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Hyperelastic modeling of the human brain tissue: Effects of no-slip boundary condition and compressibility on the uniaxial deformation

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A B S T R A C T

Being extremely soft, brain tissue is among the most challenging materials to be mechanically quantified. Despite recent advances in mechanical testing of ultra-soft matters, there still exists a need for robust procedures to analyze their behavior at large deformation. In this paper, it is shown how failing to taking into account the precise boundary conditions can result in substantial variation from the "assumed" ideal behavior, even for the case of simple loading conditions such as the uniaxial mode. For an accurate analysis, the mathematical modeling is combined with the finite element simulation to interpret the mechanical behavior of the brain tissue based on the comprehensive experiments conducted by Budday et al. (2017). It is demonstrated herein that only an Ogden hyperelastic model with both negative and positive nonlinearity constants can predict the mechanical behavior of the brain tissue in tension and compression, and the tension-compression asymmetry might arise from the difference in compressibility behavior in tension and compression. This hypothesis is utilized for modeling the mechanical behavior of the brain tissue in uniaxial loading condition and exhibits excellent agreement with the experiments. This study also provides a comprehensive explanation for nonlinear analysis of soft matters, in general, and the brain tissue, in particular, with thoroughly describing the concept of hyperelasticity and modeling incompressible or compressible behaviors utilizing the Ogden strain energy function.

1. Introduction

Mechanical interaction of the human body with its surrounding environment is known to be a major factor in its heath or disease conditions. Due to the hierarchical nature of the living matters, their mechanical investigation entails a multiscale analysis that expands a wide range of length scales from the subcellular to tissue levels (Cloots et al., 2013; Cowin and Doty, 2007; Fung, 2013; Humphrey and O'Rourke, 2015; Mofrad and Kamm, 2006). Such analyses can be performed via physical/mechanical modeling that give valuable insight into the mechanisms involved in the biomechanics of the tissue deformation and provide predictive patterns that aim at reducing the injuries in traumatic conditions and increasing the remodeling rate during healing (Holzapfel and Ogden, 2017). A proper implementation of such models, however, requires a thorough understanding of the mechanical properties of the tissue and the constitutive relations that govern its deformation behavior.

Among human body organs, brain is arguably the most vulnerable unit during mechanically induced trauma (Ahmadzadeh et al., 2014; Faul et al., 2010; Giordano et al., 2014; Prabhu et al., 2011). In addition, some aspects of its growth and folding processes have been recently shown to be mechanically driven (Bayly et al., 2014; Kuhl, 2016; Tallinen et al., 2016). Nevertheless, the extremely soft nature of this tissue has made its mechanical testing a challenging task. Mechanical quantification of the brain tissue was started more than half a century ago, however, the results for mechanical stiffness of this tissue based on earlier studies are very scattered within a range of several orders of magnitude (Chatelin et al., 2010). With recent developments in experimental techniques and increase in accuracy and rate of data acquisition of the instruments, it seems that the results based on different studies with different testing methods and procedures are demonstrating a better agreement with one another (Budday et al., 2015, 2017; Franceschini et al., 2006; Huston III, 2014; Johnson et al., 2013; MacManus et al., 2015; Miller and Chinzei, 2002; Moran et al., 2014; Rashid et al., 2012, 2013, 2014; Samadi-Dooki et al., 2017, 2018; Van Dommelen et al., 2010; Weickenmeier et al., 2016). Accordingly, the constitutive models that relate the deformation of the tissue to the force can be confidently calibrated using the recent experimentally obtained data.

Brain tissue exhibits a nonlinear mechanical behavior with notable rate and regional dependency. To address its nonlinearity, hyperelastic constitutive laws have been extensively used for this tissue (Budday

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Table 1
Ratio of volume change ($\frac{V^f}{V^i}$) in terms of Poisson's ratio and the longitudinal stretch for uniaxial deformation of a cuboidal sample with four different measures of strain.

Strain measure	Deformed height	Deformed side length	Volume ratio $\frac{V^f}{V^i}$
Biot	$l_1(1+E_1^{(1)})$	$l_2(1-\nu E_1^{(1)})$	$\lambda_1(1+\nu(1-\lambda_1))^2$
Swainger	$\frac{l_1}{1-E_1^{(-1)}}$	$\frac{l_2}{1+\nu E_1^{(-1)}}$	$\frac{\lambda_1^3}{(\lambda_1+\nu(\lambda_1-1))^2}$
Hencky	$l_1 exp(E_1^{(0)})$	$l_2 exp(-\nu E_1^{(0)})$	$\lambda_1^{1-2\nu}$
Bažant ($m=1$)	$l_1\left[\Pi_1^{(1)}+\sqrt{1+(\Pi_1^{(1)})^2}\right]$	$l_2\left[-\nu\Pi_1^{(1)}+\sqrt{1+(\nu\Pi_1^{(1)})^2}\right]$	$\frac{1}{8\lambda_1^3}\left(-1+\lambda_1^2+\lambda_1\sqrt{2+\frac{1}{\lambda_1^2}+\lambda_1^2}\right)\left(\nu+\lambda_1\left(-\nu\lambda_1+\sqrt{4-2\nu^2+\frac{\nu^2(1+\lambda_1^4)}{\lambda_1^2}}\right)\right)^2$

