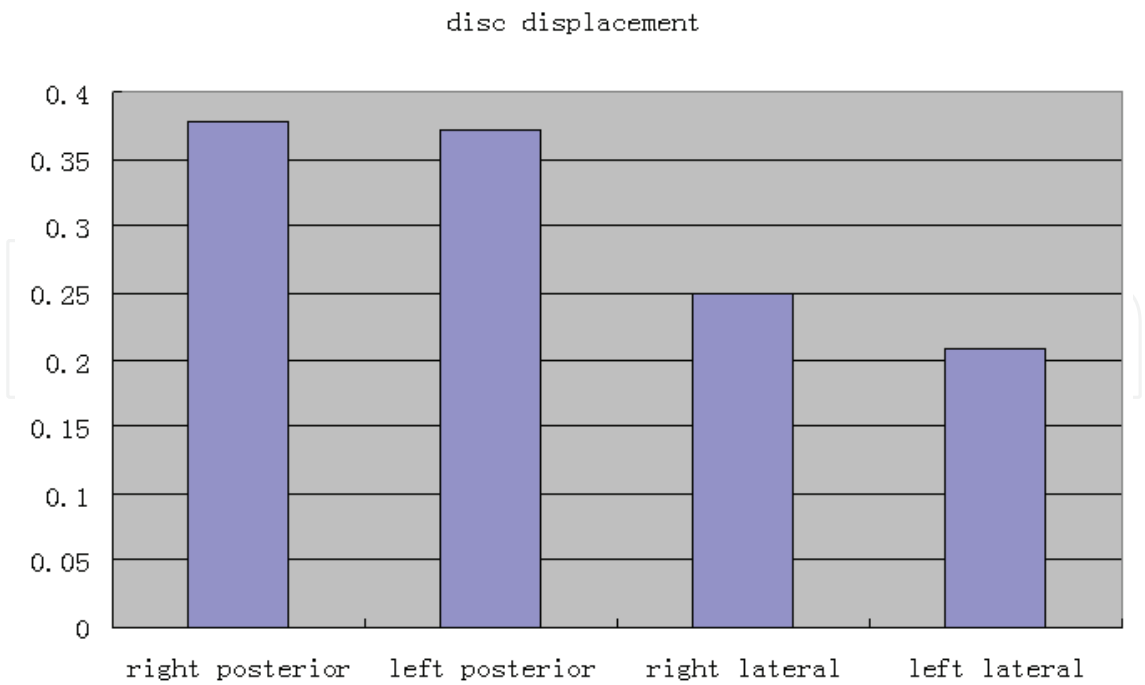


Table 4. Disc bulge displacement at axial load of 500N

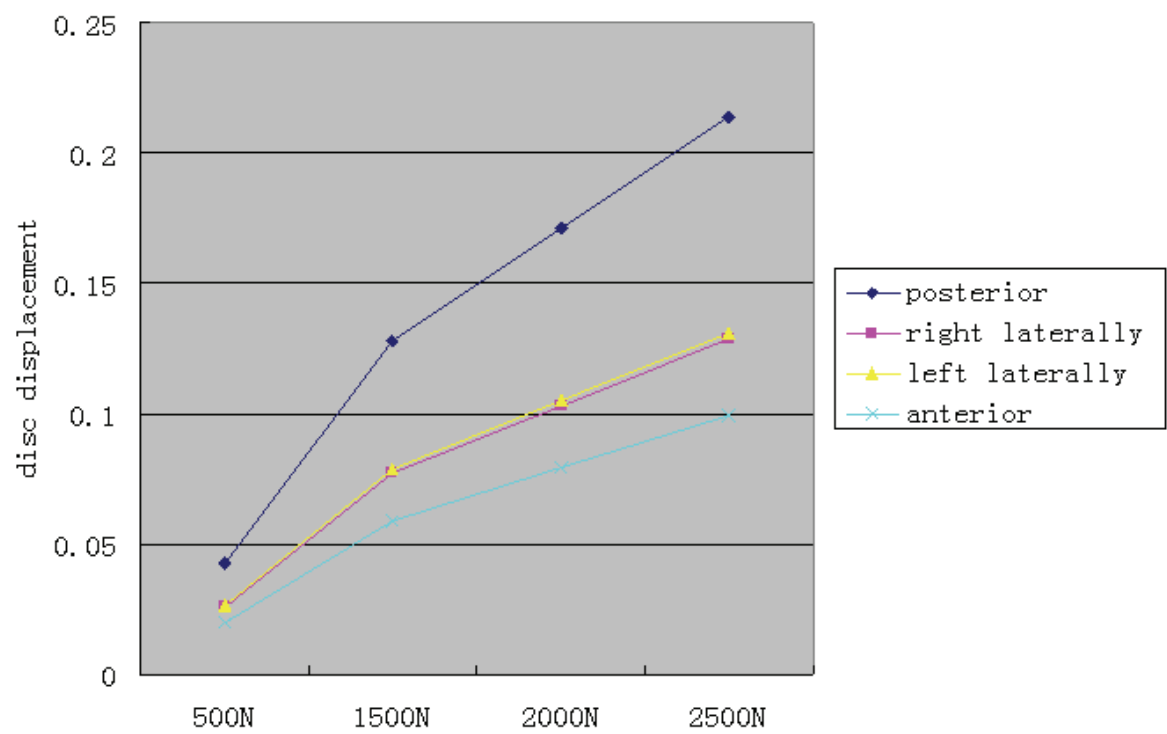


Table 5. Disc displacement at different axial loads

with what we observed on the patients. Our clinical experience tells us that posterior-lateral of the disc is the region where annulus fiber ruptures and nucleus popoulos extrusion occurs easily. As we reviewed above, this is also the region with higher stress. It indicates stress focusing on the region defined has certain relation with disc herniation.

3.4 Removal of the nucleus

The normal disc consists of the annulus fiber and the nucleus popoulos. The nucleus popoulos was considered incompressible and was modeled as 3D tetrahedral solid elements [11,14]. The denucleated disc only includes annulus fiber tissue (Fig. 7).

