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THE ROLE OF POROELASTICITY ON THE BIOMECHANICS OF THE INTERVERTEBRAL DISC: A FINITE ELEMENT STUDY

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ABSTRACT: *The major goal for the present work is to evaluate a biomimetic Finite Element (FE) model of the Intervertebral Disc (IVD). Recent studies have emphasized the importance of an accurate biomechanical modeling of the IVD, which is a highly complex biphasic medium. A novel biphasic poroelastic model was implemented and coupled with Wilson's model (2005) for biphasic osmotic swelling behavior. Numerical tests were devoted to the analysis of the time-dependent behavior of the IVD. The results show good agreement with literature experimental data (Heuer et al., 2007 or O'Connell et al., 2011) and also with other numerical studies (Galbusera et al., 2011). In brief, this in-development IVD FE model aims to be a valuable tool to study the biomechanics of the IVD and its pathways for degeneration.*

1 INTRODUCTION

The Human spine is composed by twenty four motion segments (MS). Each one of these load-sharing units is composed by two vertebral bodies (VB), connected by an IVD and two facet joints. The IVDs are fibro-cartilaginous cushions serving as a shock absorbing system of the spine, which protect the vertebrae, brain, and other structures, providing both flexibility and load support. They are composed by three major components: the nucleus pulposus (NP), the annulus fibrosus (AF) and the cartilaginous endplate (CEP). Each IVD deals with complex loads, and such loads leave it vulnerable for both acute and chronic injuries. In fact, spine problems are a major cause of disability on western societies [1, 2].

In the middle of the 80s, Simon and co-workers and Huyghe and co-workers proposed pioneer finite element (FE) approaches aiming to describe the already recognized poroelastic behavior of soft-tissues, pursuant to the theoretical and experimental study of Mow and co-workers (1980) about the biphasic behavior of articular cartilage [3]. On the one hand, Simon and co-workers (1985) devoted their work to the study of the IVD biomechanical behavior. They analyzed factors as strains, fluid flow, fluid pressure and stresses, in order to evaluate the creep response of the IVD. The major outcome of this work was the description of the association between the increase of permeability and disc degeneration. The healthy IVD depends on the flow of nutrients from the adjacent structures, so the daily cycles, which include both loading and recovery phases, are highly dependent of the poroelastic phenomena [4]. On the other hand, the work of Huyghe and co-workers established a FE implementation of a biphasic model [5], which evolved to triphasic and quadriphasic formulations, between 1986 and 2003 [6, 7]. The primary field of application was the cardiovascular tissue [5], but this model became appropriate to describe IVD biomechanical behavior and even to propose new replacements methods for the IVD [6, 7, 8, 9].

2 MATERIALS AND METHODS

2.1 FORMULATION

From the aforesaid it is worth noting the paramount role of poroelasticity on the biomechanics of the IVD. The IVD encloses both solid and fluid phases. Biphasic, triphasic or quadriphasic theories may be applied to model such tissues [6, 7]. Biphasic approaches only consider the influence of solid and fluid parts, while triphasic and quadriphasic theories also include the influence of some ionic fluxes [6, 7, 10]. Most of the IVD FE studies were performed using biphasic formulations, which are an effective choice, as they mean smaller number of constitutive parameters and thus lower complexity of the constitutive modeling [11, 12].

A home-developed FE solver is used. The former version of this solver comprised the most relevant features of biomechanics of soft-tissues, namely their almost incompressibility, the most general isotropic and anisotropic hyperelastic laws, viscoelastic effects. A total Lagrangian formulation with a fully implicit time integration scheme is also adopted [13].

The key novelty of this work is the implementation of a new biphasic poroelastic formulation. This formulation consists of an innovative coupling between the strain energy density potential (eq. 1) adopted by Alves et al. (2010) [13] and Darcy law (eq. 2), which was represented on the formulation proposed by Huyghe (1986) [5] for a biphasic medium:

$$W(\mathbf{C}) = \overline{W}(\overline{\mathbf{C}}, \mathbf{a}_1, \mathbf{a}_2) + \overline{W}_H(J) + Q^0(J) \qquad (1)$$

