Characterization of the Biomechanical Properties of the Lower Esophagus for Surgical Simulation

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Abstract. In this work a method to characterize soft tissue properties for mechanical modeling is presented. Attention is especially focused on developing a model of the lower esophagus to be used in a surgical simulation, which shows a promise as a training method for medical personnel. The viscoelastic properties of the lower esophageal junction are characterized using data from animal experiments and an inverse FE parameter estimation algorithm. Utilizing the assumptions of quasi-linear-viscoelastic theory, the viscoelastic and hyperelastic material parameters are estimated to provide a physically based simulation of tissue deformations in real time. To calibrate the parameters to the experimental results, a three dimensional FE model that simulates the forces at the indenter and an optimization program that updates new parameters and runs the simulation iteratively are developed. It was possible to reduce the time and computation resources by decoupling the viscoelastic part and elastic part in a tissue model. The comparison of the simulation and the experimental behavior of pig esophagus are presented to provide validity to the tissue model using the proposed approach.

1. Introduction

Over the last few years, laparoscopic Heller myotomy has emerged as a viable surgical treatment for esophageal motility disorder [1]. This procedure involves palpations of the long tubular-shaped organ as well as skin incisions of the outer muscle layer at the lower esophageal junction (Fig. 1). The success of the procedure depends on the training of the surgeon, thus it is reported that multiple trials of the procedure significantly reduced the reoperation and complication rates. However, the rate of occurrence of conditions requiring such a procedure is relatively rare; therefore doctors have limited opportunities to hone their skills in this area. To provide more opportunities without sacrificing the safety of patients, virtual reality-based surgical simulators for the procedure are being developed. However, the lack of biomechanical data of the esophagus has been a great impediment to the development of surgical simulator with acceptable fidelity [2]. This paper presents an integrated framework for measuring, modeling, and calibrating the material properties of the lower esophagus as a step toward developing a soft tissue model. Data were collected from anesthetized pigs using a robotic indenter and a force transducer. The constitutive model was fit to the experimental data using an optimization algorithm combined with a finite element simulation. The soft tissue was modeled as continuous, viscoelastic, non-linear, incompressible and isotropic. These bio-mechanical models are suitable for computing accurate reaction forces on surgical instruments, and computing deformations of the esophagus during the surgical procedure.
2. Methods

2.1 Animal experiments

The pigs’ lower esophaguses were measured under open surgical conditions to measure their mechanical properties at the Harvard Center for Minimally Invasive Surgery [3]. A total of 10 pigs were used in these experiments. The pigs were first put under general anesthesia and placed on the surgical table. A midline incision was then made at the abdominal region and dissection carried out on the anatomical structures in order to expose the organs. The indentation stimuli were delivered using a haptic interface device, Phantom Premium-T 1.5 (SensAble Technologies, www.sensable.com) that was programmed to perform as a mechanical stimulator. Reaction forces were measured using a six-axis force transducer, Nano 17 (ATI Industrial Automation), which was attached to the tip of the haptic interface device. The transducer had a force resolution of 0.781 mN along each of the three orthogonal axes when connected to a 16-bit A/D converter. The indenter was a 2 mm diameter flat-tipped cylindrical probe that was fixed to the tip of the haptic interface device with the force transducer mounted in-between to accurately measure the reaction forces.

2.2 Material modeling

The quasi-linear viscoelasticity (QLV) framework proposed by Fung [4] was used. This approach assumes material behavior can be decoupled into two effects: a time-independent elastic response, and a linear viscoelastic stress relaxation response. These models can be determined separately from the experiments. The stresses in the tissues, which may be linear or nonlinear, are linearly superposed with respect to time.

The three-dimensional constitutive relationship within the framework of QLV is given by [5],

\[
S(t) = G(t)S^*(0) \int_0^t (t - \tau) \frac{\partial S^*(E(\lambda))}{\partial \tau} d\tau
\]

where \(S(t)\) is the second Piola-Kirchhoff stress tensor and \(G(t)\) is termed the reduced relaxation function. \(S^*(E(\lambda))\) is termed the pure elastic response of the material and can be nonlinear or linear. The reduced relaxation function \(G(t)\) is a scalar function of time and can be often expressed by the Prony series,

\[
G(t) = G_0(1 - \sum_{i=1}^{N} \overline{g}_i^\sigma (1 - e^{-t/\tau_i}))
\]

where \(\overline{g}_i^\sigma\) are the Prony series parameters.

