3-D Non-Linear Finite Element Analysis of Normal Pressure Hydrocephalus

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Abstract. This paper presents the mechanics of Normal Pressure Hydrocephalus (NPH) growth using a computational approach. We generated a generic 3-D mesh of a healthy human brain and treated the brain parenchyma as single and biphasic continuum with non-linear constitutive law undergoing finite deformations. Contact boundary conditions constrained the brain which is enclosed in a skull. We loaded the brain using transmantle pressure difference. Non-linear, implicit, Finite-Element (FE) procedures in time domain were used to obtain the deformations for the brain and ventricles. We propose that for modelling NPH, there is no significant advantage gained by using biphasic continuum to model brain parenchyma and that single phase continuum is adequate. We obtained almost equal ventricular volume for both single and biphasic treatment of brain parenchyma under same loading condition. The use of single phase continuum simplified the mathematical description for the model and resulted in large saving of computational time.

1 Introduction

Overlap of symptoms and diagnostic findings between Normal Pressure Hydrocephalus (NPH) and other neurodegenerative diseases (Alzheimer's etc) makes diagnosis of NPH a reoccurring problem for clinicians. Hakim and Adams [1] were the first to identify the condition of NPH. Currently, clinical and diagnostic findings of neurosurgeons in combination with engineering principles enhance the diagnosis of NPH [2, 3, 4 and 5], but these approaches offer no insight into NPH growth mechanics.

Hakim [6] proposed a "sponge" type model of brain parenchyma for NPH growth but without any mathematical formulations. Nagashima et al. [7] and Péna et al. [8] utilised coupled pore fluid diffusion and stress analysis (biphasic approach) for a linear elastic model of brain parenchyma (porous medium) undergoing infinitesimal deformations on a 2-D horizontal brain slice obtained from a brain atlas. Kaczmarek et al [9] used finite deformation biphasic theory on simplified brain geometry (cylindrical) and obtained an analytical solution. Taylor and Miller [10] utilised reassessed brain parenchyma elastic modulus and finite deformation biphasic theory on realistic 2-D brain geometry. Apart from NPH analysis, Miga et al. [11], Miga et al. [12], Paulsen et al. [13], Platenik et al. [14] and Lunn et al. [15] used biphasic

approach for intra-operative image registration of brain deformation during neurosurgery. They treated brain parenchyma as linear elastic and used infinitesimal deformation theory [11, 12, 13, 14 and 15].

In all these works [7, 8, 9, 10, 11, 12, 13, 14 and 15], the brain parenchyma was treated as linear elastic. The assumption regarding infinitesimal deformation [7, 8, 11, 12, 13 14 and 15] was violated during NPH formation and brain deformation during neurosurgery, due to large deformations in the brain parenchyma. For correct understanding of NPH growth mechanics, finite deformation formulations and constitutive law (e.g. hyperelastic) which can handle large strains (> 20%) encountered during NPH is required. The outer surface of the brain parenchyma was assumed to be fixed to the skull [7, 8, 9 and 10]. As a result, displacement of the brain outer surface was not possible. This is an oversimplification of the brain-skull interaction. For complete understanding of NPH, proper boundary conditions between the brain and the skull should be included in the model [16]. We addressed the deficiencies pointed above by using fully non-linear (geometric, material and boundary) model for our simulations. To the best of our knowledge, this is the first 3-D, non-linear model investigating NPH growth mechanics.

Section 2 includes descriptions of the generic brain mesh as well as loading and boundary conditions for both single and biphasic cases used in our simulation. Element types and formulations for single and biphasic continuum is given in section 3. We detail the results in section 4. Comprehensive discussions and summary of our main findings is in section 5.

