ABSTRACT

Surgical simulation is a field that shows much promise in terms of permitting surgeons to learn new skills, as well as enabling the development of new procedures and instruments while minimizing the use of animal subjects. Essential to simulation with realistic force feedback are material property data for the wide variety of soft tissues, both in healthy and diseased conditions. Properties are known to change significantly after death, so measurements should ideally be made in vivo. This paper reviews a number of methods in use to perform in vivo testing. It focuses on ongoing work using a minimally invasive instrument that uses normal indentation to acquire the force-displacement response of solid organ tissues such as liver, spleen and kidney. The design and characterization of the instrument as well as some of the in vivo and other test results will be presented.

I Introduction

Flight simulators aid in the training of pilots without risk to themselves or their passengers. Surgical simulators have the same potential to permit surgeons to acquire basic skills or practice unusual or advanced procedures without risk to their patients. Software-based simulations employing visual and force-feedback are beginning to approach the sophistication required for such training and practice, but face a number of challenges before they will become useful on a broad scale.

Before generating curricula and validating the training approaches, technical hurdles need to be overcome including developing force-feedback instruments, creating algorithms for generating force-feedback data at speeds up to one kilohertz, and of relevance here, determining the values of real tissue material properties, in vivo. Tissue properties in vitro are available from a number of sources [1, 2], however it is well recognized that such properties may be significantly different from those of living tissue. In addition, properties for diseased or otherwise abnormal tissue may be required, as well as variations due to age or other normal physiological conditions.

To acquire these properties, researchers are using a wide variety of techniques to measure force-displacement or stress-strain responses of tissues. The tissue parameters can be extracted by assuming some model of tissue behavior and performing inverse calculations to determine the parameters given the observed response and the known boundary conditions. A variety of the measurement techniques will be described in the next section.

II In vivo measurement techniques

These measurements fall into two categories: non-invasive imaging-based which typically examine the strain field within tissue; and invasive (minimally or otherwise) methods that measure local force-displacement responses of tissue. The former include techniques in which fixed or oscillatory displacements are applied to the exterior surface of tissues and the changes in internal structure are observed. The latter include a broad range of techniques in which tissue is indented, stretched, twisted, or otherwise locally deformed.

II.1 Non–invasive methods

A review of these methods, typically referred to as “elastography” can be found, for example, in [3], but the principles will be covered here. Essentially, some known displacement, which may include static or dynamic compression or shear displacement, is applied to the exterior of the tissue under consideration. Since the tissue is continuous, the surface displacement produces a strain field within the tissue, which is a function of the surface displacement, the elastic properties of the tissue and other boundary conditions. The strain field is measured through the use of some non-invasive imaging technique, including, but not limited to, magnetic resonance imaging (MRI) and ultrasound. In the static case, “before” and “after” scans are made, and the strain can be calculated based on the local displacements measured between them. For dynamic testing, vibration amplitude or velocity can be used instead of static strain.

[4] describes an ultrasound-based, static deformation technique which has been used to image breast tissue in vivo. The paper shows the utility of the technique in identifying tumors, which may often not be distinguishable from healthy tissue from ultrasound imaging alone, however they did not report quantitative data on the material properties of the various structures. [5] and [6] are examples
of the use of dynamic excitation, which use MRI to examine the propagation of shear waves within the tissue, and provide some data on the shear modulus of tissues. [6] employs vibration frequencies from 10 to 1000Hz, and provides results not only on porcine liver and bovine muscle in vitro, but also human breast and brain tissue in vivo. The latter was obtained by vibrating thermoplastic bars held in the teeth of the volunteers and show that the white and gray matter have stiffnesses of 14.2 and 5.3kPa respectively.

[7] uses a pair of ultrasound units to simultaneously send waves through tissue in one direction, and measure the motion of the tissues from an approximately orthogonal direction. Given transit times and the dimensions of the various structures within the tissue, the sonic velocity can be calculated, which is related to the stiffness of a given material. They have been able to distinguish between fat and other tissues, but do not yet have finer discriminability.

Non-invasive techniques have shown some success, but are subject to certain limitations. Tissues are known to exhibit non-linearity in their stress-strain relationships, and the majority of the non-invasive techniques can only examine the small strain, linear portion of the response. Further, the static techniques cannot examine the visco-elastic behavior of tissues, while the dynamic techniques require excitation frequencies large enough so that the wavelength of sound is small enough compared with the structures under consideration. Of interest to surgical simulation developers is the behavior of tissue under large strain conditions, and over frequency ranges from the static to a few tens of Hertz. This is because of the large deformations (up to and beyond failure) involved in surgery in the former case, and because of the bandwidth of human motor responses in the latter. Human beings can generate controlled motion up to approximately 3 or 4 Hz, and reflexes up to about 10Hz [8], so property data of tissue in this range is especially desired.

For these and other reasons, various researchers are also pursuing invasive measurement techniques in which deformations are directly applied to the tissues in question.

