

## Intraoperative brain shift prediction using a 3D inhomogeneous patient-specific finite element model

JINGWEN HU, M.S.,<sup>1</sup> XIN JIN, M.S.,<sup>1</sup> JONG B. LEE, PH.D.,<sup>1</sup> LIYING ZHANG, PH.D.,<sup>1</sup> VIPIN CHAUDHARY, PH.D.,<sup>2</sup> MURALI GUTHIKONDA, M.D.,<sup>3</sup> KING H. YANG, PH.D.,<sup>1</sup> AND ALBERT I. KING, PH.D.<sup>1</sup>

<sup>1</sup>Bioengineering Center, <sup>2</sup>Institute for Scientific Computing, and <sup>3</sup>Department of Neurosurgery, Wayne State University, Detroit, Michigan

**Object.** The aims of this study were to develop a three-dimensional patient-specific finite element (FE) brain model with detailed anatomical structures and appropriate material properties to predict intraoperative brain shift during neurosurgery and to update preoperative magnetic resonance (MR) images using FE modeling for presurgical planning.

**Methods.** A template-based algorithm was developed to build a 3D patient-specific FE brain model. The template model is a 50th percentile male FE brain model with gray and white matter, ventricles, pia mater, dura mater, falx, tentorium, brainstem, and cerebellum. Gravity-induced brain shift after opening of the dura was simulated based on one clinical case of computer-assisted neurosurgery for model validation. Preoperative MR images were updated using an FE model and displayed as intraoperative MR images easily recognizable by surgeons. To demonstrate the potential of FE modeling in presurgical planning, intraoperative brain shift was predicted for two additional head orientations.

Two patient-specific FE models were constructed. The mesh quality of the resulting models was as high as that of the template model. One of the two FE models was selected to validate model-predicted brain shift against data acquired on intraoperative MR imaging. The brain shift predicted using the model was greater than that observed intraoperatively but was considered surgically acceptable.

**Conclusions.** A set of algorithms for developing 3D patient-specific FE brain models is presented. Gravity-induced brain shift can be predicted using this model and displayed on high-resolution MR images. This strategy can be used not only for updating intraoperative MR imaging, but also for presurgical planning.

**KEY WORDS** • finite element model • brain shift • model-updated magnetic resonance imaging • presurgical planning

**B**RAIN shift during neurosurgery can significantly compromise the integrity of neuronavigational systems, which are currently based on preoperative images. Substantial intraoperative brain shift has been well documented in recent literature. Cortical displacement of 10 mm, due predominantly to gravity, has been reported by Roberts et al.<sup>13</sup> Nabavi and colleagues<sup>12</sup> have reported that a very extensive range of brain surface shift (0–50 mm) can occur throughout surgery and is mainly due to gravity. Subsurface deformation occurred in the vicinity of the resection and gravitationally above the surgical site.<sup>6,12</sup>

To improve surgical accuracy, MR imaging systems have been used to visualize intraoperative brain shift. However, the resolution of existing intraoperative MR imaging systems is relatively low, the associated costs are very high, and the current systems cannot be used as a presurgical

planning tool. Given these shortcomings, numerical models designed to estimate intraoperative brain shift are viable alternatives to improve surgical outcomes at a very low cost.

Finite element modeling is the most suitable method to estimate the extent of brain shift, because of its versatility in modeling detailed anatomical structures with multiple material compositions and complex loading and boundary conditions. By using the cortical surface, ventricles, and tumor deformations measured during surgery as boundary conditions, Ferrant and colleagues<sup>3,4</sup> have been able to calculate subsurface brain deformation via a linear elastic brain model. Note, however, that this method requires intraoperative MR imaging data to provide input. Similarly, Castellano-Smith et al.<sup>2</sup> have utilized ventricular shrinkage, obtained from intraoperative MR imaging studies, as boundary conditions in their patient-specific models, constructed using a nonrigid registration strategy. Although their method reduces the time needed to construct patient-specific FE models, intraoperative MR imaging data are still needed to calculate brain shift. Assuming brain shift to be mainly due to a loss of fluid content, Miga and associates<sup>10,11</sup> modeled the brain as a “porous medium” represent-

Abbreviations used in this paper: AC = anterior commissure; CSF = cerebrospinal fluid; CT = computed tomography; FE = finite element; MR = magnetic resonance; PC = posterior commissure.

ed by an elastic body containing interstitial fluid, without simulating the change in the CSF space during surgery. It is noteworthy that this research group simulated brain shift caused by surgical retraction and resection based on the same principle.<sup>11</sup> Skrinjar and associates<sup>14,15</sup> have developed a method of FE model mesh generation with only hexahedral elements and used the intraoperative information from a pair of stereo cameras as the boundary conditions to estimate subsurface brain shift. Both brain sinking and bulging were simulated in their study, and the short calculation time made this approach practicable in real-world neurosurgery.

