

Three-dimensional nonlinear finite element model of the human lumbar spine segment

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The objective of this paper is a three-dimensional modelling of vertebral segment (L4–L5), which can be used for numerical simulation of surgery, analysis of spinal equilibrium and stability. Because of the extreme complexity of finite element modelling we propose to carry out analyses on the simplified model. The geometry of vertebrae is known due to the computer tomography (CT) or nuclear magnetic resonance (NMR). CAD model is built and then imported into FEA program. The model under consideration, i.e. the model of spinal segment, consists of two bones and an intervertebral disc. The mechanical properties of tissues, boundary/interaction conditions and loadings accepted for computations are based on the literature and our own studies. The simplified model was proposed, developed and validated for several loading schemes, including axial compression, bending and torsion.

Key words: biomechanics, nonlinear FEA, lumbar spine

1. Introduction

The finite element method (FEM) is successfully applied to simulations of biomechanical systems [2], [3], [14], [23]. As is well known, FEM belongs to a group of few methods which allow advanced computations taking into account such essential features as material inhomogeneity and anisotropic mechanical properties of the bone tissue as well as extreme complexity of bone geometry. It is proved [4], [7] that failure or prediction of the changes in biomechanical parameters and remodelling of the motion of spinal segment are related to the stress and strain fields in the tissue and

may be computed using FEM. To estimate the stress and strain fields the computational models have to incorporate material properties determined experimentally, realistic geometry and appropriate boundary conditions. FEM is an efficient tool for testing biomechanical sets in terms of their strength, but its proper application and success of the following estimation depend greatly on a deep understanding of physical properties of the element used and its proper geometry. The scientific literature on the assessment of the bone physical parameters appears to be extensive [11]–[13], [16], [23], however, it is difficult to use basic information on physical parameters, because the mechanical data referring to the very tissue are sometimes grossly inconsistent [11], [16], [23]. The methods of defining mechanical parameters of biomechanical sets used up to date are inaccurate and cannot be used for research on alive people. The osseous tissues of the spine have very heterogeneous structure and their changes depend on their position in the vertebra. The heterogeneity of the tissue structure becomes still more conspicuous while testing the material taken from various donors, because the same element of various people has different size. Practical application of FEM to individual biomechanical sets will be possible when the actual basic data describing an individual's traits are provided. We developed the method that is based on the relation between the mechanical properties of osseous tissue and its radiological density, which allowed us to make a computer tomography in order to estimate the tissue rigidity [12], [13].

2. The aim of the study

The aim of the study was to build and validate fully 3-D and simplified models of the lumbar spinal motion segment. This simplification is based on the replacement of the lumbar spinal motion segment by a connector-type elements, whose elasticity behaviour is spring-like in available components of relative motion. The required behaviour of spinal segment is represented by the plots of a force or moment versus relative displacements. Alternatively 6×6 elasticity matrix (as a function of field variables) can be constructed and optionally it can be used in more complex analyses of the whole spine studies. To follow this approach we examine the lumbar spinal motion segment (figure 1) which is subjected to six loading schemas: compression, shearing in two directions, bending in two plane and torsion. By means of applying the unit loads (axial, shear and moments) and recording displacements at the appropriate points, where relative motion can be recorded, we build a compliance matrix and then the stiffness matrix for the model. If the behaviour of the segment is in the range of interest, i.e. small rotations and small displacements, we can use a tangent stiffness matrix because of the linearity of load–displacement relations.

3. Anatomy of a L4–L5 motion segment

The motion segment L4–L5 under consideration consists of two vertebral bodies and the intervening facet joints, intervertebral disc, posterior elements and spinal ligaments [1]. The intervertebral disc and ligaments play the essential role in the spine kinematics and contribute to general flexibility of spinal segment. The vertebral body is composed of two cancellous bones and cortical one.