$$\mathbf{w} = -\mathbf{K}^* \cdot \nabla p^f \qquad (2)$$

On eq. 1, $\overline{W}(\overline{\mathbf{C}}, \mathbf{a}_1, \mathbf{a}_2)$ and $\overline{W}_H(J)$ are, respectively, the isochoric and volumetric strain energy densities, while $Q^0(J)$ has the merit of coupling both displacements and pressure fields. On eq. 2, $\mathbf{w}$ is the flux of the fluid relative to the porous solid matrix, ∇p^f is the gradient of the pore (or fluid) pressure and $\mathbf{K}^*$ is the hydraulic permeability tensor, which is here defined by [14]:

$$\mathbf{K}^* = K_0^* J^M \mathbf{I}, \qquad (3)$$

where K_0^* is the initial permeability, J^M is the strain-dependent term (related to the tissue's deformation, as $J = \det(\mathbf{F})$ and $\mathbf{F}$ is the deformation gradient), and $\mathbf{I}$ is the second order unity tensor [10, 15].

In addition, several studies proved the importance of osmotic swelling behavior to the IVD biomechanics, namely for the height recovery during recovering periods and also for the maintenance of healthy IDP levels [9, 16, 17]. The biphasic osmotic swelling model here implemented is the one adopted by Wilson et al. (2005) [10], in the following form:

$$\boldsymbol{\sigma}_{tot} = -(\mu^f + \Delta\pi)\mathbf{I} + \boldsymbol{\sigma}_s \qquad (4)$$

The total Cauchy stress ($\boldsymbol{\sigma}_{tot}$) results from the contribution of the solid ($\boldsymbol{\sigma}_s$) and fluid phases ($\mu^f + \Delta\pi$), where $\boldsymbol{\sigma}_s$ is the effective solid stress tensor, μ^f is the water chemical potential and $\Delta\pi$ is the osmotic pressure gradient, given by the following expression [10]:

$$\Delta\pi = \phi_{int} RT\left(\sqrt{c_F^2 + 4c_{ext}^2}\right) - 2\phi_{ext} RT c_{ext} \qquad (5)$$

To reach this osmotic swelling model, which is consistent with the adopted biphasic formulation, some assumptions have to be addressed. In fact, the temperature (T), the external salt concentration (c_{ext}) and the osmotic coefficients (ϕ_{int} and ϕ_{ext}) were assumed to be constant. Therefore, the only non-constant in this equation is the fixed charge density (c_F), which can be expressed as a function of the tissue deformation [10]:

$$c_F = c_{F,0} \frac{n_{f,0}}{n_{f,0} - 1 + J}, \quad (6)$$

where $n_{f,0}$ is the initial fluid fraction and $c_{F,0}$ the initial fixed charge density.

2.2 NUMERICAL SIMULATIONS

In order to evaluate the importance of poroelasticity for the IVD biomechanical behavior, a validated 3D full lumbar MS FE model [17] is used in two different numerical tests. The FE model was built with 27-node hexahedra, which are u/p-c finite elements. For the sake of simplicity, in fig.1 only vertex nodes are visualized. The material properties of the five MS components (NP, AF, CEP, cortical bone and trabecular bone) are listed in table 1 (Annex A).

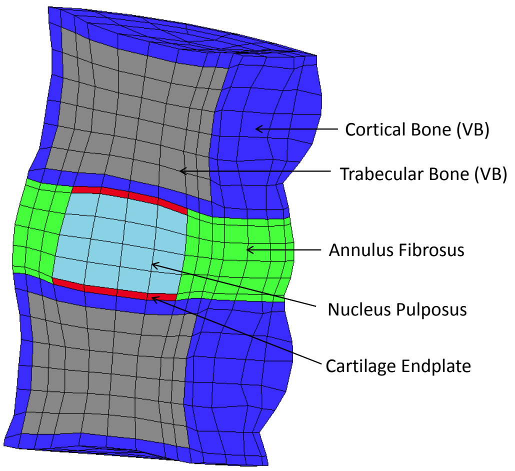

Fig. 1 Anterior-posterior cut of the FE model of a lumbar spine MS

The first test intended to evaluate the response of the MS FE model to three different loading magnitudes, in uniaxial unconfined compression. The magnitudes were 1000, 1500 and 2000N. All the loads were applied during 5min and then held for 15min (creep period). These loads are in the scope of daily activities, from normal walking to more acute efforts, such as carrying or lifting an object [18, 19].