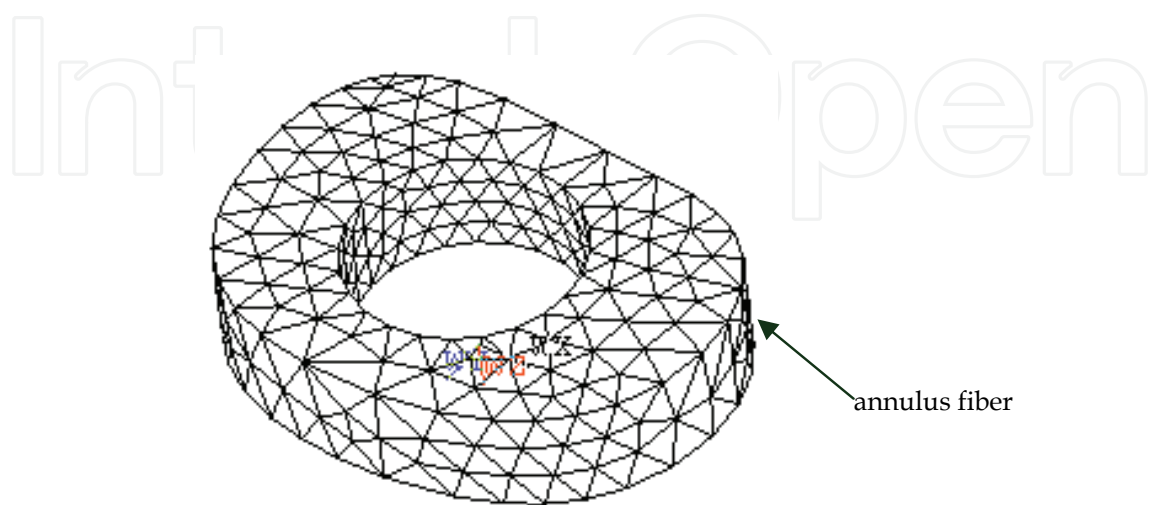


Fig. 7. The finite element model of an intact denucleated disc

Consequently, the process of modeling is similar to that of a normal disc except setting IDP of nucleus to zero. The whole model of the L1-L2 segment with a denucleated disc was developed as shown in Fig. 8.

The disc plays an important role in the stabilization of lumbar. The nucleus populous is the fulcrum during lumbar activity [15,16]. It maintains the height of the lumbar, and distributes the stress evenly around the annulus fiber, as shown in Fig. 9.

