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Beyond Finite Elements: A Comprehensive, Patient-Specific Neurosurgical Simulation Utilizing a Meshless Method

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Abstract

To be useful in clinical (surgical) simulations, a method must use fully nonlinear (both geometric and material) formulations to deal with large (finite) deformations of tissues. The method must produce meaningful results in a short time on consumer hardware and not require significant manual work while discretizing the problem domain. In this paper, we showcase the Meshless Total Lagrangian Explicit Dynamics Method (MTLED) which meets these requirements, and use it for computing brain deformations during surgery. The problem geometry is based on patient-specific MRI data and includes the parenchyma, tumor, ventricles and skull. Nodes are distributed automatically through the domain rendering the normally difficult problem of creating a patient-specific computational grid a trivial exercise. Integration is performed over a simple, regular background grid which does not need to conform to the geometry boundaries. Appropriate nonlinear material formulation is used. Loading is performed by displacing the parenchyma surface nodes near the craniotomy and a finite frictionless sliding contact is enforced between the skull (rigid) and parenchyma. The meshless simulation results are compared to both intraoperative MRIs and Finite Element Analysis results for multiple 2D sections. We also calculate Hausdorff distances between the computed deformed surfaces of the ventricles and those observed intraoperatively. The difference between previously validated Finite Element results and the meshless results presented here is less than 0.2 mm. The results are within the limits of neurosurgical and imaging equipment accuracy (~1mm) and demonstrate the method's ability to fulfill all of the important requirements for surgical simulation.

Keywords

Neurosurgical Simulations; Patient-Specific Simulations; Registration; Meshless Methods; Total Lagrangian Formulation

1 Introduction

In their influential review (Nakaji and Spetzler, 2004) list the “accurate localization of the target” as the first principle in modern surgical approaches. As only the preoperative anatomy of the patient is known precisely from medical images, it is now recognized that

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Conflict of Interest Statement

The authors have no conflicts of interest in this study.

the ability to predict soft organ deformation (and therefore intraoperative anatomy) during the operation is the main problem in performing reliable surgery on soft organs. Our aim is to simulate the deformation of the entire brain using information from preoperative MRIs and the knowledge of the position of the portion of the brain surface exposed by craniotomy. The simulation we present is based on MRI data which has already been used for FEM simulation and model validation in (Wittek et al., 2010). The surgical procedure involved removing a section of the skull immediately over the large anterior tumor, after which a significant retraction of the cortical surface was observed near the craniotomy.

Calculations of soft tissue deformation have historically been built on Finite Element Analysis (Cotin et al., 1999; Ferrant et al., 2001; Luboz et al., 2005; Picinbono et al., 2003) with promising results. In (Wittek et al., 2010) we demonstrated that a high level of precision can be achieved in patient-specific simulations of surgical procedures using non-linear (both geometric and material) biomechanical models. The accuracy of the Finite Element calculations depends heavily on the mesh which discretizes the geometry, so it is desirable to use only good quality hexahedral elements (Yang and King, 2011). With highly irregular geometry, an experienced analyst is required to create a patient-specific mesh, which consumes valuable human time. From our experience (Wittek et al., 2010) tedious manual mesh correction is still necessary even with recently developed software for generation of anatomic Finite Element meshes such as IA-FEMesh by (Grosland et al., 2009).

One solution to this challenge is to use a numerical method that does not require so strict a spatial discretization, such as a meshless method. Various meshless (or Element-Free) methods have been presented as alternatives to Finite Element methods (Babuska and Melenk, 1997; Belytschko et al., 1994; Li and Liu, 2004; Liu, 2003; Melenk and Babuška, 1996). These have primarily been motivated by the study of fracture mechanics and large deformations. Efforts to apply these methods in surgical simulation have been very limited (Doblare et al., 2005; Horton et al., 2007) and, to date, inadequate for clinical use.

