

# Anisotropic response of the Holzapfel's constitutive model for the lumbar spine considering degenerative conditions

**Moustafa MOSBAH**

PhD candidate in biomechanics under the supervision of Dr. BENDOUKHA at the laboratory of numerical and experimental modeling of mechanical phenomena. Obtained the licence and Master's degree from the University of Mostaganem in Mechanics.

**Mohammed BENDOUKHA**

Research Professor at the University of Mostaganem. Obtained his Engineering degree in Mechanical Engineering from the University of Science and Technology of Oran (USTO). After postgraduation studies he obtained Magister at the USTO IN 1991. Obtained his PhD in tribology speciality in 2010 at the University of Mostaganem, and subsequently his habilitation to direct the research of the same university in 2011. He is active in the field of research at the laboratory of numerical and experimental modeling of mechanical phenomena.

**MOUSTAFA MOSBAH** • Laboratory of numerical and experimental modeling of mechanical phenomena, Department of Mechanical Engineering, University Abd El Hamid Ibn Badis of Mostaganem, Algeria

**MOHAMMED BENDOUKHA** • Laboratory of numerical and experimental modeling of mechanical phenomena, Department of Mechanical Engineering, University Abd El Hamid Ibn Badis of Mostaganem, Algeria • bendoukham@yahoo.fr

Érkezett: 2018. 02. 26. • Received: 26. 02. 2018. • <https://doi.org/10.14382/epitoanyag-jsbcm.2018.20>

## Abstract

In order to study the risk of disc degeneration under load carriage, an osseo ligamentous FE model of the L4-L5 segment extracted from the validated lumbosacral model L1/S1 was used. Starting from CT images to simplified geometry, many programs and codes were used to simplify the geometry, assemble the mesh and formulate the bony structures of the segment. Abaqus 6.14 was used to pre and post analysis of the FE-model, which included a detailed calibrated model of intervertebral disc (IVD), The range of motion (ROM) curves, the intradiscal pressure, load-strain amount under pure physiological loads were considered for investigation. The model developed in this study considered the holzapfel anisotropy hyperelasticity of the annulus as well as a realistic description of the nucleus geometry, which allowed an improved representation of in vitro and in vivo experimental data during the validation process.

Keywords: intervertebral disc, L4-L5, range of motion finite element model, CT images, intradiscal pressure, Holzapfel anisotropy, hyperelasticity

Kulcsszavak: intervertebrális lemez, L4- L5, mozgásterjedelem vége-selelemes modellje, CT felvétel, intradiszkalis nyomás, Holzapfel anizotrópia, hiperelaszticitás

## 1. Introduction

The disc degeneration is characterized by changes in the morphology and biochemistry of the IVD [16]. These biologic changes of disc degeneration are associated with back pain and other spinal disorders, such as disc herniation, spondylolisthesis, facet arthropathy, and stenosis. FE method is able to simulate a variety of clinical situations in a more way, and has been acting a fully matching partnership based on experimental approaches for spine biomechanics research [12]. The FE-models are widely used to provide an assessment of spine with spinal instrumentation and assist in the design, development and optimization of that spinal instrumentation. Advanced to other methods, these FE-models can predict the changes of stresses and strains in the IVD, vertebrae and ligaments and very comprehensive ROM data [8, 20]. Our main objective is to review the most recent computational studies in the application of FE models that report the issue of research better understanding of low back pain and eventually treat the spinal disorders. Little experiments were performed on intact motion segments (Miller 1986, Lin 1978, Hirsch & Nachemson 1954) and some on FSUs without posterior elements [2, 10, 13]. In the compression load case, the data from Brown 1957 is slightly lower than the generated values and the other experimental data. Some FE-models simulate the effects of degeneration [16], by simply removing some elements from the annulus fibrosus. Where others [18, 19, 20] disclose the influence of geometrical, mechanical, or poromechanical parameters on IVD behavior without showing change in

mechanical properties of degenerated IVD. More recently, recent poroelastic finite element model of the lumbar spine was developed to assess spinal response during physiological functions and behavior of degenerative disc [15, 16].