2 Biomechanical Model

2.1 Brain Mesh

The brain mesh is shown in Fig 1. We created the generic mesh of the healthy human brain by modifying person specific brain mesh [16] using Hypermesh (Altair Engineering, USA) pre-processing software. Table 1 presents the values for brain and ventricular volume of a healthy human brain [17]. Brain and ventricular volumes in our mesh were consistent with the values given in Table 1. As the brain is approximately symmetrical, half of the brain was simulated. Ventricular volume for a healthy human in our simulations was $14cm^3$. NPH was deemed developed when ventricular volume increased from $14cm^3$ to more than 58 cm^3 (Table 1).

Table 1. Brain and Ventricular Volumes for Healthy and NPH Cases (adapted from Matsumae et al. [17])

Case	Brain Volume (cubic cm)	Ventricular Volume (cubic cm)
Healthy Brain	1188±104	27±10
NPH	1163±129	116±42

2.1.1 Modified Hyperelastic Material Parameters

The stress-strain behaviour of the brain parenchyma is non-linear with stiffness in compression significantly higher than tension with strong stress-strain rate dependency [18, 19]. To account for these complexities in the brain parenchyma, we chose the hyper-viscoelastic constitutive model proposed by Miller and Chinzei [19].

The time required for NPH growth is relatively long (typically 4 days) [7] when compared to surgical interventions and loading of the brain occurs very slowly. Thus, the strain rate dependency of the brain parenchyma disappeared [10]. Hence, the brain parenchyma was modelled as hyperelastic (Ogden form [20]) given by:

$$W = \frac{2\mu}{\alpha^2}(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3) \quad (\mathbf{1})$$

where, W is the potential function, λ_i*'s* are the principal stretches, μ is the relaxed shear modulus and α is the material coefficient which can assume any real value without any restrictions. The value μ was 155.77 Pa [10] and the value of α was -4.7 [10]. We considered the brain parenchyma to be homogenous and isotropic for simulation purposes [21] as the brain tissue does not exhibit directional structure, unless the behaviour of very small tissue specimens is of interest [22].

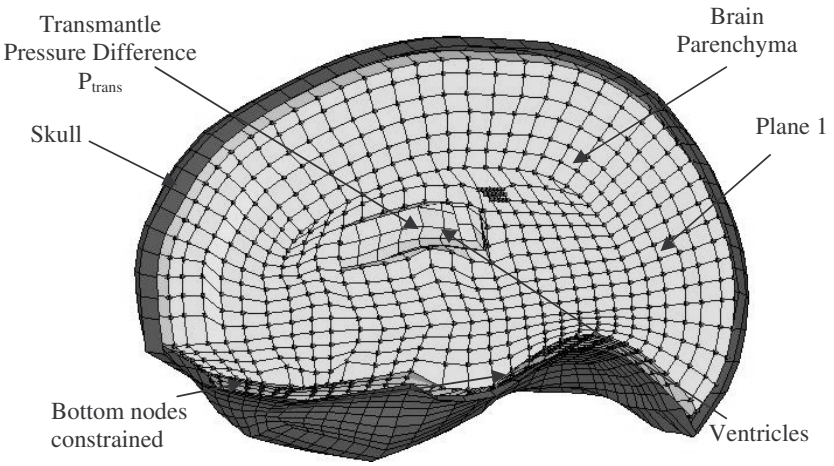

Fig. 1. Brain geometry, pressure loading and applied boundary conditions

2.1.1.1 Biphasic Continuum

Brain was considered to be a sponge like structure with the solid matrix corresponding to neurons and neuroglia and voids being extracellular space [1, 6] occupied by CSF. This is referred to as biphasic [7, 8, 9, 10, 11, 12, 13, 14, 15 and 22] continuum because of presence of two phases: brain parenchyma (solid or porous phase) and CSF (fluid phase). To understand the interaction between brain parenchyma and CSF when loaded by a given transmantle pressure difference, we performed a coupled pore fluid diffusion and stress analysis. The reader may refer to

the works of Nagashima et al. [7], Miller [23], Biot [24], and Bowen [25] for detailed discussions on the mathematical treatment of biphasic continuum.

