Computational modeling of an adolescent's reduced functional spine unit: a comparative study between isotropic and anisotropic hyperelastic intervertebral discs

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Abstract. Computational finite element analysis has been used in investigations of the spine, contributing to the understanding of its biomechanical behavior. Intervertebral discs (IVD) are structures that make the connection between the vertebrae, allowing movement between it and, consequently, increasing the mobility of the spine and been one of the most critical components of the spine. Therefore, the present paper has as main objective to propose two analyzes of a reduced thoracic functional spine unit: in the first, the IVD is modeled as a hyperelastic rubber like isotropic structure and, in the second, a hyperelastic anisotropic property is adopted to represent the fibers of the annulus fibrosus. Finally, these models were compared with each other to assess the contribution of anisotropy hyperelastic analysis to the biomechanical behavior of the spine. Through the results obtained, it is concluded that the hyperelastic anisotropic models of the annulus fibrosus can bring more authentic results to the computational models of the spine, considering the contribution of fibers and the large deformations to which soft tissues are subjected. Furthermore, the reduced model can predict the spine's biomechanical behavior, being useful in spine mechanical studies.

Keywords: Functional spine unit, Finite elements model, Intervertebral disc, Hyperelastic disc, Anisotropic model.

1 Introduction

The spine is a mechanical structure in which the vertebrae articulate with each other through the articular facets, discs, ligaments and muscles. Its function is to support the weight of the head and trunk, to protect the spinal cord and to provide a partially rigid and flexible axis for the body, in addition it's playing a fundamental role in locomotion [1].

In the vertebral column, the discs are hyperelastics structures that make the connection between adjacent vertebrae. The disks are composed by nucleus pulposus and annulus fibrosus. The intervertebral disc predominantly supports compression stresses. However, it is also subject to moment, tensile and shear stresses, depending on the type of movement that the spine is subjected [2]. The nucleus pulposus is white, shiny and semi-gelatinous tissue; it occupies the center of the disc in about 30 to 50% of the volume. It works as a movement axis between adjacent vertebrae and absorbs the forces acting on the spine [3]. The annulus fibrous consists of concentric lamellae of collagen fibers oriented at 30º, arranged in a spiral shape, with the opposite direction from layer to layer. Among the functions of the annulus fibrous are helping to stabilize the adjacent vertebral bodies; allow movement between the vertebrae; act as an accessory ligament; ensure that the nucleus pulposus remains in position; and dampen acting forces [2,3].

Computational analyzes based on the Finite Element Method (FEM), described in terms of partial differential equations and approximations, have been widely used in investigations of the spine, contributing to the understanding of its biomechanical behavior. With the increase in computational capacity, increasingly complex models can be analyzed, making it possible to study the displacements, deformations and stresses of the spine when subjected to different types of loads, and allowing for a better understanding of various diseases. According to Kurutz [4], the intervertebral disc is the most critical component of the spine, both in its mobility and in its load-bearing capacity. Therefore, its modeling is of great importance. Ruberté, Natarajan and Andersson [5] investigated the behavior of the lumbar spine under disc degeneration at the level of the L4 and L5 vertebrae and concluded that the degeneration of only one disc can compromise the biomechanics of the adjacent segments, increasing the risk of injury. Little and Adam [6] studied the behavior of an adolescent's spine with variations in the properties of soft tissues and concluded that changes in the physical properties of the ligaments and intervertebral discs directly influence the flexibility of the spine.

Thus, the present paper proposes to make two analyzes of an adolescent's reduced thoracic functional spine unit (FSU). In the first analysis, called model 1, the intervertebral annulus fibrosus is modeled as a hyperelastic isotropic material. A second analysis, entitled model 2, consider the annulus fibrosus as a hyperelastic anisotropic material, in other words, it will consider the concentric lamellae of collagen fibers of the annulus fibrosus. We chose to model a spinal unit with adolescent geometry due to the low level of spine analysis in this age group, although many pathologies are present in this public, such as adolescent idiopathic scoliosis. The aim of this study is to evaluate the contribution of collagen fibers from the annulus fibrosus in the functions of the intervertebral disc, considering the great importance of this component in the flexibility and stability of the human spine.

2 Materials and methods

2.1 Creation of 3D reduced FSU geometry

In this topic, a proposal for reduced 3D geometric modeling of the adolescent T7-T8 thoracic spinal unit is presented. The objective of the adopted construction methodology is to provide a simple modeling, since the focus of this study is on the intervertebral disc and not in the posterior structures of the vertebra (lamina, pedicle, transverse and spinous processes). Furthermore, the present model may be capable of replication, which may help new researchers to initiate 3D analysis of biomechanical structures. The SolidWorks® program was adopted to construct the 3D model geometry.