## 2. Methods and materials

A three-dimensional finite element (FE) model of the L4-5 FSU segment (*Fig. 2*) by modifying the mechanical properties of the intervertebral disc constituents [3]. All the bony structures including cortical and trabecular bone, whereas facet cartilage layers, annulus ground substance, nucleus pulposus and cartilage endplates were modeled by using solid hexahedral elements. The nucleus pulposus composed 43% of the whole volume of the disc. The symmetric FE-model of the human lumbar disc is developed to investigate the optimized properties required to mimic the intact by matching model predictions to experimental results. Facet joints simulated by a cartilaginous layer extruded in Abaqus along the vectors normal to the inferior face of the superior articular process to the central of thickness of 0.4 mm were modeled to be multi-linear elastic in compression [19] by surfaces-to surface contact with softened contact in the normal direction. [20] An initial typically gap of 0.2 mm was specified as reported [20]. The seven major ligaments were modeled as three dimensional, 4 noded Quad elements (T3D2) and allocated were as piecewise linear functions in stress-strain relationship [7, 8] as shown in *Table. 1*. This formulation allows simulation of changing ligament characteristics with different stress at a different level as a function of stress-strain.

Part	Structure	Young's modulus E (MPa)		Poisson's ratio $\nu$		Element type	Reference			
Vertebrae	Trabecular bone	100		0.2			[8, 14]			
	Cortical bone	12,000		0.3			[8]			
	Posterior element	3,500		0.25			[8, 14]			
	Cartilaginous End plate	23.8		0.4		C3D8	[20]			
	Facet cartilage	11		0.4			[5, 20]			
	Facet contact	Nonlinear soft contact				Contact	[21,22]			
Annulus Fibrosus	Healthy	$C_{10}$	$C_{20}$	$K1$	$K2$	$\kappa$				
	Degenerated	1.5715	9.2044	12.2	39.7	0.113	C3D8 [22]			
Nucleus Pulposus	Healthy	1		8.6		18.56	82.31	0.226		
	Degenerated	Incompressible fluid						F3D4	[6, 19, 21]	
Ligaments	Nonlinear stress-strain curve						S4R	[7, 8, 21]		

Table 1. Material properties used in FE Model  
1. táblázat A végeleemes modellben használt anyagjellemzők

The entire model consists of approximately 424 S4R shell elements and 33360 hexahedral elements C3D8

### 2.1 The intervertebral disc (IVD)

The intervertebral disc was modeled using two separate structure (Figs. 2 and 5).

### 2.2 The Nucleus Pulposus (NP)

The Nucleus Pulposus (NP) exhibits a gelatinous core and, therein describes the mutually coupled behavior of both solid deformation and viscoelastic characteristics of fluid flow [7]. An initial hydrostatic pressure of 0.1 MPa was prompted in the fluid filled cavity in the nucleus pulposus to simulate unloaded motion segments (Fig. 2) [17]. The healthy nucleus pulposus was modeled as incompressible fluid while the degenerated nucleus was defined as compressible fluid [10]

### 2.3 The Annulus Fibrosus (AF)

The Annulus Fibrosus (AF) was modeled as an anisotropic continuum structure, with deformation tensor [8], and the material coefficients  $C_{10}$ ,  $C_{20}$ ,  $k_1$ ,  $k_2$  and  $\kappa$  were based on biaxial tension tests performed on healthy and degenerated annuli fibrosi [9]. The progression of degeneration was simulated using three model variants:

$$W_{matrix} = C_{10} (I_1 - 3) + C_{20} (I_1 - 3)^2 \quad (1)$$

$$W_{fibers} = \frac{k_1}{2k_2} \sum_{\alpha=4,6} (e^{k_2 [(1-\kappa)(I_1 - 3) + \kappa(I_\alpha - 1)]^2} - 1) \quad (2)$$

The human healthy and degenerated intervertebral disc were modeled using three structures:

(a) Healthy intervertebral disc (grade I) defined by healthy annulus and healthy nucleus;

(b) Moderate degeneration disc (grade III) defined by healthy annulus and degenerated nucleus;

(c) Severe degeneration disc (grade IV) defined by degenerated annulus and degenerated nucleus. The grading is based on the Thompson degeneration scale Fig. 1.