In most of the aforementioned FE examples, tetrahedral elements were used to construct the model. Theoretically speaking, strains within a tetrahedral element are constant, rendering the mesh stiffer. Hence, more tetrahedral elements are needed to achieve the same accuracy attained with a handful of hexahedral elements. Moreover, the human brain is an inhomogeneous and highly complex structure. Inhomogeneous brain models have been described by Miga and colleagues.<sup>8,9</sup> They incorporated the falx cerebri into the FE model to predict brain shift and found as much as a 4-mm difference between cases with and those without a falx.<sup>8</sup> Data in that study demonstrated that brain substructures play important roles when predicting brain shift. Other than Miga and colleagues, no authors have offered detailed descriptions of the intracranial structures when calculating brain shift. Note also that CSF drainage may create voids for very compliant brain tissues to shift into. Therefore, we believe that the space between the pia and dura mater, which has not been considered previously, should be simulated in brain shift calculations. Lastly, most reported methods of patient-specific model development are not robust enough, according to a review by Viceconti and Taddei.<sup>17</sup> Note also that some authors<sup>2-4,8-11,14,15</sup> have provided FE model-updated MR images for visualization by neurosurgeons because standard FE results could not be easily interpreted without engineering training.

The objectives of the present study were as follows: 1) to derive a new algorithm to develop robust 3D patient-specific brain models with detailed anatomical structures by using membrane and hexahedral elements; 2) to predict gravity-induced brain shift using these models; and 3) to update the preoperative MR imaging data based on FE-predicted brain shift results. We believe that this strategy is not only suitable for updating intraoperative brain shift but also useful for presurgical planning.

## Materials and Methods

### Development of a Patient-Specific FE Model

Two patients with preoperative MR images and CT scans were selected retrospectively to demonstrate the algorithms used in the development of patient-specific FE meshes. For model-validation purposes, intraoperative MR images obtained immediately after patient positioning and immediately after dura opening were also retrieved from our databank. Figure 1 shows the new algorithm devised to construct patient-specific models. The CT images were used to identify the dura mater, whereas MR images were used to outline the pia mater so that the CSF space could be simulated accurately. To form the patient-specific FE model, a baseline FE brain model (Fig. 2), previously developed by Zhou et al.,<sup>19</sup> was deformed to conform to patient-specific geometry after coregistration of the baseline model with the CT and MR images.

The basic geometry of the baseline model was taken from an atlas used by McGrath and Mills<sup>7</sup> and represents a 50th percentile male human brain. It includes detailed descriptions of the dura, pia, falx, tentorium, CSF, ventricles, cerebrum (gray and white matter), cerebellum, and brainstem. Only four-node shell/membrane and eight-node brick elements were used in this model. The entire model consisted of 14,297 nodes, 9824 solid elements, and 6578 shell/membrane elements. The length, width, and height of most of the elements ranged from 2.5 to 5 mm. In Fig. 2, the left upper hemisphere of the brain model has been removed to display the relative position and connectivity of the various components of the brain.

We utilized a landmark-based registration method that was established using an in-house toolkit developed with Scilab (ENPC; INRIA). We first rigidly registered the baseline model according to segmentation data to reorient and scale the baseline model so that it closely approximated preoperative imaging data. This step reduces the complexity of deforming the baseline model to fit the patient-specific geometry. Eight Talairach reference points—the AC; the PC; posterior to the PC (PPC); and anterior to (AAC), left of (LAC), right of (RAC), superior to (SAC), and inferior to (IAC) the AC—were selected as landmarks on both the model and preoperative images. To match the mesh of the baseline model to the segmented surface, the dura and pia mater of the former were projected onto the segmented dura and pia surfaces, respectively, using the “projection” function in HyperMesh (Altair, Troy, MI). The deformation vectors obtained through this process were used as boundary conditions (prescribed displacements) to deform the baseline model into the patient-specific model.