In this paper, we showcase a suit of meshless algorithms which can deal with irregular 3D geometries composed from multiple parts, large deformations, non-linear, almost incompressible materials and contacts. It allows automatic discretization of the problem domain and is sufficiently fast to produce clinically useful results in a short time using only easily-available consumer hardware.

2 Methods

2.1 Problem Geometry

This is a real-world, patient-specific 3D geometry with all the irregularities that can be expected in clinical simulations. Figure 1 shows the anatomical components as surfaces generated from the segmented MRIs. The parenchyma, tumor and ventricle volumes were filled with a total of 31,753 nodes with average spacing of 3.5mm. Discretizing the irregular geometry in this way is a trivial exercise, whereas creating a good quality Finite Element mesh for the same geometry was a tedious and time-consuming, semi-automatic effort requiring a lot of manual intervention. The rigid skull was represented by its internal surface (Figure 2).

2.2 Background Integration Grid

One way to integrate across the volume is to use a background mesh, but this requires manual work and largely defeats the purpose of using a meshless method. For this simulation, a regular Cartesian grid of 53,672 integration points was created within the problem domain with 3 mm spacing between points. Such a grid can contain a small volume

error (at most half of a volume of a hexahedron intersected by a domain boundary) because the grid is not volume-conforming but this has proven to be too small to have any significant effect on the results in all simulations to date (Horton et al., 2010; Horton et al., 2007). The freedom and simplicity offered by such a background grid outweigh the small inaccuracy.

2.3 Boundary Conditions and Loading

We defined a contact interface between the skull and the parenchyma surface. Nodes on the parenchyma surface could not penetrate the skull, but could slide without friction as in (Wittek et al., 2010). The tumor and ventricle surfaces were tied (by shared nodes) to the internal surfaces of the parenchyma.

To load the model, we prescribed displacements on the portion of the parenchyma surface which had been exposed by the craniotomy (see Figure 2). The displacement for each loaded node was determined from distances between the preoperative and intraoperative parenchyma surfaces in the segmented MRIs.

2.4 Material Properties

Following (Wittek et al., 2009; Wittek et al., 2010) we used Neo-Hookean constitutive model with parameters given in Table 1. From solid-mechanical perspective our simulation belongs to the special class called displacement-zero traction problems whose solution is known to be weakly dependent on the unknown patient-specific properties of tissues (Ciarlet, 1988; Neal and Kerckhoffs, 2010; Wittek et al., 2009).

Following (Wittek et al., 2007), the skull was modeled as rigid since its stiffness is several orders of magnitude greater than that of all the other simulated parts.

2.6 Suite of algorithms

Solution algorithm—We used Meshless Total Lagrangian Explicit Dynamics solution algorithm. (Horton et al., 2010) describe and verify the algorithm as well as provide recommendations about node placement and relative ratio of nodes to integration points.

Time-step selection—In this simulation we used constant time step estimated based on the linearized theory (Joldes et al., 2012).

Termination criterion—To terminate the simulation we used a displacement error estimation as described in (Joldes et al., 2009a).

Contact algorithm—We used frictionless finite sliding contact as described in (Joldes et al., 2008)

Determination of correspondence between pre- and intraoperative brain surfaces—This correspondence was determined by applying a vector-spline regularization algorithm (Arganda-Carreras et al., 2006) to the surface curvature maps (Joldes et al., 2009b).

3 Results

3.1 Cross Sections

We show six 2D sections, taken at 5mm intervals, through regions of the brain which involve all the major anatomical components (Figure 3). We also compare our results to those obtained with a high quality FE mesh which was presented and validated in (Wittek et al., 2010).

3.2 Hausdorff Distances

Following (Archip et al., 2007; Oguro et al., 2011; Wittek et al., 2010), we used the 95% Hausdorff distance to objectively measure the differences between the intraoperative surfaces of the major anatomical structures of the brain predicted using the MTLED-based method and the surfaces obtained through segmentation of the intraoperative MRIs, Figure 4.

