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A study of biphasic models to represent the viscous mechanical behavior of tendons

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Abstract

*In this work is presented a study ofthecapabilityto represent the viscous mechanical behavior of tendonsusingtwo biphasic models. Poroelastic and poroviscoelastic models were implemented in to a finite element framework, and numerical tests were performed. Also,it is presenteda numerical-experimental characterization of the biphasic models, using transversal isotropy for the solid phase.The results shownthat the PE model is highly dependent on the Poisson's ratio of the solid matrix, and it is capable of reproducing viscous effects only when high compressibility of the tissue arises. Both models can reproduceexperimental results of a cyclic test using a porcine flexor tendon specimen submerged in water, where the poroviscoelastic model performed better due to the inclusion of the viscosity on the solid phase. However, the poroelastic model was capable of reproducing the experimental observation only when itis assumed Poisson's ratio close to zero, which could represent a physical violation of the incompressibility assumption commonly used in literature.**DOI**:https:/doi.org/10.24243/JMEB/3.5.207*

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1 Introduction

Numerical simulation of biological tissues has attracted the interest of researchers due to the possibility of predicting the mechanical response of complex biomechanical structures and systems. However,the simulation of these materials presentsmajor difficulties when compared to traditionalengineering materialssince they present different inelastic and nonlinear phenomena in their natural range of operation **[1]**. For instance, it is well-known thatbiological tissues have a significant dependence on loading rate**[2]**and hydration**[3]**,besides the non-linearity and anisotropy elastic behavior**[1]**, **[4]**.Among these tissues, it is possible to highlight the tendon, which attaches muscle to bone or more complex structures.Tendons are important biological tissuesfor the human body that may besubject to high loads and several types of injuries. Therefore, understanding their mechanical behavior, and the best way to numerically simulate their response, could assist engineers and medical professionals to develop of implant devices, tendon reconstruction techniques, treatment and prevention of injuries**[5], [6]**.

The tendon belongs to the category of connective tissue. This type of tissue presents as basic composition: cells, solid extracellular matrix and interstitial fluid **[7]**. It is ahighly hydrated structure that shows a complex mechanical behavior. However, despite the significant presence of fluid, most works found in literaturestill use classic

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viscoelastic models associating the viscosity only to the solid matter [4], [8], [9]. Aninteresting constitutive familythatis capable toincorporatefluidmechanical behavior into the response of solid materialsare the biphasic models.

Biphasic poroelastic (PE) models were initially developed for the study of soil mechanics [10], being later employed in the constitutive modeling of hydrated tissues[11]-[13]. Linear biphasic models havebeen widely used for the analysis of hydrated tissues in the region of physiological behavior (linear, preconditioned) [14], [15].Despite the simplicity of this model, which presents a small number of parameters, it includes a fluid phaseinto the mechanical response.Then, the desirable viscous response is obtained from the fluid phase while the solid phase is modeledwith a simple linear elastic model.Models incorporating viscosity in the solid phase were proposed in the literature to extend the capability to represent viscous effects into this constitutive framework[16], [17]. These models, denominated poroviscoelastic(PVE) models, may improve the capacity to represent the mechanical behavior of hydrated biological tissues,however, the number of material parameters increase significantly.

Thus, the objective of this work is to study the application of poroelastic and poroviscoelastic biphasic models to reproduce the behavior of tendons in the linear physiological region of operation(when fibers are aligned with theirlongitudinalaxis). Also, the capability to represent the mechanical behavior of a real tendon submitted to anexperimental cyclic tensile test is investigated.

2 Biphasic Models

In the biphasic theory, the effects of viscosity are included through a diffusive body force proportional to the relative velocity between the fluid and the solid, allowing the fluid exudation of the tissue [14], [15]. In the PE model, it is assumed that permeability and solid content are independent of deformation and the inertial effects are negligible. The equilibrium equations of linear momentum for the solid and liquid phase are, respectively,

$$\nabla . \boldsymbol{\sigma}^{s} + \mathbf{b}^{s} = 0;\ \nabla . \boldsymbol{\sigma}^{f} + \mathbf{b}^{f} = 0 \tag{1}$$

