Comparison of constitutive models of brain tissue for non-rigid image registration

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Abstract
Non-rigid registration of magnetic resonance images (MRI) for neurosurgery requires the use of biomechanical brain models to accurately capture the internal deformations that result from brain shift. The objective of this study is to compare deformation fields obtained by means of a non-linear finite element model of the brain using various constitutive models of brain tissue. A patient-specific brain shift problem is analysed to allow qualitative and quantitative assessment of the computed deformation field, through comparison with intraoperative MRI. For geometrically non-linear analysis, selection of material model did not significantly alter the results. However, the linear elastic model reduced the computation time by 30%. The infinitesimal deformation analysis yielded incorrect results. Thus, a biomechanical model using geometrically non-linear analysis and linear constitutive tensors is suggested for computing deformations for non-rigid registration problems.

1. Introduction

1.1. Non Rigid Registration

Image Guided Neurosurgery involves the production of intraoperative images and/or update of pre-operative images to track the current position of internal structures in the brain during surgery. This is necessitated by brain deformation during surgery known as the "brain shift".

a) b) c)

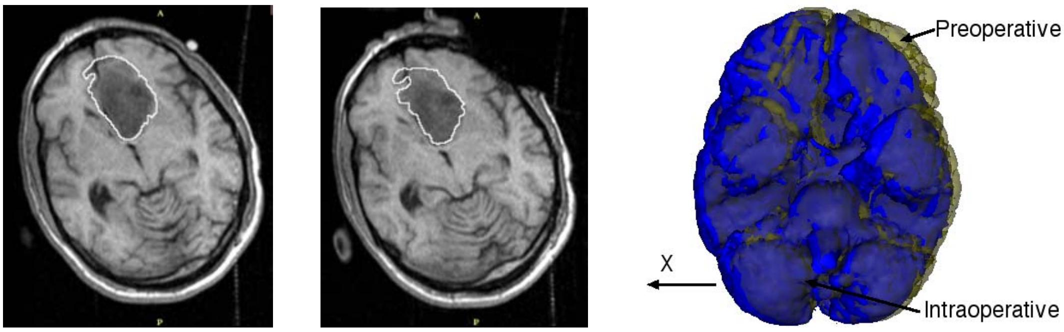

Figure 1: Brain shift during craniotomy (surgical opening of the skull). Magnetic Resonance Images (MRI) acquired (a) Preoperatively, (b) Intra-operatively. (c) Cortical surfaces derived from the pre-operative and intraoperative MRIs overlaid to highlight brain shift and deformation (images courtesy of Dr Simon Warfield, Harvard Medical School)

In most biomechanical models for image registration, linear finite element formulation is used (Castellano-Smith et al., 2001; Ferrant et al., 2002; Hagemann et al., 1999; Warfield et al., 2005). In this formulation, infinitesimal deformations (i.e. geometric linearity) and linear stress–strain relationship (i.e. material linearity) are assumed. Our results show that this approach does not yield accurate results. Instead, appropriate finite deformation procedures must be used so that large displacements of the brain during the surgery can be accounted for. When strains and stresses are referred to the current configuration, the following formula should be used to derive finite element equations:

$$\int_{{}^tV} {}^t\tau_{ij}\delta\varepsilon_{ij}\,{}^t dV = \int_{{}^tV} {}^t f_i^B \delta u_i\,{}^t dV + \int_{{}^tS} {}^t f_i^S \delta u_i\,{}^t dS \tag{1}$$

where ${}^t\varepsilon$ is the current Almansi strain (i.e. Almansi strain at time *t*), ${}^t\tau$ is the current Cauchy stress (i.e. current forces per unit areas in the deformed geometry), ${}^tV$ is the current volume, ${}^tS$ is the current surface, $\int_{{}^tV} {}^t\tau_{ij}\delta\varepsilon_{ij}\,{}^t dV$ is the internal virtual work in the current configuration, $\int_{{}^tV} {}^t f_i^B \delta u_i\,{}^t dV$ is the virtual work of the external body forces in the current configuration, and $\int_{{}^tS} {}^t f_i^S \delta u_i\,{}^t dS$ is the virtual work of the external surface forces in current configuration.

In addition to geometric non-linearities caused by large deformations, there is also strong experimental evidence that brain tissue does not obey the linear stress–strain relationship (Miller and Chinzei, 1997; Miller and Chinzei, 2002). They behave as hyperviscoelastic materials (i.e. non-linear stress–strain relationship with strain rate dependency) with different (i.e. bi-modular) responses in tension and compression.