A typical vertebra is formed by the body, arch and vertebral processes. The body, usually in a cylindrical shape, is the anterior part of the vertebra and consists of a denser external part formed by cortical bone and a more porous internal part formed by trabecular bone. A reduced FSU consists in only two vertebrae body interconnected by the intervertebral disc, that is, the presence of the vertebral arch, processes, joints and ligaments was not considered.

To model the vertebra, its cross-sectional contour was made using the upper view of a typical thoracic vertebra obtained from the Atlas of Human Anatomy [7]. This contour was made with the use of curved lines (splines), since these are more adaptable to the format sought. After creating the outline of the cross section of the body of the vertebra, the dimensions were adapted to fit the dimensions of adult thoracic vertebrae obtained in the literature [8][9]. Then, these values were scaled to the dimensions of the thoracic vertebrae of a 10-year-old [10].

The next process consisted of dividing these solid bodies. The vertebrae were divided into cortical bone and trabecular bone and the discs divided into nucleus pulposus and annulus fibrous. The region of cortical bone is 1.5 mm thick [11] and the nucleus pulposus occupied 40% of the total volume of the disc (Figure 1).

2.2 Numerical analysis of reduced FSU

For the creation of meshes and numerical analysis of the FEM models, the Abaqus software was adopted. The machine used to develop the segments contains the following specification: Intel Core i5-10210U CPU 1.60GHz 2.11 GHz, 8.0 GB RAM and NVIDIA GeForce MX250 graphics card.

The geometric model generated in SolidWorks was imported into Abaqus as an "Assembly" and then the material assignments, loading and contour conditions, meshing and processing were carried out.

The vertebrae body, both in the region of cortical and trabecular bone, are modeled as linear elastic isotropic material. There was a great difficulty in finding the physical parameters of adolescent spine, for this reason the data from the adult spine were used and it is shown in Table 1.

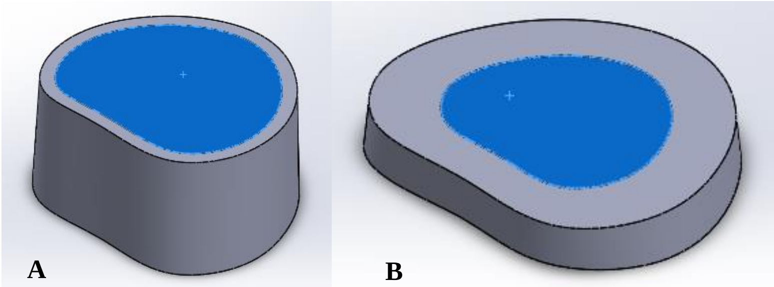

Figure 1: A- Representation of the body of the vertebra divided into cortical bone (grey) and trabecular bone (blue). B - Representation of the intervertebral disc divided into annulus fibrous (grey) and nucleus pulposus (blue).

Table 1. Properties of vertebrae materials.

Region	E (MPa)	Poisson (v)
Cortical bone	1200 [5]	0,3 [5]
Trabecular bone	350 [12]	0,3 [12]

The nucleus pulposus was modeled as an incompressible hyperelastic material in both models, its behavior was represented by the constitutive two-parameter Mooney-Rivlin model. The annulus fibrous, in model 1, was established as an isotropic rubber like hyperelastic material using the Neo-Hooke constitutive model. For the model 2, the annulus fibrosus was model as an anisotropic hyperelastic material by the Holzapfel-Gasser-Ogden constitutive model to represent the collagen fibers presents in the annulus. All constitutive models are available in Abaqus 6.14 and are well describe in the Abaqus documentation [13]. The material parameters used in both models are shown in Table 2 followed by the references.

Table 2. Properties of disc materials.

