Incorporation of Surface–based Deformations for Updating Images 
Intraoperatively
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ABSTRACT
Patient–to–image misalignment becomes exacerbated by common surgical events such as brain sag, drug interactions, retraction, and resection. One strategy to remedy this mis–registration is to employ computational models in conjunction with low–cost intraoperatively–acquired data (e.g. surface tracking, and co–registered ultrasound) to deform preoperative imaging data to account for OR actions. In this paper, we present preliminary data from a cortical surface scanning system and study the impact of surface–based information on model–updates. Preliminary data is presented using a three–dimensional laser scanning technology in conjunction with an iterative closest point (ICP) algorithm to register and track phantom and ex vivo data. Simulations are presented to analyze the direct use of displacement data versus modeling the underlying physical load in a clinical example of gravity–induced deformation. Results demonstrate dramatic differences in subsurface deformation fields highlighting that the nature of the surgical load (i.e. surface or body force) must be thoughtfully discriminated to accurately update images. Furthermore, the results suggest that the application of surface displacements to update image volumes must be consistent with the physical origin of deformation rather than applied in a direct interpolative sense.

Keywords: Brain shift, surface deformation, intraoperative data, finite element, non–rigid registration

1. INTRODUCTION

The realization of co–registration among patient, OR space, and preoperative image series has become somewhat commonplace in the neurosurgical theatre. As can be expected, improved computer–based surgical systems are increasingly on the horizon and their integration into treatment and reliance on for navigational assistance has made this technology essential. The major technical issues addressing image–guided surgery are the complexity and degree of feedback that characterize these advanced systems. In addition, the correct balance among cost, practicality, versatility and feedback becomes an important concern in the development of these systems and numerous approaches are being investigated.

One of the most challenging problems in the pursuit of accurate image–guided surgical systems has come to light in the past five years. The foundation for many of the systems developed in the 80’s and 90’s was the reliance on transformations between coordinate spaces that assumed objects were rigid structures. Although fixed–body registration may be appropriate in the early stages of surgery, the neurosurgical community has largely acknowledged that intraoperative tissue deformation may make these transformations inadequate for accurate guidance throughout surgery. Independent investigations have revealed that during surgery the brain can deform significantly and that preoperative–based image guidance may be compromised if these movements are not taken into account [1–6]. For example, one detailed study documenting both surface and subsurface shift was recently reported by Nimsky et al. [6]. Using image processing techniques and the shift scale proposed by Bucholz et al. [7] (low, 0–2.9 mm; moderate, 3.0–6.9; high, > 7.0 mm), they classified surface and subsurface deformation on 64 patients undergoing tumor resection. This investigation revealed 64% of all cases experienced brain shifts greater than 7 mm. With respect to the deep tumor margin, low shifts were experienced in 34% of cases, moderate shifts in 42%, and high shifts in 24% [6].

In response to these data, there has been a concerted effort to account for intraoperative shifting brain structures during surgery. Several medical centers are now deploying intraoperative magnetic resonance (iMR) imaging [6, 8–10] and are developing sophisticated methods of visualization in the OR [6,11]. Although conceptually appealing, the exorbitant cost and cumbersome nature of such systems have left their widespread adoption unclear at this time [12]. Alternative imaging strategies such as co–registered intraoperative ultrasound (iUS) are also under investigation [7, 13–14]. Although not capable of whole–volume imaging, the locally reconstructed image volumes generated by iUS can provide real–time guidance feedback during surgery. The quality of that feedback during surgery is still being assessed.
In the recent study by Nimsky et al. [6], they have attributed the failure of complete tumor removal in preoperative-based neuronavigational systems to intraoperative brain shift and that iMR can provide the essential feedback to eliminate residual tumor. As an alternative to whole-brain MR imaging, it seems that an equally elegant solution could be provided by the integration of low-cost readily available intraoperative data in a framework which updates the preoperative image database. In this approach, the solution fidelity of computational models in conjunction with non-intrusive intraoperative data acquisition could serve as a means for updating all high resolution preoperative-based spatially encoded data (i.e. positron emission tomography, electroencephalography, functional MR imaging, and MR spectroscopy) to reflect current OR conditions [15]. Preliminary studies with retrospective updating of gravity-induced brain shifts have shown that computational models can capture 70–80% of cortical surface deformations [16]. A more quantitative validation with subsurface data was performed in a series of repeat-experiments in a porcine system and demonstrated similar accuracy [17].

