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Publisher's version / Version de l'éditeur:
https://doi.org/10.1109/WHC.2007.12
Proceedings of the Second Joint EuroHaptics Conference and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems, 2007-03-24

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A viscoelastic soft tissue model for haptic surgical simulation

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Abstract

Brain surgeons distinguish tumors from healthy tissue by the way the tissue deforms. Accurately simulating these deformations at haptic update rates is still a challenge. Biological tissues exhibit complex viscoelastic behavior and undergo large deformation during surgery. For this purpose we evaluate the accuracy and computational performance of the Christensen viscoelastic material model in combination with non-linear finite elements. Our results show that realistic viscoelastic properties of brain tissue can be simulated with haptic feedback at 70% of the update rate of the non-viscoelastic large deformation Neo-Hookean model.

1. Introduction

Computer simulation with force feedback is believed to play an increasingly important role in the training of surgeons. Such simulators should be able to recreate important aspects of the procedures. In brain surgery the sense of touch is important, in particular to distinguish healthy tissues from tumors. Biological tissues are non-linear, viscoelastic, non-homogenous, and anisotropic. Their deformation can be accurately computed with an appropriate constitutive model and a large deformation algorithm. However, the computational requirements imposed by haptic update rates have limited its use in haptic systems.

Currently used deformation algorithms in haptic rendering often greatly simplify tissue properties to gain speed. For example, mass spring systems have limited fidelity and are hard to tune. Methods that have a good foundation in solid mechanics, such as the finite element method (FE), are gaining in popularity, but algorithms that account for large deformation are rare. In addition, viscosity is either simplified as a velocity dependent drag force on the nodes, or as dampers between the nodes, as in [1].

The objective of this work is to evaluate the accuracy and computational efficiency of a large deformation viscoelastic constitutive model in the context of the development of a haptic training system for brain surgery.

2. Brain tissue modeling

The viscoelastic behavior of brain tissue can be modeled by a generalized Maxwell model: a spring parallel to spring-damper combinations (see symbols in Table 1). Gefen *et al.* [2] performed indentation tests on porcine brains to obtain the constants for this model (see Fig. 1).

In this work, we propose to use the Christensen viscoelastic constitutive equation [3]. The true stress σ is given by:

$$\boldsymbol{\sigma}(t) = \kappa\left[\det(\mathbf{F})-1\right]\mathbf{Id} + C_0\mathbf{F}\mathbf{F}^T + \mathbf{F}\int_0^t G(t-\tau)\frac{\partial \mathbf{E}(\tau)}{\partial \tau}d\tau\mathbf{F}^T \quad (1)$$

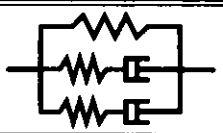
where κ is the compressibility modulus, $\mathbf{F}$ is the deformation gradient, C_0 is the hyperelastic modulus, $\mathbf{E}$ is the Green-Lagrange strain, and $G(t)$ is the material memory function:

$$G(t) = \sum_{k=1}^{n}\left(C_k e^{-t/\tau_k}\right) \quad (2)$$

where C_k are relaxation moduli and τ_k are relaxation times.

Representative curves from Gefen *et al.* were fitted with a 5 parameters Christensen model by inverse FE modeling (see Fig. 1). The shape of the porcine brain was approximated with a spherical mesh of 40 tetrahedrons and a diameter of 3cm. Table 1 shows the material parameters that we obtained.

TABLE I
Obtained parameters for the Christensen model.

Symbols	C_k [Pa]	τ_k [s]	k
	100		0
	15000	2	1
	5000	90	2

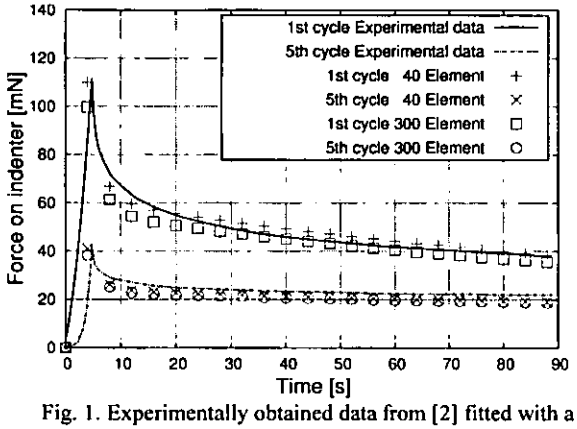


Fig. 1. Experimentally obtained data from [2] fitted with a Christensen model. Both experiment and simulation consisted of repeated 4mm indentations with v=1mm/s. The end position was held for 90 sec. After a 45sec waiting period the next cycle was started.

To account for the effect of deformation during previous cycles, repeated measurements were fitted with a single set of parameters whereas Gefen et al. fitted each cycle with a different set of parameters. The tissue was assumed compressible (κ = 500Pa) which explains higher values of C_1 and C_2 in our case.

Fig. 1 shows the experimental data, the results of the 40 elements mesh used for the fit, and the results of a 300 elements mesh used for validation. The forces simulated with the 40 element mesh are within 7% of the experimentally measured forces (below perceivable difference for humans [4]). The same set of parameters fits the experimental data equally well at both experimental velocities: v = 1 mm/s and v = 3 mm/s (Plot for v = 3 mm/s omitted due to lack of space).

3. Performance evaluation

Non-linear finite element simulations using the Verlet explicit integration scheme were performed to evaluate the computational costs and performance of the Christensen model. The FE code is multithreaded, which results in a 30% performance gain on a dual-core machine. Element masses are lumped on the nodes to avoid inverting the mass matrix at each time step.

In order to damp numerically induced high frequency vibrations a third relaxation time (C_3=10000Pa, and τ_3=0.001sec) was added to those given in Table 1. Since its short time constant targets high frequencies, it did not affect the quality of the fit.

When compared to a Neo-Hooken model, which does not evaluate the last (viscoelastic) term of (1), the additional cost of the Christensen model is relatively small: the haptic update rate decreases by 10% for each added relaxation time. Updating 1000 elements takes 1.2 ms on a 2.2GHz 64 bit AMD Athlon dual core PC and computation time scales linearly with the number of elements. The current model can still benefit from speed optimizations, for example the exponential functions in the Christensen model can be approximated by Taylor expansion if a constant time step is assumed.

We evaluated haptic performance with a Freedom6 haptic interface (MPB Technologies Inc, Montreal) and a 3 DOF haptic rendering algorithm which ran at 1000 Hz in a separate thread. The viscoelastic effects are perceivable both graphically and in the force feedback.

The current bottleneck in rendering refined meshes has proven to be the critical time step for the stability of the Verlet algorithm. For example, the critical time step is 0.1ms for a mesh with elements of 1 mm. We are currently evaluating different strategies to increase the critical time step.

4. Conclusion

A good fit for in-vivo experimental data of brain tissue was obtained with a 5 parameters Christensen model. The computational burden of this model is relatively small: the update rate with 2 relaxation times is 80% that of the Neo-Hookean equivalent. The advantage is a perceivable more accurate haptic and graphic rendering of the viscoelastic behavior of the tissue. Additionally, numerically induced high frequency vibration can be filtered with an additional relaxation time without affecting behavior at low deformation rates.

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