31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Real-time, Haptics-enabled Simulator for Probing *ex vivo* Liver Tissue

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Abstract— **The advent of complex surgical procedures has driven the need for realistic surgical training simulators. Comprehensive simulators that provide realistic visual and haptic feedback during surgical tasks are required to familiarize surgeons with the procedures they are to perform. Complex organ geometry inherent to biological tissues and intricate material properties drive the need for finite element methods to assure accurate tissue displacement and force calculations. Advances in real-time finite element methods have not reached the state where they are applicable to soft tissue surgical simulation. Therefore a real-time, haptics-enabled simulator for probing of soft tissue has been developed which utilizes preprocessed finite element data (derived from accurate constitutive model of the soft-tissue obtained from carefully collected experimental data) to accurately replicate the probing task in real-time.**

I. INTRODUCTION

THE development of complex surgical procedures coupled with more stringent regulations on medical education has recently promoted the need for developing reality-based surgical simulators for medical training. Regulations on the maximum working hours permitted for medical residents limits the amount of time available to practice common surgical tasks such as needle insertion, probing, cutting, dissecting and electrocauterizing tissue. Additionally, fatalities related to surgical procedures constitute the seventh highest mortality rate in the United States. In light of these facts, surgical simulators need to be developed to provide thorough training for common surgical tasks.

Surgical simulators are now commonly classified into three categories: anatomy, physics and physiology based simulators [1]. The first category consists of simulators based on the geometrical structure of the anatomy involved in the simulation tasks. Simulators of this kind have been mainly used for endoscopic procedures where little interaction with the environment is required [2]. Anatomy-based simulators do not incorporate any physical components such as tissue deformations. Physics and physiology based simulators have recently become the focus of several research groups due to the drive for more comprehensive surgical simulators. It is essential to replicate a surgical procedure as accurately as possible; therefore physics based approaches must be utilized to maximize the realism of the surgical task since visualization alone is insufficient.

Many researchers are currently studying the material behavior of soft biological tissue in an effort to develop constitutive models that can be used for simulation purposes. Precise characterization of the interactions between surgical instruments and soft tissue are the framework from which simulators are derived. Therefore it is essential to model the material response of the tissue as accurately as possible. Detailed studies of biological samples shows that the tissue (brain, liver, kidney, etc.) behave in nonlinear fashions under different loading conditions tested [3]-[6]. These nonlinear characteristics vary between tissue types and loading conditions; however it has been observed that most tissue will act in a linear manner for only small displacements. In surgical procedures, the tissue often experiences very large deformations; therefore linear models such as the mass-spring representation are insufficient to properly characterize the material deformation during surgical tasks.

Finite element models have become the focus of current surgical simulation applications due to their increased accuracy and continuous representation of volumetric tissue deformation. The improvement in realistic deformation however, comes at the cost of computational complexity. New techniques are required to solve the finite element problem at speeds fast enough to produce real-time visual (30Hz) and haptic (1000Hz) displays. Linear elastic finite element methods have been implemented in studying tissue deformation and hepatectomy simulation [7],[8]. However, simulations built upon linear elastic models can only be applied to small deformations and are not applicable to general surgery procedures.

Real-time finite element methods involving nonlinear material characteristics have begun to gain attention in the research community. Picinbono *et al* developed a simulator for laparoscopic liver surgery based on a St. Venant-Kirchoff model [9]. Although this is a nonlinear application, it is the simplest hyperelastic model and is only well suited for small strain and high rotation analysis. A Mooney-Rivlin model was utilized in the simulation of eye surgery by Sagar *et al*; however, even with a simple geometry, update rates were not sufficient to relay real-time information [10]. Wu *et al* also implemented a Mooney-Rivlin model for the process of liver probing. By using adaptive meshing techniques and 4,500

Manuscript received April 7, 2001. This work was supported in part by the National Institutes of Health under Grant 1R01EB006615 and the National Science Foundation under Grant 0704138.

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elements an update rate of 20 frames/s was achieved [11]. An endoscopic simulator was also developed utilizing a Neo-Hookean material; however, in order to calculate the analysis in real-time, a custom built, highly parallel computer was needed [12]. To the authors' knowledge no real-time finite element methods have been implemented for materials governed by the Ogden model or the more detailed models developed for accurate characterization of soft-tissue deformation.