This paper reports on preliminary experiences using surface scanning technology in the context of registering and tracking the brain. Additionally, a simulation study is reported which tested the validity of applying surface deformations to a model-updating framework in a clinical case. Our preliminary experience with topographic scanning technology indicates comparable accuracy to more traditional point-based techniques. Our experiments designed to simulate OR-like conditions suggest that surface scanning is capable of cortical-surface tracking during surgery. Simulation studies show that variations in the application of surface displacement data can result in tumor boundary discrepancies of approximately 3–4.5 mm among the different surface-data deployment strategies.

2. PRELIMINARY DATA

In this section, preliminary data representative of three-dimensional surface scanning systems is reported with some qualitative measures of accuracy. The major objective for employing these scanner systems is to provide an abundance of data to both register and drive model computations. Using the detailed spatial information provided by the scanner, algorithms to perform patient-to-image registration and deformation tracking are being developed. The tracked deformation can then be used to constrain the computational model to more accurately reflect the surgical field environment.

Surface-based registration has been an alternative to fiducial registration for several years. The predominant problem with surface versus fiducial or implantable fiducial registration is the lack of significant geometric features on the scalp or the tendency of those surfaces to deform. However, using the geometric patterns associated with the outer brain, it seems apparent that a unique and very accurate registration methodology can be developed. Using features from the cortical surface to register does have precedent. Nakajima et al. demonstrated an average of 2.3 +/- 1.3 mm fiducial registration error using cortical vessels for registration [18]. More recently, Nimsky et al. reported using a deformable surface approach to quantify surface shifts using a variation on the iterative closest point (ICP) algorithm [6]. Also, some preliminary work using scanning based system for cortical surface registration has been reported and has demonstrated encouraging results [19].

<table>
<thead>
<tr>
<th>Geometry Measure</th>
<th>Scanner (Polaris)</th>
<th>Tracking Measure</th>
<th>Average Error (max)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Diagonal Measures</td>
<td>43.0, 43.1 mm (45.5, 44.5 mm)</td>
<td>Dx Tracking</td>
<td>0.3 (0.5) mm</td>
</tr>
<tr>
<td>Square Lengths</td>
<td>29.9 x 30.4 mm (31.6 x 31.4 mm)</td>
<td>Dy Tracking</td>
<td>0.8 (1.0) mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Dz Tracking</td>
<td>0.5 (0.8) mm</td>
</tr>
</tbody>
</table>

Figure 1. 3D RealScan setup for generating surface data from blocks.

Table 1. Geometry (columns 1,2) and tracking (columns 3,4) measures for comparison between scanning and stylus-based digitization.

In an effort to ascertain the potential capability of topographical scanning technologies, a limited series of experiments is presented to test various aspects of one particular laser scanning system (RealScan 3D, 3D Digital Corporation, Bedford Hills, NY). In the first experiment, small blocks with raised lettering were scanned (Figure 1) to
To ascertain some measure of registration accuracy, the cube phantom surfaces generated in the pre–post displaced positions (Figure 2c) were registered using the ICP algorithm. This transformation matrix was then compared to the transformation provided by the Polaris tracking system. The fiducial registration error between the transformations was approximately 0.5 +/− 0.4 mm. At a radius of approximately 5 and 10 cm from the cubic surfaces, the average target localization error between the two rigid transformations was 0.6 +/− 0.2, and 0.9 +/− 0.3 mm, respectively. Figure 3 shows the combined surfaces of the pre–and–post translated surface when using the transformation provided by the ICP algorithm (notice the integrity of the surfaces).
of the letter on the top surface, i.e. no blurring due to misregistration).

However, another important step in determining efficacy of such a system is to generate accurate topographic maps of surfaces found in the OR, i.e. the brain. In a second experiment, data was acquired from an ex vivo intact human brain (Figure 4a) in an effort to test the scanner’s ability to resolve sulcal patterns. The detail in surface Figure 4b is impressive and sulcal patterns are easily distinguished (note that the brain had glistening areas which did not affect the laser−based reconstruction). To add further realism, a 10 cm diameter model craniotomy (Figure 5a) was placed over the brain with the scanned geometry results shown in Figure 5b. Figure 5b demonstrates the initial quality in which the brain surface was captured relevant to the existence of the craniotomy. It should be noted that some artifact at the craniotomy/brain interface is present and is due to a line−of−sight requirement which would be expected in any scanning technology. To simulate tracking of the cortical surface, the brain was translated away from the craniotomy opening as could happen in gravity−induced brain shift. Figure 5c illustrates a side view overlay of the pre−post translated brain surface. The results indicate a clear translation of the brain surface away from the craniotomy while maintaining the sulcal resolution found in Figure 5b.