Model 1	Model 2
Nucleus pulposus	**Nucleus pulposus**
Mooney-Rivlin C_{10}= 0.4 C_{01}= 0.1 D=0.0004 [14]	Mooney-Rivlin C_{10}= 0.4 C_{01}= 0.1 D=0.0004 [14]
Annulus fibrosus	**Annulus fibrosus**
Neo-Hooke C_{10}= 0.34 D=0.306 [15]	Holzapfel-Gasser-Ogden C_{10}= 0.34 D=0.306 $K_1 = 1.8$ $K_2 = 11$ Kappa $= 0$ [15]

Gravitational forces are primary loads sustained by the spine in an upright position. In addition, efforts of muscular origin and external loads contribute to this load, which leads to a load greater than the portion of body weight located above it, in each mobile segment [16]. The force acting on the third lumbar vertebra of a 70 kgf adult in the standing position is about 700 N [2]. In addition, to each vertebra segment, in the caudal direction, 2.6% of the load of the anterior vertebra is added [17]. Thus, the adoption of a 400 N load (direction of gravity) distributed on the upper plateau of the T7 vertebrae, about 0.9 N / mm², was satisfactory for an adolescent spine.
The base of T8 was completely embedded and the contact between disc and vertebrae was guaranteed through tie constraints.
The choice of the mesh to be used went through a process of analysis. Five meshes with different sizes were analyzed (Table 3). Based on the study of convergence (table 4) and mesh quality it was decided to use the mesh 4 which has average mesh size of 0.375 mm in the discs and 0.5 mm in the vertebrae. Altogether there were 179176 linear 8-node hexahedral elements with reduced integration. It is important to emphasize the need to use a hybrid formulation for the elements of the nucleus pulposus to represent its almost incompressible behavior, since when the material is incompressible, the volume does not change during the application of a load, so there is no way to relate the stresses with the displacements. Then the hybrid formulation uses independent functions for the stress and displacement fields. Finally, the model mesh is seen in figure 2.A.

Table 3. Average mesh size in the different analyzes (mm).

	Mesh 1	**Mesh 2**	**Mesh 3**	**Mesh 4**	**Mesh 5**
Cortical bone	1.5	0.7	0.5	0.5	0.375
Trabecular bone	1.5	0.7	0.5	0.5	0.375
Intervertebral disc	1.5	0.7	0.5	0.375	0.375

Table 4. Convergence analysis – Minimum principal stress.

	Minimum principal stress (MPa)				
	Mesh 1	Mesh 2	Mesh 3	Mesh 4	Mesh 5
Vertebrae	-2,206	-2.650	-2.853	-2.854	-3.060
Discs	-1,816	-1.824	-1.828	-1,829	-1.829

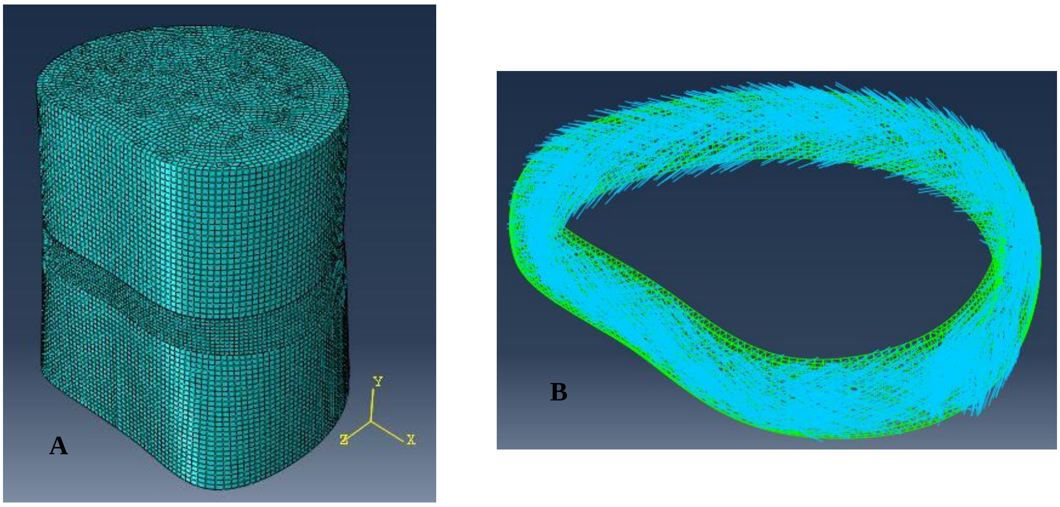

Figure 2. A- FSU mesh; B- Fibers oriented at ±30 in the first two annulus fibrosus lamellae (Model 2).

In model 2, the annulus fibrosus was divided into 5 lamellae to represent the fibers oriented at ±30º. The constants K1 and K2 of the Holzapfel-Gasser-Ogden model refer to the physical properties of the collagen fibers and the orientation of these fibers is virtually established for each annulus fibrosus layer through the material orientation in the Abaqus. Figure 2.B shows the two outermost lamellae of the ring, in blue it is possible to clearly see the fibers oriented at ±30 degrees.