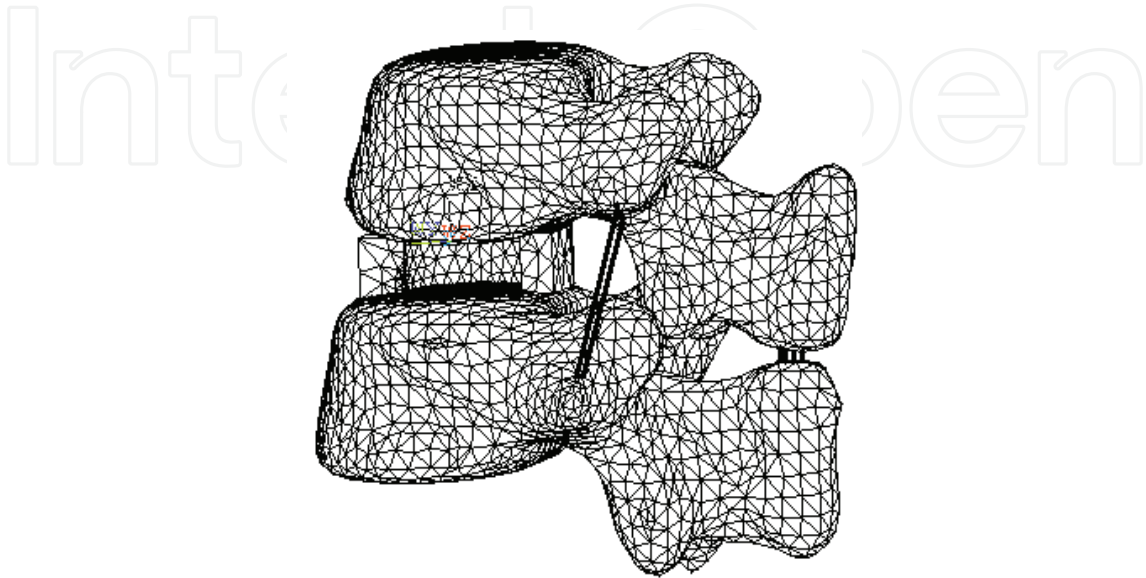


Fig. 8. The L1-L2 segment model with denucleated disc

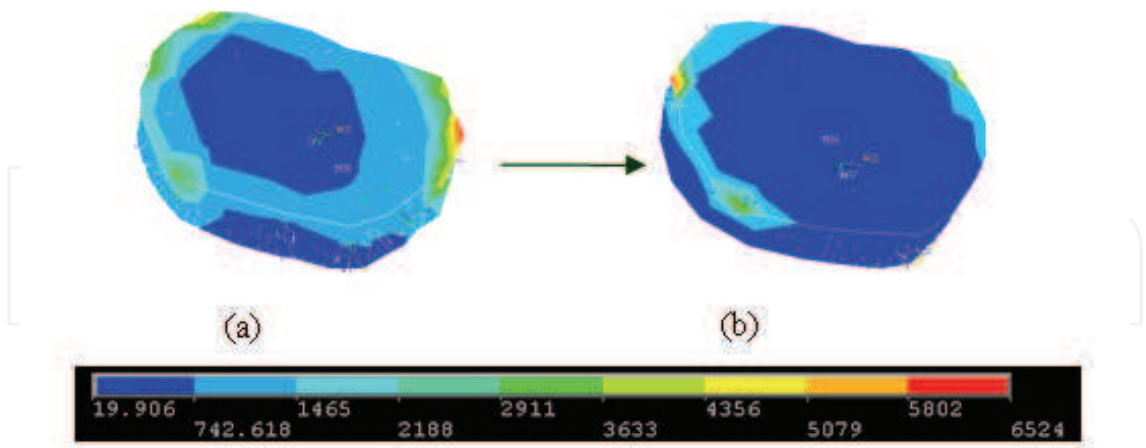


Fig. 9. Von Mises of the intervertebral disc: (a) Intact Disc,(b) Intact denucleated Disc.

We applied 500N load on the L1 upper surface body of the normal disc and observed changes in the stress. As is shown in Fig.9a, the central area of the disc shows low stress indicated in blue. The area outside the disc bears higher stress indicated in green. Fig. 9b shows the results of the Von Mises stress on the denucleated disc, we can see almost the

whole disc bears low stress indicated in blue, illustrating that the IDP on the denucleated disc falls without the nucleus populous' function of sustaining surface tension of annulus fibers near the nucleus. The results from the analysis show that resection operations can relax the IDP of the disc, relieve the pain in the patients, and attain the goal to cure disc herniation through operation.

Table 6 lists the axial displacement of normal disc and denucleated disc. Three nodes in different locations representing three directions (anterior, posterior, lateral) are selected to determine the displacement of disc. The histogram in blue in the table represents the axial displacement of normal disc, and the red represents that of denucleated disc.