The registration analysis and techniques highlighted in these studies are focused on rigid body registration and in some sense are inappropriate for situations involving substantial brain deformation (unless iMR or iCT is being performed where re−registration of the deformed image volume can be achieved). Based on the results in Figures 1−5, registering and tracking the cortical surface using laser−based scanning technology seems viable and in cases where minimal deformation has occurred may be amenable to fixed body registration techniques. Our goal is to develop a single registration system capable of alignment initialization using scalp−based features followed by further spatial refinement provided by the cortical surface. For proof of concept, a phantom head (Figure 6a) was used as the source for initial rigid body registration. Figure 6b shows the results from the scanner and Figure 6c shows the surface rendering from CT (surface rendered in Analyze AVW, Mayo Clinic, Rochester, MN). Using a standard iterative closest point algorithm [20], Figure 7a illustrates the registration of these surfaces while Figure 7b demonstrates the registration of the cortical surfaces shown in the simulated craniotomy experiment (Figure 5).

3. METHODS

The general framework presented by Roberts et al. is an excellent starting point for developing image−guided systems augmented by model−based intraoperative reconstructions for enhanced navigation [15]. The methodology is centered around a three−dimensional model of the brain using a biphasic porous media description [21]. Given intraoperative data, boundary conditions are applied to the patient−specific model of the brain and a complete volumetric field of displacements are produced and subsequently used to register all forms of preoperative−based data. In principle, any source of intraoperative data may be used for the deformation process, iMR, iCT, iUS, digitization data, etc. However, if one could use sparse intraoperative data (i.e. small data sets) and still provide adequate navigation, a cost−effective solution to the brain shift problem could be formulated which would be minimally invasive to an already over−crowded OR environment.
The model used in this paper is based on consolidation physics and was first presented in the context of finite element and brain deformation mechanics by Nagashima et al. [22]. This work concentrated on modeling brain pathophysiology, specifically vasogenic edema, and was validated qualitatively in a feline experiment. This framework was extended by Paulsen and Miga et al. to three-dimensional calculations and validated in a series of repeat-experiments in a porcine system subjected to loads comparable to neurosurgical applications [17,23]. The equations describing the model for brain deformation are:

\[ \nabla \cdot G \nabla u + \nabla \left( \frac{G}{1-2\nu} \nabla \cdot u - \alpha \nabla p \right) = (\rho_f - \rho_t) g \quad (1) \]

\[ \alpha \frac{\partial}{\partial t} (\nabla \cdot u) + \frac{1}{S} \frac{\partial \rho}{\partial t} - \nabla \cdot k \nabla p = 0 \quad (2) \]

where \( G \) is the shear modulus \((G=E/(1+2\nu))\) where \( E \) is Young’s Modulus, \( \nu \) is Poisson’s ratio, \( \alpha, 1/S \) are saturation constants, \( k \) is the hydraulic conductivity, \( \rho \) is density with \( f \) and \( t \) corresponding to fluid and tissue respectively, and \( g \) is gravitational acceleration. The dependent variables are displacement in the x, y, and z direction represented by \( u \) and the interstitial pressure \( p \).

For the calculations presented in this paper, it was desirable to understand the implications of applying intraoperative data directly to drive model calculations in lieu of applying a load in a physically consistent manner. In some sense, this paper represents a comparison between using model-based calculations in a pure interpolative manner versus trying to model the true physical events impacting the continuum. To demonstrate the difference in these approaches, a series of calculations has been performed under three specific load applications which elucidate why the need for physically consistent boundary conditions is equally important as to having a representative mathematical description of brain deformation.

Given the prominence of gravity-induced deformation in the literature [1–6], it was chosen as the starting point for comparison. The clinical example involved a 54-year old woman who had developed a large contrast-enhanced right frontal lobe mass. The patient was brought to surgery supine, secured in three-point fixation, and with head in the neutral position (gravity acting anterior to posterior, i.e. right to left in Figure 8). During the case,

<table>
<thead>
<tr>
<th>Tissue Type</th>
<th>Young’s Modulus (E), Poisson’s ratio</th>
<th>Hydraulic Conductivity, Density</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gray matter</td>
<td>4000 Pa, 0.45</td>
<td>5x10^-12 m^3/s/kg, 1x10^3 kg/m^3</td>
</tr>
<tr>
<td>White Matter</td>
<td>2100 Pa, 0.45</td>
<td>1x10^-10 m^3/s/kg, 1x10^1 kg/m^3</td>
</tr>
<tr>
<td>CSF (ventricles)</td>
<td>100 Pa, 0.49</td>
<td>1x10^-7 m^3/s/kg, 1x10^3 kg/m^3</td>
</tr>
<tr>
<td>Tumor</td>
<td>21000 Pa, 0.45</td>
<td>5x10^-12 m^3/s/kg, 1x10^3 kg/m^3</td>
</tr>
</tbody>
</table>