### Material Laws and Properties

To study brain shift, accurate in vivo material properties of the various anatomical features of the brain are needed. Unfortunately, these properties are largely lacking. Although a number of experimental studies have been conducted to obtain brain tissue material properties, in vivo material properties of human tissues are generally not available. Consequently, assumptions or approximations are needed in the development of such a model. Additionally, experimentally measured material properties of biological tissues generally possess a wide range of variation in terms of their stiffness and strength. Therefore, the selection of these material constants was somewhat subjective. Three material models were selected from the material library of the FE solver to approximate brain behavior needed in this study. Isotropic elastic shell/membrane elements were used to represent the pia and dura mater, falx, and tentorium. Ventricles were modeled using isotropic elastic solid elements with a fluid option. A Zener model, which is a representation of linear viscoelastic materials, was used to model the gray and white matter, brainstem, and cerebellum. According to the theory manual provided by the LS-DYNA3D developer, the shear relaxation behavior in a Zener model (Fig. 3) is described by Equation 1,<sup>5</sup>

$$G(t) = G_{\infty} + (G_0 - G_{\infty}) \cdot e^{-\beta \cdot t} \quad (\text{Eq. 1})$$

where  $G_0$  is the short-term shear modulus,  $G_{\infty}$  is the long-term shear modulus,  $\beta$  is the decay constant, and  $t$  is the duration. The deviatoric stress rate,  $\dot{\sigma}_{ij}'$ , depends on the shear relaxation modulus and is given by Equation 2,

$$\dot{\sigma}_{ij}' = 2 \int_{\tau=0}^t G(t - \tau) \dot{D}_{ij}(\tau) d\tau \quad (\text{Eq. 2})$$

where  $\dot{D}_{ij}$  is the deviatoric velocity strain tensor. The deviatoric strain is approximated by a piecewise constant function. The material responds to each step function following the relaxation law.

Material properties used in this study were taken from the published literature. The skull was assumed to have no deformation and was not simulated to reduce computational time. To simulate brain shift without the skull, the dura acted as the outermost boundary to restrict any brain motion beyond it. A Young modulus of 8.0 GPa, a thickness of 1.0 mm, and a Poisson ratio of 0.22 were assumed for the combined dura-skull layer. A bulk modulus of 2.19 GPa, the same as that of water and similar to that used by Zhang et al.,<sup>18</sup> was used to simulate the nearly incompressible behavior of brain tissue.

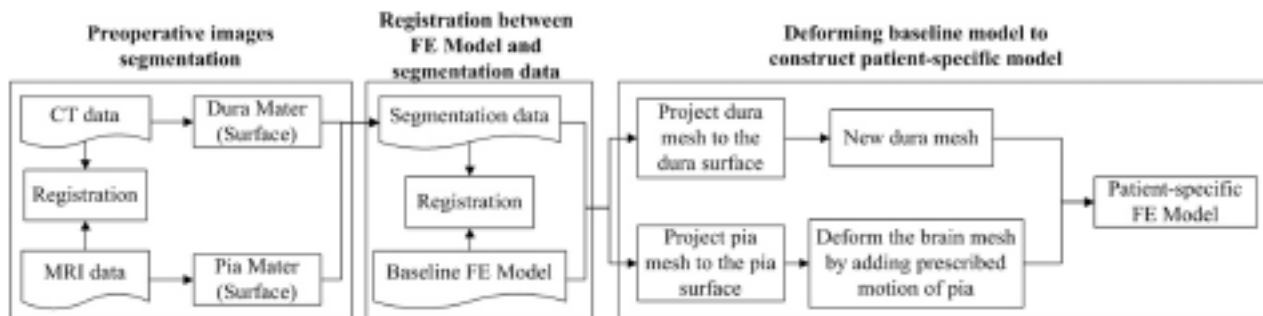


FIG. 1. Flow chart showing the patient-specific FE model generation algorithm.