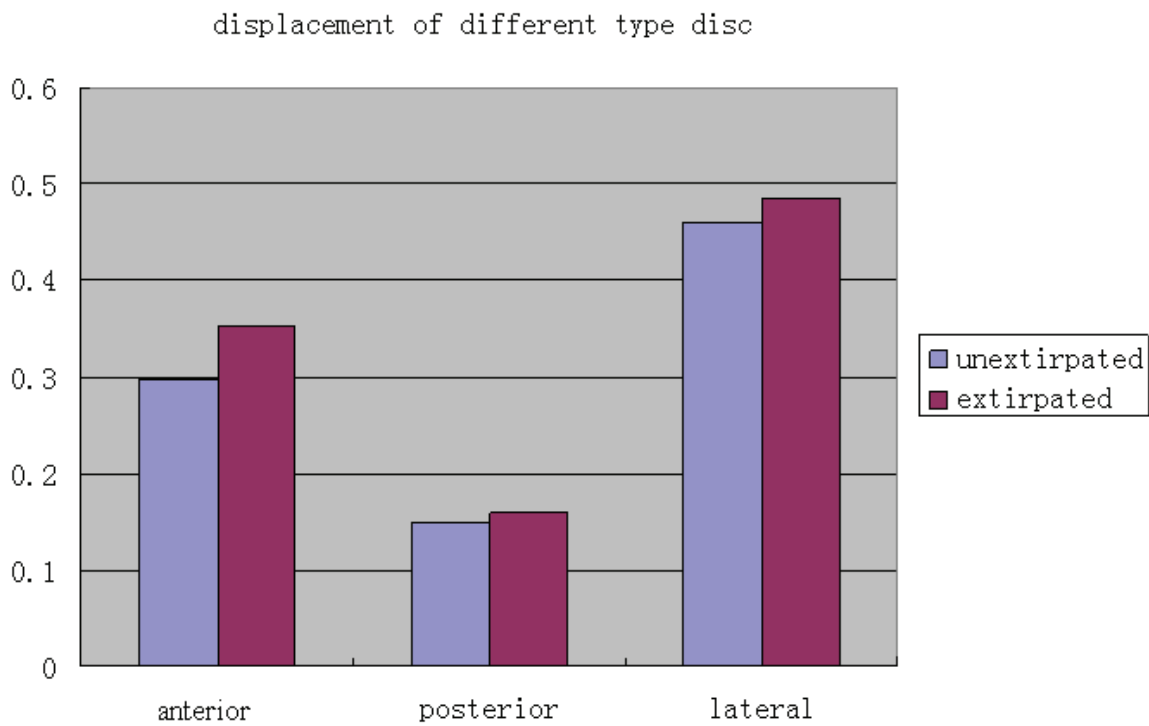


Table 6. Axial displacement of two types of discs

From the table, we can see that the value of axial displacement of of the denucleated disc is larger than that of the normal, illustrating that the height of intervertebral disc will become smaller when the nucleus pulposus is resected. The disease or the operation on the disc will make the cubage of the disc and the height it maintains disappear. They also destroy the compact structure between upper and lower intervertebral bodies. Then, the stress on small joints on both sides may change. Finally, the cartilage on it degenerates, leading to deformation of the disc.

As we know, the nucleus populous serves as the fulcrum of the spine and maintains the stability of the lumbar [17]. Once resected, the function of fulcrum disappears, and the disc will tend to extrude frontal, resulting in instability of the spine.

4. Discussion

The CT image based finite element modelling can provide a thorough understanding of the biomechanical influence on the lumbar and may offer reasonable explanations to

biomechanically linked pathologies. It allows us to implement simulation calculation of biomechanical parameters of lumbar tissues, thus to analyze the influence of a single component within the construct investigated. It is useful in analyzing stress and strain distribution patterns of lumbar, leading to an optimal design of the surgeon[3]. It does, however, not mean that biomechanical *in vitro* approaches should be replaced by such a model. The current finite element model also has limitations, even if its modeling is based on the characteristics of physiological material and the geometric shape of lumbar. The anatomic structure of spine is complicated, and such properties of the small articulation as friction coefficient are not very clear. So all the material parameters adopted for the model have been simplified or based on hypothesis on some degree. It is a simplified model. In the future study, more nonlinear materials will be modeled. Any finite element model does only represent a mathematical model and thus is only an approximation to the specimen and even further from real life conditions[18-20]. It cannot reflect the variability of shape and material properties of the bone within an individual or among individuals. The interface between two bones only simulates appropriately the