Computing Reaction Forces on Surgical Tools for Robotic Neurosurgery and Surgical Simulation

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Abstract

The objective of our research is to create a system computing brain deformations. In this paper we concentrate on assessing the feasibility of using non-linear finite element computation for patient-specific simulations. As an example we use computation of reaction forces on surgical tools, with application in e.g. surgical robot control system, virtual reality operation planners, etc. We specifically address issues related to creating geometrically and mechanically precise representations of the brain. The method comprises of the following steps: 1) development of a "generic" brain mesh; 2) conversion of the generic brain mesh to patient-specific brain mesh; 3) selection of the appropriate mathematical model of the brain biomechanics; and 4) development of an efficient computational scheme. As an illustration of the presented concepts we provide an example of 3D meshing, and calculation of reaction force acting on a surgical tool using a single-phase mathematical model solved using an explicit, non-linear finite element procedure.

1 Introduction

The advantages of surgical robots and manipulators are well recognised in the clinical and technical community. Precision, accuracy and the potential for telesurgery are the key motivators for applying advanced robots in surgery [Chinzei et al., 1999; Chinzei and Miller, 2001]. Surgical robots require trajectory planning, which in practice relies upon the preoperative images. However, if the organ deforms, the trajectory needs to be updated during the procedure. Although nuclear magnetic resonance images (NMRI) can provide rich information of tissue deformation [Grimson et al., 1999; Kaus et al., 1999], currently tens of seconds are required to produce and analyse a new set of images. One possible method of dealing with these delays can be predicting the organ deformation by means of computer simulation. When the organ deformation can be predicted and the tissue mechanical properties are known, the interaction force between the end-effector (i.e. surgical tool) and organ surface during surgery can also be computed. Thus, simulation can be used to predict reaction forces acting on surgical tools (equal to forces with which the tool acts on the organ) as well as to provide realistic force and tactile feedback for virtual reality surgical simulators.

To achieve a realistic, clinically acceptable computer simulation of human body organ deformation during surgical procedure, one requires an adequate model, capturing the intrinsic physical properties of the organ considered, intervention performed, and surgical accessories used. Neurosurgery is particularly demanding, as the brain is arguably the most complicated object in the known universe. Modelling of physical properties of the brain is still an uncovered area pioneered by a few only [Miller and Chinzei, 1995; Bilston et al., 1997; Paulsen et al., 1999; Prange and Margulies, 2002].

The models used in surgery simulation should contain detailed anatomical (geometrical) information. Such information can be provided by suitable anatomical atlases [Talairach and Tournoux, 1988; Visible Human, 1995; Nowinski, 2003]. However, simulation of surgical procedures typically requires patient-specific geometric information. Therefore, the information provided by the atlas has to be individualised through an appropriate scaling process to fit to a particular patient, referred to as registration, see e.g. [Warfield, 1999; Ferrant, 1999; Ferrant, 2000; Ruiz-Alzola, 2000]. In this process the medical radiographic images of the patient brain are utilised.

Next, a computational grid has to be created on the domain of interest. In most practical cases this amounts to producing a finite element mesh.

Mathematical models governing the deformation behaviour of continua consist of sets of partial differential equations supplemented by constitutive relations, boundary and initial conditions. Numerical methods are needed to solve such sets of equations. These methods require appropriate discretisation of the domain of interest. The most common and probably the most effective numerical method for sets of partial differential equations is the finite element method [Bathe, 1996]. After creating the mesh, the partial differential equations of a chosen mathematical model are solved and the evolution of variables of interest obtained.

It must be noted here that because human organs during surgical procedures undergo large deformations, fully non-linear finite element formulations and material models have to be used.

The final objective of our research is to create a system computing brain deformations (Figure 1). In this paper we focus on three aspects of such system:

1) Using *Visible Human* [1995] electronic atlas and patient specific NMRI to obtain geometric information about the brain.
2) Building a patient specific finite element model for predicting the brain deformation and reaction forces acting on surgical tools.
3) Choosing an appropriate finite element algorithm to accurately and effectively compute these forces.

