

A TECHNIQUE FOR MEASURING LOCAL INTERNAL MECHANICAL PROPERTIES OF PERFUSED SOLID ORGANS

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INTRODUCTION

Mechanical simulation of soft tissue is increasingly important for the development of new minimally invasive surgical applications. For surgical planning, the prediction of tissue deformation in response to instrument manipulation can improve accuracy of image-guided procedures. For surgical training, simulations permit new surgeons to safely practice a wide variety of procedures. To be effective in these applications, mechanical simulations must be realistic, so that the tissue behaves as in the living organism. This requires accurate *in vivo* constitutive models for tissues under the large deformations typical of surgical procedures.

The complexity of soft tissues makes characterizing their behavior a formidable challenge. Soft tissues exhibit viscoelastic properties, anisotropy, and inhomogeneity [1]. These tissues are drastically altered when no longer perfused, maintained at constant temperature, or bounded by the original surrounding structures as in the commonly tested *ex vivo* state [1, 2].

To make meaningful measurements requires an experimental environment that is close to the *in vivo* state and provides force and displacement measurements of the local tissue volume over time. *In vivo* devices have recently been developed that probe the tissue of interest from its surface and measure the corresponding force response [2-5]. However, these devices typically apply small perturbations to the organ surface and thus only measure superficial regions and do not account for large surgical strains. At present, imaging techniques (e.g. elastography) provide only qualitative or relative material property information for arbitrary organ geometry [6].

We have developed a new technique for probing the local mechanical properties of solid organs such as the liver, kidney, and spleen at arbitrary internal locations while approximating *in vivo* conditions. A pair of cylindrical bars is carefully inserted into the organ on each side of a local volume causing minimal damage to the volume of interest and allowing for a nearly uniform stress distribution. In the following sections we describe the apparatus and preliminary tests on porcine liver under physiologic conditions.

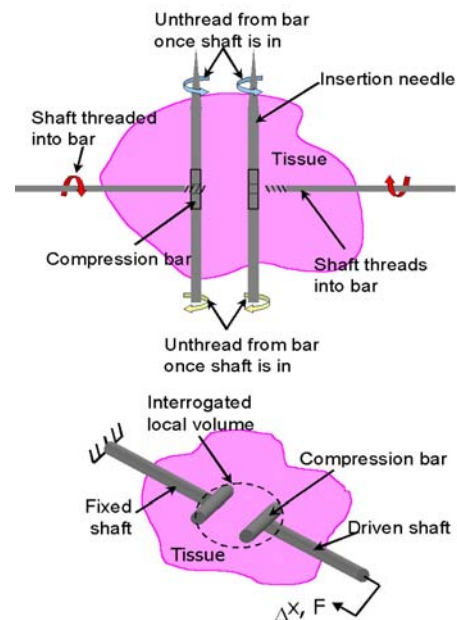


Figure1: (Top) Inserting the T-needles into the tissue. (Bottom) T-Needles probing local volume of interest.

METHODS

Experimental Apparatus

The device uses a pair of “T-needles” as mechanical probes for force-displacement measurements within an organ (Figure 1). To minimize damage to the tissue, the shafts and crossbars of the needles are inserted separately using a block alignment system, and assembled within the tissue. First, extensions are threaded onto the ends of the bars: the leading extension has a sharp point to cut through the tissue as it is inserted and the trailing extension provides the insertion force and controls the insertion direction. This needle assembly is inserted into the tissue and advanced to the correct position for mating with the

shaft. Second, the shafts are advanced into the tissue perpendicular to the insertion needles and threaded into the bars. The shafts are locked into place on their blocks and the bar extensions are removed. A cannula with a beveled end is inserted over the shaft on the driving side to minimize the effects of friction. The volume "resolution" of the device is defined by the diameter of the crossbars (3 mm in this prototype) and the spacing between the bars (19 mm).

The device applies controlled displacements between the two compression bars using a motor and linear potentiometer while measuring forces using a strain gauge-based force sensor. A computer controls the desired trajectory. The force and displacement data are recorded using a 16-bit analog-to-digital converter sampling at 5kHz. The device can test tissue under large strains (>30%) and low frequencies (<10 Hz) typical of surgical manipulations. Operation of the T-Needle prototype was confirmed using silicone tissue phantoms.