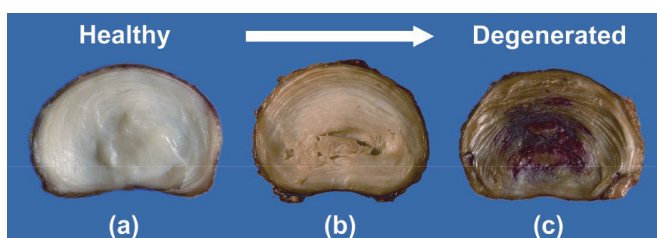


Fig. 1. (a) Healthy intervertebral discs; (b) moderate degeneration; (c) severe degeneration [23]

1. ábra (a) Egészséges intervertebrális lemez; (b) közepes mértékű károsodás; (c) súlyos károsodás [23]

## 3. Model formulation

The algorithms proposed in the present study is implemented into Abaqus as a User Material UMAT code embedded in an Python and Intel Fortran Compiler environment to simulate the biomechanical behavior of Annulus. This numerical approach consists on Holzapfel's hyperelastic constitutive models of the Annulus and a constitutive model of the seven major Ligaments, which were implemented in Abaqus via subroutines. The coefficients employed in all three conditions are listed in Table 1. While the material properties and in situ nuclear pressure were altered, osteophytes, local tears or similar macroscopic abnormalities were not modeled. Rather than attempting to simulate these defects of an independent individual and random nature, the focus was kept on the effects of measured changes in tissue behavior.

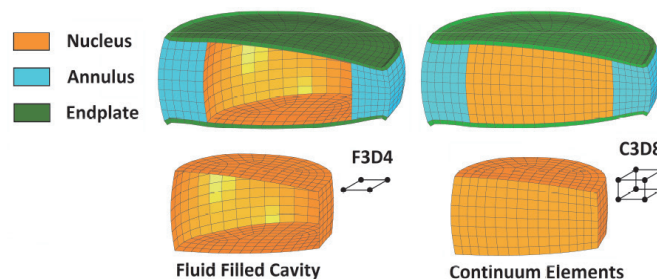


Fig. 2. Finite Element Model of the Intervertebral Disc IVD  
2. ábra Intervertebrális lemez IVD végeleemes modellje

## 4. Loading and Boundary Conditions

The IVD model comprises the intervertebral disc and the adjacent vertebral bodies. The lower vertebral body was rigidly fixed. Pure unconstrained axial compressive of 2000 N was applied to the upper vertebral body to the reference point which coupled all points on upper surface of the disc. The resulting stress-strain distribution at the center of the disc were estimate and the intradiscal Pressure IDP was predicted using the cavity point. Theoretically speaking, we used the concept of nucleus pulposus pressure measurement; by means of in vitro experiments on vertebra-disc-vertebra preparation whither, we found that the incompressible fluid pressure within the nucleus is directly related to the axial compression applied to the disc. The lower portion of the L5 vertebral body and its inferior facets were rigidly fixed. The load was applied through a reference node constrained to the upper surface of the L4 vertebrae using Abaqus Coupling Constraints elements (CCE). The L4/L5 Functional Spinal Unit (FSU) move in six different directions (six degrees of freedom (DOF)). A pure bending moment was incrementally increased 0, 1, 2.5, 5, 7.5, 10 Nm and applied to the model in all three planes of motion.

## 5. Results

### 5.1. ROM results

The finite element model predicted an increase in hydrostatic stress in the middle regions of the annulus by nearly seven fold. The intradiscal pressure IDP is considered to be very significant component to approximate internal stress behavior of the intervertebral disc. The in vitro measurements of the nucleus pressure in human intervertebral discs have shown proportion between IDP pressure and applied Load on the superior surface motion (ROM) in all loading directions except for lateral bending comparing to experimental.

