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A patient-specific, finite element model for noncommunicating hydrocephalus capable of large deformation

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ABSTRACT

A biphasic model for noncommunicating hydrocephalus in patient-specific geometry is proposed. The model can take into account the nonlinear behavior of brain tissue under large deformation, the nonlinear variation of hydraulic conductivity with deformation, and contact with a rigid, impermeable skull using a recently developed algorithm. The model was capable of achieving over a 700 percent ventricular enlargement, which is much greater than in previous studies, primarily due to the use of an anatomically realistic skull recreated from magnetic resonance imaging rather than an artificial skull created by offsetting the outer surface of the cerebrum. The choice of softening or stiffening behavior of brain tissue, both having been demonstrated in previous experimental studies, was found to have a significant effect on the volume and shape of the deformed ventricle, and the consideration of the variation of the hydraulic conductivity with deformation had a modest effect on the deformed ventricle. The model predicts that noncommunicating hydrocephalus occurs for ventricular fluid pressure on the order of 1300 Pa.

1. Introduction

Cerebrospinal fluid (CSF) plays an important role in the physiological activities and protection of the brain. Under the classical theory of cerebrospinal fluid hydrodynamics (Orešković and Klarica, 2011), this fluid is produced at a constant rate in the choroid plexuses of the lateral and third ventricles. Most of the CSF drains through the Sylvius aqueduct to the fourth ventricle, while a small amount flows through the cerebrum into the subarachnoid space adjacent to the skull. If the Sylvius aqueduct becomes obstructed, such as caused by a growing tumor adjacent to it, CSF accumulates in the ventricles and an abnormally high transmantle pressure gradient develops. As a result, the ventricles expand significantly, leading to a medical condition known as noncommunicating, or obstructive, hydrocephalus (Corns and Martin, 2012).

Over the past 25 years, numerous mathematical models have been proposed to analyze hydrocephalus. Many of these models have represented the brain as a single-phase material in planar geometries (Fritz and Drapaca, 2009; Roy et al., 2013) and in cylindrical geometries (Drapaca et al., 2006; Sivaloganathan et al., 2005a, 2005b; Wilkie et al., 2010, 2011, 2012a, 2012b). In contrast, since the brain is immersed in and permeated by the CSF, others have represented the brain as a poroelastic or biphasic material in planar geometries (Momjian and Bichsel, 2008; Nagashima et al., 1987; Peña et al., 1999; Shahim et al., 2010; Taylor and Miller, 2004), in cylindrical geometries (Kaczmarek et al., 1997; Stastna et al., 1999; Tenti et al., 1999; Wilkie et al., 2012b), and in spherical geometries (García and Smith, 2010; Levine, 1999; Mehrabian and Abousleiman, 2011; Shahim et al., 2012; Smillie et al., 2005; Sobey and Wirth, 2006; Tully and Ventikos, 2009, 2011; Vardakis et al., 2013; Wilkie et al., 2012c). However, as the ventricles are not well represented as cylindrical or spherical cavities, recent efforts have focused on modeling hydrocephalus in anatomically realistic geometries (Cheng and Bilston, 2010; Clatz et al., 2007; Dutta-Roy et al., 2008) or quasi-realistic geometries (Wirth and Sobey, 2006).

While many of these models have given relatively good correlations of clinical observations of hydrocephalus, only a few single-phase models (Drapaca et al., 2006; Fritz and Drapaca, 2009; Roy et al., 2013; Wilkie et al., 2011) and two biphasic models (Dutta-Roy et al., 2008; García and Smith, 2010) have considered the nonlinear stress–strain response documented experimentally under finite deformation (Franceschini et al., 2006; Kaster et al., 2011; Miller, 1999; Miller and Chinzei, 1997, 2002). Considering that displacements occurring during hydrocephalus can be large, it would appear that nonlinear stress–strain curves under finite deformations should be taken into account. However, despite accounting for such behavior, the biphasic, noncommunicating model of Dutta-Roy et al. (2008) was not

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capable of producing either the ventricular volume associated with communicating, normal-pressure hydrocephalus or the larger ventricular volume associated with noncommunicating hydrocephalus found in clinical studies (Matsumae et al., 1996a).