Table 2. Material properties for brain tissue. For all runs \( \alpha=1, L/S=0, \Delta t=180 \) sec, \( t_{final}=1800 \) sec, \#nodes=24,484, \#elements=134,870.
three dimensional digitization technology was used to track cortical surface features in the direction of gravity. These points were later compared to a retrospective model calculation and were presented in [16]. For the computations presented here, a more refined model was generated which included heterogeneous brain structures, i.e. tumor and lateral ventricles as shown in Figure 8 (given the positioning of the brain, the falx was not modeled for this case). Further refinement was achieved by using an image-to-grid segmentation scheme to capture white-gray matter heterogeneity (Figure 9 right). The identical boundary conditions used in [16] were applied to the new model and used as the base calculation for comparison with subsequent alternative boundary conditions. Briefly described, the boundary conditions in [16] allow the brain to slip along the cranial cavity with areas surrounding the brain stem being constrained. With respect to the drainage of interstitial fluid, no drainage was allowed in regions below a user-specified cerebrospinal fluid (CSF) level while above this level, drainage to the surface was allowed. The principle mode of deformation is from the term on the right-hand-side of equation (1). In areas above the user-defined CSF level, the difference in density between tissue and fluid was that of tissue and air, respectively, while below, the density was between tissue and CSF which was assumed equal. This creates a volumetric body force term which deforms the entire geometry under the weight of the tissue above the user-defined CSF level. Using this calculation as the basis for comparison, a second calculation series was then performed in which the gravity-based body force was not prescribed, but rather, the surface deformations from the gravity-based computation were directly applied as displacement boundary conditions in the region of the craniotomy. The final calculation series for comparison used the same conditions as the second, except that lateral ventricular deformations were also prescribed as direct displacements simulating the ability to acquire subsurface deformation measurements using alternative imaging modalities, i.e. iUS or iMR. The material properties used for all simulations are reported in Table 2.

4. RESULTS

Figure 10 demonstrates the difference in subsurface shift in the direction of gravity among the different boundary conditions. The property-based corresponding sagittal slice is also shown to aid in visual reference to the MR volume (Figure 10 corner). In the gravity-induced shift distribution (Figure 10 left), we see a distinct displacement pattern with maximum movement occurring in areas of highest elevation with a gradual decrease in movement spatially associated with moving more posterior. With the surface-based application of displacement (Figure 10 middle), a markedly different distribution is seen with maximum shift located at the application area and displacement decreasing spatially associated with moving superior-to-inferior. In the surface/ventricular application (Figure 10 right), there is more subsurface displacement in the tumor region, yet outside of the region some noted differences are apparent, i.e. less shift in anterior regions. Figure 11 is a close-up of the tumor region (Figure 11 top left) with a planar displacement distribution shown for each case in Figure 10 (Figure 11 top right, bottom left, bottom right are planar displacement fields for gravity-induced shift, applied surface displacement, and applied surface/ventricle deformation, respectively). In Figure 11, the same observations from Figure 10 can be observed but a better appreciation for planar movement can be obtained. Figure 12 illustrates the segmented cortical surface (left) with the changes in cortical surface shape shown as a difference image resulting from the three simulations, i.e. changes in cortical surface shape induced by gravitational loading (Figure 12 middle-left), applied surface deformation (Figure 12 middle-right), and applied surface/ventricle displacements (Figure 12 right).
5. DISCUSSION

The preliminary data provided in Section 2 offers valuable insight into the power of laser−scanning technology. At a moderate cost, this digitization technology affords detailed sub−millimetric capturing of cortical surface subtleties which will ultimately lead to new registration approaches in the OR. The limited tests shown here demonstrate an
impressive ability to capture surface topography and that registrations performed with these surfaces are at least comparable to other 3D digitization technologies. The challenge for surface−based registration systems in the OR still remains in the development of fast iterative registration algorithms which can compete with the near−instantaneous point−based methods.