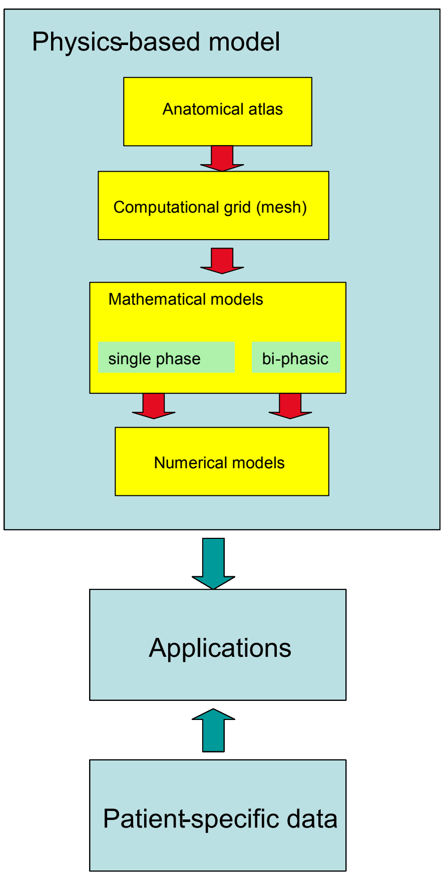


Figure 1 Framework for developing the system computing brain deformations. Some applications, such as e.g. surgical simulators for medical training, do no require patient specific information. Other, such as e.g. surgical robot control system or operation planning system, do.

2 Methods

2.1 Construction of Mesh for Patient Specific Brain Model

Suitable grids are required so that computational analysis of anatomical and geometrical information contained within NMRI can be conducted. One can attempt to construct patient specific meshes either anew, directly from the NMRI, or by utilising the NMRI to modify the pre-existing generic meshes (i.e. mesh templates) to match the patient specific data. In the present study the second approach was used, as it is believed that the "template-based" meshing may be amendable to full automation in future.

We chose the brain mesh consisting of hexahedron elements (i.e. 8-node "bricks") previously built for the Total Human Model for Safety (*THUMS*) [Iwamoto et al., 2002] developed by Toyota Central R&D Labs., Nagakute, Japan, with the help of Wayne State University, Detroit, Michigan, USA (Figure 2). The hexahedron finite elements are known to be the most effective ones in non-linear finite element procedures using explicit time integration. The *THUMS* brain mesh was developed using the anatomical and geometrical data obtained from the *Visible Human* [1995] electronic atlas of the human body and the Gray's Anatomy textbook [Berry et al., 1995], and it represents a brain of a healthy adult. Although it distinguishes between the grey and white brain matter, it disregards important components of the brain anatomy such as ventricles.

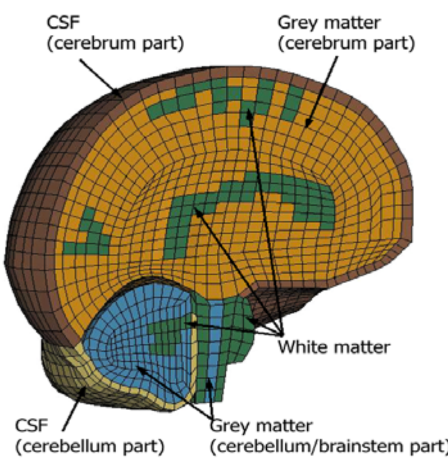


Figure 2 Hexahedron mesh for *THUMS* brain model. CSF is the cerebrospinal fluid. Courtesy of Toyota Central R&D Labs.

Patient Specific Anatomical and Geometric Information for Brain Mesh

As in the present study we intended to demonstrate how finite element models can be used in simulation of the actual surgical procedure, patient specific geometric data needed to be incorporated. Such data were obtained from a set of sixty pre-operative NMRIs of a patient undergoing brain tumour surgery at the Department of Surgery, Brigham and Women's Hospital (Harvard Medical School, Boston, Massachusetts, USA).

In order to distinguish between the ventricles and the brain parenchyma, the images were segmented using the *SLICER* (http://www.slicer.org/) program developed by the Surgical Planning Laboratory of Harvard Medical

School (Figure 3). After the segmentation, the digital models of 3-D surfaces of the brain and ventricles were created in *SLICER* using the *Visualization Toolkit* (*VTK*) binary format [Schroeder et al., 2002] (Figure 4).