The objective of the present study is to quantify the effects of material non-linearities on the accuracy and computation time when applying biomechanical models to predict deformation field due to brain shift for non-rigid registration of images. To achieve this objective, registration results obtained using a patient-specific finite element brain model together with three constitutive models: i) hyperviscoelastic material model for tumour and brain parenchyma; ii) hyperelastic material model for tumour and brain parenchyma; and iii) linear elastic material model for tumour and brain parenchyma; are evaluated on the basis of qualitative and quantitative comparison with intra-operative MRI data, with a particular focus on the brain internal structure.

2. Methods

Patient-specific brain mesh developed in the previous studies by Wittek et al. (2004, 2005) was used. The mesh comprises of 15050 hexahedron (8-noded bricks) elements defining the brain parenchyma, ventricles and tumour. The skull was discretised with 4-noded shell elements and treated as a rigid body.

The model was loaded through the enforced motion of nodes (i.e. through prescribed motion of a boundary) at the brain surface in the craniotomy area, Figure 2 as described in (Wittek et al. 2005). To prevent free rigid body motion of the entire model, the bottom-most brainstem nodes were rigidly constrained in all directions to simulate interaction with the remaining part of the spinal cord. The remaining nodes on the brainstem surface were constrained only in transverse direction.

a) b)

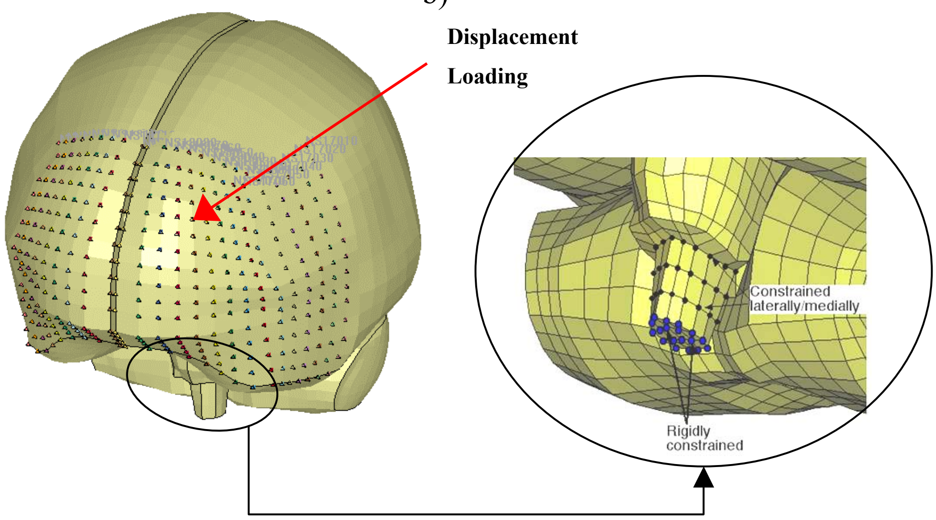

Figure 2: a) Model loading. Nodes at which the enforced boundary motion was prescribed are indicated as dots. b) Boundary conditions for the brainstem.

To investigate the effects of material model on image registration, the behaviour of the healthy brain parenchyma was determined using three distinct material models:

Hyperviscoelastic material (Miller and Chinzei, 2002)

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

$$\mu = \mu_0[1 - \sum_{k=1}^{n} g_k(1 - e^{-\frac{t}{\tau_k}})], \tag{3}$$

where W is a potential function, λ_i's are principal stretches, μ is the instantaneous shear modulus in undeformed state, τ_k's are characteristic times, g_k's are relaxation coefficients, and α is a material coefficient, which can assume any real value without restrictions. Material constants are given in Table 1.

Table 1: Material constants for hyperviscoelastic model of brain tissue (Miller and Chinzei 2002)

Instantaneous Response	$k = 1$	$k = 2$
μ_0 = 842 Pa	τ_1 = 0.5 s	τ_2 = 50 s
α = -4.7	g_1 = 0.450	g_2 = 0.365

Hyperelastic Model

$$W = \frac{\mu}{\alpha}\left(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3\right) \tag{4}$$

The constants $\mu = \mu_0$ = 842 Pa and α = -4.7 were the same as for the hyperviscoelastic material model (Table 1).

Elastic Model

For assumed incompressibility (i.e. ν=0.5), $E = (2\cdot1.5)\cdot\mu_0 = 3\cdot842$=2526Pa for brain parenchyma.

We used the same constitutive models for the tumour and ventricles as in (Wittek et al., 2005).

Commercial finite element solver LS-DYNA 970 revision 5434 (LSTC, 2003, 2004) was used in all the simulations. The computations were performed on personal computer with 3.0GHz Pentium 4 processor and 1024MB of internal memory.

3. Results

3.1. Displacement fields

In simulations using geometrically non-linear analysis, the differences between the computed and MRI-derived intra-operative cross sections were very small (Figures 3a-c). In these simulations, varying the material model exerted very little effect on the computed deformation field. On the other hand, geometrically linear (infinitesimal deformations assumed) calculation yielded highly localised, unrealistic deformations, confirming that this approach cannot grasp rigid body translation and finite rotation modes, and therefore should not been used for image registration.