Figures 8 and 9 illustrate the remarkable detail to which computational models can now be expected to entail. Although the complete ventricular system has not been created, this is not a limitation of the mesh generation but more in keeping the size of the computations tractable. Given the biphasic nature of the brain, consolidation theory would seem an appropriate starting point for capturing brain tissue mechanics and the finite element method is an excellent choice for capturing the brain’s geometric complexities. As more complexity is needed (i.e. capturing sulcal patterns), the division between mesh generation and segmentation will be readily blurred as is already evident in the image−to−grid heterogeneity results shown in Figure 9.

The results in Figures 10 and 11 illustrate a compelling case for the necessity of understanding intraoperative loading conditions. In each figure, significantly different displacement gradients are present. With the gravity−induced shift calculation, Figure 10 (left) shows a gradually decreasing displacement field moving anterior−to−posterior. In the close−up in Figure 11 (top right), we see a somewhat uniform translation of the tumor backward towards the occipital lobe. By applying measured surface displacements solely (calculation 2) in lieu of gravitational forces, a dramatic difference in deformation can be observed (Figure 10 middle) with displacements decreasing in a superior−to−inferior direction (orthogonal to the gravity−induced gradient). In the close−up (Figure 11 bottom left), the model does capture surface movement but it’s accuracy decreases progressively more subsurface. Although this may be sufficient for early tumor resection, there are displacement discrepancies on the order 3−4.5 mm at the deep tumor margin which would be a significant error for image−guided navigation. By using measured surface movement and subsurface ventricular deformation, this error can be reduced as shown in Figure 10 (right) and 11 (bottom right). In both of these images, the added constraint of deforming the lateral ventricular structure has created displacement fields in the region of the tumor boundary similar to the gravity−induced calculation. However, there are some regions mid−tumor which have displacement errors on the order of 2 mm. In this particular surgery, this discrepancy would have little effect; but in the case of delivering a focal lesion to the brain, this could be important if this central region was the focus. In Figure 12, an appreciation for the amount of cortical shift predicted by each calculation can be observed. From this figure, gravity−induced shift is more uniform and larger in magnitude than the other calculations. Constraining the ventricular system to move in accordance with gravity−induced shift markedly improves the calculation and significantly improves the capture of anterior−to−posterior movement.

6. CONCLUSIONS

The improvement of image−guided surgical systems to account for surgically−induced brain deformations has important implications for the performance of difficult surgical techniques. Digitization technology, such as that provided by the scanner in Section 2, affords rich yet limited data sets regarding topographic features and progressive distortion due to intraoperative deformation. Given the increasing performance of computational resources, numerical modeling of complex intraoperative events is extending beyond its former predictive role to one of feedback thus providing fast, efficient, and valuable assistance to surgeons during surgery.

This paper illustrates that it is important to thoughtfully discriminate boundary conditions occurring in the OR environment. Comparing the gravity−induced shift calculation to the other two simulations (i.e. surface displacement application, and surface/ventricle displacement application) illustrates that the distribution of displacement is fundamentally dissimilar and applying measured deformation in a direct−interpolative sense may not be sufficient for accurate navigational assistance. Moreover, if one was to measure the state of strain energy within the continuum, a completely different elastic potential is present among the three calculations. With the gravity−induced brain shift, unless all cerebrospinal fluid levels are restored to their preoperative state, the brain will remain in this deformed condition. In the other two calculations, elastic potential has been generated within the domain from the artificial compression/stretching of tissue. As surgery progresses, boundary conditions concerned with retraction and resection [24] will be affected by this stored potential and could ultimately precipitate more error in navigation, i.e. understanding how much elastic recovery is anticipated during resection is critical for predicting the location of tumor margin.

The advantage to using a model−based update for brain shift compensation is that the field equations help constrain the deformation in accordance with known constitutive laws. With a model which closely represents the continuum, computational results will accurately reflect the state of the brain subjected to user−prescribed boundary
conditions. In the case of using sparse data, it is more critical that careful administration of load type be achieved since less volume constraints are known. In the case of using sparse data, this may not be as critical for gravity-induced deformation; but, future work must look at the impact of elastic potential in the context of other surgical loads (retraction, resection, drug effects, etc.) and the degree of pre-processing necessary to resolve the application of large mMR data sets. In some sense, the abundance of data from mMR may be more appropriate to non-rigid intensity-based methods such as optical flow [25] or mutual information [26]. These methods are not bound by constitutive laws in the same sense. Early surgical events will not effect future surgical predictions; updates are only generated from scan-to-scan. However, other issues exist with these approaches in that often the removal of tissue volumes can adversely affect these algorithms.

7. ACKNOWLEDGMENTS

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8. REFERENCES