Mesh Construction

Construction of patient specific mesh started from scaling the *THUMS* brain mesh to ensure that the distances between the anatomical landmarks in the model equal those in the patient's brain. The distances between the patient's brain anatomical landmarks were determined from the raw (i.e. not segmented) NMRI. The landmarks' definitions according to Nowinski [2001] were used.

One of the problems encountered when generating patient specific meshes of human organs is that the format of medical images is incompatible with the one used by engineering mesh generators, i.e. conversion is needed to transfer the geometric data from the images or *VTK* surface models to mesh generator. In the present study, *PATRAN* mesh generator developed by PDA Engineering (Costa Mesa, California, USA) was used [PATRAN, 1998]. When using this mesher to add ventricles to the *THUMS* brain model, we dealt with the geometric data conversion problem in the following way. The digital model of the ventricle surface built using *SLICER* was processed by means of *ParaView* program (http//: www.paraview.org/) to create the ventricle sagittal sections. The sectioning interval was 3 mm, which resulted in 28 contour lines (Figure 5). The lines were saved in the *VTK* ASCII format and converted to the *PATRAN* session format using software developed in-house. Then, from the contour lines, the ventricle surface and volume patches were manually created using *PATRAN* geometry builder module to represent the entire ventricle volume. The *THUMS* brain model elements located inside the volume patches were removed, and the *PATRAN* automatic mesh generator was applied to discretise the patches using hexahedron elements (Figure 6). The mesh generator parameters were manually selected to ensure that the ventricle mesh density equals that of the remaining part of the brain model. This resulted in a brain hemisphere mesh consisting of 12600 nodes and 7485 solid elements. Finally, the element quality check was performed, and the elements with poor quality rating (i.e. Jacobian close to zero, too large aspect ratio and/or face skew) were modified by moving their nodes. In order to simulate the pia matter, the brain surface was covered by a layer of 2070 thin membrane elements (thickness of 0.4 mm).

Thus, in the present study, the patient specific brain mesh was generated using a semi-automatic method that strongly relied on the analyst skills. This is a typical process when building hexahedron meshes of human organs. Fully automatic hexahedron meshing of structures with complex geometry cannot be achieved as yet.

a)

b)

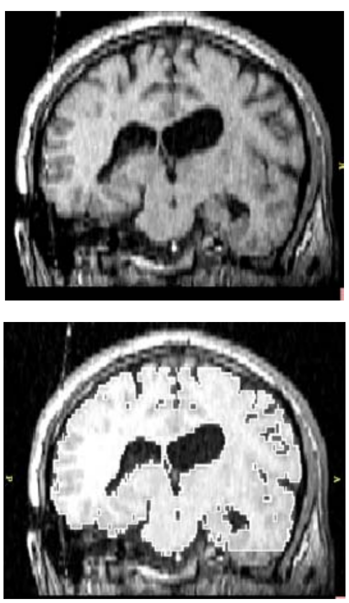


Figure 3 Example of raw and segmented NMRI of the head used in the present study when building patient specific brain mesh. The images were taken before the surgery was conducted.

a)

b)

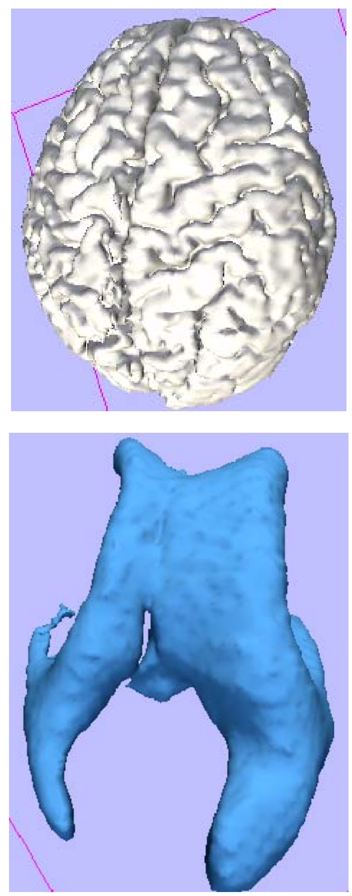


Figure 4 a) Brain and b) ventricle surface models created from NMRIs. These models were applied to build the patient specific brain mesh.