For the preliminary testing of porcine liver, the organ is placed on a floating plate in a bath of Lactated Ringer's solution, allowing free translation during testing to minimize static forces. To approximate *in vivo* conditions, a perfusion apparatus uses hydrostatic pressure generated by containers suspended at appropriate heights to maintain a constant, nonpulsatile flow throughout the liver. Flow is established to the hepatic artery at a pressure of approximately 120 mm Hg and to the portal vein at 50 mm Hg. The fluid is allowed to freely exit the liver at the hepatic vein into the basin where it is circulated by a pump to the suspended containers. An immersion heater maintains the temperature of the bath at 36 C.

Preliminary Protocol

Two preliminary tests using the T-needles were performed on both unperfused and perfused whole organs to determine the best protocol for future tests. Freshly harvested porcine livers were perfused with a mixture of normal saline and heparin and then iced for transport to the testing lab. The unperfused test used the liver from a 42.5 kg pig. Data was collected 3.5 hours after death. To test loading rate dependencies, the tissue was compressed at different ramp speeds, and then the force response was measured over time to determine the relaxation time constants. The ramp speeds ranged from 1 to 80 mm/s at a displacement of 2mm (11% strain) and then held for 60 sec. The perfused test used the liver from a 46 kg pig. Perfusion began 2.5 hours death, and testing began 20 minutes later. The ramp speeds ranged from 0.5 to 100 mm/s at a displacement of 7 mm (36% strain). Three data sets were collected at 2 and 6 hours post perfusion at a ramp speed of 1 mm/s. The last test for both organs compressed the tissue at a steady 1 mm/s rate until failure.

The data was filtered in MATLAB using a moving weighted average. Fung's quasilinear viscoelastic theory was chosen as the model to fit to the force relaxation data using a simplex method [1].

PRELIMINARY RESULTS

Results for both perfused and unperfused livers show a viscoelastic response with hysteresis, force relaxation, and nonlinear force-displacement. The unperfused organ was stiffer than the perfused organ, with a force of 1.2 N versus 0.6 N at 2 mm compression. The unperfused organ showed lower failure loading of 52% strain at 5 N versus 84% strain at 6.5 N for the perfused organ

Figure 2 shows the ramp and hold filtered data and model fit for the perfused organ at a ramp speed of 60 mm/s. The quasilinear model for force relaxation was fit using a sum of discrete time constants. The best fit was found for three time constants plus a steady state term. The fitted time constants for this data are 102 ms, 1.47 s, and 27.76 s with weights of 167, 140, and 225 mN respectively and a steady state value

of 1.23 N. Lastly, between 2 and 6 hours into perfusion (4.5 hours post mortem) little or no change in the observed behavior was noted.

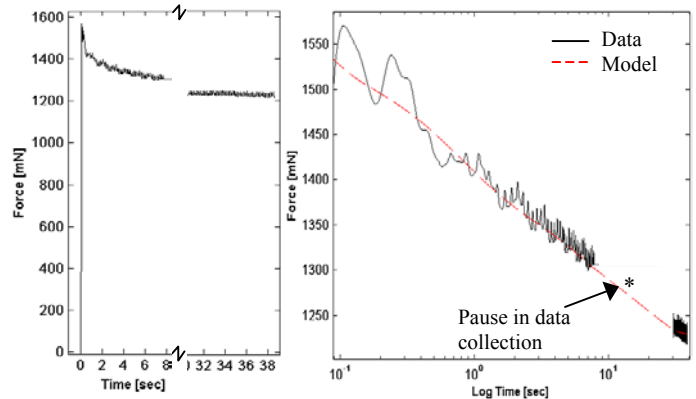


Figure 2: (Left) Ramp and hold of perfused organ at ramp speed = 60 mm/s. (Right) Hold phase with model fit

DISCUSSION

The preliminary results on porcine liver constitute a proof of concept test. Although the values for single specimens are not definitive, they do serve to demonstrate the approaches ability to collect relevant data under approximately *in vivo* conditions. These data are also collected from a relatively small region of the liver thus minimizing the volume over which inhomogeneities are averaged. The scale of the device can be reduced as needed to improve resolution by decreasing the bar diameter and the spacing between the two bars.

The force-displacement data illustrated by the preliminary tests can be directly used to evaluate aspects of the time response of the tissue. To determine the elastic response, however, requires more elaborate analysis. In particular, the variation of strain levels within the test volume (between and around the bars) makes it difficult to determine the nonlinear constitutive law from the data. One approach to this problem is the use of iterative finite element modeling. In this method, a constitutive law is assumed and a finite element simulation of the T-needle and tissue produces a prediction of the force-displacement relationship. Comparison with the experimental data is then used to update the constitutive parameters until a good fit is obtained. Issues of uniqueness of the parameters and adequacy of the experimental excitation must be addressed and validated against different loading conditions.

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