Modeling Large Deformation in Soft-tissues: Experimental results and Analysis*

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Abstract: Biomechanical model of soft tissue derived from experimental measurements is critical for developing a reality-based model for minimally invasive surgical training and simulation. In our research, we focus on developing a biomechanical model of the liver under large tissue deformation. This paper presents the experimental apparatus, experimental data, and methods to model the experimental data through finite element simulation and also compare it with the hyperelastic models in the literature. We have designed and developed tissue indentation equipment for characterizing the biomechanical properties of the liver and compared the local effective elastic modulus (LEM) derived from experimental data with that from plane stress and plane strain analysis in ABAQUS and hyperelastic models in the literature. Our results show that the experimentally derived LEM matches closely with that derived from ABAQUS in plane stress and plane strain analysis and the Ogden hyperelastic model.

1. Introduction

In surgery, probing soft tissue is one of the most common tasks to ascertain the tissue characteristic as being hard or soft. Hence, reality-based modeling of soft tissues is critical for providing accurate haptic feedback to the surgeon in surgical training and simulation. By reality-based modeling, we are interested in modeling tissues as accurately as possible by determining the mechanical properties experimentally. In the literature, most modeling efforts assume the mechanical properties of the soft tissue and develop methods to efficiently solve the tissue simulation problem for robot-assisted surgery/training. The goal of this paper is to derive the local effective modulus (LEM) of the liver tissue as it is compressed over a large range and compare the experimental data with the finite element simulation in ABAQUS and some hyperelastic models in the literature. In our study, pig liver was used as the sample tissue for deriving the material properties. The technique developed in this paper can be easily extended to characterize the material properties of other soft tissues as well.

"Global" elastic deformations of real and phantom tissues have been studied extensively in previous work, through simple poking interactions [1]. However, these

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methods are simplistic since they do not consider the complex boundary conditions that are normally present, both internal to the organ and on the exterior surface. Howe and colleagues [2] have developed a "truth cube" for validation of models, however they have not yet extended this model to tool-tissue interactions for common surgical tasks such as probing tissues. Ottensmeyer [3] and others have performed tissue experiments to characterize force vs. displacement for pig liver tissue, however the tissue displacement was small and they have not characterized the mechanical properties of the liver.

The quantitative knowledge of the biomechanical property of tissue is essential for soft tissue modeling. Fung [4] showed that the elasticity property of rabbits' mesentery could be simply expressed as an exponential function. While this characterization is important, it is necessary to understand the deformation of the surrounding region as well when the tissue is probed in surgical simulation. Therefore, it is necessary to derive the biomechanical properties of the liver both at the point of interaction and the surrounding region, which is valid for both small and large deformation [5].

The rest of this paper is organized as follows. In section 2, we describe the materials and methods used to derive LEM for the liver over a large deformation range. We present the mathematical formulation of the large probe model and the computational method using ABAQUS. In section 3, we present our experimental results and compare the various hyperelastic models with the experimental and ABAQUS data. Finally in section 4, we make some concluding remarks and discuss our future work in this area.

2. Materials and Methods

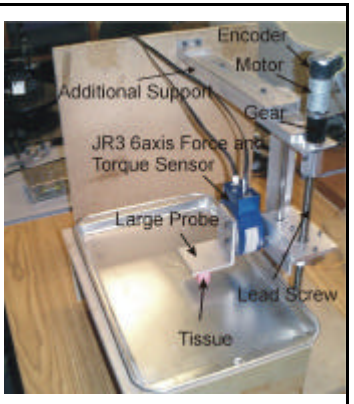

Fig. 1. Experimental setup for measuring force and displacement during tissue compression tests.

We have designed and developed a soft-tissue compression apparatus, which can measure the compressive force and displacement. Figure 1 shows the configuration of our experimental system. The system consists of a motion control part, a force measuring part, and a post-data processing part. The motion control part is a lead screw assembled with a geared DC motor and encoder (Maxonmotor, Inc.), which is supported by two horizontal supports. The anti-backlash nut in the lead screw prevents any backlash in the mechanism. A precision JR3 6 axis force/torque sensor (model 85M35A-I40) was attached to the probe and it travels along the lead screw as shown in Figure 1. The position of the probe was controlled by the dSPACE DS1103 controller board (dSPACE, Inc.) and it also records the force and displacement data. The size of the probe was 50mm x 50 mm and the surface was polished and covered with petroleum jelly to minimize the contact friction force. The size of the probe ensured that the liver sample fully contacted with the probe surface

and contact over the entire surface was maintained as the tissue was compressed. We also assumed that over the entire range of compression the volume of the sample was conserved (though in reality this is not generally true). The liver samples were taken from freshly slaughtered pigs and transported to the lab within 2 hours post mortem. A cubic sample with the size of 10 mm x 10 mm x 10mm was cut from the liver. In the compression experiment, the liver sample was compressed to 30% of its nominal thickness, and the probe speed was 6.096mm/min.