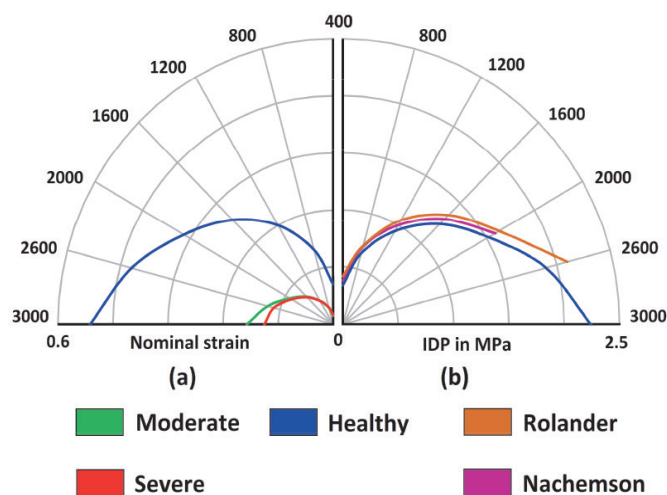


Fig. 3. (a) Comparison of intradiscal pressure (IDP) of the Healthy model with experimental results (b) Comparison of nominal strain of fibers against the Load (N) of Healthy, Moderate and Severe models  
 3. ábra (a) Kísérleti intradiszkalás nyomás (IDP) összehasonlítása az egészséges modell alapján számított értékkel; (b) Szálmegnyúlás és terhelőerő összehasonlítása egészséges, közepes és súlyos modellben

The greatest changes between Healthy and Moderate and Severe models were predicted in axial rotation by 30 % and 36 %. In lateral bending, a 27% increase in ROM was predicted in the Severe scenario (with respect to the Healthy condition), while the change in the Moderate scenario was approximately 32% (Fig. 4).

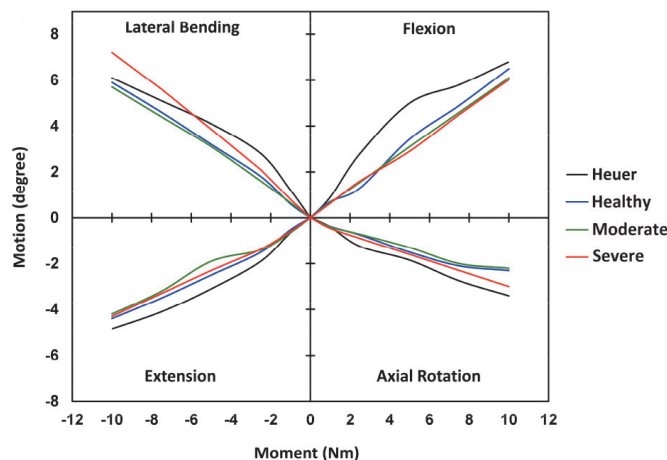


Fig. 4. ROM of L4-L5 FSU  
 4. ábra L4-L5 funkcionális szegmentális egység (FSU) mozgásterjedelme (ROM)

### 5.2. Stress and strain results

The analysis of the stress results allowed us to prove the most loaded region. As a most critical region for of spinal diseases, the presence of disc degeneration under physiological load steer to the alteration of the spine compartment and the pain function. The results for the body functions (Figs. 5-6) indicate that the superior values of the stresses were situated at posterior region which reaches stress of 2.9 MPa for healthy disc and 2.4 MPa for the degenerated disc; as expected, the results highlight that the most risky load condition corresponds to posterior side. For the degenerated nucleus the results presented in Moderate degeneration a decreased stress according to strain decreasing comparing to Healthy IVD. Furthermore, the magnitude of stress predictions were reduced in the anterior region of the disc while they increased in the posterior region as the simulated degeneration progressed. In all cases, the highest stresses were concentrated in the posterior region of the disc in extension loading. The highest strain were concentrated in the posterolateral region of the disc in lateral bending loading as degeneration progressed (Healthy-Moderate-Severe). The compressive strain prediction in the annulus increased in anterior extremities in extension and in the posterior extremities in flexion Fig. 5.

## 6. Discussion

The aim of this work was to construct an accurate FE-model to characterize the mechanical behavior of healthy and degenerated IVD. This model takes into account the IVD nature and also the clear preferential orientation of the collagen fibers in the annulus. The reduction in nonlinearity behavior with degeneration could suggest a diminished compaction effect of the degenerate tissues at large deformations which could be related to structural changes in annulus. The obtained