The second test simulated the behavior of the MS through a daily cycle, including both loading and recovery phases. The first 8h of the day corresponded to a 200N load, while the other 16 were replicated by a 500N load, following the protocol of an analogous FE analysis performed by Galbusera and co-workers (2011) [20].

3 RESULTS AND DISCUSSION

The results from the first test are shown in Figs. 2 and 3. Disc height variation (DHV) and intradiscal pressure (IDP) were assessed and compared between each loading level.

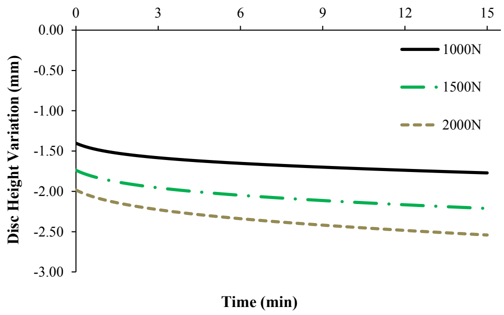

Fig. 2 Creep test during 15min, with different loads applied: Disc Height Variation

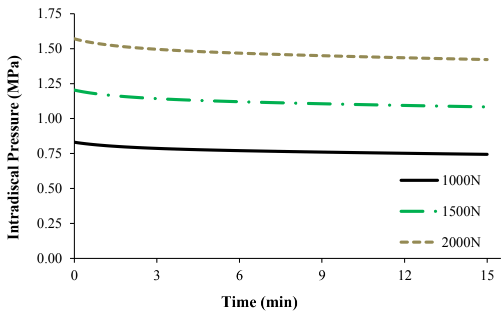

Fig. 3 Creep test during 15min, with different loads applied: Intradiscal Pressure

The results here obtained seem to be in good agreement with the data available from several experimental works. Before the creep period, the DHV correspondent to 1000N is about -1.40mm, -1.74mm for 1500N and -1.98mm for 2000N. These results are inside the range of the studies of Schmidt et al. (2010) and O'Connel et al. (2011) [18, 19]. The creep response seems to be consistent with the work of Heuer et al. (2007) [21]. The measured IDP varies from 0.83MPa to 0.74MPa for 1000N, 1.20MPa to 1.08MPa for 1500N and 1.57 to 1.42MPa for 2000N, during the creep period. These results are in agreement with the work of Schmidt et al. (2010) [18], whose numerical model was also validated by in vivo and in vitro data. The relative decrease of the IDP in 15min is about 10%, which is also consistent with the data available on the literature [21]. The overall behavior of the FE model shows that the IVD response is directly proportional to the increase of the load, as expected. In addition, poroelastic phenomena are noticed on the creep effects, independently of the loading magnitude.

The results from the second test are shown in Figs. 4 to 6. DHV, IDP and osmotic pressure (in the center of the NP) were assessed and compared with the results from Galbusera and co-workers (2011) [20]. Their work included a pre-conditioning period, prior to the 24h test. Therefore, the present simulation was also preceded by the same testing scheme (8h at 200N and 16h at 500N), which first served as a pre-conditioning period.