We used ABAQUS to build a finite element model which simulated the compressing process. The LEM values were iteratively obtained by matching the computational results with the experimental force data.

3. Results

3.1 Comparison of LEM computed with 0.5% and 1% strain increment

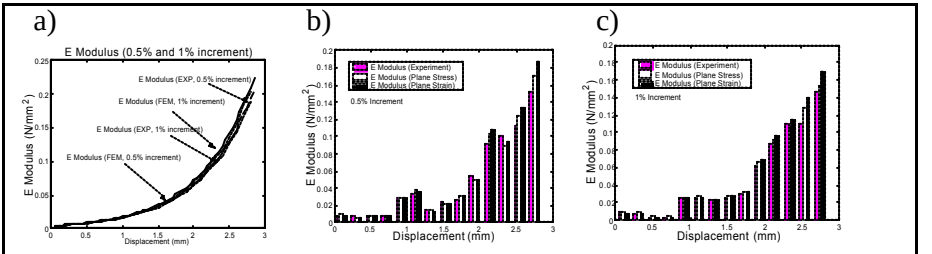

Fig. 2. a) Representative plot of the LEM computed with 0.5% and 1% strain increment. Local effective modulus computed from FEM with plane stress and plane strain analysis for: b) 0.5% increment in strain and c) 1% increment in strain in FEM simulation.

Figure 2(a) shows the representative plot of LEM computed with 0.5% and 1% strain increment at a given displacement of the probe. As seen from the plot, the LEM values for 0.5% and 1% strain increments are very close to each other for a given displacement. Based on the observed closeness in the local effective modulus for 0.5% and 1% strain increment from ABAQUS simulation, in subsequent plots we assume that 1% strain increment for analyzing other tissue samples should be adequate.

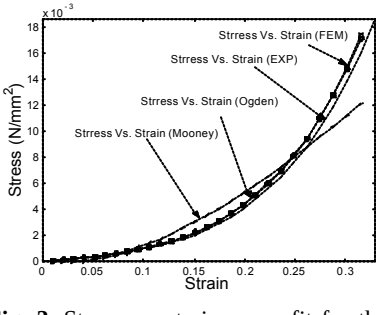

Fig. 3. Stress vs. strain curve fit for the Ogden and Mooney-Rivlin model with the ABAQUS data. Experimental stress vs. strain curve is also shown.

3.2 Comparison of Hyperelastic models with Experimental Analysis and ABAQUS Computations

Finally, we were interested in comparing the various hyperelastic models with the experimentally observed and ABAQUS computed stress vs. strain curve for large tissue deformation. Figure 3 shows the plot for a particular sample. We did this analysis for all the four samples. Table 1 shows the various parameter values for the Mooney-Rivlin and Ogden model. The experimentally observed and ABAQUS computed stress vs. strain curve fits well with the Ogden model. However, it should be noted that agreement of the Ogden

	Parameter	Sample 1	Sample 2	Sample 3	Sample 4
Mooney-Rivlin Model	C_{10}	0.039	0.067	0.052	0.063
	C_{01}	-0.041	-0.067	-0.052	-0.066
Ogden Model	μ_1	-0.228	-0.221	-0.179	-0.380
	μ_2	0.162	0.230	0.177	0.272
	μ_3	0.067	-0.002	0.007	0.109
	a_1	11.085	7.625	6.663	13.514
	a_2	11.265	7.913	7.062	13.616
	a_3	10.996	-25.000	5.846	13.465

Table 1. Mooney-Rivlin and Ogden model parameters for different tissue samples.

model is valid only for quasi-static analysis presented in this paper. Validity of this model for normal probing speeds is the area of future work.

4. Conclusions and Future Work

We have designed and developed a tissue compression apparatus for characterizing the mechanical response of liver tissue using large probe analysis. In our initial work, the liver tissue is assumed as incompressible, isotropic, and homogeneous elastic material. The total displacement of the tissue is up to 30%. We used 0.5% and 1% strain increment to calculate the experimental and ABAQUS computed LEM. The results showed that the experimental and computational LEM were very close to each other for a given displacement. We concluded that the two dimensional ABAQUS model can provide a reasonable estimate of the LEM in the tissue over the range of compression. Finally, we compared the various hyperelastic models with the experimentally observed and ABAQUS computed stress vs. strain curve for large tissue deformation. Based on the results of all four samples, we concluded that the experimentally observed and ABAQUS computed stress vs. strain curve fits well with the Ogden model.

The work presented in this paper represents the initial step toward developing a reality-based model for tool-tissue interaction during probing. The results presented in this paper are valid only for quasi-static analysis. Our next step would be to compute the LEM over a range of probing speeds. The computational model developed with this research will be the basis for solving the complex small probe problem.

References

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