A new, patient-specific model of noncommunicating hydrocephalus is proposed to more accurately represent the ventricular expansion found in clinical studies. The model takes into account the biphasic nature of brain tissue and the nonlinear behavior of the tissue under finite deformation. Different from other anatomically realistic models (Cheng and Bilston, 2010; Clatz et al., 2007; Dutta-Roy et al., 2008), the model also accounts for variation of the hydraulic conductivity with deformation and for contact between the expanding cerebrum and the rigid skull that encloses it using a recently developed algorithm (Ateshian et al., 2010). With this model, we estimate the necessary ventricular fluid pressure to cause noncommunicating hydrocephalus.

2. Methods

2.1. Geometry

The cerebrum and ventricle geometries were generated in 3D Slicer (slicer.org) using magnetic resonance (MR) imagery of a 72-year-old male from the Visible Human Dataset (National Library of Medicine, Bethesda, MD). The geometries were smoothed and simplified in MeshLab (meshlab.sourceforge.net), and then, due to the approximate symmetry of the brain, divided in half along the sagittal plane in ParaView (Kitware, Inc., Clifton Park, NY), yielding the surfaces shown in Fig. 1.

The portion of the skull above the cerebrum was also reconstructed from the same MR image dataset. In addition, a surface corresponding to the soft tissue below the brain was created by offsetting the bottom surface of the cerebrum a fixed distance of 5 mm (Fig. 2), which is a representative distance consistent with the reconstructed skull from the MR images. This surface proved necessary to prevent physically unrealistic anterior/posterior deformation of the base of the cerebrum.

To compare against the model of Dutta-Roy et al. (2008), we also created a set of artificial skulls by offsetting the entire outer surface of the cerebrum by a fixed distance. The offset distance between the cerebrum and skull ranged from 3 mm, which was used by Dutta-Roy et al. (2008), to 7 mm, which is representative of the maximum distance between the cerebrum and skull recreated from the MR images.

2.2. Mathematical model

The cerebrum was modeled as a homogeneous biphasic medium representing the solid tissue and cerebrospinal fluid. This theoretical framework assumes that both the solid and fluid phases are intrinsically incompressible but that the medium may compress by expulsion of the fluid. The governing equations are force equilibrium and conservation of mass of the mixture, which may be expressed, respectively, as

$$\nabla \cdot (\boldsymbol{\sigma}_e - p\mathbf{I}) = \mathbf{0}, \tag{1}$$

where $\boldsymbol{\sigma}_e$ is the effective stress that results from the deformation of the solid matrix and p is the fluid pressure, and

$$\nabla \cdot (\mathbf{v}_s + \mathbf{w}) = 0, \tag{2}$$

where $\mathbf{v}_s$ is the velocity of the solid matrix and $\mathbf{w}$ is the flux of the fluid relative to the deforming solid (Ateshian et al., 2010; Smith et al., 2012). The relative fluid flux

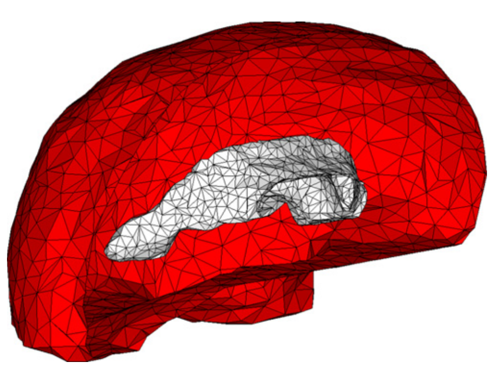

Fig. 1. Cerebrum and ventricle geometries reconstructed from magnetic resonance imagery from the Visible Human Dataset.