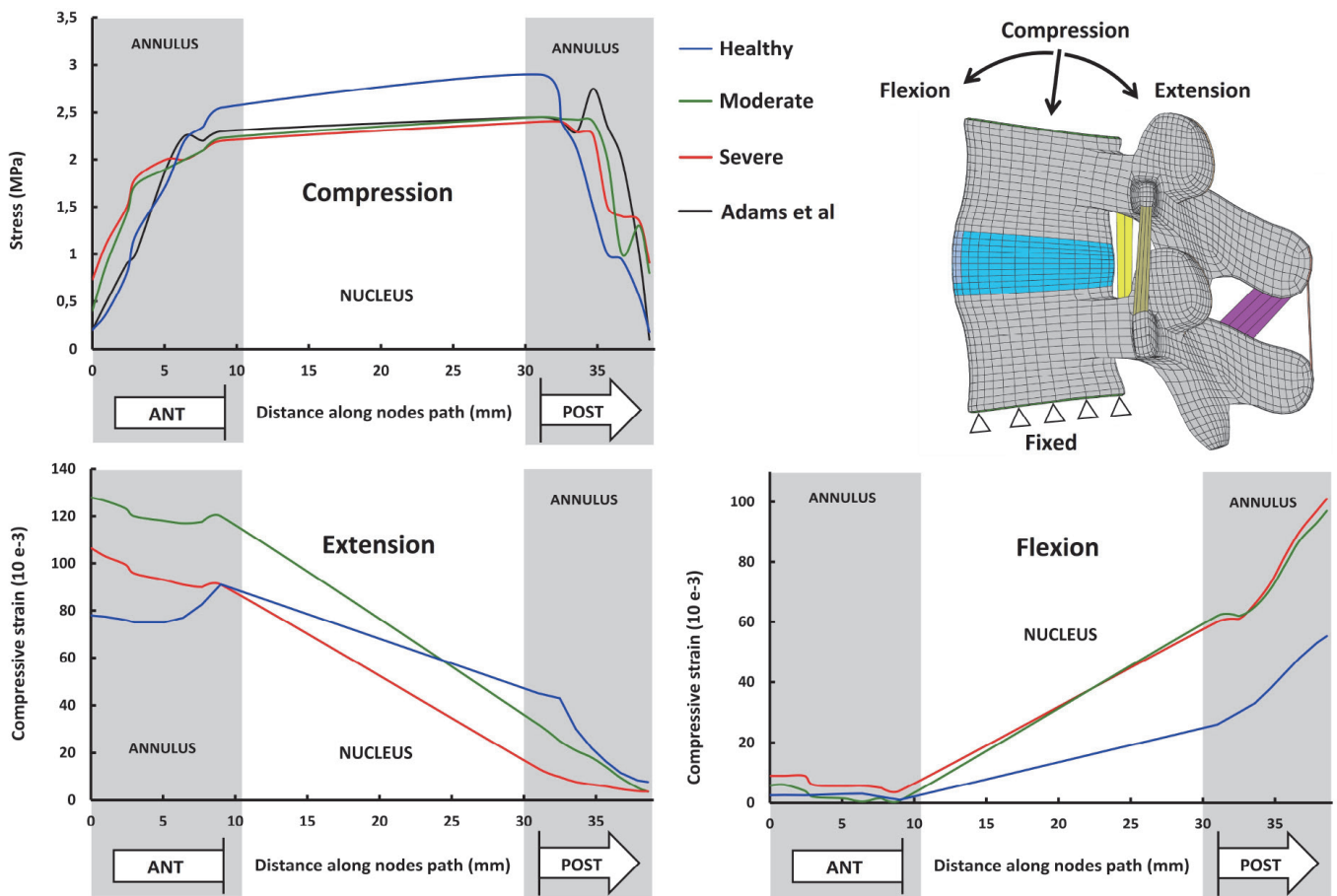


Fig. 5. Axial stress distribution and compressive strain predictions obtained from FE Models in the mid-height plane of the IVD from posterior to anterior midline under Compression (2000 N), Flexion and Extension (10 Nm).

5. ábra Tengelyirányú feszültségeloszlás és összenyomódás a végeleemes modell alapján az intervertebrális lemez középsíkjában, a posteriorból az anteriorba, nyomás (2000 N), hajlítás és megnyúlás (10 Nm) hatására