The assessment of computational efficiency of discussed models is given in Table 2. The right-hand side column summarises the percentage variation of CPU time with respect to the simulation, in which the non-linear hyperviscoelastic material model and geometrically non-linear analysis were used.

Table 2: Computation times and their comparison with the computation time in the simulation in which hyperviscoelastic material model and geometrically non-linear analysis were used

Material Model/Analysis Type	Computation Time	% Variation
Hyperviscoelastic material/ Geometrically non-linear analysis	2411 s	—
Hyperelastic material/ Geometrically non-linear analysis	2358 s	2.20%
Linear elastic material/ Geometrically non-linear analysis	1713 s	28.95%
Linear elastic material/Linear analysis	496 s	79.43%

4. Discussion and Conclusions

For geometrically non-linear analysis, selection of material model (i.e. hyperviscoelastic, hyperelastic or linear elastic) did not affect accuracy when predicting brain deformations. Consequently, a linear elastic material model is suggested for image registration as, without compromising the accuracy, it reduces the computation time by up to 30% in comparison to hyperelastic model. It is important to note that this conclusion is unlikely to be valid for models in which loading is defined through body and surface forces. These are not displacement – zero traction problems and thus accurate representation of constitutive behaviour of the brain tissue through non-linear hyperviscoelastic constitutive tensor would be necessary to obtain reliable results.

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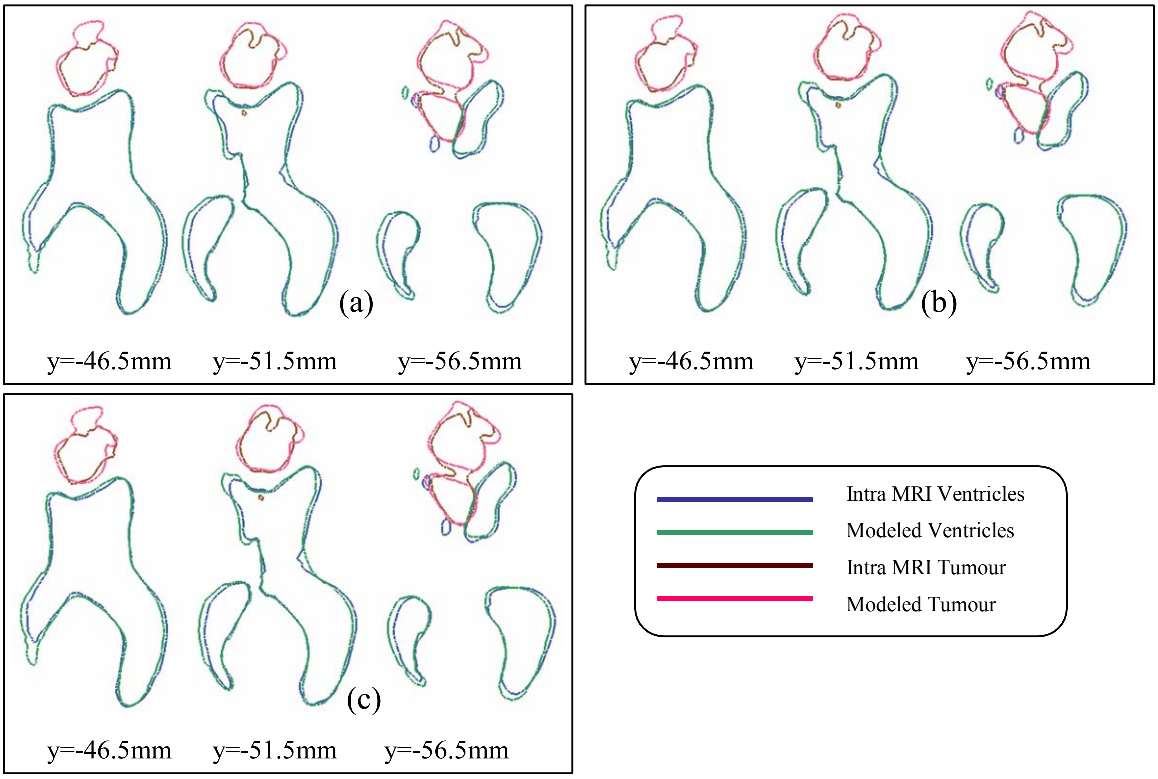

Figure 5: Transverse contour sections: (a) Hyperviscoelastic material model and geometrically non-linear analysis, (b) Hyperelastic material model and geometrically non-linear analysis, (c) Linear elastic material model and geometrically non-linear analysis.

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