Arbogast and Margulies<sup>1</sup> reported a short- and long-term shear modulus of 1.04 and 0.68 kPa, respectively, for the white and gray matter of the porcine brain. Additionally, they reported that the shear modulus of the brainstem was higher than that of other brain tissues tested. In their in vitro study, however, they could have underestimated the tethering and pressurization effect of blood vessels.<sup>16,19</sup> Consequently, shear moduli used in the model to represent the brain tissues were assumed to be higher than those reported and are listed in Table 1.

The ventricles were modeled using isotropic fluid elements based on the theory manual provided by the software developer. A bulk modulus of 2.19 GPa, a viscosity coefficient of 0.2/second, and a shear modulus of zero were assumed.<sup>18</sup> The Young modulus, Poisson ratio, and thickness selected for the pia mater were 20 kPa, 0.45, and 0.2 mm, respectively. Additionally, the Young modulus and Poisson ratio of the falx and tentorium were assumed to be 31.5 MPa and 0.45, respectively.

*Numerical Simulations*

Total drainage of the CSF between the dura and pia mater was assumed while no changes were made to the ventricles. The volume of the intervening trabeculae was neglected. Contact interfaces were defined between the dura and pia mater to simulate the CSF drainage. Gravitational force was applied according to the orientation of the patient. For the case simulated in this study, the model was rotated 45° to the patient’s right with fixed boundary conditions assumed for the dura/skull to simulate firm fixation of the patient’s head on the operating table. The model was executed to predict the magnitude and location of brain shift using LS-DYNA3D. After comparing the simulation results with the intraoperative MR imaging data,

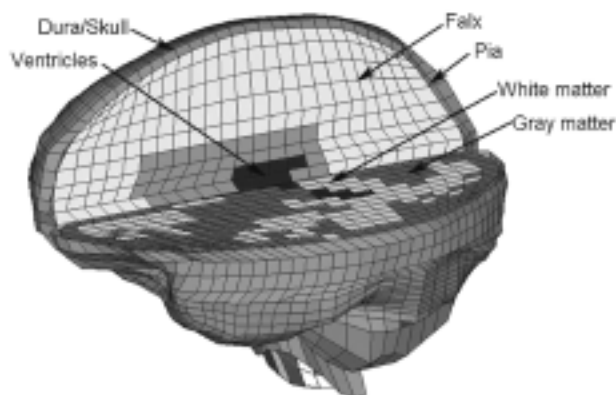


FIG. 2. Oblique view of the baseline FE human brain model featuring detailed anatomical structures, including gray and white matter, ventricles, pia mater, dura mater, falx cerebri, tentorium cerebelli, brainstem, and cerebellum. The left upper hemisphere of the brain model has been removed to display the relative position and connectivity of the various components of the brain.

the model was used to predict the magnitude and location of brain shift by assuming that the patient was oriented supine and in a side-facing position.

*Intraoperative Image Model-Updating Algorithm*

A set of algorithms was developed to update preoperative images based on FE-calculated brain shift (Fig. 4). First, displacement fields for the entire brain were calculated using the FE model. The displacement for each voxel of the MR images was then interpolated from the model-predicted displacements. Finally, the position and gray scale of each voxel of the preoperative images was translated and interpolated, respectively, to new voxel positions and a new gray scale as an MR image at dura opening. Note that the patient-specific FE model was developed based on segmentation data; therefore, it was not necessary to do the model–image registration again.

**Results**

Two patient-specific models have been developed using algorithms described in *Materials and Methods* and shown in Fig. 5. Note that only one layer of the solid elements in each patient-specific model is shown in a similar position. White, gray, and black represent the white matter, gray matter, and ventricles, respectively. Even though the two patients selected for this study were very different in terms of head geometry, the algorithms used were able to accommodate the differences and generate high-quality meshes for FE analysis. Note that the shape of the ventricle can be improved if the resolution of the baseline model is increased.

Panels a and b of Fig. 6 show the model-predicted brain shift. The lighter grid lines represent the preoperative brain, whereas the solid, darker mesh indicates the displaced brain (Fig. 6a). Displacement contours of the brain are shown in Fig. 6b; the larger brain shifts are depicted by the darker contours. The intraoperative MR imaging data before and after craniotomy are shown in Figs. 6c and 6d, respectively. A slice of the preoperative MR image and FE model–updated MR image at dura opening are featured in Fig. 6e and f,

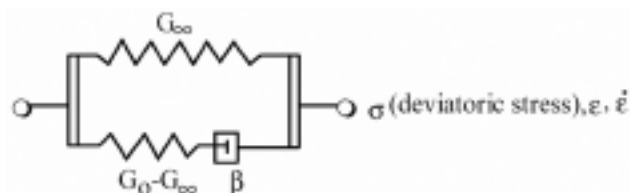


FIG. 3. Schematic depicting a linear viscoelastic material model (Zener model) available in LS-DYNA3D.