et al., 2017; Feng et al., 2013; Giordano et al., 2014; Kaster et al., 2011; Mihai et al., 2017, 2015; Moran et al., 2014). In this way, the stress-strain relations can be obtained by taking partial derivatives of the strain energy function which is assigned to the tissue behavior. Among different hyperelastic energy functions, the Ogden model has been shown to appropriately predict the nonlinear behavior of soft tissues including the brain (Budday et al., 2017; Mihai et al., 2017, 2015; Ogden, 1997). The generalized energy function of the Ogden hyperelastic material may be presented as:[1]

$$\Psi_{Ogd}=\sum_{j=1}^{N}\left[\frac{2\mu_j}{m_j}I_1\left(\boldsymbol{E}^{(m_j)}\right)+\frac{1}{D_j}(J-1)^{2j}\right] \tag{1}$$

in which μ_j, m_j, and D_j are the material parameters pertaining to the shear modulus, degree of nonlinearity, and compressibility, respectively, J is the determinant of the deformation gradient, and $I_1(\boldsymbol{E}^{(m_j)})$ represents the first invariant of the m_jth order Seth-Hill strain tensor $\boldsymbol{E}^{(m_j)}$. The parameter N in this equation determines the number of terms required to appropriately fit the material behavior. Obviously, the Ogden hyperelastic energy function depends on the form of the strain that is considered for the material behavior. This form depends substantially on the magnitude and sign of the nonlinearity parameter m_j, and as will be discussed later in this paper, can control the tension-compression asymmetry of the material. More importantly, for a hyperelastic energy function to be physically meaningful in general, there are some mathematical criteria that need to be satisfied as described by Attard and coworkers (Attard, 2003; Attard and Hunt, 2004). The most important conditions which are relevant to the analysis of soft tissues include:

1. The energy function cannot attain a negative value for any deformation.
2. At the undeformed state (zero principal strains), the strain energy function must possess a zero value, which according to condition 1, is the minimum value as well.
3. At singularities (zero or very large principal stretches), the strain energy function should approach positive infinity.
4. The stresses derived from such energy functions must approach negative and positive infinity for deformation with zero or very large principal stretches, respectively.

These conditions have been elaborately discussed in a recent publication by Moerman et al. (2016) in which it has been described how selecting a certain form of the Seth-Hill class of strains can affect the validity and appropriateness of the resulting strain energy function. Accordingly, the authors of that article came up with a hybrid form of the strain tensor that satisfies the aforementioned criteria and allows a better control over tension-compression asymmetry without altering the nonlinearity degree.

The last term in the right-hand-side of Eq. (1) pertains to the compressibility of the material. In most of the publications on the mechanical analysis of the brain, this tissue has been considered to be incompressible (Franceschini et al., 2006; Laksari et al., 2012; Mihai et al., 2017). Accordingly, this term is usually eliminated from the mathematical formulation for Ogden hyperelastic modeling of the brain tissue. Since the brain tissue possesses a considerable interstitial and intracellular fluid content (up to 0.8 g/ml (Whittall et al., 1997)), the incompressibility condition seems to be reasonable, especially in compression. However, recent numerical analysis of the hyperelasticity of the brain tissue has shown that this assumption is not necessarily correct. For example, Moran et al. (2014) calibrated different hyperelastic models based on the experiments performed by Jin et al. (2013). Although Moran et al. (2014) did not discuss the concept of compressibility in their publication, the volume change during deformation can be readily investigated by rebuilding the same 3D object in ABAQUS (with the same boundary conditions, number and type of elements, and loading condition) and incorporating the numerical values for the Ogden hyperelastic model as presented in Table 1 of that publication. The variation of the volume ratio based on this analysis is demonstrated in Fig. 1 which shows considerable changes for both tensile and compressive loading conditions for all regions of the brain. Although this level of volume changes seems unrealistic (especially in compression), the results indicate that the incompressibility assumption for the brain might need to be revisited.

Another important consideration in model calibration is to appropriately consider the boundary conditions of the experimental setup and their conformity with the modeling assumptions. For uniaxial experiments on the brain tissue, the size of the samples excised from the tissue is usually small with aspect ratios close to unity (Budday et al., 2017; Jin et al., 2013). For such sample size and shape, one needs to appropriately consider the deviation from a homogeneous deformation. For tensile experiments, top and bottom of the sample are usually glued to the instrument crosshead surfaces which causes a non-even lateral deformation of the sample (see Fig. 7(a)–(c)). The same scenario happens for samples glued to the crosshead surfaces in compression (Fig. 7(d)–(f)). Even for non-glued samples in compression, the friction between the tissue and the crosshead surfaces can cause a level of inhomogeneity in compressive deformation of the sample (Miller, 2005). To appropriately compensate for deviation from a homogeneous elastic field assumption, Miller proposed a method in which the shape of the lateral deformation of the sample during no-slip boundary condition uniaxial deformation is calculated and incorporated in the deformation function for obtaining the material constants of the brain tissue (Miller, 2001, 2005; Miller and Chinzei, 2002). Some other researchers have used numerical analysis using finite element method and minimizing the objective function for the difference between simulation results and experimental data in an iterative loop to optimize the model parameter values (Moran et al., 2014). While the former method is mathematically rigorous and can be used for cylindrical samples only, the latter is time consuming and might result in numerical outputs for model parameters

[1] There are alternative presentations for the Ogden hyperelastic energy function, however, the authors utilized the one which is used in ABAQUS in order to maintain the consistency between the modeling and the simulation processes.