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A 3D Brain Deformation Model Experiencing Comparable Surgical Loads

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ABSTRACT

During the past 10 years, the finite element method (FEM) has been successfully employed to model the neuroanatomy under varying load conditions. Loading conditions used in previous work can be predominantly separated into two categories. The first category concerns large accelerations of the head followed by a sharp deceleration or impact. The second category considers maladies of the brain such as edema and hydrocephalus. Our research is focused on creating a third category which involves the application of surgical loads. In neurosurgical procedures, various instruments are used which purposely retract/resect or inadvertently move tissue. In addition, other sources of brain shift include reduction in brain buoyancy due to cerebrospinal fluid (CSF) drainage and administered drugs such as mannitol which causes brain volume to decrease by transporting fluid away from the tissue via the brain vasculature. The concern is that intraoperative loads such as these will move designated subsurface operative areas and lead to surgical error. The ultimate goal of our research is to model surgical loads, predict subsequent deformation and update the surgeon's navigation fields in real time. However, the scope of this paper is limited to the development of an initial 3D model of brain deformation.

INTRODUCTION

Neuroanatomical models of tissue mechanics are principally multi-phase systems. One common approach found in the literature patterns the brain after a sponge-like material using the theory of consolidation. Deformation by consolidation is characterized by an instantaneous deformation at the contact area followed by additional deformation from exiting pore fluid.

Over the past decade, there has been a surge of interest in applying consolidation theory to soft tissue mechanics. Taylor et. al. used the theory to create a mathematical model of interstitial transport taking into account plasma protein movement, and interstitial swelling [1], [2]. Basser also recognized the theory's potential and used it to model infusion-induced swelling in the brain which included a solute transport equation [3]. Nagashima et. al. were the first to employ the theory in conjunction with FEM for modeling brain tissue [4], [5], [6]. Their goal was to understand maladies of the brain such as edema and hydrocephalus by creating a two-dimensional model of vasogenic edema within the feline cranium. Nagashima and Tada went on to simulate similar maladies in the human brain and concluded that consolidation theory was able to yield results comparable to clinical conditions [5], [7].

Although more complex theories exist, consolidation physics as described by Maurice Biot [8] is much more useful than simple linear, single phase elasticity due to the added fluid component. Allowing for a fluid component provides the model with access to more realistic boundary conditions dealing with intracranial pressures and CSF drainage. Consolidation also has the advantage of being linear which significantly reduces the computational effort. Understanding the strengths and limitations of consolidation as a modeling theory for brain tissue has become our starting point for assessing the possibility of exploiting computational modeling in computer-assisted stereotactic surgery.

THEORY

The theory of general consolidation proposed by Biot [8] assumes the medium is a linearly elastic isotropic solid experiencing small strains. The theory also assumes that the pore fluid is incompressible. Biot's complete equilibrium equations with some modification for brain tissue modeling are represented by

$$G\nabla^2\mathbf{u} + \frac{G}{1-2\nu}\nabla\varepsilon - \alpha\nabla p + (\rho_{tissue} - \rho_{fluid})\mathbf{g} = 0. \quad (1)$$

where independent variable: G is the shear modulus, α is a saturation constant, ρ is density, $\mathbf{g}$ is the gravity vector, and dependent variable: $\mathbf{u}$ is the displacement vector, ε is the volumetric strain, and p is the interstitial pressure. Equation (1) relates mechanical equilibrium to the fluid pressure gradient across the medium, to changes in medium buoyancy, and to deformation resulting from applied surface stresses to the medium.

In order to complete the continuum model, a constitutive relationship relating volumetric strain and fluid pressure is required [8]. The final modified constitutive relationship is

$$\nabla \cdot k\nabla p - \alpha\frac{\partial \varepsilon}{\partial t} - \frac{1}{S}\frac{\partial p}{\partial t} = \psi, \qquad (2)$$

where the independent variable: k is the hydraulic conductivity, $1\backslash S$ is a void compressibility constant, and ψ is the body fluid load. This equation predominantly relates the time rate of fluid transport with volume deformation. An additional compressiblity term appears in Biot's theory, but generally the brain can be considered a saturated medium which eliminates the time derivative of pressure (i.e. $1\backslash S = 0$). Finally, surgical loads effecting fluid content in the brain, such as the drug mannitol, are considered and implemented. These terms are lumped into a surgical fluid load leading to the complete constitutive relationship shown in (2).