The intervertebral disc is essentially composed of three different parts. Its main component is annulus fibrosus which consists of 10–20 collagen sheets called lamellae. The second component of the intervertebral disc is nucleus pulposus, a hydrated gel at the centre of the disc. Both components maintain the stiffness of the disc in the conditions of compression loading and allow, to some degree, movement between vertebral bodies. The next components of the intervertebral disc are the superior and inferior vertebral endplates. These are the cartilaginous plates that cover the superior and inferior aspects of the discs and bind the disc to their respective vertebral bodies. Besides enabling bending movements between the vertebral bodies, the intervertebral disc allows twisting and small sliding movements. Their amplitudes depend on elasticity and tensile stiffness of the annulus. The posterior elements control the position of the vertebral bodies such as: a pair of stout pillars of bone called the pedicles with two transverse processes, two superior articular processes and two laminae ending with the spinous process. These several processes serve as an attachment for the muscles that control and stabilize the lumbar vertebral column. Many components of lumbar spinal segment are connected by ligaments. The most definitive ligament is the ligamentum flavum which consists of elastin that connects the lower end of the internal surface of one lamina with the upper end of the external surface of the lamina below and closes the gap between consecutive laminae. The transverse processes are connected by thin sheets of collagen fibers. The opposing edges of spinous processes are connected by collagen fibers referred to as the interspinous ligament and the supraspinous ligament. In addition to the ligaments of the posterior elements, the lumbar vertebral column is reinforced with ligaments that connect the vertebral bodies. These are: the posterior longitudinal ligament and the anterior longitudinal ligament. The vertebral bodies, the disc and the posterior elements create a complete structure of the lumbar vertebral column which maintains stability and control of movements. The consecutive lumbar vertebrae are fixed together with the zygapophysial joints which create a locking mechanism between them. They block axial rotation and forward sliding of the lumbar vertebrae. All the biological elements mentioned above are to some extent modelled in the segment.

4. Geometry acquisition

Creation of a computational model of the spinal motion segment requires precise geometrical data of a real object. Besides topology, some additional data such as volume

density, surface texture, etc, are also in focus of our interest. Different methods of acquisition of geometrical data can be used, namely: contact scanning and non-contact scanning. A detailed description of these methods is presented in [13], [17], [19]. Recall briefly some information that allows us to understand better the goal of the studies. The 3-D scanner transforms geometry of the real object into CAD system. After reconstruction the virtual model can be used for preparing the visualization of tissues (the models describe the shape and size of the individual organs). This can be used in medical diagnostics (virtual endoscope), pre-surgical planning (simulations of surgical interventions) and also in FEA modelling (e.g. to calculate durability of bones affected by osteoporosis). In medicine, the most popular techniques allowing us to recognize the 3-D geometry of a living body are based on computer tomography (CT) scanning and magnetic nuclear resonance (MNR). Final results of these techniques can be itemized as follows: surfaces grid, solid or volume FE grid. The methods mentioned above have also their limitations and the results obtained not always can be used in FEA models.

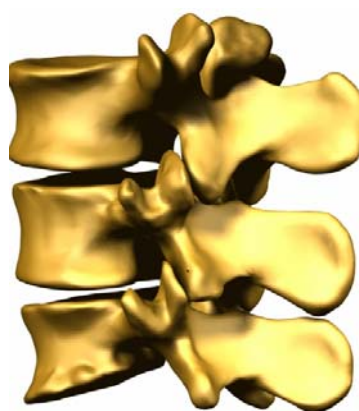


Fig. 1. 3-D geometrical model of lumbar vertebra bones (L3–L5)

Advanced CAD codes are able to transform surface grid into solid model. In practice, its use is limited only to the objects of relatively simple shapes. When the object geometry is such complex as in vertebrae, a lot of errors may occur during generation of surface grid and when forming a solid model and thereby a mesh of finite elements. Because of the complexity of the geometry and wrongly connected contours obtained, it is often impossible to automate the process of FE grid generation. Therefore, this solid model is mainly used for visualization purposes, but not in structural analysis. In this work, the CAD model is constructed using 3D-Doctor software from Able Software Corporation that converts the data from NMR or CT to 3-D geometrical model. The result is visible in figure 1.

