# Model-Updated Image Guidance: Initial Clinical Experiences with Gravity-Induced Brain Deformation

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Abstract-Image-guided neurosurgery relies on accurate registration of the patient, the preoperative image series, and the surgical instruments in the same coordinate space. Recent clinical reports have documented the magnitude of gravity-induced brain deformation in the operating room and suggest these levels of tissue motion may compromise the integrity of such systems. We are investigating a model-based strategy which exploits the wealth of readily-available preoperative information in conjunction with intraoperatively acquired data to construct and drive a three dimensional (3-D) computational model which estimates volumetric displacements in order to update the neuronavigational image set. Using model calculations, the preoperative image database can be deformed to generate a more accurate representation of the surgical focus during an operation. In this paper, we present a preliminary study of four patients that experienced substantial brain deformation from gravity and correlate cortical shift measurements with model predictions. Additionally, we illustrate our image deforming algorithm and demonstrate that preoperative image resolution is maintained. Results over the four cases show that the brain shifted, on average, 5.7 mm in the direction of gravity and that model predictions could reduce this misregistration error to an average of 1.2 mm.

*Index Terms*—Brain deformation model, brain shift, consolidation, finite element model, image guidance, porous media.

#### I. INTRODUCTION

WITH advancement in high-resolution magnetic resonance (MR) and computed tomography (CT) imaging has come the ability to perform stereotactic tasks in the operating room using patient-registered image guidance [1]–[6]. However, recent literature has highlighted a potential problem regarding the fidelity of such systems. During the course of neurosurgical procedures substantial intraoperative tissue deformation has been documented, which ultimately compromises the registration between the patient's preoperative

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image series and the current state of the operating field in the operating room (OR) [7]–[11]. Cortical surface shifts of 1 cm have been reported [7], [8] as well as subsurface tissue motion of 4–7 mm at the interhemispheric fissure and lateral ventricles [10], [11]. These studies also suggest that gravity-induced deformation is a primary source of intraoperative movement [7], [8].

Solutions to this problem have been predominantly limited to intraoperative MR and ultrasonography. The former has been under evaluation by several groups [12]-[14]. While intuitively appealing and conceptually powerful, intraoperative MR presents certain challenges. For example, Wirtz et al. noted the cumbersome nature of the technique, as well as the need to evaluate its cost effectiveness. Steinmeir et al. have proposed the use of an open scanner in the context of a twin operating theater, in order to address some of the cost issues. However, problems with patient transport time and reregistration of surgical fields in some cases need to be overcome. Coregistered ultrasonography does have a distinct cost advantage, as well as the capability of collecting fast multiplanar data with three-dimensional (3-D) reconstruction capability [15]-[18]. However, image clarity with respect to soft tissue structures is often difficult, due to limited contrast resolution. Additionally, image quality tends to degrade during surgery as the imaging field becomes progressively invaded. Despite these shortcomings, intraoperative ultrasound can be expected to play an important role in neurosurgical image guidance, but likely as a means of augmenting other methods of correcting for misregistration errors which become exacerbated during surgery due to tissue motion. For example, Trobaugh et al. [15] have developed the means of providing the surgeon with a second view of brain structures, using ultrasound which has been used to estimate intraoperative shift [16].

Another strategy for retaining registration fidelity between the preoperative image and OR spaces, despite tissue movement subsequent to the initiation of surgery, is to exploit a computational model which updates the preoperative image database by estimating intraoperative tissue displacement [19]. This approach is not only cost effective, but retains use of the wealth of high-resolution preoperative information which is routinely available, while affording the opportunity to integrate sparse intraoperatively acquired data (e.g., intraoperative ultrasound and cortical surface tracking) short of complete volumetric imaging. Deformation models of this type have been used successfully to match pre- and postcondition images or to conform patient-specific images to a reference image template [20]–[23]. In these situations, the primary task of interest has been to transform one known image into the shape of another known image, without knowledge of the physical driving forces involved or, in some cases, to better understand pathological progression, e.g., tumor growh [24]. Intraoperatively, one has the potential opportunity of being able to model the physical events which take place during surgery, in order to account for tissue motion.