# Brain shift prediction using a 3D patient-specific model

TABLE 1

Viscoelastic material properties used in the brain model\*

Structure	Shear Modulus (kPa)		Bulk Modulus (GPa)
	$G_0$	$G_\infty$	
gray matter	6.00	1.20	2.19
white matter	7.50	1.50	2.19
brainstem	13.50	2.70	2.19
cerebellum	6.00	1.20	2.19

\* The density of each structure was 1040 kg/m<sup>3</sup>, and the decay constant was 200 msec<sup>-1</sup>.

TABLE 2

Brain shift results of intraoperative imaging and FE model calculations\*

Data	Landmark					
	1	2	3	4	5	6
MRI data	2.6	6.2	0.9	2.0	1.0	6.2
FE model results	2.5	7.3	1.4	4.1	3.1	6.5
error	1.4	1.5	0.5	2.2	2.2	2.3

\* All values are in millimeters.

respectively. The light-colored lines distinguish the boundaries of the brain before and after dura opening to highlight the magnitude of the brain shift. Six anatomical landmarks, which can be identified on both preoperative and intraoperative MR images, were manually selected for quantitative validation purposes. Table 2 shows the landmark displacements calculated from intraoperative MR images and FE model simulations. The maximum brain shift based on intraoperative MR imaging data was 6.2 mm, whereas the maximum brain shift calculated using the FE model was 7.3 mm. For most of the landmarks, the brain shifts predicted using the FE model were a little larger than those predicted with the intraoperative MR imaging data, and the maximum error was 2.3 mm.

## Discussion

A technique, similar to the “template matching methods” described by Viceconti and Taddei<sup>17</sup> has been developed to construct patient-specific FE models easily and automatically by deforming the baseline model to conform to the geometry of several key landmarks identified from preoperative images. The newly added procedure of rigid registration between the baseline model and preoperative images before template-matching made for a more robust strategy in models with complex geometries. We used the deformation field produced by the projection of surface nodes instead of the general grid-projection method for building patient-specific models to ensure high mesh quality during the presimulation phase. For a typical model, fewer than 2 hours were needed to deform the baseline model to construct a patient-specific model on a dual-processor 2.0-GHz central processing unit workstation. Further reductions in time can be achieved if a parallel computing strategy is used. Additionally, the main strength of the proposed method was in predicting brain shift prior to the actual surgery. Thus, the procedure can be used as a presurgical planning tool by surgeons.

For the two cases tested using this algorithm, the mesh quality was similar to that of the original model. However,

it is not clear if mesh quality will remain high in extreme cases. The deformation from the baseline to the patient-specific model was established on the principle of minimizing the total strain energy. As long as equilibrium can be achieved, the final mesh will be good quality. Moreover, the resulting patient-specific model consisted of only computationally efficient four-node quadratic and eight-node hexahedral elements and includes critical anatomical structures such as the gray and white matter, ventricles, cerebellum, brainstem, falx, tentorium, pia mater, and dura/skull to improve the accuracy of the predicted brain shift.

As expected, the brain shift predicted using this method was slightly higher than that predicted using intraoperative MR imaging because total drainage of the CSF was assumed. In other words, our model predictions represent the maximal brain shift that could occur. However, results calculated from this study are still very useful. First, opening the dura is a relatively long process, thus making our model predictions relevant. Second, a viscoelastic material was used to represent various cerebral structures. The short- and long-term shear moduli as well as the decay constant (Table 1) of the viscoelastic model could account for continued brain deformation over time. In other words, as long as the time after dura opening is known, our model can calculate the deformation due to viscoelastic creep. In a patient, the amount of CSF drainage depends on the incision size, the time elapsed after dura opening, and whether drugs are used to reduce CSF volume. Future studies need to include the time history of CSF drainage to better predict brain shift. Furthermore, the mass density, shear modulus, bulk modulus, and decay constant of the brain tissues play important roles when calculating gravity-induced brain shift. Although the material constitutive laws and material properties used in this study were consistent with those in the published literature, the reader must be aware that these reported properties have varied greatly and generally were obtained from nonhuman in vitro specimens. For these reasons, no other cases were selected for model validations. Other studies, such as those conducted by Ferrant and associates,<sup>3,4</sup> Castellano-Smith et al.,<sup>2</sup> and Skrinjar and colleagues,<sup>14,15</sup> also involved a limited number of cases for model validation.



