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Short communication

Intervertebral disc creep behavior assessment through an open source finite element solver

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ABSTRACT

Degenerative Disc Disease (DDD) is one of the largest health problems faced worldwide, based on lost working time and associated costs. By means of this motivation, this work aims 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, as it is a highly complex multiphasic medium. Poroelastic models of the disc are mostly implemented in commercial finite element packages with limited access to the algorithms. Therefore, a novel poroelastic formulation implemented on a home-developed open source FE solver is briefly addressed throughout this paper. The combination of this formulation with biphasic osmotic swelling behavior is also taken into account. Numerical simulations were devoted to the analysis of the non-degenerated human lumbar IVD time-dependent behavior. The results of the tests performed for creep assessment were inside the scope of the experimental data, with a remarkable improvement of the numerical accuracy when compared with previously published results obtained with ABAQUS®. In brief, this in-development open-source FE solver was validated with literature experimental data and aims to be a valuable tool to study the IVD biomechanics and DDD mechanisms.

1. Introduction

The intervertebral discs (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. IVDs are composed of three fundamental structures: the nucleus pulposus (NP), the annulus fibrosus (AF) and the cartilaginous endplate (CEP) (Raj, 2008). Creep behavior and stress-relaxation effects have been exhaustively described in the literature, as the IVD has a rate-dependent behavior, grounded on poroelastic and viscoelastic effects (Ellingson and Nuckley, 2012; Heuer et al., 2007; Nerurkar et al., 2010). The motivation for this work comes from the fact that spine problems are a major cause of disability in western societies, with a special emphasis on Degenerative Disc Disease (Adams and Dolan, 2012).

The key novelty of this work is a new constitutive modeling and FE implementation in a home-developed open-source software of a biphasic poroelastic formulation applied to the IVD, coupled with strain-dependent osmotic swelling behavior (Wilson et al., 2005). The adoption of a fully home-developed FE solver offers major advantages over programming in commercial software packages, because the drawbacks associated with the rigidity of a proprietary commercial code hamper the freedom of the researcher, when the complexity of the model increases. Given that the researcher has direct view on the source code, the verifiability of the software is also taken to a higher level. The main tasks to be fulfilled are to briefly demonstrate the implementation of this formulation and to compare the behavior of the open source IVD FE model with experimental data from Heuer et al. (2007), O'Connell et al. (2011) and Wilke et al. (1999), as well as numerical data from ABAQUS® software (Schroeder et al., 2010), through creep assessment tests.

2. Materials and Methods

2.1. Formulation

This work focuses on the evaluation of a novel poroelastic biphasic model, which is implemented on a home-developed open source FE solver. The almost incompressible nature of biphasic soft-tissues is taken into account by assuming a multiplicative decomposition of the deformation gradient into volumetric and isochoric parts, and by using a mixed interpolation of the displacement and pressure fields. The strain energy density potential $W(\mathbf{C})$ is the one adopted by Alves et al. (2010):

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

where $\overline{W}(\overline{\mathbf{C}}, \mathbf{a}_1, \mathbf{a}_2)$ and $\overline{W}_H(J)$ are, respectively, the isochoric and volumetric strain energy densities, while the additional term Q^0 has the merit of coupling the mixed

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formulation, i.e., displacements and pressure fields (Alves et al., 2010; Sussman and Bathe, 1987):

$$\overline{W}(\overline{\mathbf{C}},\mathbf{a}_1,\mathbf{a}_2)=\overline{W}_{iso}(\overline{\mathbf{C}})+\overline{W}_{aniso}(\overline{\mathbf{C}},\mathbf{a}_1,\mathbf{a}_2) \quad (2)$$

$$\overline{W}_H(J)=+\frac{\lambda_k}{2}(J-1)^2 \quad (3)$$

$$Q^0(J)=-\frac{1}{2\lambda_k}(\overline{p}-\tilde{p})^2 \quad (4)$$

The contribution of the isochoric strain energy density can be divided between isotropic and anisotropic parts, as stated in Eq.(2), as a function of the adopted hyperelastic constitutive model. The Mooney–Rivlin model is applied here for the isotropic part, which comprises the matrixes of the NP and the AF, along with the CEP (Bonet and Wood, 1997; Schmidt et al., 2007). The Holzapfel model is applied for the anisotropic contribution of the AF fibers (Holzapfel et al., 2005). About Eqs. (1)–(4), $\mathbf{C}=\mathbf{F}^T\mathbf{F}$ is the right Cauchy–Green strain tensor, $\mathbf{F}$ is the deformation gradient, $J=\det(\mathbf{F})$ and $\mathbf{a}_1,\mathbf{a}_2$ are unit vectors representing fiber directions. λ_k is a penalty parameter playing the role of a bulk modulus, $\overline{p}$ is the pressure computed from the (unknown) displacement fields and $\tilde{p}$ is the pressure interpolated from the (unknown) pressure field (Alves et al., 2010; Castro et al., 2013).