Then the geometrical model is imported into FE preprocessor and, after some repairing steps, meshed. A special effort is required to succeed this step because a real geometry of the segment analyzed is very complex. Partitioning and smoothing

techniques are necessary to apply and to maintain at the same time the most important geometrical parameters. In fact, it is done in a few steps and the intermediate results are compared with some available data. To validate the geometry we compared the models obtained with the result of 3-D scanning of single vertebral segment. Then the mechanical properties of tissues as well as of the system of loading are recognized, and finally, computations are performed by FEM using the ABAQUS/Standard environment. This allows us to obtain the results concerning the fields of deformation, the fields of strains and stresses as well as the contact surfaces with their interactions.

5. FEA modelling

The first step of providing the simulation of the behaviour of biomedical system is a proper definition of the model geometry. The complexity of this step was described in some details above. For further information we refer to [17].

The second step is a definition of the material properties. The material complexities should be considered when choosing material constants for the finite element model. Thus, the development of a set of experimental methods that allow defining such parameters as longitudinal elasticity module becomes a matter of basic importance for any numerical application, especially for solving problems in biomechanics. There is an extensive literature on the assessment of bones' physical parameters [10], [11], [15], [20]. Unfortunately, the methods of defining biomechanical parameters are inaccurate and cannot be used in practice on alive people. The osseous tissue of the spine is very heterogeneous in its structure and its properties change depending on the place of sample in the vertebra. On the basis of the existing relationship between bone mechanical properties and radiological density, Young's modulus of vertebral cancellous bone FE model of motion segment can be estimated [12]. The other data was taken from the literature [9].

In the next step, an appropriate finite element is chosen and assigned to each part of the segment considered. In the context of the kinematics of spinal motion segment, intervertebral disc and ligaments play the most essential role. In numerical simulation of L4–L5 segment, our attention will be focused on these two parts.

The models of intervertebral discs were developed based on the achievements reported in the study [7]. The annulus fibrous of the intervertebral disc is a highly structured material made up of alternating tissue layers with collagen fibers oriented at $+30^\circ$ and -30° with respect the circumferential axis. The disc can be modelled as anisotropic material of hyperelastic properties [4] or by structural elements which induce this anisotropy. The ground matrix of the disc annulus is modelled using 3-D solid elements. The collagen fibers can be modelled by truss elements carrying only tensile stresses [15], [18] and/or by rebar-type elements embedded in 3-D solid elements. When using the first approach each truss element in each layer of the annulus fibrous has to be connected to

3-D solid element nodes in order to keep up the orientation mentioned above. Development of such a model is quite tedious and time-consuming work. Additionally, this approach is mesh-dependent because any remeshing requires re-defining of the data concerning truss elements. In this study, we have used the second method which has some advantages over the previous one. The rebar elements are uniaxial reinforcements in solid elements and can be defined as surface layers with uniformly spaced reinforcing bars (such layers are treated as a smeared layers of a constant thickness equal to the area of each reinforcing bar divided by the reinforcing bar spacing).

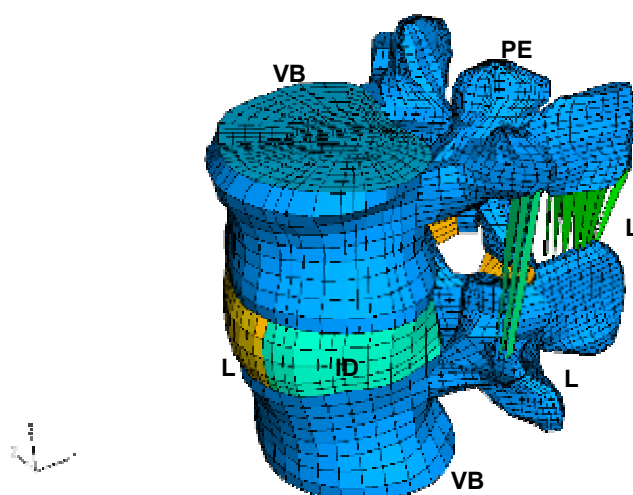


Fig. 2. FE model of human lumbar spinal segment:
 VB – vertebral bodies, ID – intervertebral disc, L – ligaments, PE – posterior elements