where$\boldsymbol{\sigma}^{\mathrm{f}}$and $\boldsymbol{\sigma}^{\mathrm{s}}$are, respectively, the tension of the fluid and solid phase, and $\mathbf{b}^{\mathrm{f}}$and$\mathbf{b}^{\mathrm{s}}$are, respectively, the body force of the fluid and solid phase. The linear constitutive equations for the solid and fluid phase are, respectively,

$$\boldsymbol{\sigma}^{\mathrm{s}} = -\boldsymbol{\phi}^{s} p\mathbf{I} + \boldsymbol{\lambda}^{s} e^{s}\mathbf{I} + 2\boldsymbol{\mu}^{s}\boldsymbol{\varepsilon}^{\mathbf{s}};\ \boldsymbol{\sigma}^{\mathrm{f}} = -\boldsymbol{\phi}^{f} p\mathbf{I} \tag{2}$$

where p is the pressure, $\mathbf{I}$is the second-order identity tensor, λ^{s} and μ^{s} are the Lamè constants, $\boldsymbol{\varepsilon}^{s}$ is the infinitesimal deformation tensor of the solid phase, and e^{s} is correspond to the volumetric part of $\boldsymbol{\varepsilon}^{s}$, ϕ^{s} is the solidity ratio (ratio of solid volume and total volume)and ϕ^{f} is the porosity ratio (ratio of fluid volume and total volume), such as $\phi^{s} + \phi^{f} = 1$ The equilibrium between phases is given bythe momentum exchange that couples the problem, using a term proportional to the relative velocity between them, i.e.:

$$\mathbf{b}^{s} = -\mathbf{b}^{f} = \frac{\left(\phi^{f}\right)^{2}}{k}\left(\boldsymbol{v}^{f} - \boldsymbol{v}^{s}\right) \tag{3}$$

Where k is the permeability, $\boldsymbol{v}^{f}$the fluid phase velocity, and$\boldsymbol{v}^{s}$the solid phase velocity. In addition to the momentum equilibrium equations, the continuity equation (conservation of mass) must also be satisfied.

$$\boldsymbol{\nabla}.\left(\phi^{s}\mathbf{v}^{s} + \phi^{f}\mathbf{v}^{\mathrm{f}}\right) = 0 \tag{4}$$

where the boundary conditions are solid displacements $\mathbf{u}^{s}$, fluid velocity $\mathbf{v}^{f}$, solid traction force $\mathbf{t}^{s}$ and pressure p. The number of material parameter for a PE model is four, where two of them are the elastic solid parameters, such as the Lamè constants, or any equivalent pair of elastic parameters, as Elastic modulus E and Poisson's ratio υ. Detailed information about this model can be found in [14], [15]. The poroviscoelastic models can be achieved using a classical viscoelastic relationship into the solid elastic model [18]:

$$\boldsymbol{\sigma}^s = \lambda^s \int_{-\infty}^{t} G(t-\tau) \frac{\partial}{\partial \tau} tr(\boldsymbol{\varepsilon}^s)\mathbf{I} d\tau + 2\mu^s \int_{-\infty}^{t} G(t-\tau) \frac{\partial(\boldsymbol{\varepsilon}^s)}{\partial \tau} d\tau \quad (5)$$

where$G(t)$is the relaxation function. This model has more three parameters that correspond to two discrete relaxation times (τ_1, τ_2) and the spectrum magnitude$G$. More details about this model can be obtained in [19].

3 Methodology

In this study, axisymmetric quadrilateral elements with 4-node were implemented intoa homemade finite element code to simulate the PE and PVE models. Crank-Nicholson trapezoidal scheme was used for the solution into time domain **[20]**. Further details on the system of equations for a finite element implementation can be found at **[14], [19]**. Besides the classic linear isotropic elasticity, in this work,a constitutive matrix for axisymmetric transversal isotropy is also implemented to allow to perform a more realistic investigation of the linear behavior of a tendon, which is anisotropic **[21]**.

Since PE models still are commonly used to simulate biological tissue, first is presented a case of study to understand the capacity of the PE model to simulate viscous phenomena of a typical relaxation test, when volumetric parameters are changed. Then, a numerical-experimental characterization is performed to understand the capability of the studied models to represent real data obtained from an experimental test.