In the macroscopic point of view, the biphasic medium is assumed to be homogeneous and continuous, and characterized simply by the initial fluid (f) and solid (s) volume fractions, n_α, with $\alpha=\{f,s\}$ and $\sum n_\alpha=1$. The fluid is assumed to completely fill the pores of the solid phase, by flowing through it. The fluid flux relative to the solid matrix is commonly modeled by the Darcy's law (Ehlers et al., 2009; Wilson et al., 2005), written as:

$$\mathbf{w}=n_f(\mathbf{v}_f-\mathbf{v}_s)=-\mathbf{K}^*\cdot\nabla p^f, \quad (5)$$

where $\mathbf{w}$ is the flux of the fluid relative to the solid matrix, and n_f is the current fluid fraction, $(\mathbf{v}_f-\mathbf{v}_s)$ is the relative velocity of the fluid with respect to the solid matrix. $\mathbf{K}^*$ is the hydraulic permeability tensor which, in case of isotropic permeability, is simply defined as $\mathbf{K}^*=K^*(J)\ \mathbf{I}$. $K^*(J)$ is the strain-dependent permeability, $\mathbf{I}$ the second order unity tensor and ∇p^f is the gradient of the pore (or fluid) pressure (Ehlers et al., 2009; Wilson et al., 2006, 2005). The permeability is strain-dependent and $M>0$ is the unique material parameter, accordingly to (van der Voet, 1997):

$$K^*(J)=K_0^*J^M \quad (6)$$

The biphasic formulation here adopted consists of an innovative implementation that couples the strain energy density potential shown in Eq. 1 with the Darcy law (Eq. 5), embedded on the formulation proposed by Huyghe (1986) for a biphasic medium. In the authors' best knowledge, no other biphasic formulation with the aforesaid coupling has been yet published. Validation of this formulation was previously reported (Castro et al., 2013). After some mathematical developments, the new elemental stiffness matrix and the corresponding elemental system of equations is written as:

$$\begin{bmatrix}\mathbf{K}_{UU} & \mathbf{K}_{UP}\\ \mathbf{K}_{PU} & K_{PP}-K_{K^*}\end{bmatrix}:\begin{bmatrix}\Delta\mathbf{u}\\ \Delta p\end{bmatrix}=\begin{bmatrix}\mathbf{R}\\ U+T_1\end{bmatrix}-\begin{bmatrix}\mathbf{F}_U\\ 0\end{bmatrix}, \quad (7)$$

where indexes U,P are associated with the unknown displacement ($\mathbf{u}$) and pressure fields (p), respectively, and $\mathbf{R}$ is the vector of the external forces. The most relevant terms for the biphasic behavior are defined as follows (Huyghe, 1986; van Loon et al., 2003):

$$K_{K^*}=\int_{\Omega_0}\nabla_0\mathbf{\Psi}^T\cdot K^*\cdot\nabla_0\mathbf{\Psi}\cdot\Delta t d\Omega_0 \quad (8)$$

$$U=\int_{\Omega_0}\mathbf{\Psi}(J-J_n)d\Omega_0 \quad (9)$$

$$T_1=\int_{\Omega_0}(K^*\cdot\nabla_0\tilde{p})\nabla\mathbf{\Psi}\cdot\Delta t d\Omega_0 \quad (10)$$

Indexes n and 0 refer to the beginning of the time increment ($t=t_n$) and to the reference configuration ($t=0$), respectively. The term U acts as quantifier of the volume variation due to fluid flowing in time increment $[t_n,t_n+\Delta t]$. $\mathbf{\Psi}$ are the shape functions for pressure field interpolation.

Several studies have shown the importance of osmotic swelling behavior to the IVD biomechanics, namely for the height recovery during rest periods and for the maintenance of healthy IDP levels (Riches et al., 2002; Huyghe et al., 2003; Schroeder et al., 2010). The biphasic osmotic swelling model implemented is the one adopted by Wilson et al. (2005), in the following form:

$$\boldsymbol{\sigma}_{tot}=-(\mu^f+\Delta\pi)\mathbf{I}+\boldsymbol{\sigma}_s, \quad (11)$$

where the total Cauchy stress ($\boldsymbol{\sigma}_{tot}$) results from the contribution of the solid ($\boldsymbol{\sigma}_s$) and fluid phases $(\mu^f+\Delta\pi)$,$\boldsymbol{\sigma}_s$ is the effective solid stress tensor, $\mu^f$is the water chemical potential and $\Delta\pi$is the osmotic pressure gradient, as defined by Wilson et al. (2005):

$$\Delta\pi=\pi_{int}-\pi_{ext}=\phi_{int}RT\left(\sqrt{c_F^2+4c_{ext}^2}\right)-2\phi_{ext}RTc_{ext} \quad (12)$$