This technique allows us to alter easily the number of layers, section properties and its orientation. Moreover, it is independent of the element re-meshing mentioned above. In our study, we have calibrated the 3-D embedded elements based on the data published [9], [22]. So, the numbers of fiber (rebar) elements and cross-sectional areas were calculated based on the available data published [9]. The nucleus pulposus assumed here as an incompressible body was modelled as fluid-filled cavity using hydrostatic fluid elements with the initial pressure equal to 2 MPa. Hydrostatic fluid elements cover the boundaries of the nucleus pulposus. They share the nodes at the boundary between cavity and standard elements of annulus fibrosus.

In order to model vertebral bodies, solid elements were used (figure 2), but the material was described by orthotropic properties [11], [12]. Two kinds of bones were taken into consideration: cortical and cancellous ones. Shell elements were used for the cortical bone of the vertebral body, which is really very thin sheet. For the cancellous part solid 3-D elements were used. The posterior elements were meshed with hexahedral elements, too. The cartilaginous endplates were modelled by thin shell elements.

The current finite element model (figure 2) consisted of 24277 elements, 20334 nodes and 84385 degrees of freedom. For seven spinal ligaments we used truss and membrane elements with the description of material devoid of compression.

6. Results and conclusions

The model presented was validated for compression, two bendings in vertical planes, and torsion [6]. The boundary conditions and values of the forces were adapted based on the data published [9], [18], [22]. On the basis of the results collected it was possible to verify the validity and quality of the model definition. The axial displacement and disc bulge under compression and rotation at bending and an axial rotation at torsion were compared with the values reported in literature [9], [18]. The analyses in [6] were performed for this validation for the segment without facets and ligaments.

The lumbar spinal segment with facets and ligaments accounted for was analyzed to build its stiffness matrix. The six loading cases with the unit loads/moments were applied to the spinal segment to calculate the required coefficients of the stiffness matrix. The loads were applied to the reference point P (figure 3) which couples all points on the upper surface of the vertebral body; the bottom surface of the vertebral body was protected against any movement. The relative displacements recorded allow us to build the compliance matrix, and as inverse of it – the stiffness matrix. Figures 3 and 4 present the shape of deformed segment at sagittal bending, both positive and negative, while figure 6 pertains to the case of torque moment.

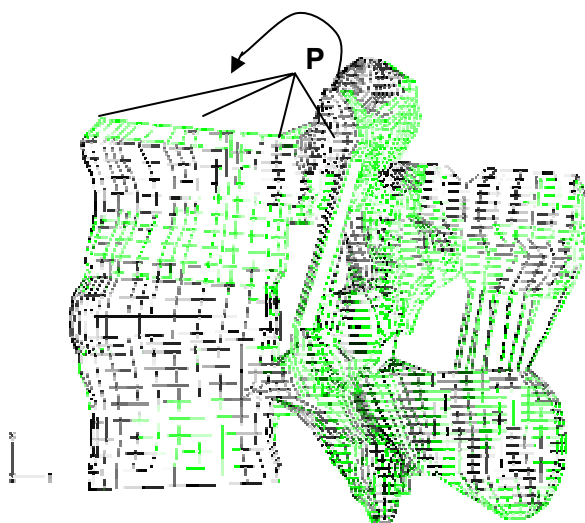


Fig. 3. Deformed shape (with undeformed light sites) for sagittal bending (positive)

The bulging of the disc as well as its structured behaviour are noticed easily. Figures 6 and 7 show the moment versus rotation/displacement curves representing the case of sagittal bending, while figure 8 refers to the case of torsion of the segment. The linear response within the examined range of deformations is evident, so the load independence of the stiffness coefficients has been approved. It can be seen from the figures that the responses of the lumbar spinal segment due to shear loads as well as due to bending in two vertical planes are different in both directions, so the stiffness matrix will be in fact sensitive to the sign of the loads. Hence the stiffness matrix will not be symmetric, as is expected, due to unsymmetry of real geometry. The connector-type element used in the studies allows accounting for this phenomenon and it will be implemented in the modelling of lumbar spinal segment.