condition *in vitro* or *in vivo*. The model is just one segment of the whole spine. The results may vary if done with the whole spine. There are lots of differences and uncertain factors induced by the individual diversity during modeling. Based on the above reason, even though a finite element model has some limitations, it simulates the biomechanical characteristics of the lumbar preferably[19-21].

5. Conclusions

The lumbar is an important organ for bearing the weight. Wherein, the intervertebral disc can keep the height of the spine, linking the adjoining vertebral body. Experiment tests show that the nucleus pulposus bears a load of 45–60 kg when one lies down with muscles relaxed. A normal disc can bear a load of 30 kg while it goes up to several times when exercising violently or lifting heavy things. The nucleus pulposus is more likely to be injured and extrude out while the loading threshold is exceeded. We apply the load from 500 to 2500 N, which are in the range that the human being can bear. The result accords with the biomechanical characteristics in the normal disc[22–24]. In our research, we applied small distortion parameters to simulate the stress and strain distribution of the lumbar. The height of L1 is about 23 mm. When a load is applied on the L1, small distortion appears, reflecting the flexion properties, which indicates that the cancellous bone and cortical bone bear the force together, and the high stress is concentrated on the pedicle. The high stress is on the lateral and posterior region of the normal disc. From the tables, we learn that the extrusive magnitude for the posterior and the lateral is larger than in other directions. The tendency is more distinct with the increase of the load. A 3D nonlinear finite element model of lumbar motion segment was established to simulate the loading state of the spine. The study indicates the biomechanical characteristics as follows:

1. The strain of L1-L2 segment under axial compressive load increases with the performed load.
2. Large stress concentrations are found in the pedicle region, a common place for injuries. Under the axial compressive load, the body of vertebrae and articular process was compressed downward. The magnitude of stress in the pedicle region depends on the proportion of load applied on the superior articular processes. The stress on the pedicle region is higher with greater proportion.