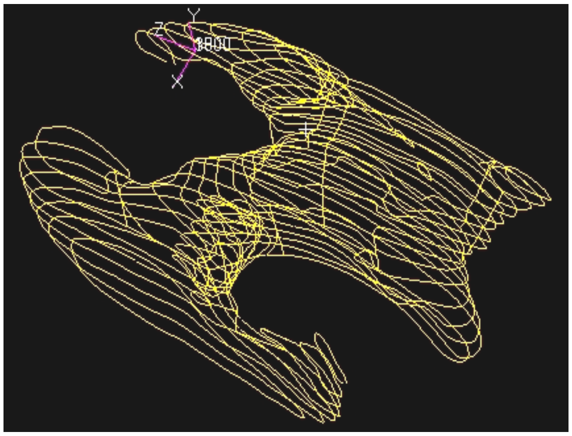
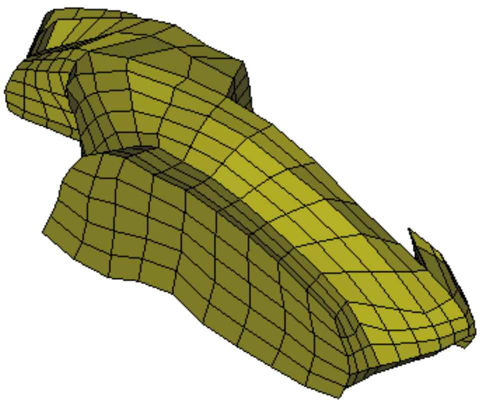


Figure 5 Ventricle contour lines constructed from the surface model shown in Figure 4b.

a)



b)

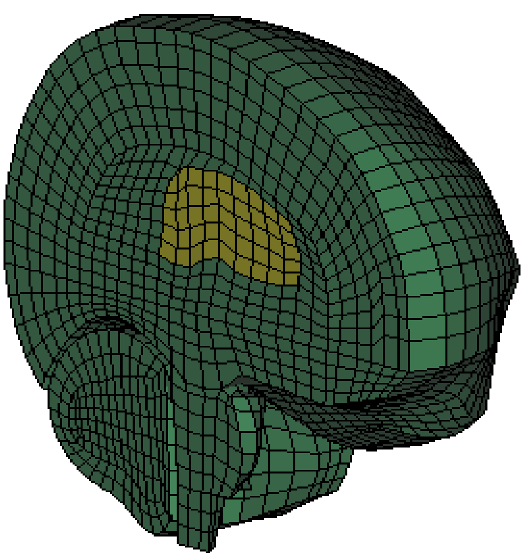


Figure 6 Patient specific brain mesh constructed in the present study. a) Left-side lateral ventricle; b) Entire left brain hemisphere.

2.2 Continuum Mechanics of the Brain

Continuum Mechanics Equations for the Brain and their Integration

Detailed account of the biomechanics of the brain including relevant equations was given by Miller [2002]. In this section the basic ideas are summarised.

From the perspective of surgical simulation, the brain can be considered a single-phase continuum. The main reason for neglecting effects of cerebrospinal fluid flow within the brain tissue is that the time scale of this flow is much different (larger) to the time scale relevant to surgical procedures. This relevant time scale is tens of seconds to minutes.

The mathematical model of the deformation behaviour of a single-phase continuum consists of the standard solid-mechanical partial differential equations of equilibrium (or dynamics), boundary and initial conditions, and a constitutive model for the brain tissue. It should be noted, however, that because during surgery parts of the brain undergo large deformations (including finite rigid-body translations) full non-linear equations must be used, and not, as in some previous studies, equations of linear elasticity.

Depending on whether one decides to measure stresses and strains with respect to the deformed or undeformed configuration, different (but equivalent) formulations of equations of equilibrium should be used [Fung, 1965; Miller 2002]. If Cauchy stress and Almansi strain, both measured with respect to the deformed (current) configuration, have been chosen the equation of equilibrium can be written in the following way:

$$\boldsymbol{\tau}^{ij}_{,i} + \rho F_i = 0 \qquad (1)$$

where $\boldsymbol{\tau}$ denotes Cauchy stress, ρ is a mass density, F_i is a body force per unit mass in direction i, and comma indicates covariant differentiation with respect to the deformed configuration. Repeated index summation convention was used.