In our model, brain parenchyma (solid phase) had a Poisson's ratio (υ) of 0.35 [7, 8, 9, 10, and 22] with relaxed hyperelastic material properties (section 2.1.1), initial void ratio of 0.2 [7, 26] and permeability of 1.59×10^{-7} m/sec [9]. It was fully saturated with CSF. CSF (fluid phase) was incompressible, non-viscous with mechanical properties of water.

2.1.1.2 Single Phase Continuum

Due to long development time for NPH, there existed possibility of Cerebrospinal Fluid (CSF) to be absorbed or evacuated in the brain parenchyma, resulting in CSF flow within it and change in brain and ventricular volume. We treated the brain parenchyma as compressible single phase continuum with non-linear constitutive law [19] (generalisation of Ogden rubber [23]), relaxed hyperelastic shear modulus (section 2.1.1) and a low Poisson's ratio of 0.35 [7, 8, 9, 10, and 22] and investigated this effect of compressibility.

2.2 Loading

Load was a transmantle pressure difference (P_{trans}) in form of pressure on the ventricular surfaces as shown in Fig 1. There was no pressure acting on the outer surface of the brain. It is a widely held view that transmantle pressure difference (P_{trans}) of 1mm of Hg (133.416 Pa) produced the clinical condition of NPH [27, 28] and the same was applied to the ventricular surface to investigate this claim.

2.2.1 Biphasic Continuum

Even though material strain rate effects were absent due to use of hyperelastic constitutive law for the brain parenchyma, rate effects were present because of relative motion between brain parenchyma (solid phase) and CSF (fluid phase). The time period of load application was of importance and transmantle pressure difference (P_{trans}) was applied over the development time of NPH (4 days) using a polynomial which provided zero velocity and acceleration respectively at the beginning and end of the loading.

2.2.2 Single Phase Continuum

The time period of load application was not important as we seek a static solution for the single phase continuum and material strain rate effects were absent due to use of hyperelastic constitutive law for the brain parenchyma. Hence, time period of transmantle pressure difference (P_{trans}) application was arbitrarily taken to be 10 seconds.

2.3 Boundary Conditions

As the brain is approximately symmetrical about the mid-sagittal axis, half of the brain for both single and biphasic continuum was simulated. The nodes on plane 1 (Fig 1) had symmetrical boundary conditions in YZ plane (no motion allowed for X translation) applied to them. As the brain was resting in the skull, we constrained the

brain bottom nodes in Y and Z translation (Fig 1). A skull enclosed the brain and frictionless, finite sliding; node-to-surface penalty contact between the brain and skull constrained the nodes on the brain outer surface. Following Wittek et al. [16], Sub-Arachnoid's Space (SAS) was accounted by a 3mm gap between the skull and the brain outer surface.

2.3.1 Biphasic Continuum
There exists a pressure gradient between the ventricles and Sub-Arachnoid's Space (SAS) resulting in flow of CSF from ventricles to SAS. We set the pore pressure on the ventricular surface equal to the transmantle pressure difference (P_{trans}) and 0 Pa on the outer surface of the brain and implemented this pressure gradient.

3 Computational Model

3.1 Brain Mesh

3.1.1 Biphasic Continuum
5858 porohyperelastic type C3D20PH (20 node triquadratic displacement, trilinear pore pressure, mixed formulation with linear pressure, pore pressure) [29] and 89 type C3D10H (10 node quadratic tetrahedron, mixed formulation with linear pressure, stress displacement) [29] elements discretised the brain parenchyma. We used mixed formulation quadratic tetrahedrons to complete the brick dominated mesh. Volumetric locking was not shown by both C3D20PH and C3D10H type elements.