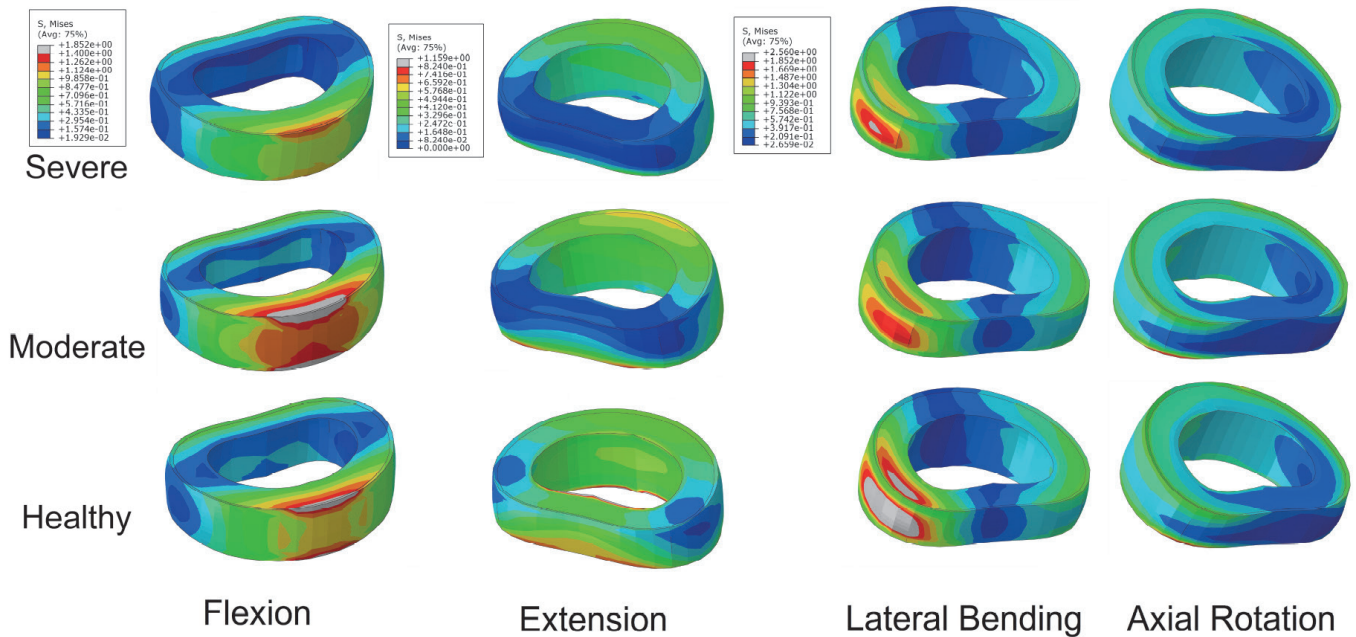


Fig. 6. Von-Mises Stress distribution obtained from FE Models: Healthy, Moderate, Severe

6. ábra Von-Mises feszültségeloszlás a végeleemes modell alapján egészséges, közepes és súlyos modellben

results are also in agreement with the experimental behavior [9, 20]. Experimental results, showed an increase of the elastic modulus in the toe part, thus there is a big role of fibers therefore when the fibers act there is no additional stiffening. In this analysis it has been obtained that the degeneration affects the biomechanics of this FSU; in particular an increase of the ROM in extension and partial decrease in flexion and lateral bending has been noticed.

## 7. Conclusions

In conclusion, the developed FE-model is capable of generating the mechanical behavior of normal and degenerated NP tissue with a favorable approximation. But on the degeneration of the other segments is a very complex procedure which should be identified in the degeneration of the AF tissue with accurate mechanical properties. Furthermore, the current work provides a qualitative analysis of the influence of single level disc degeneration on the mechanics of the segments under flexion/extension, lateral bending and axial rotation moments. It has been seen that degeneration modified the degree of motion and loading of the degenerated level. These changes could increase the risk of progression of degeneration to the nearest segments of the spine. The results obtained with the FE model seem to be in good agreement with in-vitro data as showed in Figs. 2 to 4. Additionally, nonlinear analytical functions used in the IVD ligaments facet joint may be used as input data to perform more accurate computational simulations of the full lumbosacral spine.

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### Ref:

**Mosbah, Moustafa – Bendoukha, Mohammed:** *Anisotropic response of the Holzapfel's constitutive model for the lumbar spine considering degenerative conditions* Építőanyag – Journal of Silicate Based and Composite Materials, Vol. 70, No. 4 (2018), 110–114. p. <https://doi.org/10.14382/epitoanyag-jsbcm.2018.20>