In case Second Piola-Kirchoff stress tensor and Green strain (both measured with respect to undeformed configuration) are preferred the equations of equilibrium have to be rewritten:

$$(\mathbf{S}^{ij} x^j_{,k})_{,i} + \rho_0 F_{0_i} = 0\,, \qquad (2)$$

where ρ_0 is a mass density in undeformed configuration and F_{0i} is a body force per unit mass in undeformed configuration in direction i measured with respect to undeformed configuration. Comma denotes covariant differentiation, this time, with respect to the original configuration.

If one prefers Lagrange stress, the equations of equilibrium would look as follows:

$$\mathbf{T}^{ij}_{,i} + \rho_0 F_{0_i} = 0\,. \qquad (3)$$

The formulation of appropriate boundary conditions supplementing the above equations constitutes a significant problem in biomechanics of soft tissues. In the case of the brain it is possible to assume the rigidity of the skull, certain gap between the brain and the skull, and no friction sliding boundary condition at the skull-brain interface [Miller et al., 2000]. However, the suggested approach is only a crude approximation. Research on boundary conditions, in the authors' opinion, is at least equally important to the investigation of the mechanical properties of brain internal structures.

In this contribution we are most interested in demonstrating the feasibility of application of finite element modelling in predicting the forces acting on surgical robot end-effector performing brain surgery. This requires application of an efficient numerical scheme when integrating equations of equilibrium (or dynamics) in time domain. Such integration can be done using either implicit or explicit methods [Bathe, 1996; Crisfield, 1998]. When using implicit methods, equations of dynamic equilibrium (i.e. equilibrium including internal, external and inertial forces) at both given t and following $t+\Delta t$ time points are directly used. Under the assumption that this equilibrium is exactly satisfied at time t, the displacements, velocities, accelerations and internal forces at time $t+\Delta t$ are predicted. Then, the predicted variables are substituted into an equation of dynamic equilibrium at time $t+\Delta t$. Finally iterations are performed to minimise residuals resulting from such substitution. The implicit integration methods are unconditionally stable but can be time consuming as iterations are conducted at each time step. Therefore, in the present study an explicit integration was used. In the explicit integration, no iteration is needed as the displacement at time $t+\Delta t$ is solely based on the equilibrium at time t. The explicit time integration has been proved to be reliable and efficient in automotive industry when simulating car structure deformation, e.g. Pipkorn [1996], Tabiei and Wu [2000], http//:www.ncac.gwu.edu, and predicting injury resulting from car accidents, e.g. Darvish and Crandall [2002], Hyncik [2002], Wittek and Omori [2003]. The present study appears to be one of the first attempts to apply explicit integration in medical biomechanics when the modelled system undergoes deformation with moderate strain rates.

The computations were conducted using *LS-DYNA* code (Livermore Software Corporation, Livermore, California, USA) [Hallquist, 1998; LS-DYNA, 2003], which is one of the explicit finite element codes routinely applied in car crash simulation.

Constitutive Model of the Brain

To compute force acting on surgical robot end-effector when performing surgery, one must use the appropriate constitutive model for the brain tissue. As shown by Miller et al. [2000] and Miller and Chinzei [2002], the stress-strain behaviour of the brain tissue is non-linear. The stiffness in compression is significantly higher than in extension. One can also observe a strong stress – strain rate dependency. The distinguishing feature of the mathematical model of the brain intended for the simulation of neurosurgery is the strain rate range (the loading speed range) considered — 0.001 s^{-1} - 1.0 s^{-1} — orders of magnitude lower than that experienced in situations leading to injury. To account for these complexities, Miller and Chinzei [2002] suggested the following constitutive model:

$$W = \frac{2}{\alpha^2}\int_0^t [\mu(t-\tau)\frac{d}{d\tau}(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3)]d\tau, \qquad (4)$$