3.3 Computational efficiency

The simulation presented here was performed on a machine with an Intel Core i7 930 2.8GHz processor and 4GB of physical memory. The calculation time of our fully nonlinear problem with almost 100,000 degrees of freedom was 19.2 seconds for the approximately 1000 time steps required to obtain convergence. Such an excellent computational efficiency is due to the fully explicit nature of our algorithm and the use of Total Lagrangian formulation.

4 Discussion and Conclusions

We used MTLED-based suite of algorithms to perform a comprehensive patient-specific surgical simulation. The contours in Figure 3 show good similarity between the MTLED and FE results. Only in a few places are slightly different results seen and the differences are not larger than the voxel size of the intraoperative brain MRI (which is of an order of $0.86 \times 0.86 \times 2.5 \text{ mm}^3$). From this comparison, it appears that from the perspective of image-guided surgery, the results obtained using the MTLED method are as useful and accurate as those obtained with the FE method. Considering the reduced complexity of model generation, this is an excellent result and makes algorithms of this sort a viable option for clinical use.

Looking at the color codes in Figure 4, we see that the majority of the surfaces (blue) agree to within 1mm. Since this is the level of accuracy in image-guided surgery (Bourgeois et al., 1999; Warfield et al., 2005), we can make the claim that the results are accurate enough to be useful in clinical situations. Note however the localized region of high discrepancy in the right-hand side image in Figure 4 and the resulting larger value for $h_{95}(B, A)$. This is a consequence of the entire third ventricle in the segmented intraoperative MRI that was not present in the segmented preoperative MRIs. Comparison of the two images in Figure 4 and good agreement of the predicted intraoperative contours of the tumor and ventricles with the intraoperative images (Figure 3) suggest that the localized region of discrepancy in the right-hand side image in Figure 4 is due to the differences in ventricle segmentations of the preoperative and intraoperative images rather than inaccuracies in predicting the intraoperative deformations.

We simulated a craniotomy-induced brain shift with over 30,000 nodes and 50,000 integration cells and obtained results within half a minute. This is certainly fast enough to be used in clinical situations. However, there is excellent scope to increase the speed of this algorithm by employing GPUs to handle the very parallel calculations (Joldes et al., 2010). Therefore clinically useful results can be obtained without the need for supercomputers.

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Figure 1. Surface visualizations of the problem geometry; Parenchyma (yellow), Tumor (blue) and Ventricle (Red).

Figure 2. Triangulated skull surface (blue). The parenchyma surface (yellow) is visible through the craniotomy. Part of the brain stem (yellow) is also visible where it protrudes from the base of the skull. Also shown are the loaded nodes on the surface of the parenchyma exposed by the craniotomy.

Figure 3.

Intraoperative MRIs overlaid with contours (green lines) of the deformed tumor and ventricle surfaces as generated by our MTLED-based suite of algorithms. The three transverse sections (top row) and three sagittal sections (bottom row) were taken at 5mm intervals. For reference, the red lines represent the contours of the deformed ventricles and tumor computed by FEM, but these are almost entirely obscured by the very similar MTLED results.

Figure 4.

Differences between calculated (Surface *A*) and intraoperatively observed (Surface *B*) ventricle surfaces. The left image shows the calculated ventricle surface *A* (based on the segmented preoperative MRIs and deformed by the MTLED algorithm). The colors represent the distance $d(a, B)$ from point *a* on Surface *A* to the nearest point on Surface *B* generated from the segmented intraoperative MRIs. The scale reaches 95% Hausdorff Distance $H_{95}(A, B)$. Similarly, the right image shows the surface *B*, with colors representing the distance $d(b, A)$.

Table 1

Material properties for each component in the simulation. μ and λ are the Lamé parameters.

Parameter	Parenchyma	Tumor	Ventricles
μ (Pa)	1.007×10^3	2.013×10^3	4.56
λ (Pa)	49.33×10^3	98.66×10^3	1.14
Poisson's ratio	0.49	0.49	0.1
Mass Density (kg m^{-3})	10^3	10^3	10^3