4 Results and Discussion

4.1 Study of PE model into a compressive relaxation test

Geometry used in thiscase of study was a cylinderwith a diameter of 3 mm and height of 1.22 mm. It was applied boundary conditionsconsistent with a non-confined compressive relaxation test, with a linear increaseof prescribed displacements until obtaining a compressive deformation of 10% at the time15 s. Then, the displacement was kept constant until the time instant 300 s. Numeric relaxation tests were performed varying the Poisson's ratioto investigate volumetric changes. The other parameters for this study was E = 0.46 MPa, k = 0.76E-14 m^4/N.s and ϕ^s = 0.2.

The resultsof the reaction force and relative velocity over time, for different values of the Poisson's ratio,are shown in *Fig.1*. For the relative velocities, it is shown the results of the node withthe highest velocity value at the initial time. The results showed that the Poisson's ratio presenthigh sensitivity to relaxation, and, consequently, to viscous effects. Also, it should be noted that when Poisson's ratio is about 0.5, no significant viscous effects are obtained.

Since the tendon is usually considered incompressible due to the high quantity of fluid **[9], [10], [22]**, this physical observation could represent a limitation in the use of this modelto simulate viscous phenomenon. In other words, to numerically obtain significant viscous effects observed in biological tissues, the Poisson's ratio should be the lowest possible value, which may represent a violation of the incompressibility hypotheses usually assumed for this material. Low values of Poisson's ratio are expected to soil material but not for biological tissues.

4.2 Numerical-experimental characterization with PE and PVE models

An experimental test was performed on a flexor tendon from a swine paw to study the response of the models in a more realistic situation. After dissection and sample collection, a specimen was submitted to a cyclic test into a TA ElectroForce 3200 testing machine. After the preparation of the sample, the test was carried out with the tendon submerged in water at the controlled temperature of 32ºC, seeking to reproduce *in vivo* conditions, and with boundary conditions of a case of non-confined porous materials, similarly to the real condition. *Figure 2* shows the specimen prepared and attached to the tensile test machine.

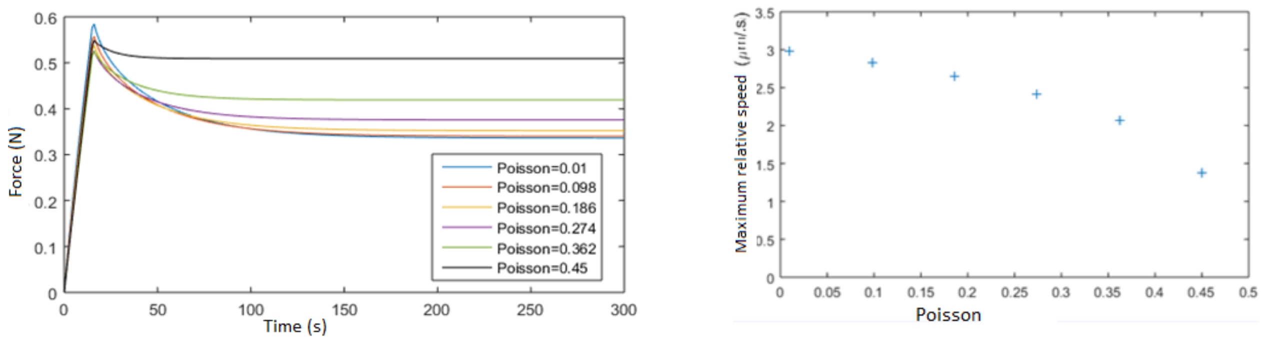

Fig.1Reaction force over time (left) and maximum relative velocities (right) for different Poisson's ratio.

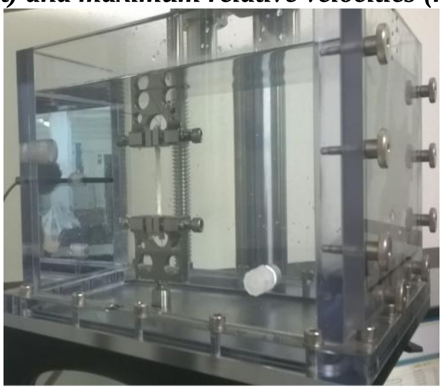

Fig.2 Sample attached to the test machine

Before the testing, preconditioning of the sample was performed. A 50 N preload was imposed for 100 s in order to reach the linear region of the tendon (aligned fibers) **[23]**, seeking consistency with the hypothesis of the implemented model (linearity). After preconditioning, the load is carried to 100 N,and two cycles of loading-unloading are carried out for 10 seconds, with a maximum force peak of 180 N. For this case of study, transversal isotropy was used in the PE and PVE modelsusingan axisymmetric transversal isotropy constitutive matrix. Constitutive characterizationof the material parameters was performed using the least squares optimization technique. Displacements obtained in the test were imposed in the numerical model, and the adjustments of the parameters were performed by minimizing the quadratic difference between the numerical and experimental force. Parameters obtained for the PE and PVE models are presented in *Table 1*. The comparison of experimental and the numerical force, resulting from the fitting procedure, are presented in *Fig.3*.