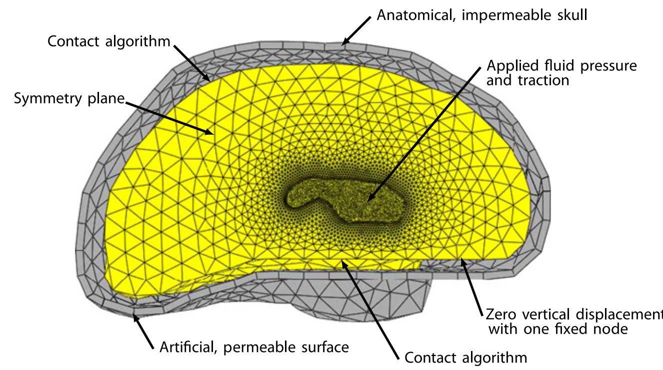

Fig. 2. Computational mesh of the cerebrum, rigid and impermeable skull, and rigid but permeable soft tissue. The boundary conditions associated with the ventricle and outer surface of the cerebrum are annotated.

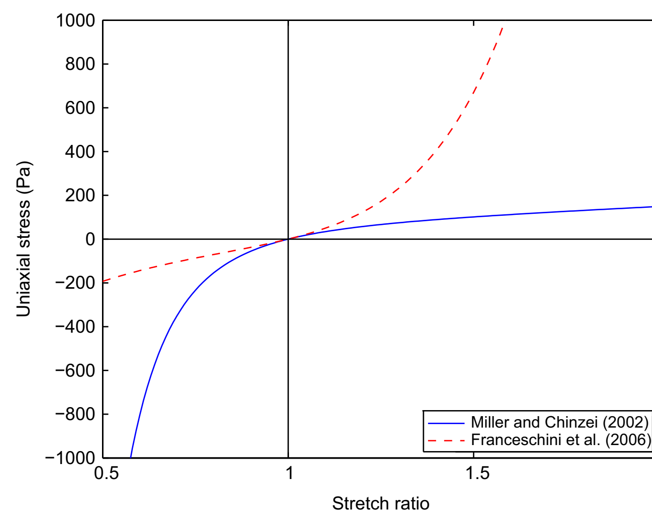

Fig. 3. Uniaxial stress response for the two sets of strain energy function material parameters. The Miller and Chinzei (2002) model represents softening behavior in tension, while the Franceschini et al. (2006) model represents stiffening behavior.

$\mathbf{w}$ may be expressed in terms of the fluid pressure gradient by Darcy's law as

$$\mathbf{w} = -\kappa \nabla p, \tag{3}$$

where κ is the hydraulic conductivity.

The solid phase of the cerebrum followed an Ogden material model, wherein the strain energy has the form

$$W = \sum_{k=1}^{n} \frac{\mu_k}{\alpha_k}\left(\lambda_1^{\alpha_k} + \lambda_2^{\alpha_k} + \lambda_3^{\alpha_k} - 3 - \alpha_k \ln J\right) + \frac{\mu'}{2}(J-1)^2, \tag{4}$$

where λ_1, λ_2, and λ_3 are the principal stretch ratios, α_k describes the shape of the stress–stretch curve, μ_k and μ' are material parameters, and J is the determinant of the deformation tensor (Ogden, 1984). To assess the sensitivity of results with respect to material nonlinearities, two sets of parameters were used in this study. Based on the experimental study of Miller and Chinzei (2002), in the first set only one term of Eq. (4) was considered with $\alpha_1 = -4.7$. In addition, based on the experimental study of Franceschini et al. (2006), in the second set two terms of Eq. (4) were considered with $\alpha_1 = 4.31$ and $\alpha_2 = 7.74$. While both nonlinear sets have the same slope at zero deformation, the former predicts a softening in tension, whereas the latter yields a significant stiffening effect in tension (Fig. 3). Henceforth, these will be referred to as the softening and stiffening models, respectively.

To be consistent with other studies of hydrocephalus (Dutta-Roy et al., 2008), a Young's modulus E of 420.6 Pa and a Poisson's ratio ν of 0.35 were used, from which values for μ_k and μ' can be determined (García and Smith, 2010; Smith et al., 2012). The chosen Young's modulus represents a relaxed value that corresponds to an instantaneous modulus of 2273 Pa (Miller and Chinzei, 2002). Thus, while viscoelastic effects were not directly included in the model, the relaxed material properties account for the viscoelastic nature of brain tissue found in experimental studies (Cheng and Bilston, 2007; Miller, 1999) and for the slow development of hydrocephalus, an approach recommended by Taylor and Miller (2004).