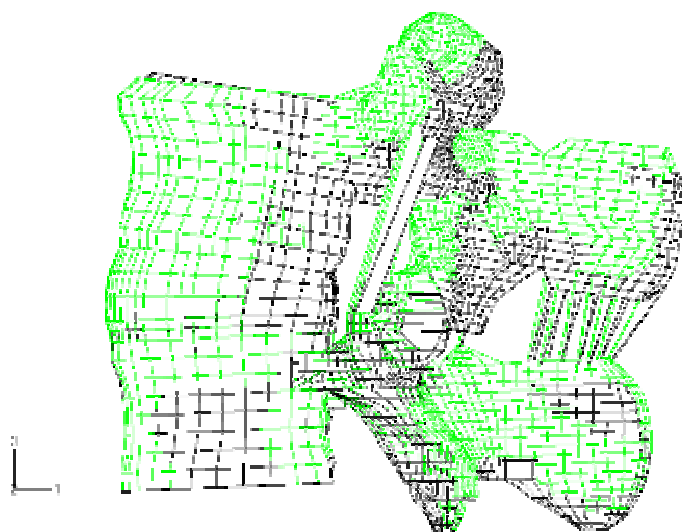


Fig. 4. Deformed shape (with undeformed light sites) at sagittal bending (negative)