3 Results

3.1 Model 1

Compressive stresses in the vertebrae ranged from approximately 1 MPa in the T7 vertebra to 2.85 MPa at the base of T8 vertebrae, both in cortical bone regions. The disc underwent low compressive stresses in the outermost part of the annulus fibrosus, with a gradual increase towards the nucleus pulposus, where the maximum stress of approximately 1.83 MPa occurs (Figure 3-A). The magnitude of displacement remained constant across the entire T7 vertebra, about 0.31 mm while the T8 vertebra suffered almost zero displacement. The vertical displacement (U2 displacement - y axis) was greatest at the top of the disk, about -0.33 mm, and smallest at the bottom of the disk, about -0.02 mm. Disc displacement in the frontal plane (U1 displacement - x axis) was ±0.39mm. In the sagittal plane (displacement U3 - z axis), the disc underwent displacements of ±0.5mm.

3.2 Model 2

Compressive stresses in the vertebrae ranged from approximately 1 MPa in the T7 vertebra to 3.20 MPa at the base of T8 vertebrae, both in cortical bone regions. The disc underwent low stresses in the outermost part of the annulus fibrosus, with a gradual increase towards the nucleus pulposus, where the maximum stress of approximately 1.81 MPa occurs (Figure 3-B). The magnitude of displacement remained constant across the entire T7 vertebra, about 0.32 mm while the T8 vertebra suffered almost zero displacement. The vertical displacement (U2 displacement - y axis) was greatest at the top of the disk, about -0.34 mm, and smallest at the bottom of the disk, about -0.015 mm. Disc displacement in the frontal plane (U1 displacement - x axis) was ±0.39 mm. In the sagittal plane (displacement U3 - z axis), the disc underwent displacements of ±0.51 mm.
Table 5 shows the maximum values for stress and displacement in both models.

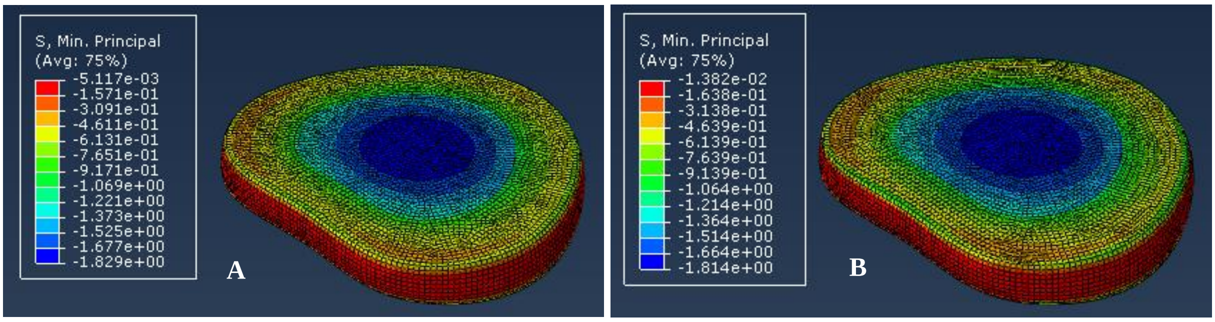

Figure 3. A- Minimum principal stress on disc – Model 1; B- Minimum principal stress on disc – Model 2.

Table 5. Maximum compressive stresses (MPa) and maximum displacements (mm) in models 1 and 2.

Parameter	**Model 1**	**Model 2**
Maximum compression of vertebrae	-2.854	-3.204
Maximum compression of disc	-1.829	-1.814
Maximum FSU magnitude displacement	0.533	0.536
Maximum frontal disc displacement	0.391	0.389
Maximum sagittal disc displacement	0.501	0.508

4 Discussion

It is known that the discs are fiber reinforced structures that behave like nonlinear hyperelastic structures. Despite this, they are targets for simplifications in the spine FEM models where sometimes the fibers are not included in the analyses. However, this type of simplification can lead to fictitious results in terms of stress and displacements of the models. So, the present work proposed to investigate the behavior of a thoracic FSU in adolescent age through a simplified geometric modeling focusing on the comparative analysis between two models, 1 and 2, when the constituent characteristics of the annulus fibrous were varied (hyperelastic isotropic or hyperelastic anisotropic model).
The vertebrae of both models showed similar behavior in terms of stress, however model 2 showed a maximum compression stress about 12% higher than the maximum found in model 1. It was noticeable in the two analyzes that the vertebrae T8 was the most requested to compression. The magnitude displacements in both models were basically equal.
Contrary to expectations, there were small differences in terms of compression and displacement in the intervertebral discs of the models. This fact must be related to the applied loading, as the annulus's fibers only