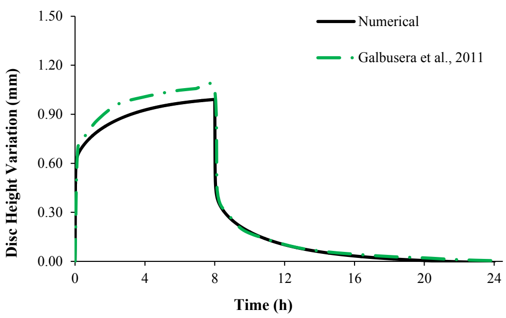

Fig. 4 Full day test: Disc Height Variation

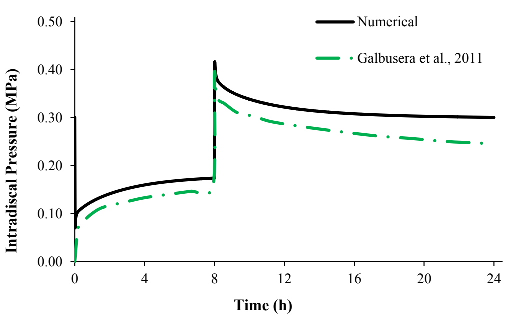

Fig. 5 Full day test: Intradiscal Pressure

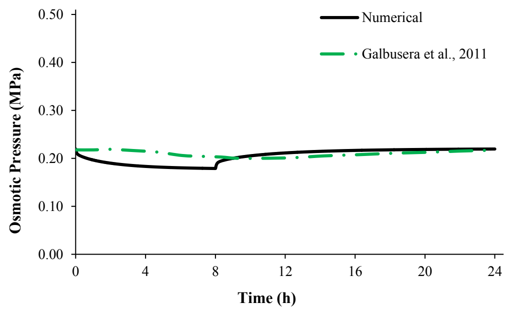

Fig. 6 Full day test: Osmotic Pressure

On the one hand, the differences between the current MS FE model and the model used by Galbusera and co-workers must be taken into account. The geometrical features of the two models are not exactly the same, even if both symbolize the lumbar Human spine. In addition, the pre-conditioning period did not follow the same protocol. Consequently, the comparison between the results of the two simulations is more qualitative than quantitative.

On the other hand, the general behavior of both models is very similar. In fact, only slight differences are noticed in DHV and IDP. However, the osmotic pressure results from the current model seem to be more sensitive to loading than the results from Galbusera and co-workers, even if the biphasic swelling formulation is the same (Wilson's model [10]). This circumstance is probably associated with the original biphasic formulation presented in this work.

4 CONCLUSIONS

From the global results produced, some conclusions may be drawn, even regarding that this is an on-going work. On the one hand, the creep behavior of the IVD, mostly given by its hyper-poro-viscoelastic properties, seems to play a major role on the IVD biomechanics. The acquired data on DHV and IDP supports this finding [19, 21]. On the other hand, the swelling behavior of the IVD seems to be determinant for the reached IDP levels, even taking into account the dissipative effect of the fluid flowing [21]. The slight differences in the comparison with the results from Galbusera and co-works support this remark, as they applied the same biphasic swelling model [10, 21]. Therefore, this in-development biomimetic IVD FE model is an appropriate tool to the description of IVD biomechanical behavior. Further tests will include longer periods of time and different loading conditions, such as flexion or rotation. Other sources of experimental data are also expectable, as well as data from degenerated IVD [19].

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ANNEX A

Table 1 Material properties of the MS components. Multiple data sources were assessed, as stated on each entry of the table. Isotropic (MS ground substances), permeability, anisotropic (AF fibers), viscoelastic and swelling properties were considered.

		NP	**AF**	**CEP**	**TB**	**CB**
Isotropy [22]	C_{10} [MPa]	0.15	0.18	1.00	41.67	3846.15
	C_{01} [MPa]	0.03	0.045	0.00	0.00	0.00
Permeability [14, 23]	K_0^* [$mm^4.N^{-1}.s^{-1}$]	7.5e-4	7.5e-4	7.5e-3	1.0e-1	1.0e-1
	M	8.50	8.50	8.50	18.0	22.0
Anisotropy[1] [24]	$\bar{k}$	-	300.0	-		
	$k_4 = k_6$ [MPa]	-	12.0	-		
Viscoelasticity [9, 25]	a_1	1.7	-	-		
	τ_1 [s]	11.765	-	-		
	a_2	1.2	-	-		
	τ_2 [s]	1.100	-	-		
	a_3	2.0	-	-		
	τ_3 [s]	0.132	-	-		
Swelling [20]	R [$N.mm.mmol^{-1}.K^{-1}$]	8.31450	8.31450	-		
	T [K]	298.0	298.0	-		
	ϕ_{int}	0.83	0.83	-		
	ϕ_{ext}	0.92	0.92	-		
	c_{ext} [$mmol.mm^{-3}$]	0.00015	0.00015	-		
	$c_{F,0}$ [$mmol.mm^{-3}$]	0.00030	0.00018	-		
	$n_{f,0}$	0.80	0.70	-		

[1] For further information on the anisotropic modeling of the AF fibers, the reader is addressed to the work of Cavalcanti and Alves ("…"), also presented in "5º Congresso Nacional de Biomecânica".