The real-time, haptics-enabled soft-tissue probing simulator presented in this paper utilizes a detailed, preprocessed finite element analysis derived from an accurate constitutive model of soft-tissue and realistic geometry. Though other researchers are mainly interested in developing methods to efficiently use preprocessing techniques in surgical simulation; the focus of this paper is the ability to utilize the more complicated nonlinear material models in conjunction with realistic tissue geometry to obtain a more accurate representation of soft tissue response to tool interactions. Preprocessing the tissue response is a means to incorporate the more detailed Odgen model and complex geometry into the simulation for improved accuracy of tissue response to a probing action.

II. Materials and Methods

A. Experimental Data and Constitutive Modeling

A test apparatus was developed to record the force and displacement data for soft tissue samples in tension, unconfined compression and pure shear deformation modes. Using this experimental data various constitutive models were analyzed; the Ogden model was selected for the simulator due to the accurate fit over the range of desired deformation (0-50%) and the simplicity of implementation into the ABAQUS finite element environment. For details in experimental procedures and constitutive modeling please refer to the authors' previous work [13].

B. Finite Element Analysis

To increase the deformation and haptic accuracy, finite element methods have been utilized in determining the tissue response to a probing action. Due to the complex nature of the finite element analysis (FEA), the time required to accurately simulate the deformation process limits its direct use in the simulator. However, by preprocessing the data, the deformation and force information obtained during the analysis can be directly loaded into the simulator for display in real-time.

1) FEA Parameter Verification: Before performing the FEA on realistic soft-tissue geometry it is important to check the accuracy of the settings with the finite element analysis setup. An analysis of an unconfined compression test was completed for which the force vs. displacement data obtained from the simulation was compared with the theoretical results. This will verify that all settings, such as element type and size, are appropriate for use in the analysis of the realistic tissue geometry. All FEA simulations were conducted using ABAQUS software.

The simulation setup consists of a cylindrical test sample that undergoes loading identical to an unconfined compression test. The material properties of the test sample were defined by the Ogden model generated from the experimental data. The geometry was meshed with 1329 quadratic tetrahedral elements (C3D10H).

In the theoretical analysis, the principal Cauchy stresses for the Ogden model are defined as:

$$\sigma_j = \sum_{i=1}^{N} \mu_i \lambda_j^{\alpha_i} - p \qquad (j = 1,2,3) \tag{1}$$

where N, μ_i and α_i are experimentally defined constants, λ_j is the stretch ratio and p is the hydrostatic pressure.

For uniaxial compression, let $\lambda_1=\lambda$ be the stretch ratio in the direction of compression, and $\sigma_1=\sigma$ the corresponding principal Cauchy stress. The other two principal stresses are *assumed to be* zero, since no lateral forces are applied in the simulation. Because the material is incompressible, we have $\lambda_2 = \lambda_3 = \lambda^{-1/2}$. The stress can be expressed as:

$$\sigma = \sum_{i=1}^{N} \mu_i \lambda^{\alpha_i} - p \qquad \text{and} \qquad 0 = \sum_{i=1}^{N} \mu_i \lambda^{\frac{-\alpha_i}{2}} - p \tag{2}$$

Utilizing the relationship between nominal force and stress ($F = A_{CS}\sigma\lambda^{-1}$) in combination with (2) results in the theoretical nominal force applied to the tissue sample:

$$F = A_{CS}\left(\mu_1\left(\lambda^{\alpha_1 - 1} - \lambda^{\frac{-\alpha_1}{2} - 1}\right) + \mu_2\left(\lambda^{\alpha_2 - 1} - \lambda^{\frac{-\alpha_2}{2} - 1}\right)\right) \tag{3}$$

where F is the nominal force and A_{CS} is the cross sectional area of the tissue sample.

2) FEA of Ex-Vivo Tissue Sample: A lobe of soft tissue was modeled using Pro/Engineer to develop the geometric representation of the tissue sample to be analyzed (figure 1). The size and shape of the sample is within the range of lobes obtained during the experimental tests (approximately 180cm X 115cm X 40cm at extremes).