work by tension while the isotropic matrix absorbs the compression stresses [2]. In the case of the model presented, only compression load was used, so it is possible to imagine that the largest contribution is given by the isotropic matrix of the annulus, making it difficult to notice the contribution of the fibers.
The results of compression in the vertebrae of model 2, stress 12% higher than in model 1, proves that there is, in fact, a contribution of the fibers in the stress distribution of the FSU, although in a subtle way due to the type of loading adopted.
There are several works on spine by the MEF modeling in the literature, but each study deals with loads and simplifications according to the objectives that are proposed in them. Therefore, validating this type of analysis becomes one of the biggest challenges of this biomechanical analysis of the spine, even more because it is an FSU of adolescents with reduced geometry.
Aroeira [12], analyzed an adult thoracic FSU T7-T8 with complete geometry. The load implemented in her model was an axial 400 N distributed in the top of T7. The maximum values of compression stress obtained in the vertebrae was about 6 MPa and 2.8 MPa on discs. The maximum disc displacement in the sagittal plane was approximately 0.02 mm. It is important to highlight that no FEM modeling of the reduced adolescent thoracic spine was found in the literature to serve as a basis for comparison and validation of the model presented. But despite that, it was observed that the behavior of the models presented in this work was like that presented by Aroeira [12]. However, the differences in terms of displacements were substantial, but such differences can be explained by the geometric difference between the models and, even more, by the adoption of different constitutive models for the intervertebral discs. Although it is known that biological structures such as the intervertebral disc present large deformations, Aroeira [12] adopted an isotropic elastic modeling for the entire disc, as its focus was to present a detailed geometry and evaluate the contributions of the joints between vertebrae.
Additionally, the ultimate stress of human bone depends on the type of bone, the amount of bone mass, the type of load imposed and other factors, which prevents a standardized value. According to Hamill, Knutzen and Derrick [18], bone fracture can happen due to an isolated traumatic event or an accumulation of microfractures. Besides that, some data found in the literature indicate that the cortical bone may show ultimate resistance to compression close to 200 MPa [18,19,20] and the trabecular bone may have ultimate resistance to compression between 2.45 and 4.6 MPa [20,21,22,23]. Therefore, the model proved to be satisfactory and coherent, dealing with compression stresses in the vertebrae. Furthermore, although there are no studies that provide concrete results on the maximum compressive strength of intervertebral discs, Nachemson [24] concludes from experimental evaluations that the stress inside the intervertebral discs is 1.5 times that which was applied, thus, the model is coherent.
However, despite the geometric simplifications imposed on the model of the present work, it proved to be coherent and useful to represent the biomechanical behavior of the spine. Still, it was clear from the results that, the use of hyperelastic constitutive models for the intervertebral discs, gives more accurate results for the FEM models of the spine, since greater flexibility is guaranteed and, consequently, it allows a better movement between the vertebral bodies and acts as a better damper, two of the main functions of intervertebral discs [2,3].

5 Conclusion

Computational analysis of the spine by FEM offers several alternative tools to simulate various parameters and conditions of the spine. Thus, the experimental and morphological parameters can be modified and tested in a reproducible way. The human spine is a complex structure to model. Therefore, much of the work involving spine models by the FEM makes simplification assumptions to facilitate simulation.
Through the results obtained, it is concluded that the hyperelastic anisotropic models of the annulus fibrosus of the discs can bring more authentic results to the computational models of the spine, considering the large deformations that the soft tissues are subject and considering the contributions of the fibers present in the structure. In addition, the reduced model predicts the biomechanical behavior of the spine and can be useful in pre-surgical analysis, being able to simulate the effects of different types of treatment, helping in more assertive decision making, because simplified analysis requires less processing time when compared to full geometry models. In addition, the model becomes a source of study for researchers in the field of biomechanics, given the clear methodology employed in the construction of geometry, simulation and analysis.
Still, it would be interesting to impose loads of typical daily tasks to better evaluate the model in order to better assess the contribution of collagen fibers in the behavior of the disc and spine.