Interestingly, there have been only a few attempts to model brain deformation during surgery as an aid to image guidance. Examples include recent reports by Edwards et al. [25] and Skrinjar et al. [26]. The Edwards study represented the brain as a three compartment system consisting of bone, fluid, and soft tissue where rigid body transformations were applied to bone, fluid regions were unconstrained, and smooth deformation was applied to the parenchyma. Several energy models were compared in two dimensions with data from an epilepsy patient where preoperative MR and postoperative CT were available for analysis. The Skrinjar study represented the brain as a homogeneous linear viscoelastic medium, using relatively coarse discretizations of the tissue continuum. Simulations of an artificial parietal craniotomy were reported in two and three dimensions, illustrating time sequences of the computed deformation field which showed settling effects, due to gravity, that cause not only posterior movements near the craniotomy, but also motion in the superior and inferior directions.

We have been developing a brain deformation model for use as an aid in image guidance as well. Our approach models brain mechanics with consolidation theory, which represents tissue as a solid matrix that is saturated with an interstitial fluid [27]. Nagashima et al. [28]-[32] recently exploited consolidation physics to study various pathophysiologies of the brain, such as hydrocephalus and hemorrhage, and we have adapted this computational framework for the neurosurgical image guidance setting. In previous work, we presented the details of the numerical underpinnings of this model and have investigated its computational accuracy and stability [33], [34]. We have also performed a number of in vivo validation studies in the porcine brain where we have shown that the model can recapture 75-85% of the registration error associated with brain tissue motion, under a variety of loading conditions [35]-[37].

In this paper, we extend the model to include gravitational effects and deploy it retrospectively in four clinical cases involving the human brain. The incorporation of gravity is timely because several recent studies on cortical surface motion have documented the importance of gravity-induced brain shift. Further, this brain sag occurs early in an OR case, can be measured reasonably easily intraoperatively, and accounts for a sizable registration error that should be corrected. Hence, gravity-induced brain shift is an initial neurosurgical event that makes an excellent test case for evaluating the prospects of model-based updating of preoperative images for image guidance. The results we present here are encouraging and show that registration errors on the order of 6 mm can be reduced

to approximately 1 mm with the aid of the model. They are also significant because they demonstrate that all of the essential elements associated with generating 3-D high-resolution patient-specific models, computing volumetric gravity-induced brain shift and deforming the preoperative MR imaging study based on estimates of intraoperative tissue motion, are in place. This appears to be the first application of complete 3-D high-resolution consolidation theory modeling of actual human clinical cases where resulting brain deformation estimates are used to update preoperative images. While the experience reported herein is considered very positive, it represents a small series of surgeries where limited intraoperative data on the actual tissue motion was available, making further validation of the technique in humans essential.

## II. COMPUTATIONAL MODEL

In this section, we briefly describe the computational model we are using and focus attention primarily on the addition of gravity effects which we have not modeled previously. Our approach is based on a finite element rendering of consolidation physics where the mechanics of the continuum are characterized by an instantaneous deformation at the contact area, followed by subsequent additional displacement over time as interstitial fluid drains in the direction of straininduced pressure gradients (i.e., from high to low pressure) when subjected to load.

Given its biphasic nature, consolidation theory would appear to be a more realistic description of the continuum than simple elasticity. Specifically, acute transient deformation associated with neurosurgical loading conditions seems more closely related to the brain's hydrodynamic nature, rather than its viscoelastic behavior [38]–[40]. Viscoelastic response is undoubtedly important in brain mechanics, but likely at longer time scales, and its incorporation into the consolidation framework can be readily accomplished. There are also major computational advantages resulting from the linearity of the Hooke's law and Darcy's law assumptions used in a consolidation formulation. Clearly, inaccuracies in the model will be introduced under large deformations. However, we have found that the model still generates reasonably good updates even under gross deformation.