For the nonlinear elastic response, a linear elastic material model is utilized, as the hyperelastic material did not appear to have any advantage over the linear-elastic model in this case. It is important to mention here that the hyperelastic model may show a better performance in the case of larger indentations on solid organs, such as the liver. As an analytical solution considering the above material law and experimental conditions is very difficult, the Finite Element Method (FEM) has been widely used in these simulations.

2.3 Inverse FEM simulation

During the characterization, the compared quantities are the simulated forces from the FEM simulation and the associated experimental forces measured at the indenter. Therefore, we can minimize a nonlinear sum of squares given by
\[ E = \sum_{i=1}^{m} (F_s(t_i) - F_e(t_i)) (F_s(t_i) - F_e(t_i)) \quad t_i = (t_1, t_2 \ldots t_m) \] (3)

where \( F_e, F_s, t_i, \) and \( m \) are measured forces, simulated forces, time and the total number of data points, respectively. Among several optimization algorithms that could be used, the nonlinear least square optimization known as the Marquardt-Levenberg algorithm was adopted. Fitting the experimental data to the mechanical model is necessary in order to complete the modeling procedure. An analytic solution based on the boundary value problem was not a good candidate, given the complexity of the material behavior, the organ geometry, and the three-dimensional deformation imposed on the surface. To circumvent these difficulties, inverse finite element estimation was considered for the characterization of the soft tissue properties [6]. This method estimates unknown material parameters for a selected material law by minimizing the least-squares difference between predictions of a finite element model and experimental responses, as shown in Fig. 1. In many cases, the FEM model does not require a geometrical model, as the size of indentation is much smaller than the organ size. However, unlike for solid organs, for hollow organs such as esophagus it is necessary to incorporate the geometrical information in the simulation. For this reason, it was necessary to incorporate the data of the organ geometry and boundary conditions. Fig. 2 shows the modeling of the esophagus structure. The esophagus was modeled as a tubular structure 60 mm in length. As the abdominal esophagus is well attached to the diaphragm at the upper end and to the fundus (the upper part of stomach) at the lower end, as shown in Fig. 2, both ends can be modeled as fixed ends. 30mm of Hg internal pressure is also modeled in the simulation.

3. Results

Fig. 2 shows the predicted forces from the FE simulation with the estimated parameters and experimental forces for the selected experiments of the lower esophagus. The Young’s modulus of the esophagus was estimated, from the inverse FEM modeling, to be 5.222kPa. The relaxation time constant and the first order of the Prony parameter are 6.372 and 0.363sec, respectively. The force responses of the hyperelastic model in ABAQUS match the experimental data well.

4. Concluding Remarks

In this paper, the mechanical properties of the lower esophagus have been characterized from animal experiments. To calibrate the parameters to the experimental results, a three dimensional FE model that simulates the forces at the indenter as well as an optimization program that updates new parameters and runs the simulation iteratively were developed. Key assumptions in the present approach are that the organs are incompressible, homogenous, and isotropic, and that the deformations imposed were small compared to the size of the organ. With these limitations in mind, the material models presented in this study offer two basic uses in a VR-based medical simulation. First, they can be used directly in the simulator to compute visual deformations and interaction forces that are displayed in real time. Second, the mathematical models presented here can be used as a standard for the evaluation of new real-time algorithms for computing deformation. Similar to other applications that use optimization, it is difficult to guarantee the uniqueness of the solution due to the locality of the algorithm. One of the possible solutions to this difficulty, which was utilized in this study, is to optimize from a range of initial conjectures, and to verify if the final estimate is physically plausible. Ex vivo tests of organ properties remain helpful for formulating the initial conjecture and checking the final estimate. If the initial estimates are chosen without considering a priori knowledge, the algorithm may generate a physically unacceptable solution and lead to a failure to converge in the FE simulation. Therefore, accumulating as much knowledge as possible is still important in order to reach a physically meaningful solution as well as for reducing the number of iterations needed in order to achieve a convergence.
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References


Figure 1. Anatomical location of the lower esophagus (L). Schematic of the inverse FEM parameter estimation (R).

Figure 2. FE simulation of the indentation experiment on the esophagus muscle tube (L). Simulated and experimental responses of the esophagus indentation considering the geometry (R).
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