FIG. 4. Flow chart revealing the intraoperative image model-updating algorithm.

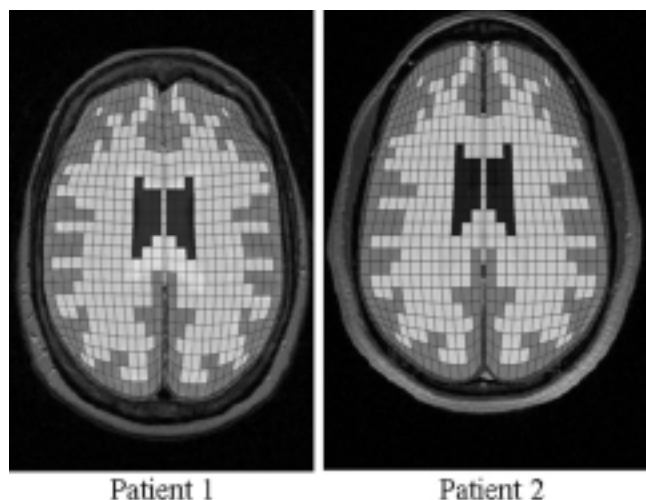


FIG. 5. Patient-specific meshes obtained in two patients, demonstrating one layer of the solid elements in each patient-specific model. White, gray, and black represent the white matter, gray matter, and ventricles, respectively.

Figure 7 demonstrates the magnitude of the deformation and the deformed shape of the brain in one patient, which was predicted by the patient-specific FE model for two other head orientations. As the patient orientation changed, there was a significant change in the location and magnitude of the model-predicted brain shift due to changes in the direction of the gravitational force.

The method developed to produce updated MR images

based on model simulation results addresses the critical issue raised by some surgeons that FE models are not useful to them. We believe that providing model-updated pseudo-MR images is a critical step in convincing neurosurgeons that FE models are useful in presurgical planning.

A couple of limitations of this study should be noted. First, the material properties used in our model were based on data obtained in *in vitro* human or animal specimens. Unfortunately, *in vivo* material properties are not available, posing a significant challenge in using the FE modeling technique to predict brain shift. Future studies should be designed to obtain relevant data from volunteers or patients to overcome this shortcoming. Second, in the current study we only focused on predicting the magnitude and location of brain shift immediately after dura opening. No effort was made to determine the magnitude of brain shift after tumor resection, which may be more clinically relevant. Although the FE method is capable of simulating tumor resections performed using various techniques, it was not used in the current study because of the unavailability of biomechanical properties of various kinds of brain tumor. Nevertheless, our method provides a reasonable estimate of gravity-induced brain shift after dura opening, which accounts for the greater part of brain shift before tumor resection<sup>6,9,12,13</sup> and can be used for presurgical planning. Moreover, even with the tumor resection, the subsurface deformation still took place primarily in the area gravitationally above the surgical site.<sup>12</sup> Thus, for the current stage, we considered only gravity as the major factor causing brain shift. More work is needed to measure the material properties of different types of brain tumor and to incorporate patient-specific tumors into the FE brain model to simulate tumor resection.