In this study of four clinical cases, we focus on gravityinduced deformation which results from the drainage of cerebrospinal fluid (CSF) from the cranial cavity. With the intracranial loss of CSF, buoyancy forces acting on the brain to counteract gravity are reduced, causing gravitational forces to deform the brain. These forces have been applied in a simple manner by designating a gravitational force in each element in the finite element mesh of the patient-specific brain anatomy due to the difference in density of the tissue and surrounding fluid (e.g., in the case of CSF drainage, elements above the fluid line use the density of air). The gravitational effect adds the additional (last) term to the mechanical equilibrium expression

$$\nabla \cdot G \nabla \mathbf{u} + \nabla \frac{G}{1 - 2\nu} \nabla \cdot \mathbf{u} - \alpha \nabla p + (\rho_t - \rho_f) \mathbf{g} = 0 \quad (1)$$

where  $\rho_t, \rho_f$  are the density of the tissue and surrounding fluid, respectively, and **g** is the gravitational acceleration vector. The displacement vector **u**, interstitial pressure *p*, and material constants  $G, \nu, \alpha$  remain defined as before ([33, (1a)]). Finite element treatment of (1) and its coupled evolution equation in the pore fluid pressure leads to a modification of the discrete matrix system for time advancement of the solution reported in [33]

$$AU^{n+1} = BU^n + C^{n+\theta} \tag{2}$$

by adding the weighted residual volumetric integration of gravitational forces into column vector  ${\cal C}$ 

$$C_{i}^{n+\theta} = \begin{cases} \hat{x} \cdot \oint \sigma_{s}(t_{n+\theta}) \cdot \hat{n}\phi_{i} \, ds + \langle (\rho_{t} - \rho_{f})g_{x}\phi_{i} \rangle \\ \hat{y} \cdot \oint \sigma_{s}(t_{n+\theta}) \cdot \hat{n}\phi_{i} \, ds + \langle (\rho_{t} - \rho_{f})g_{y}\phi_{i} \rangle \\ \hat{z} \cdot \oint \sigma_{s}(t_{n+\theta}) \cdot \hat{n}\phi_{i} \, ds + \langle (\rho_{t} - \rho_{f})g_{z}\phi_{i} \rangle \\ \Delta t \oint k \nabla p(t_{n+\theta}) \cdot \hat{n}\phi_{i} \, ds \end{cases} \end{cases}.$$

Numerically, this results in a body force term which is added to the right-hand side at each time step that is nonzero when the density between tissue and the surrounding fluid is nonzero, i.e., elements above the fluid line.

# III. CLINICAL CASE STUDIES

### A. Intraoperative Data Acquisition

In previous work we presented a detailed study where cortical landmarks were tracked in 28 neurosurgical cases [7]. Data acquisition involved digitizing and tracking cortical features (i.e., typically three or four blood vessel bifurcations) identified postcraniotomy, using the Surgiscope stereotactic system (Elekta AB, Stockholm, Sweden). This unit integrates a Leica operating microscope (Model M695, Leica USA, Rockleigh NJ) with a robotic platform spatially coregistered with preoperative imaging studies. Additionally, video images of the field of view were recorded on an intraoperative computer system. In the cases presented here, the surgical focus was located by comparing sulci formations in the video images with the preoperative segmented MR brain surface. The accuracy with which cortical features identified in the video position can be related to their equivalent position in the segmented MR rendering is based on anatomical landmark recognition, which is somewhat subjective and relies on the experience of a trained neurosurgeon. Nonetheless, despite some uncertainties, the locations designated are representative of the surgical focal area. For the pilot series reported here, surgical procedures were selected in which the predominant mode of tissue deformation was anticipated to be gravitational sagging. In each of the four cases presented, the cortical surface was tracked with respect to gravity and then compared to calculations made using our computational model. The following subsection briefly describes each clinical condition and associated surgical intervention.

## B. Clinical Cases

Patient 1 is a 35-year-old male with a history of medically intractable epilepsy, associated with a posterior orbito-frontal tumor. Based on electrode strip recordings, he was brought to the OR for resection of tumor and surrounding epileptogenic cortex. At the time of surgery general anesthesia was administered and the patient was supine with the head secured in three-point fixation and turned  $60^{\circ}$  to his right.