$$\mu = \mu_0[1 - \sum_{k=1}^{n} g_k(1 - e^{-\frac{t}{\tau_k}})], \qquad (5)$$

where: W is a potential function, λ_i's are principal stretches, μ_0 is the instantaneous shear modulus in undeformed state, τ_k are characteristic times, g_k are relaxation coefficients, and α is a material coefficient, which can assume any real value without restrictions. The constitutive model by Miller and Chinzei [2002] is not explicitly available in the *LS-DYNA* code, and the constitutive behaviour summarised in Eqs. (4) and (5) was simulated using the *LS-DYNA* Ogden rubber model [LS-DYNA, 2003]. The model parameters were taken from Miller and Chinzei [2002] as summarized in Table 1. The brain parenchyma was assumed to be almost incompressible.

The pia matter was assigned stiffness of 100 kPa – the value consistent with previous studies on brain injury [Wittek and Omori, 2003].

Table 1. List of material constants for constitutive model of brain tissue, Eqs. (4) and (5), n=2.

Instantaneous response	μ_0=842 [Pa];
k=1	characteristic time τ_1=0.5 [s]; g_1=0.450;
k=2	characteristic time τ_1=50 [s]; g_1=0.365;

2.3 Modelling Example: Simulation of Brain Indentation

During a surgical procedure, surgical tools driven by surgeon hands or a manipulator exert forces on the brain, which results in displacement of the brain surface. To simulate such situation, a motion at constant velocity of 2 mm/s was applied to the selected nodes in direction approximately normal to the brain surface (Figure 7). The sum of forces at these nodes was a close approximation of the reaction force between the brain surface and indentor. Only half of the brain was simulated as the sagittal symmetry was assumed.

To prevent the rigid body motion of the model, the cerebellum and brainstem nodes were rigidly constrained (Figure 7).

The integration time step of 0.2 ms was chosen.

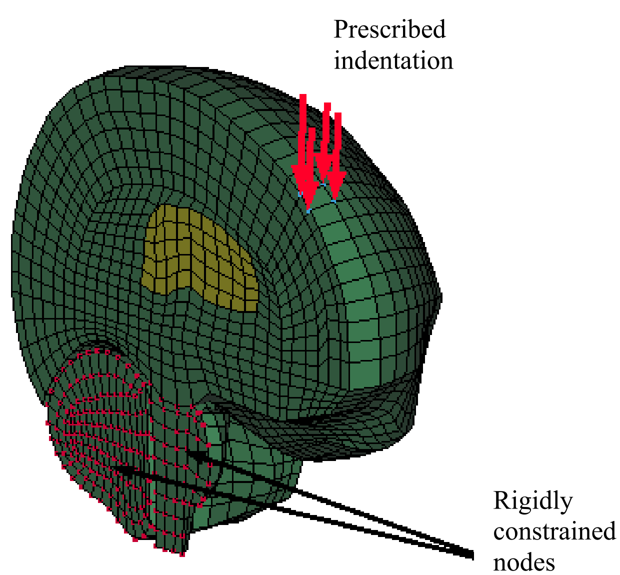


Figure 7 Simulation of brain indentation. Arrows indicate nodes at which the motion was applied. The cerebellum and brainstem nodes were rigidly constrained to prevent the rigid body motion.

Validation

As the present study focuses on demonstrating feasibility of application of finite element modelling in predicting the end-effector force rather than complete simulation of the actual surgical procedure, only limited validation was performed. The results of present computation were compared with those obtained by Miller et al. [2000] in which in-vivo indentation of the swine brain was conducted (Figure 8). The basis for this comparison was that the brain material parameters used in the present study were also derived from the experiments performed on swine brains. The strain rate used in the present model was very close to that used in the experiments by Miller et al. [2000].

The cross-section area of the element to which the displacement was applied differed from the one of the indentor used by Miller et al. [2000]. Therefore, in order to enable comparison with the experiments by Miller et al. [2000], the present nodal forces were normalised using the following formula:

$$F_{norm} = F\frac{A_{ind}}{A_{el}}, \qquad (10)$$

where F is the sum of the computed nodal forces, F_{norm} is the sum of normalised nodal forces, A_{ind} is the indentor cross section area in the experiments by Miller et al. [2000] (A_{ind} =78.54 mm^2), and A_{el} is the cross section area of the element to which the displacement was applied in the present brain model (A_{el} =37.35 mm^2).