METHODS

Initial studies are underway to evaluate a 3D finite element consolidation model for computing brain deformation under surgical loads using a porcine model. Prior to any implantation or deformation of the brain tissue, a complete set of magnetic resonance images (MRI) are taken of the cranium. It is these scans that serve as the basis for the FEM discretization process. We use **ANALYZE Version 7.5 - Biomedical Imaging Resource** (Mayo Foundation, Rochester, M.N.) to segment the 3D volume of interest. We then use **MATLAB** (Math Works Inc., Natick Mass.) to render the surface boundary description of the brain and any surgical implants. Following the surface rendering, we use mesh generation software [9] to produce a tetrahedral grid on the interior which completes the discretization process. Figure 1 displays a typical mesh where volume elements inside the tissue are approximately 1 mm^3 and for elements near the implanted catheter, 0.025 mm^3. The mesh contained 11923 nodes and 62,439 tetrahedral elements.

In exercising the model, calculations have been obtained against some of the standard soil mechanics benchmarks for consolidation and we have found the solutions are highly accurate (less than 0.1% of applied load). Some preliminary results have also been performed on brain tissue. Simulations of the effects of CSF drainage and mannitol delivery have been performed and are described in the next section. In addition a pilot series of experiments in an actual pig brain has been initiated in order to investigate model behavior under comparable surgical loads.

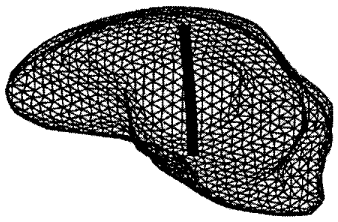

Figure 1: Sagittal boundary of volume mesh

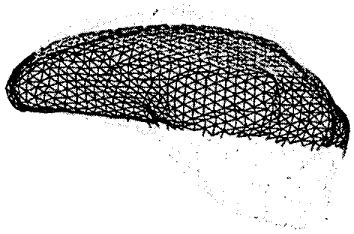

Figure 2: Brain shift due to CSF drainage and reduction in buoyancy (Sagittal View)

During the surgical procedure, a pig brain is implanted with small 1 mm beads in a grid like fashion in the parenchyma which serve as tracking markers. Following the implantation, a balloon catheter filled with contrast agent is also inserted into the cranium. All objects are easily observed in a computed tomographic (CT) scanner and are monitored during balloon inflations which are intended to mimic surgical loads. In addition to tracking deformation, effort is also underway to verify the pressure component to our model. Pressure dynamics during a procedure are measured using a Camino fiberoptic sensor (Camino Laboratories, San Diego, California).

RESULTS

We have exercised the model by subjecting the brain to two simulated fluid loads. The first is illustrated in Figure 2 and is representing the effects of CSF drainage. Note that all deformation is highlighted in black. As drainage occurs the buoyancy force decreases. The over-

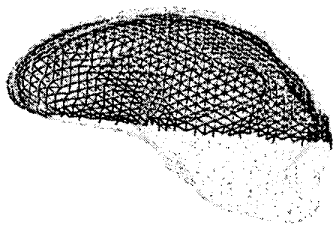

Figure 3: Brain shift due to volume reduction from administered mannitol (Sagittal View)

all effect from this buoyancy reduction is a spreading of the parenchyma which also correlates with clinical observations. The second example is shown in Figure 3 and is a representation of the effects of administering hypertonic solutions such as mannitol. Hypertonic solutions have the effect of reducing brain volume by transporting away interstitial fluid via the brain vasculature. The overall effect is a shrinking of the brain resulting in **volume reduction** contrary to Figure 2 where the volume is conserved and the subsequent shrinking on the top is accounted for by a spreading at the sides and front.

In addition to these qualitative simulations, an *in vivo* experiment using a porcine model has been performed. To illustrate the surgical intervention and subsequent bead motion described in the previous section, Figure 4 provides a volume representation of the CT scans with brain tissue thresholded out. The image was rendered in **ANALYZE**. Figure 4a is a display of the skull of the pig with the implanted catheter and bead markers prior to inflation. Figure 4b represents the same configuration with a fully inflated 4cc balloon catheter.

In our experiments, we have also made some preliminary measurements of pressure during inflation. Figure 5 represents the pressure transient during a sustained inflation. In this experiment, the balloon catheter was implanted mid-brain, interparenchymally. The magnitude of the results corresponds to Wolfla [10], [11] although our time constant appears to be somewhat longer. This

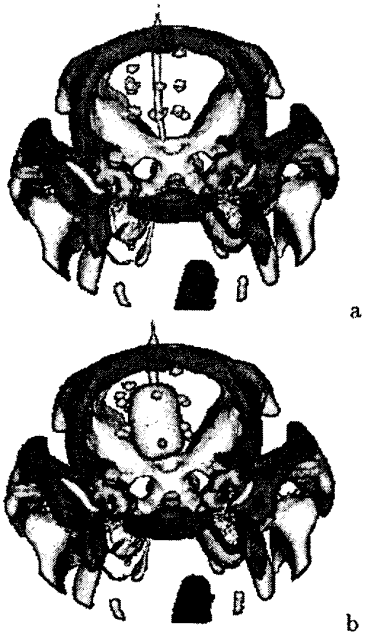

Figure 4: Experimental model (a, b - baseline, 4cc)