Patient 2 is a 33-year-old man who had previously undergone resection of a left frontal lobe arteriovenous malformation and then presented with a medically intractable seizure disorder. MRI revealed encephalomalacia in the area of the left gyrus rectus and orbito-frontal cortex. He was brought to the operating room under general anesthesia for resection of this cortex and associated gliotic scar. He was positioned supine, with the head turned 60° to his right and secured with a Mayfield clamp. The previous bicoronal bone flap was opened on the left side and dissection was carried out from the lateral fronto-orbital cortex medially.

Patient 3 is an 18-year-old female with a long-standing medically intractable, MRI-negative seizure disorder who had undergone intracranial electrode investigation without satisfactory localization of seizure onset. She was brought to the operating room for anterior 3/4 corpus callosal section. Under general anesthesia, she was positioned supine with her head in neutral position in three-point pin fixation. A right parasagittal frontal craniotomy and retraction of the right hemisphere allowed visualization down the interhemispheric fissure to the corpus callosum. The commissural section was performed with a blunt dissector and suction.

Patient 4 is a previously healthy 54-year-old woman who developed the acute onset of left-sided weakness and, on CT and MRI scans, was found to have a large contrastenhancing right frontal lobe mass. A right frontal craniotomy was performed with the patient supine under general anesthesia and the head secured unturned in three-point pin fixation. A gross total resection of the enhancing mass was accomplished; the histopathologic diagnosis was glioblastoma multiforme.

In each case, there was minimal surgical intervention immediately-post craniotomy but significant drainage of cerebrospinal fluid. A patient-specific model was generated for each person from the MR data set (models contained 15000-17000 nodes, which yields a typical spacing of 0.5 cm). Tissue mechanical properties were based on previous pig brain experiments which investigated consolidation theory modeling in vivo [36], [37]. These values are within a physiologically reasonable range, given the limited amount of in vivo data that is available on the human brain. They are softer than those used in the Nagashima studies [28], but more in keeping with the estimates by Basser based on analytical analyses [41]. The model assumes homogeneity with respect to elastic properties (E = 2100 Pa,  $\nu = 0.45$ ), heterogeneity (at this stage, we have limited heterogeneity to white and gray matter) with respect to hydraulic properties  $(k_w = 1 \times 10^{-10} \text{ m}^3 \text{s/kg}, \text{ and } k_g = 5 \times 10^{-12} \text{ m}^3 \text{s/kg}), \text{ and}$ tissue saturation. The gravitational acceleration vector g was



Fig. 1. Boundary condition templates for (a)  $60^{\circ}$  turned and (b) neutral head orientations. Surface 1 is stress free at atmospheric pressure, surface 2 allows slippage along the cranial boundary at atmospheric pressure, but no normal motion, surface 3 is fixed at atmospheric pressure, and surface 4 is the same as surface 2, except no drainage is allowed. Exact boundary delineation between surfaces varied from case to case but not significantly. CSF level was typically chosen at the height of the surface 3/surface 4 interface.

determined from OR information on patient orientation and the cerebrospinal fluid level was defined to cover the lower portion of the brain, depending on the position of the cranial opening. Fig. 1 illustrates the boundary conditions used in the model for the two surgical orientations ( $60^{\circ}$  to patient's right and supine neutral). Although the actual conditions applied are case specific, generally, the highest elevations in the brain (surface 1) reside at atmospheric pressure and are stress free, the mid-elevations (surface 2) slide along the cranial wall but are restricted in their normal direction (to the cranium) movement, the brain stem area (surface 3) is fixed at atmospheric pressure, while the lowest elevations (surface 4) are similar to the mid-elevations but do not allow fluid drainage. The CSF fluid-line was typically located along the surface 3/surface 4 interface in Fig. 1.

Subject	Point #	Measured Displ. ( <i>mm</i> )	Calculated Displ. (mm)
PATIENT-1	1	6.7	4.9
	2	4.6	5.4
	3	4.2	5.8
	4	3.5	3.4
PATIENT-2	1	10.4	5.7
	2	6.2	6.3
	3	5.9	6.2
PATIENT-3	1	6.1	5.2
	2	5.0	6.5
	3	OB	OB
	4	7.5	6.1
PATIENT-4	1	4.4	4.8
	2	3.5	3.8
	3	OB	OB



Fig. 2. Intraoperative image updating algorithm based on model calculated deformation.