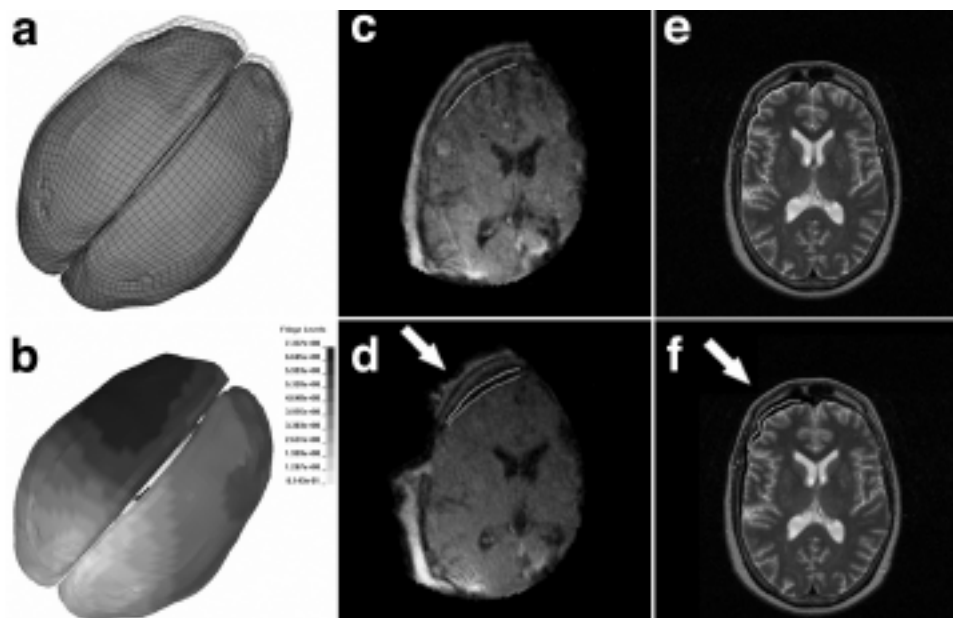


FIG. 6. Images featuring a comparison between numerical simulations and intraoperative MR imaging data. Arrows indicate directions of gravity. a: An FE model-predicted brain shift. The *light-colored mesh* is the preoperative brain, and the *solid, darker mesh* is the deformed brain. b: An FE model-predicted brain shift contour. c: Intraoperative MR imaging data immediately after patient positioning. d: Intraoperative MR imaging data immediately after dura opening. e: Preoperative MR image. f: An FE model-updated MR image.

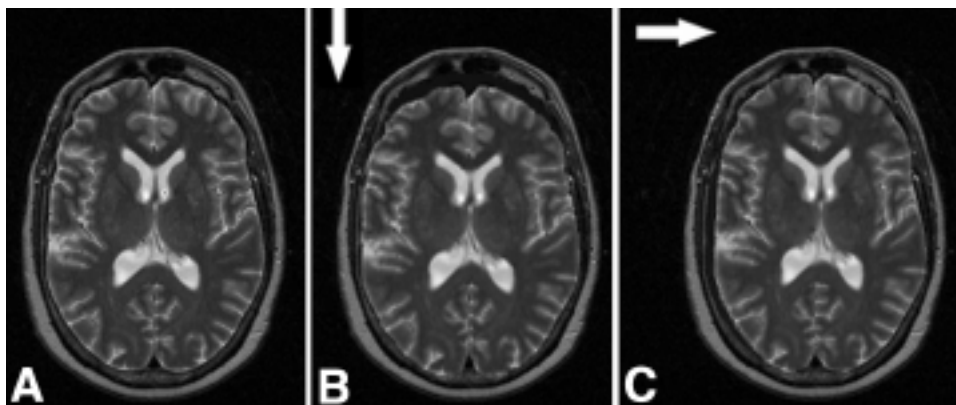


FIG. 7. Model-updated images featuring different patient orientations. Arrows indicate the directions of gravity. Preoperative (A) and model-updated (B) MR images obtained while the patient was supine. Model-updated MR image (C) obtained while the patient was on right side.

### Conclusions

In summary, two 3D patient-specific FE models were developed to predict brain shift due to gravity after dura opening. This method differs from those in previous studies because coregistration between a baseline FE model and preoperative images was conducted first before a patient-specific model was generated from CT and MR images. The model consisted of major anatomical structures of the brain; utilized separate material properties for different structures; and defined the contact between the pia and dura so that the brain is deformed and shifted into the void left by CSF drainage. One clinical case was used to validate the FE model. Model-predicted brain shift matched well with intraoperative MR imaging data. Results predicted by the FE model were further used to update preoperative MR images for two other head orientations to demonstrate the clinical versatility of the method.