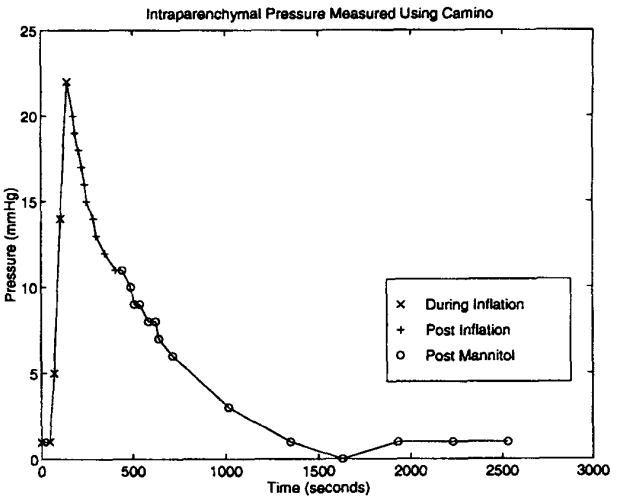

Figure 5: Interstital pressure during inflation (4cc inflation)

could be attributed to mass location. Wolfla studied interparenchymal pressure in the brain with expanding frontal epidural and extradural temporal mass lesions.

Qualitative correlations have been made to the experimental deformation. Figure 6 is an overlay of the undeformed and deformed state in which the latter has been subjected to a constant normal stress applied to the interior of the catheter. Figure 7 is a graph showing bead displacement calculated vs. bead displacement derived from analyzing the images in Figure 4. Although, the computed magnitudes are in error up to several millimeters, the general displacement trends are being predicted. There are several likely reasons for the discrepancies observed in Figure 7. First, the brain has been modeled as a single material. In future simulations, we will represent the neuroanatomy of the pig more accurately; i.e. discriminate between white and gray matter as well as other anatomical features such as the basal ganglia and ventricles, etc. Second, there are anatomical discrepancies existing in the applied boundary conditions meaning that

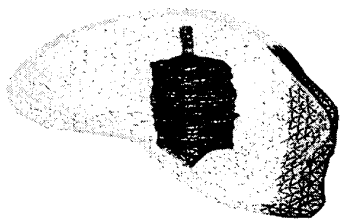

Figure 6: Deformed boundary.

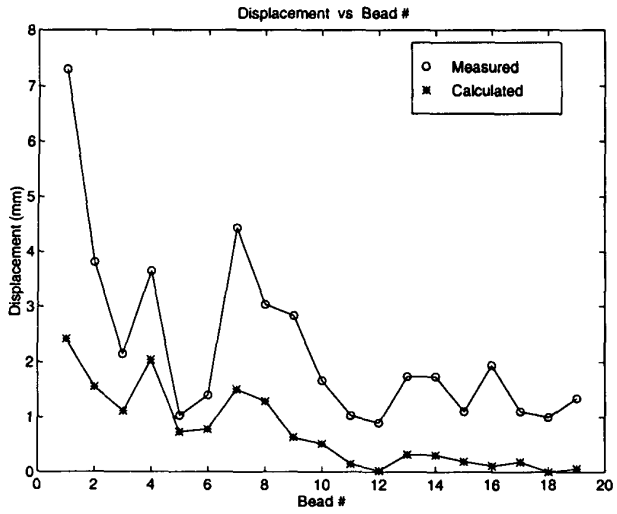

Figure 7: Displacement trend of beads.

certain surfaces in the cranium have different compliances and these areas must be designated in the boundary conditions. Third, there are uncertainties in the co-registration between CT and MRI which could also lead to inaccurate approximations. While all of these are undoubtedly responsible in part for the lack of more quantitative agreement, the overall trends in displacement seem promising.

DISCUSSION

The implications of this research are far reaching and could influence the practice of stereotactic neurosurgery. From previous research, some qualitatively predictive capabilities of two dimensional modeling theory concerning maladies of the brain have been shown [4], [5], [6]. However, in the context of surgical loads, brain shift is clearly a three dimensional problem. These preliminary results illustrate significant potential for computer assisted neurosurgery. By modeling in three dimensions and performing quantifiable *in vivo* experiments, this research may enhance the role for computer-assisted neurosurgery.

ACKNOWLEDGMENTS

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