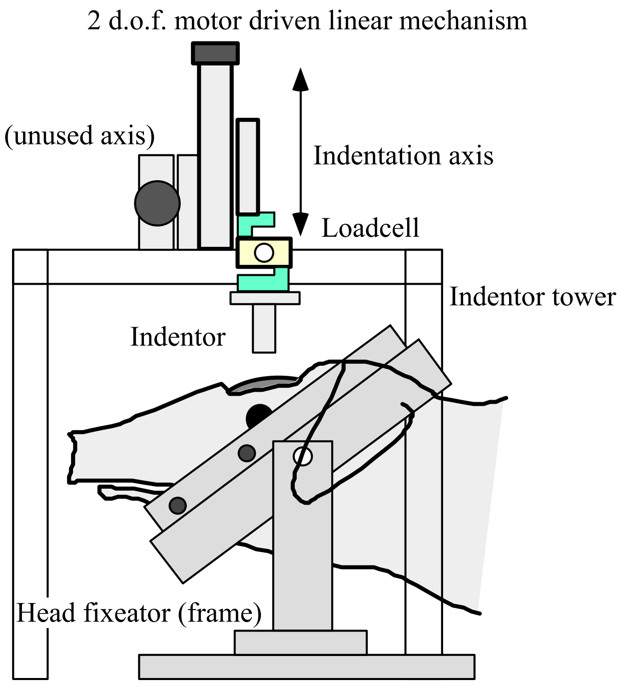


Figure 8 Schematic set-up of the in-vivo indentation experiments by Miller et al. [2000].

3 Results

The calculated reaction force between the brain surface and indentor is shown in Figure 9. It agrees well with the results measured by Miller et al. [2000] on swine brain. The calculated force is only around 20% larger than the measured one. Taking into account large variability of mechanical properties of biological tissues, the agreement between the calculated and experimental results can be considered as good.

The computation time was around 21 minutes on a single Pentium IV 2.8 GHz processor when simulating the indentation of duration of 4.2 s.

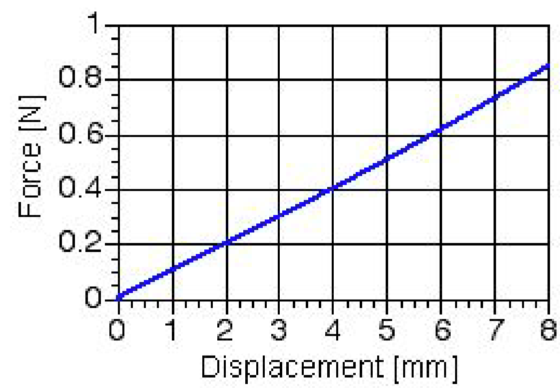


Figure 9 Force versus displacement computed in the present study. The computed force was normalised to the indentor cross section area of 78.54 mm^2.

4 Conclusions and Discussion

The present study indicated that the explicit finite element computation is a feasible method of computing the reaction forces acting on surgical tools. The results calculated when computing the brain indentation were close to those experimentally obtained by Miller et al. [2000]. Taking into account large variability of mechanical properties of biological tissues, the agreement can be considered as good and the present results as promising.

It took around 21 minutes to compute the brain indentation of duration of 4.2 s using a single 2.6 GHz Pentium IV processor computer. This time is clearly too long for intra-operative (real time) applications. However, it should be noted that the computations were conducted using the general purpose engineering code. It is reasonable to expect that the computation time can be drastically decreased by application of more specialised code with reduced capabilities but improved efficiency, and by increasing the time step used in time integration of continuum mechanics equations. We estimate that using the state-of-the-art personal computer the computational time can be reduced to about 50 seconds. Improvements in computer hardware would decrease this time even further.

When comparing the results of this paper with previous works in surgical simulation it is important to note that we used fully non-linear formulations accounting for large deformations, rigid body motions as well as non-linear material response. Almost all previous studies in this area (with notable exception of Xu and Nowinski, 2001 who used implicit non-linear formulation) used linear formulations valid only for infinitesimal deformations.

Acknowledgements

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