Temperature (T), external salt concentration (c_{ext}) and osmotic coefficients (ϕ_{int} and ϕ_{ext}) were assumed to be constants. The ionic concentrations in all areas of the disc are assumed in equilibrium with the external salt concentration c_{ext} at all times. Hence, ion flow is assumed to be infinitely fast compared to fluid flow. Otherwise, one would have to move for tri- or quadriphasic formulations (Frijns et al., 1997). The fixed charge density (c_F) is strain-dependent, and thus can be expressed as a function of the tissue's deformation (Wilson et al., 2005):

$$c_F=c_{F,0}\frac{n_{f,0}}{n_{f,0}-1+J} \quad (13)$$

where $n_{f,0}$ is the initial fluid volume fraction and $c_{F,0}$ the initial fixed charge density. Swelling behavior was considered for both NP and AF (Galbusera et al., 2011).

The initial constitutive parameters of the MS components are shown on Table 1. These parameters were based on literature data and numerical optimization. The components of the VB, trabecular bone (TB) and cortical bone (CB), were considered to be highly permeable. Viscoelastic behavior was considered for NP matrix, through Maxwell's rheological model, for which τ_n is the damper characteristic time and $a_n=E_n/E_0$ is the ratio between the viscoelastic spring E_n introduced by the Maxwell element and the elastic time-independent behavior E_0 (Ehlers et al., 2009; Iatridis et al., 1997; Kaliske et al., 2001). However, while viscoelasticity is more important to model fast events (a few seconds), poroelasticity is far more important to describe long-term phenomena like creep and/or stress relaxation (minutes to hours) (Schmidt et al., 2011). This means that viscoelasticity has a little influence on the tests performed in this work.

2.2. FE model

Fig.1 shows a section of the FE mesh used in this study, visualized on the GiD® 7.4.6b pre- and post-processing interface. The development of this model was based on a model of a Human VB and two IVDs previously published by Smit (1996). It comprises a full MS, including the most relevant features of one IVD and two VBs. The IVD has an average height of 12.8 mm and an axial cross section of 1555.3 mm^2, while the full MS has an average height of 60.9 mm. The FE mesh is discretized with 1892 quadratic 27-node hexahedron FE and 16425 nodes. In Fig.1, all the external nodes are visualized. The FE mesh convergence study permits to state that the FE mesh refinement used allows a proper convergence of the reported results; moreover, such demonstration is out of the scope of this publication.

AF fibers' mechanical properties are assumed to evolve linearly through the axial plane, both in radial and circumferential directions (Eberlein et al., 2001). In addition, fiber angle varies from $\pm 23.2°$ at ventral position to $\pm 46.6°$ at dorsal position, based on the work of Holzapfel et al. (2005).

2.3. Numerical simulations

In order to evaluate and validate this new version of the open-source FE solver, two creep behavior tests of non-degenerated human IVD were performed. For both tests, free fluid flow is allowed between the MS components and also on the MS external boundaries. In the first set of tests, 500 N were applied on the top VB during 5 min (slow loading to allow proper stabilization of the model) and then held for 15 min (as experimentally performed by Heuer et al. (2007)). The load was applied through the vertical axis of the MS model, with the bottom VB fully constrained. Lateral and sagittal movements were allowed. The importance of the osmotic swelling behavior for the IVD biomechanics was evaluated, as this test was performed with three different swelling conditions: without coupling the osmotic swelling behavior ("No Swelling"), with a pre-conditioning period of 8 h of free swelling ("Pre-Swelling") and also with osmotic swelling, but without that pre-conditioning period ("No Pre-Swelling"). The pre-conditioning periods are strictly related with the recovery times, which usually last between 1 and 8 h until swelling equilibrium (Galbusera et al., 2011). These numerical results were also compared with the outcomes of the work of Schroeder et al. (2010), which applied the same strain-dependent osmotic swelling model on ABAQUS®. Their simulations also considered similar boundary conditions and pre-conditioning period (until equilibrium).

For the second test, three stages were considered: i) a short pre-conditioning period (1 h), ii) a loading period of 2000 N at 1 N/s (in agreement with the experimental test of O'Connell et al. (2011)) and, finally, iii) a creep stage (1 h).

3. Results

Disc height variation (DHV) and intradiscal pressure (IDP, i.e., the fluid pressure inside the NP) results from literature are compared with the numerical outcomes of the current model, considering the different swelling modes (Fig.2). At the end of the 15 min, the DHV is −1.23 mm and the IDP is 0.42 MPa, after a free swelling period of 8 h ("Pre-Swelling"). When the osmotic