3. Excessive loads could cause the disc bulge. The bulge extent depends on the magnitude of the applied load. Under the same load, the disc tends to bulge laterally than posteriorly.
4. The denucleated disc shows a lower IDP than the normal disc. The resection surgery relieves the stress of the disc and alleviates the low back pain in the patients. However, after the nucleus pulposus is resected, the denucleated disc is liable to extrude outward frontally, and the height of the disc is lower than that of the normal. All of these will result in changes of biomechanics in the spine, such as lumbar instability and the inhomogeneous stress distribution on the annulus fiber, as well as the loss of the function of nucleus on the disc. This study has enriched some understandings of the biomechanical characteristics under loadings and can help surgeons make better decisions for the treatment of low back pain. In the study, an initial model of vertebral is developed, which includes solid cortical bone, disc, facet joints and several ligaments. In the further study, the models will include cortical bone, cancellous bone, and ligaments, which can also improve the accuracy of results and evaluation validity. Our next step is to study more on stress and strain distribution under torsion and shear conditions and to simulate the biomechanical characteristics of lumbar during an operation. We aim at the operation simulation and surgery navigation by developing and analyzing the finite element model. The finite element model of L1-L2 segments based on medical images can analyze biomechanical characteristics of lumbar effectively and facilitate the optimization of individualized therapy in the future.

6. Acknowledgement

This study was supported by Scientific Research Key Program of Beijing Municipal Commission of Education(Grant No. kz200810025011) and National Natural Science Foundation of China(Grant No.30670576).

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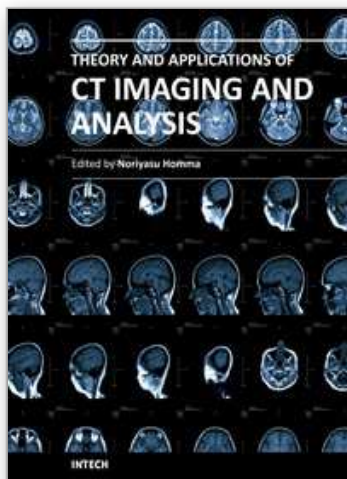
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Theory and Applications of CT Imaging and Analysis

Edited by Prof. Noriyasu Homma

ISBN 978-953-307-234-0

Hard cover, 290 pages

Publisher InTech

Published online 04, April, 2011

Published in print edition April, 2011

The x-ray computed tomography (CT) is well known as a useful imaging method and thus CT images have continually been used for many applications, especially in medical fields. This book discloses recent advances and new ideas in theories and applications for CT imaging and its analysis. The 16 chapters selected in this book cover not only the major topics of CT imaging and analysis in medical fields, but also some advanced applications for forensic and industrial purposes. These chapters propose state-of-the-art approaches and cutting-edge research results.

How to reference

In order to correctly reference this scholarly work, feel free to copy and paste the following:

Haiyun Li (2011). An Approach to Lumbar Vertebra Biomechanical Analysis Using the Finite Element Modeling Based on CT Images, Theory and Applications of CT Imaging and Analysis, Prof. Noriyasu Homma (Ed.), ISBN: 978-953-307-234-0, InTech, Available from: <http://www.intechopen.com/books/theory-and-applications-of-ct-imaging-and-analysis/an-approach-to-lumbar-vertebra-biomechanical-analysis-using-the-finite-element-modeling-based-on-ct->

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