## C. Image Updating

Since the numerical model is coregistered initially with the operating field through the preoperative image to OR space transformation, following the volumetric calculations performed by the model, the image-database is deformed accordingly to produce a more accurate registration with the current surgical field. The step-wise generation of an updated image is shown in Fig. 2. Preoperatively, the finite element discretization is created from the MR database (top left of Fig. 2). Intraoperative data is acquired which, in this case, is the patient's head position relative to gravitational forces and an estimate of the cerebrospinal fluid level in the cranium. Using that data, the finite element model computes the field



Fig. 3. Overlays of undeformed (gray) and deformed (black) volumetric meshes for each patient. (a) Patient 1. (b) Patient 2. (c) Patient 3. (d) Patient 4.

of displacements (here driven by gravitational body forces) and deforms the mesh (next clockwise subfigure shows an overlay of the undeformed mesh in gray with the deformed mesh in black). The basis function expansion for the displacement solution used in the weighted residual treatment of the field equations can then be employed to calculate an equivalent undeforming displacement vector at each voxel in the deformed volume. With the voxel displacement calculated, the voxel is undeformed to the original MR space to determine the intensity value it should have in the deformed volume. This backcasting technique produces a contiguously deformed model which averts the problem of hole formation from integer round off that results from forward propagation. It is important to emphasize that the deformed image is not an interpolation per se but, rather, it represents exactly the approximate solution to the continuum physics which is encoded into the finite element solution of the governing partial differential equations. This is a significant advantage of the finite element technique. The displacement and pressure field variables are perfectly well defined at any point in the computational space, subject to the assumptions underlying the finite element discretization process.

#### D. Results

In Table I, a comparison between measured and calculated gravity-induced shift is reported. Column 3 records the amount of measured displacement in the direction of gravity during each clinical case. Column 4 shows the model prediction on a point-by-point basis. Averaging over all points in the four surgeries produces an absolute model error of  $1.2 \pm 1.3$  mm with respect to a mean cortical shift displacement of 5.7 mm  $\pm$ 2.0 mm, which suggests that the model can account for approximately 79% of the shift induced by gravity on average. Fig. 3 is a composite, showing an overlay of the undeformed (gray) and deformed (black) volumetric mesh boundaries for each patient with gravity acting vertically down the page. Fig. 4 illustrates the distribution of total cortical displacement on the surface for each patient-specific model. Fig. 5 demonstrates the results of the algorithm shown in Fig. 2. The first column in Fig. 5 is a preoperative axial image slice, with the direction of gravity designated by the white arrow. The second column is the updated image based on finite element calculations. The last column shows the deformed image subtracted from the preoperative cross section, where areas different from the surrounding background shading represent shift in the image.



Fig. 4. Distribution of total cortical shift with dark areas correlating with maximal displacement for each patient model. (a) Patient 1. (b) Patient 2. (c) Patient 3. (d) Patient 4.

# E. Discussion

The algorithm for deforming images, as conceptualized in Fig. 2 which produced the results of Fig. 5, demonstrates that the deformed images preserve the preoperative image resolution. Table I suggests that the potential of using modelbased updates is promising with an average error of 1.2 mm relative to the average surface displacement of 5.7 mm. This indicates that the model recaptured approximately 79% of the error induced by shift, which is in remarkable congruence with the very quantitative pig experiments we have completed to date [35]. Interestingly, the only condition applied to the model is a gravitational force with no other utilization of the intraoperatively measured cortical surface shift. Assuming we can constrain the model in the future by using data from intraoperative ultrasound and cortical shift measurements, we expect the model accuracy to increase. However, even in its current form, the updates are significantly better than their preoperative counterparts. Fig. 3 shows the deformed and undeformed volumetric mesh boundary overlays, which are consistent with the gravity-induced shift observed clinically. Furthermore, Fig. 4 quantifies the distribution of cortical shift predicted by the model where areas of maximal shift (darkest areas on surface) are located at the highest brain locations, relative to the head positions shown in Fig. 3.