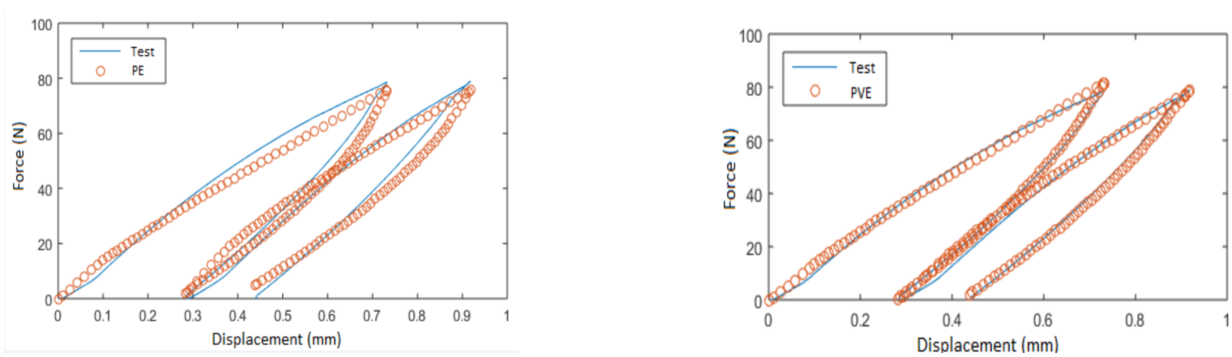

Fig.3 Fitting result for the PE model (left) and PVE model (right)

These results showed that the PVE models presented better results than the PE model. Also, should be noted that the PE model needs very low Poisson's ratios to be capable of reproducing the experimental force, which indicates that the results are violating physical observations, despite the good force representation. In order to investigate this phenomenon, transversal measurements should be obtained to study the volumetric change during the experimental test.

Table 1. Material parameter for the PE and PVE models

PE		PVE	
κ	0.3901e-14 [m^4 / Ns]	κ	2.363e-12 [m^4 / Ns]
*	*	$\overset{\circ}{G}$	0.750
*	*	τ_1	2.065 [s]
*	*	τ_2	2.221
E_1	4.102 [MPa]	E_1	0.320 [MPa]
E_3	0.205 [MPa]	E_3	3.846 [MPa]
ν_{21}	0.910	ν_{21}	0.269
ν_{13}	0.001	ν_{13}	0.247
G_{31}	2.567 [MPa]	G_{31}	3.392 [MPa]

* Parameters not used in the PE model

5 Conclusions

This paper presented a study of the application of the poroelastic and poroviscoelastic biphasic models for the characterization of the mechanical behavior of the tendon, as an alternative to classic viscoelasticity models that consider only the solid phase. The inclusion of a fluid phase with the poroelastic model allowed the representation of viscous behavior, even considering the solid phase as linear elastic. This model has a small number of parameters but presents significant viscous behavior only when it is far from the incompressibility condition (Poisson's ratio close to 0.5).This observation may not be consistent with application to biological tissues, where incompressibility is usually assumed. In the poroviscoelastic model the addition of viscosity in the solid phase, analogously to the classical models of solid viscoelasticity, brought more flexibility to the representation of viscous behavior, at the cost of more material parameters. In the study of the representation of the mechanical behavior of an experimental test, the parameter adjustment demonstrated the superiority of the PVE model in the mechanical representation of tendons, when compared to the PE model. The PE model was capable of reproducing the experimental observation only when it assumes Poisson's ratio values close to zero, which represent a physical violation of the incompressibility assumption used for this material. Finally, even if biphasic models have shown good results for the studied conditions, more experimental data and analyses are needed to validate the models.

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