### References

1. Arbogast KB, Margulies SS: Regional differences in mechanical properties of the porcine central nervous system. SAE Paper No. 973336, in **41st Stapp Car Crash Conference**. Warrendale, PA: Society of Automotive Engineers, 1997
2. Castellano-Smith AD, Hartkens T, Schnabel J, Hose DR, Liu H, Hall WA, et al: Constructing patient specific models for correcting intraoperative brain deformation. **MICCAI LNCS 2208**: 1091–1098, 2001
3. Ferrant M, Nabavi A, Macq B, Black PM, Jolesz FA, Kikinis R, et al: Serial registration of intraoperative MR images of the brain. **Med Image Anal 6**:337–359, 2002
4. Ferrant M, Warfield SK, Guttman CRG, Mulkern RV, Jolesz FA, Kikinis R: 3D image matching using a finite element based elastic deformation model. **MICCAI LNCS 1679**:202–209, 1999
5. Hallquist JO: **LS-DYNA Theoretical Manual**. Livermore, CA: Livermore Software Technology Corporation, 1998
6. Maurer CR Jr, Hill DLG, Martin AJ, Liu H, McCue M, Rueckert D, et al: Investigation of intraoperative brain deformation using a 1.5-t interventional MR system: preliminary results. **IEEE Trans Med Imaging 17**:817–825, 1998
7. McGrath P, Mills P: **Atlas of Sectional Anatomy: Head, Neck and Trunk**. Basel, Switzerland: Karger, 1984
8. Miga MI, Paulsen KD, Kennedy FE, Hartov A, Roberts DW: Model-updated image-guided neurosurgery using the finite element method: incorporation of the falx cerebri. **MICCAI LNCS 1679**:900–909, 1999
9. Miga MI, Paulsen KD, Kennedy FE, Hoopes PJ, Hartov A, Roberts DW: In vivo analysis of heterogeneous brain deformation computations for model-updated image guidance. **Comput Methods Biomech Biomed Engin 3**:129–146, 2000
10. Miga MI, Paulsen KD, Lemery JM, Eisner SD, Hartov A, Kennedy FE, et al: Model-updated image guidance: initial clinical experiences with gravity-induced brain deformation. **IEEE Trans Med Imaging 18**:866–874, 1999
11. Miga MI, Roberts DW, Kennedy FE, Platenik LA, Hartov A, Lunn KE, et al: Modeling of retraction and resection for intraoperative updating of images. **Neurosurgery 49**:75–85, 2001
12. Nabavi A, Black PML, Gering DT, Westin CF, Mehta V, Pergolizzi RS Jr, et al: Serial intraoperative magnetic resonance imaging of brain shift. **Neurosurgery 48**:787–798, 2001
13. Roberts DW, Hartov A, Kennedy FE, Miga MI, Paulsen KD: Intraoperative brain shift and deformation: a quantitative analysis of cortical displacement in 28 cases. **Neurosurgery 43**:749–760, 1998
14. Skrinjar O, Nabavi A, Duncan J: Model-driven brain shift compensation. **Med Image Anal 6**:361–373, 2002
15. Skrinjar O, Studholme C, Nabavi A, Duncan J: Steps toward a stereo-camera-guided biomechanical model for brain shift compensation. **IPMI LNCS 2082**:183–189, 2001
16. Viano DC: Biomechanics of non-penetrating aortic trauma: a review. SAE Paper No. 831608, in **27th Stapp Car Crash Conference**. Warrendale, PA: Society of Automotive Engineers, 1983
17. Viceconti M, Taddei F: Automatic generation of finite element meshes from computed tomography data. **Crit Rev Biomed Eng 31**:27–72, 2003
18. Zhang L, Yang KH, Dwarampudi R, Omori K, Li T, Kun C, et al: Recent advances in brain injury research: a new human head model development and validation. **Stapp Car Crash J 45**: 369–393, 2001
19. Zhou C, Khalil TB, King AI: A new model comparing impact responses of the homogeneous and inhomogeneous human brain, SAE Paper No. 952714, in **39th Stapp Car Crash Conference**. Warrendale, PA: Society of Automotive Engineers, 1995

Manuscript received July 27, 2005.

Accepted in final form June 27, 2006.

This research was supported by Grant No. MEDC-459 from the Michigan Life Sciences Corridor.

Address reprint requests to: Jingwen Hu, 818 W Hancock, Bioengineering Center, Wayne State University, Detroit, Michigan 48201. email: ap7826@wayne.edu.