Fig. 5 demonstrates intraoperative updating of preoperative images and quantifies the extent of shift through difference images. In each patient, a visible shift of subsurface structures is evident and, more importantly, the movement varies nonuniformly, suggesting that fixed transformation solutions to intraoperative motion should be avoided. In the four cases presented, points at the lateral ventricles were predicted to move 3-6 mm while the interhemispheric fissure is predicted to move 2-5 mm from the midline, which is in agreement with the few literature measurements which exist [10], [11]. Fig. 5(d) highlights the potential benefit of model-updated neuronavigation, where the axial plane has been selected to show a cross sectional view of Patient 4's glioblastoma. The difference image reveals that a shift of 3-4 mm of the tumor boundary has been predicted by the model. This information provided during surgery would be helpful for decision making regarding tumor excision and the sparing of healthy tissues. Clearly, the subsurface results in Fig. 5 are based only on



(a)



(b)





Fig. 5. Demonstration of intraoperative updating, based on model calculations with preoperative high-resolution MR axial slices (left column), intraoperative update (middle column), and the subtraction of the slices with shift designated by areas differing from background shading (right column) for each patient. (a) Patient 1. (b) Patient 2. (c) Patient 3. (d) Patient 4.

(d)

model calculations at this stage and validation studies in the human brain where independent measurements of tissue motion at depth are available will be needed in the future.

#### **IV.** CONCLUSIONS

Results from Table I and Fig. 5 suggest that model-updated image guidance is a promising method for correction of misregistration, due to intraoperative tissue deformation. Furthermore, by utilizing low-cost computational power, the technique is cost effective, making its widespread adoption possible. The approach effectively couples the information-rich preoperative setting with the surgical environment where more limited, but still useful, data is available.

The most significant limitation of this technique at the present time is the computational overhead associated with calculating a finite element solution for each update. However, given the linear nature of the governing equations, much of the computation could be performed preoperatively (i.e., matrix assembly and preconditioning) with time evolution of the solution executed intraoperatively. Assuming some computations are accomplished preoperatively, our nonoptimized, nonparallelized code has been able to calculate updated displacements in 5-10 min with numerical models containing 60 000-70 000 degrees of freedom (vector displacement plus pressure fields throughout the volume). Relative to a neurosurgical case which may last several hours, this time scale is acceptable, especially when compared to using intraoperative MR imaging and its transport and/or patient reregistration times. Furthermore, our empirical experience with the in vivo porcine system [35], in conjunction with recently reported human data [11], suggests that far field displacements (i.e., contralateral to the surgical focus) are sufficiently small and that a reduced volume calculation may be possible, which would accelerate the computation of updates significantly.

Another potential concern with this technique is the determination of material properties in vivo in humans. To date, there has been only a modest exploration of stiffness and hydraulic properties for brain tissue, with values being determined largely by ex vivo empirical data. However, with the advent of MR elastography [42] and MR diffusion tensor imaging [43], this may not be a long-term limitation. Nevertheless, our strategy of optimizing the data-model match by varying material properties [36], [37] has proven to be quite successful in an animal system and remains a viable avenue in the human case as well, provided a method of calibration can be found, such as intraoperative MR. Although the approach we have described is not as conceptually appealing as whole brain imaging using intraoperative MR, the method is promising, relatively inexpensive, and very effective at maximizing the use of preoperative data when the brain experiences significant intraoperative deformation. Even when intraoperative MR imaging is available, computational model estimates of volumetric tissue displacement may still be useful, for example, as an intermediate updating path between full intraoperative imaging sessions as in the case of the twin-operating theater (i.e. surgery and imaging) or when preoperative information

cannot be duplicated in the OR as in the case of functional studies (i.e., fMRI).

In future work, we intend to move from the qualitative identification of cortical features on the video/MR to a more quantitative approach which uses more comprehensive intraoperative digitization. Further, measurements regarding subsurface motion are also being pursued, using registered intraoperative ultrasound in the OR [18]. Model complexity is being increased with the addition of specific neuroanatomical structures, such as the falx cerebri [44], which have been recently reported as affecting subsurface displacement distributions [11]. Additionally, we are developing instrumented retractors as a method of acquiring intraoperative data regarding the viscoelastic behavior of brain tissue.

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