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References

[1] K.L. Moore, A.F. Dalley, A. M. R. Agur, Clinically oriented anatomy, 8 ed., LWW, (2017), 1168.
[2] A.A. White, M.M. Panjabi, Clinical Biomechanics of the Spine, 2 ed., Philadelphia: J. B. Lippincott Company, (1990), 722.
[3] J. Natour, Spine: basic knowledge, 2 ed., etcetera, (2004), 244.
[4] M. Kurutz, Finite Element Modelling of Human Lumbar Spine, Finite Element Analysis, In: David Moratal (ed) ISBN: 978-953-307-123-7, (2010), 210–236.
[5] L.M. Ruberté, R.N. Natarajan, G.B. Andersson, Influence of single-level lumbar degenerative disc disease on the behavior of the adjacent segments-A finite element model study. J Biomech, 42 (2009), 341–348.
[6] J.P. Little, C.J. Adam, The effect of soft tissue properties on spinal flexibility in scoliosis: Biomechanical simulation of fulcrum bending, Spine, 34 (2009), 76–82.
[7] F.H. Netter, Atlas oh Human Anatomy, 6 ed., Elsevier, (2015), 154.
[8] M.M. Panjabi, K. Takata, V. Goel, et al. Thoracic human vertebrae quantitative three-dimensional anatomy. Spine, 16 (1991), 888–901.
[9] M.E. Kunkel, A. Herkommer, M. Reinehr, et al., Morphometric analysis of the relationships between intervertebral disc and vertebral body heights: An anatomical and radiographic study of the human thoracic spine. Journal Of Anatomy, 219 (2011), 375–387.
[10] G. J. M. Meijer, Development of a non-fusion scoliosis correction device: numerical modelling of scoliosis correction, PhD thesis: University of Twente, Enschede, (2011), 165.
[11] M.A. Tyndyk, V. Barron , P.E. Mchugh, D. O'Mahoney, Generation of a finite element model of the thoracolumbar spine. Acta of Bioengineering and Biomechanics, 9 (2007), 35–46.
[12] R.C. Aroeira, Biomechanical study of an adolescent's thoracic spine, in kyphosis and hypocyphosis, under asymmetric ligament loading: a possible prediction of idiopathic scoliosis, PhD Thesis: Structural Engineering, Graduate Program in Structural Engineering, Federal University of Minas Gerais, Belo Horizonte, 2017, 143.
[13] SIMULIA, Abaqus analysis user's manual, Version 6.14, Dassault Systems, 2014.
[14] N.M. Lalonde, I. Villemure, R. Pannetier, et al., Biomechanical modeling of the lateral decubitus posture during corrective scoliosis surgery. Clinical Biomechanics, 25 (2010), 510–516.
[15] A. Calvo-Echenique, J. Cegoñino, R. Chueca, A.P. Palomar, Stand-alone lumbar cage subsidence: a biomechanical sensitivity study of cage design and placement. Computer Methods and Programs in Biomedicine, 162 (2018), 211-219.
[16] A. Nachemson, The load on lumbar discs in different positions of the body. Clinical orthopaedics and related research, 45 (1996), 107-122.
[17] M. Driscoll, C. Aubin, M.Alain, et al, The role of spinal concave – convex biases in the progression of idiopathic scoliosis, 18 (2009), 180–187.
[18] J. Hamill, K. M. Knutzen, T.R. Derrick, Biomechanical basis of human movement, 4 ed. Lippincott Williams & Wilkins, 2015, 490.
[19] M.J. Katzenberger, D.L. Albert, A.M. Agnew, et al. Effects of sex, age, and two loading rates on the tensile material properties of human rib cortical bone. Journal of the Mechanical Behavior of Biomedical Materials, 102 (2020), 103410-103430.
[20] E.F. Morgan, G.U. Unnikrisnan, A.I. Hussein, Bone Mechanical Properties in Healthy and Diseased States. Annual Review of Biomedical Engineering, 20 (2018), 119–143.
[21] T.M. Keaveny, E.F. Morgan, G.L. Niebur, et al, Biomechanics of Trabecular Bone. Annual Review of Biomedical Engineering, 3 (2001), 307-333.
[22] O. Lindahl. Mechanical properties of dried defatted spongy bone. Acta Orthopaedica Scandinavica, 47 (1976), 11–19.
[23] L. Mosekilde, L. Mosekilde, C.C. Danielsen, Biomechanical competence of vertebral trabecular bone in relation to ash density and age in normal individuals. Bone, 8 (1987), 79–85.
[24] A. Nachemson, Lumbar intradiscal pressure: Experimental Studies on Post-Mortem Material, Acta Orthopaedica Scandinavica, 31:sup43 (1960), 1-104.