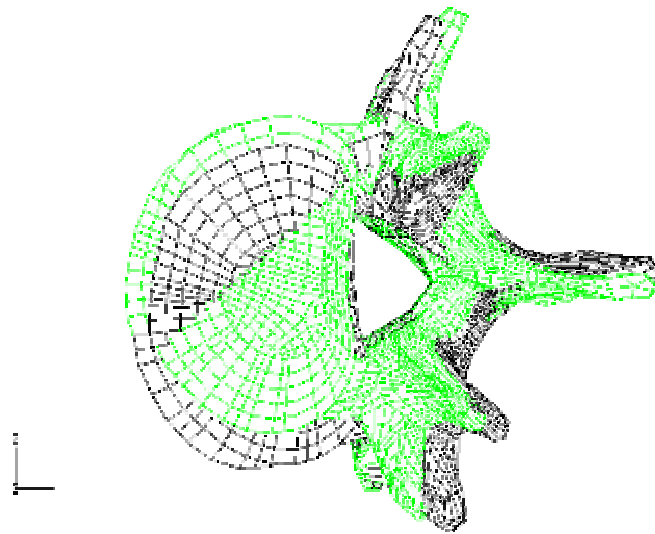


Fig. 5. Deformed shape (with undeformed light sites) at torque moment

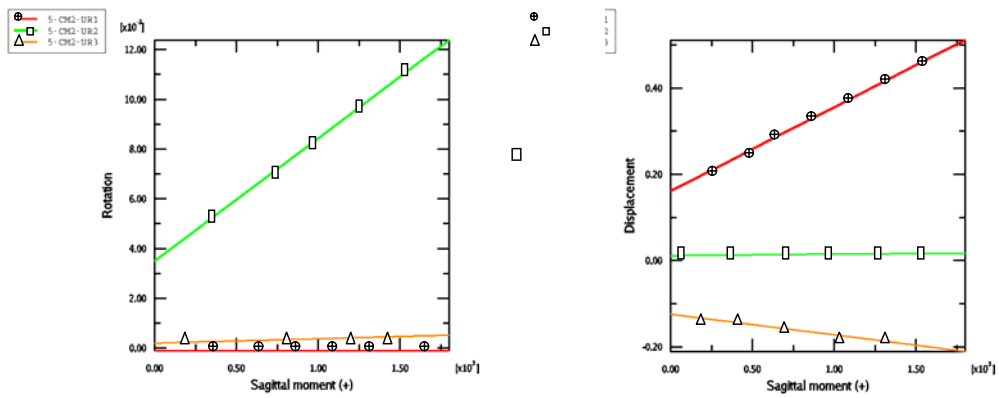


Fig. 6. Rotation/displacement plot at positive sagittal bending

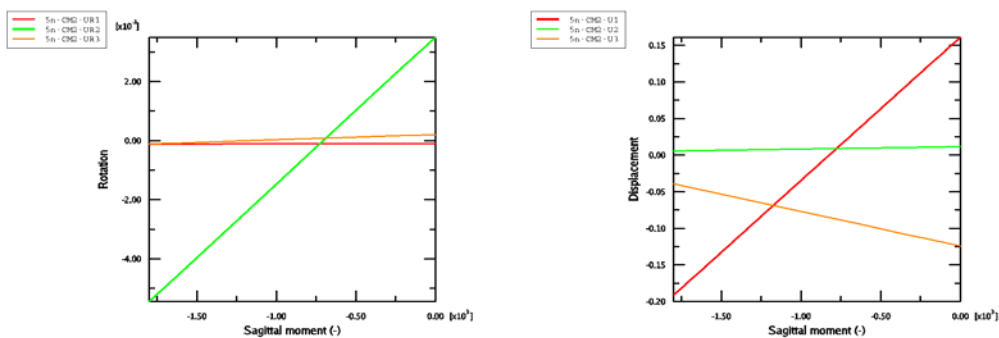


Fig. 7. Rotation/displacement plot at negative sagittal bending

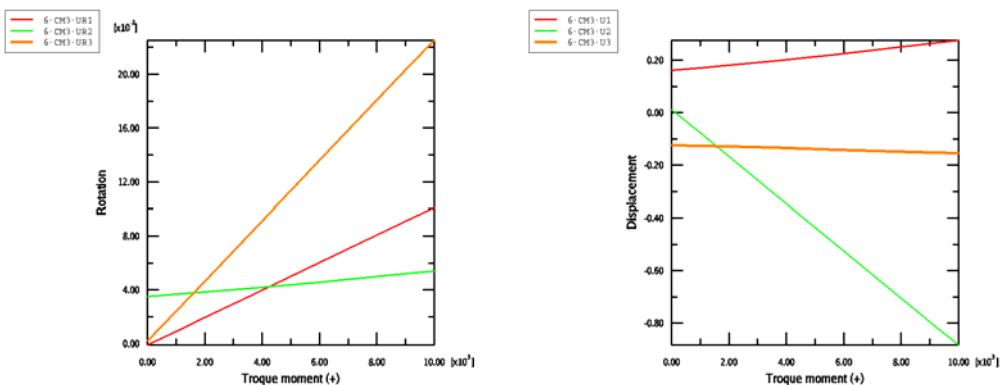


Fig. 8. Rotation/displacement plot at torque moment

7. Conclusions

The paper discusses the problem of creation of a complex 3-D numerical model making the study of the behaviour of spine segment possible. The model takes into account the major difficulties due to geometry definition, adoption of the constitutive properties of the segment components and the loading as well as boundary conditions. The mesh of FE model of a single spine segment is fully structured. It consists of 3-D elements and poses the computational problem of almost one hundred thousand degrees of freedom.

In the examined range of spine segment deformations, the behaviour of this segment seems to be almost linear. This suggested that the use of simplified connector-like elements for testing the segment response can be appropriate also for the analysis of the whole spine deformation. The characteristics of the connectors arise from these 3-D studies.

3-D and connector-like models give the qualitatively convergent results. In future applications, when we deal with the problem of the whole spine stability and deformations, it could be useful to use much simpler elements whose characteristics are computed from the complex 3-D models.

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