The geometry was imported into ABAQUS and assigned material properties defined by the Ogden model. It was constrained in all directions along the bottom surface and meshed with 28820 quadratic tetrahedral elements. A 10mm hemispherical probe was then modeled as a rigid body in

Fig. 1. Model of *ex-vivo* lobe of liver using Pro/Engineer

ABAQUS. The rigid body assumption was utilized due to the fact that a metallic probe will be much stiffer than the soft tissue. The probe was aligned perpendicular to the surface of liver in the center of the sample.

A contact analysis was performed whereby the probe is displaced in the vertical direction and introduces displacement conditions on the surface of the liver. The contact between the probe and the tissue is assumed to be frictionless.

The simulation was implemented in two steps. Initially a uniform gravitational load was applied to the tissue sample. This proved to be an essential step during the analysis because it allowed for the settling of tissue that naturally occurs due to the compliant nature of the soft-tissue. Without this step, the calculated forces would be much lower than the actual measured values. Finally, the probe was moved into contact with the tissue and continues to a distance of 1.55cm which corresponds to 47.5% strain. The simulation was run in 40 equal steps corresponding to 0.3875mm displacements where the equilibrium position is found at each displacement condition. Upon completion of the analysis the nodal displacement values and the reaction force applied to the probe can be determined.

C. Real-time, Haptics-enabled Simulator

A real-time, haptics-enabled probing simulator was developed using the data collected during the finite element analysis. The simulator consists of a graphical display, implemented through OpenGL programming framework, and a haptic display, through the use of a PHANToM haptic device (SensAble Technologies, Version 1.5A).

The graphical display is developed by rendering the surfaces of the elements generated in ABAQUS. A collection of the nodal locations and connectivity information at each step of the analysis is read into the OpenGL framework and displayed on the screen. By selecting a step size of 0.3875mm in the finite element analysis, the variation between steps is very small and visually approximates a continuous representation of the tissue deformation. This prevents noticeable jumps between states during the probing motion. Simple lighting conditions were added to the code to enhance the three dimensional aspect of the simulation as well as provide additional depth information that aids in displaying the tissue deformation.

The simulator has been programmed to allow the user to toggle the display of the probe on and off. This allows the user to get the detailed depiction of the tissue deformation without being hindered by the display of the probe. The tissue can also be rotated about all three axes and translated in all directions to provide the user will all possible views of the tissue.

The PHANToM haptic device is the interface between the user and the simulator. By moving the stylus the user is able to control the position of the probe in the simulation. Additionally, the forces acting on the probe are relayed to the user in real-time to provide the sensation that the user would feel when probing a real piece of tissue.

The force data that was collected during the finite element simulation needed to be modified prior to haptic display. Due to the sensitivity of human touch, a simulation using the 40 discrete force states would result in a discontinuous sensation that would not accurately represent the proper tissue response. Therefore the discrete force output from the ABAQUS simulation was fit with an exponential curve that relates the reaction force output to the displacement of the probe.

III. Results and Discussion

A. Experimental and Modeling Results

A full description of the results from the unconfined compression, tension and full shear tests as well as the fit of the various material models can be found in the previously published work [13].

B. FEA Results

1) Parameter Verification: Comparison of the force vs. displacement plots (figure 2) for the ABAQUS simulation and the theoretical values show a high degree of correlation. The largest error that occurs is 0.0003 N. This shows that the ABAQUS simulation is in fact quite accurate. Therefore, we can use the same parameters that were used this analysis (such as element type, boundary conditions, material properties, etc.) with a high degree of certainty in the analysis of the whole ex-vivo model.

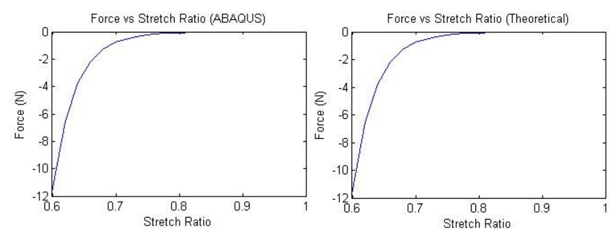

Fig. 2. Force output from ABAQUS simulation and theoretical calculations.