3.1.2 Single Phase Continuum
The brain parenchyma consisted of 5858 type C3D20H (20 node quadratic brick, mixed formulation with linear pressure, stress displacement) [29] and 89 type C3D10H (10 node quadratic tetrahedron, mixed formulation with linear pressure, stress displacement) [29] elements. As mentioned earlier, we used mixed formulation quadratic tetrahedrons to complete the brick dominated mesh. Type C3D20H did not exhibit volumetric locking for incompressible/nearly incompressible continuum (e.g. brain).

3.2 Skull
The skull consisted of 1006 type R3D4 (4 node, bilinear quadrilateral, 3-D rigid) [29] elements.

3.3 Finite Element Solver
We obtained the solution for NPH growth model using ABAQUS/Standard (Abaqus Inc, Providence, Rhode Island, USA) non-linear finite element code (ABAQUS/Standard, 2004) [32]. The code accounted for geometric, constitutive and contact non-linearities. STATIC (fully non-linear, finite deformation) procedure obtained solution for single phase continuum case and SOILS (fully non-linear, finite

deformation, porohyperelastic) procedure gave solution for the biphasic continuum case. Wu et al. [30] showed the validity of SOILS procedure for hydrated biphasic tissues.

4 Results

Table 2 gives the summary of ventricular cavity volume produced due to application of transmantle pressure difference (P_{trans}) of 1mm of Hg (133.416 Pa).

Table 2. Volume of Ventricular Cavity Subjected to Transnamtle Pressure Difference (P_{trans}) of 1mm of Hg

Case	Poisson's Ratio (υ)	Ventricular Volume (cm^3)
Single Phase	0.35	37.2
Bi-Phase	0.35	36.6

5 Discussions and Conclusions

5.1 Ventricular Volume

Application of 1mm of Hg pressure load to the ventricular surface produced almost equal ventricular cavity volumes for both single and biphasic models (Table 2). The brain parenchyma had a Poisson's ratio (υ) of 0.35 which lead to equally low (467.31 Pa) bulk modulus for both cases. Due to this, the brain parenchyma was equally compressible for each. The long development time for NPH gave adequate time for the wetting fluid (liquid phase: CSF) to flow out of the interstitial voids and subsequently the pore pressure in the biphasic continuum which should have acted against the collapse of the solid phase (porous phase: brain parenchyma) did not do so. It could be convincingly argued from the results in Table 2 that there was no significant advantage gained by modelling brain parenchyma as a biphasic continuum for NPH. Furthermore, application of single phase model significantly reduced computational time. In this study, the computation time for single phase continuum was 160 minutes as compared to 1320 minutes for biphasic continuum.

5.2 Transmantle Pressure Difference (P_{trans}) required to produce NPH

Penn et al. [27] and Czosnyka [28] reported that less than 1 mm of Hg (133.416 Pa) transmantle pressure difference was adequate to produce the condition of NPH. As mentioned earlier, NPH was deemed developed when ventricular volume increased from 14cm^3 to more than 58 cm^3 (section 2.1) [17]. Our modelling results (Table 2) clearly showed that ventricular volume was significantly less than 58 cm^3 for both single and biphasic continuum when 1mm of Hg transmantle pressure difference (P_{trans}) was applied to the ventricular surface. A higher pressure would be required to produce NPH. Thus, if hypothesis of mechanical causes of NPH needs to be

sustained, measurement of transmantle pressure difference (P_{trans}) required to produce NPH should be reassessed.

5.4 Conclusions

Our work showed that application of 1mm of Hg transmantle pressure difference (P_{trans}) resulted in almost equal ventricular volume for both single and biphasic models. Hence, we recommend use of single phase continuum model for brain parenchyma. The use of single phase continuum simplified the mathematical description of the system and lead to shorter computational time. According to our modelling results, 1 mm of Hg transmantle pressure difference (P_{trans}) as reported by other authors was not adequate to produce NPH for both single and biphasic models This suggested that measurement of transmantle pressure difference (P_{trans}) required for producing NPH needed reassessment, if hypothesis of mechanical causes of NPH was to be sustained.

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