2) Ex-vivo Tissue Sample: The representation of the tissue geometry following the application of the gravitational load and probe displacement is depicted in figure 3. This graphical display is the displacement of the tissue in the vertical direction. As can be seen on the boundary of the tissue sample, the model settled significantly (average of about 0.5cm) during gravitational loading as is expected with such a compliant material.

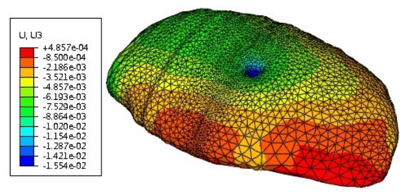

Fig. 3. Displacement, in meters, of tissue in vertical direction of final displacement induced by the probe

An irregular mesh (represented by the black lines in figure 3) was utilized in the FEA to incorporate smaller elements in the area of loading, where most displacement will occur, and larger elements on the peripheral of the geometry, where the tissue will be rather unaffected by the probe. This approach allows for the small element size required to interact with the moving probe while keeping the total number of elements as low as possible to reduce the required computational complexity.

C. Real-time, Haptics-enabled Simulation

To achieve higher resolution in the haptic display a curve was fit to the force data calculated through the ABAQUS simulation to interpolate between the discrete data points. It is given by:

$$F = -0.009991 e^{-373.6x} \tag{4}$$

where F is the reaction force in Newtons and x is the displacement in meters. This curve has an R-square value of 0.9998, corresponding to a very close fit to the data generated in ABAQUS. Equation (4) is used in the simulation code to determine a continuous representation of the force as the PHANToM stylus is moved. It is analogous to equation (3), however due to the complex geometry the derivation of an analytical expression is not feasible. Therefore the relationship is formulated from the simulation and is valid only for the given geometry and loading.

A cross-section of the soft-tissue sample taken from the simulator (figure 4) shows that the 3-D tetrahedral elements throughout the tissue results in a full volumetric study of the sample deformation. This is an important fact to consider in surgical simulation where the desired task often involves components inside a tissue volume, such as in performing biopsies and prostate brachytherapy procedures, for example.

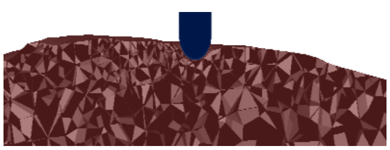

Fig. 4. Cross section of tissue sample from simulator.

The final version of the simulator is successful in displaying the graphics and force feedback to the user in real-time. An image of the setup is shown in figure 5, where a user is controlling the simulation through the PHANToM haptic device.

IV. Conclusion

The current simulator provides the framework to efficiently display preprocessed displacement and force data obtained from finite element analysis in real-time. This approach greatly increases the accuracy of the force-feedback as well as the visual deformation when compared to the current trend of linear material models commonly utilized in real-time surgical simulators. The technique is well suited for surgical procedures with a finite number of tool paths such as in prostate brachytherapy, for example. Complications in this procedure arise due to the fact that the tissue deforms as the needle is inserted resulting in bead placement errors. During prostate brachytherapy, a template is used to guide the needle into the prostate. This template is fixed and contains approximately 100 locations where the needle can be inserted. A simulator based on the methods described in this paper would present an accurate representation of the tissue deformation and force feedback that the physician would experience in a live procedure, thereby providing a valuable training tool.

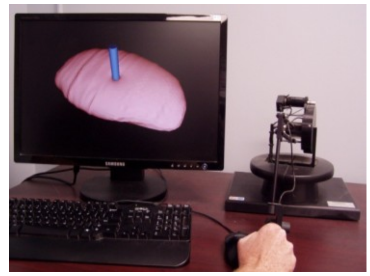

Fig. 5. Real-time, haptics-enabled probing simulator

The approach used in this paper is limited in its scope for general surgical simulation in that all desired tool paths must be simulated and preprocessed prior to constructing the simulator. Advancements in the system are planned through the development of real-time finite element analysis techniques specific to our new combined logarithmic/exponential and Ogden model. This will improve the accuracy of the simulator by using a more realistic constitutive model of the soft-tissue and also provide the framework for the training of more complex surgical tasks.

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