II.2 Invasive methods

As mentioned, there are a wide variety of techniques that have been developed to investigate the force-displacement response of tissues. This is a result of the variety of tissue types and structures within the body, including solid and hollow organs, and the type of excitation that the researcher chooses to employ. For in vitro measurements, the boundary conditions can be fairly well defined, and testing can be performed on tissue samples shaped in whatever form is convenient to the experimenter. For testing on living tissue, the instrumentation must not only acquire the desired data, but also be suitable for use in often restricted environments (e.g. minimally invasive surgery), and must not cause trauma to the tissue. A sampling of instruments that have been developed which satisfy these criteria follows.

A number of groups have developed instrumentation that applies a normal indentation to the tissue. [9] performed tests on porcine brain, using a spherical indenter and a ramp and hold displacement. The immobilized head provides rigid boundary conditions around the brain. [10] examined human liver using a hand-held instrument employing a spherical indenter and a larger concentric tube. Since the liver is not rigidly supported, the depth of indenter penetration is measured with respect to a tube, which contacts the tissue and provides a surface reference. [11] have developed an instrument which similarly uses a central probe and surrounding tube, but here the central probe is rotated, measuring the angular displacement and torque. An area of interest is the method used to ensure that the rotating probe does not slip with respect to the tissue; arrays of needles have been used, and adhesives are under investigation.

Porcine esophagus has been studied using a Phantom haptic interface as a robot arm, in which position is relative to ground using the optical encoders of the instrument, and a 6-axis force-torque sensor records the applied loads [12]. This method is more versatile than some of the other techniques, in that the Phantom can move in three dimensions, whereas the devices previously described measure only one-dimensional responses.

A number of groups have modified surgical instruments, equipping them with force and position sensors [13], as well as motors [14] to support computer control. With these instruments, the response of tissue to grasping can be measured. In the case of [14], in which the geometry of the grasper jaws is well defined, conversion from force-displacement to material property is somewhat simpler.

[15] makes use of a pair of opposing graspers, which hold and then stretch tissue between them. Extensive use of this instrument has been made on hollow porcine organs (intestine, stomach and gall bladder), some of the results of which are available in [16].

At least two groups have developed aspiration-based techniques [17, 18], in which a tube is placed against tissue, a seal is achieved, and then the pressure within the tube is lowered. The motion of the tissue surface as it is slightly deformed into the tube is measured optically. [17] uses a clear tube, while [18] employs a novel mirror arrangement, in which a profile view of the tissue surface is captured with a camera placed at the tube end, well outside the body, making it suitable for use during an open surgical procedure.

As mentioned above, of particular interest is a frequency region covering the range of human motor control. To pursue these data, another indentation instrument, called the TeMPeST 1-D, for 1-axis Tissue Material Property Sampling Tool, was constructed, and has since been used to acquire data on porcine liver and spleen in vivo, as well as a variety of rat tissues in vitro.

III TeMPeST 1-D instrument

The TeMPeST 1-D [19] (see Figure 1) was designed explicitly to investigate the visco-elastic properties of tissue under small deformations. As testing of human tissues was a long term goal, a minimally invasive form was chosen so that it could be used either during open or laparoscopic surgical settings. Portability was also an issue, so only the instrument and a laptop and docking station need to be brought into the operating theatre. Further, the shaft of the instrument, containing the force and position sensors and a
voice coil actuator, can be separated from the handle, so that the shaft can be sterilized. When in use, the instrument is fixed to the operating table with a laparoscope holder.

Figure 1. TeMPeST 1-D instrument inserted through 12mm surgical cannula

The instrument has a range of motion of +/-500 µm, can exert forces up to 300mN, and has an open loop mechanical bandwidth of approximately 80Hz. The circular indenter tip has a diameter of 5mm, and the overall diameter of the instrument is 12mm, suitable for use with standard surgical cannulas (though 10mm and 5mm versions are more common).

To verify the performance of the instrument, it was initially used to measure a variety of standard objects and materials, including mechanical springs, standard masses and a number of silicone gels. Three different grades of the latter were created so that the results from parallel plate rheometry could be compared with the TeMPeST 1-D results. Good agreement between the TeMPeST 1-D and the "gold standard" rheometer were shown.

IV TeMPeST 1-D animal tissue testing

The instrument has been used to measure the responses of porcine liver and spleen in vivo, as well as rat liver, kidney, lung, heart and abdominal muscle in vitro. For the in vivo testing, a typical animal tissue measurement sequence proceeds as follows. Ventilation with pure oxygen is suspended briefly so that breath motions are not measured superimposed on those generated by the instrument. A waveform, which may be sinusoidal, a chirp or some other arbitrary signal is commanded to the voice coil motor, which deforms the tissue with the 5mm punch. Sampling is complete in under 20 seconds, at which time ventilation resumes. A typical view through the laparoscope during testing is shown in Figure 2. Since the sampling interval is short and pure oxygen is used for ventilation, no harm comes to the animal during testing. Similarly, since the displacements of the tissue surface are small, no damage is incurred from the instrument. These issues are essential to consider early on, especially because successor instruments are planned for human use.

While the results of testing are frequency (and load) dependent stiffness values, they are dependent on the geometry of the indenter and the tissue being tested. To extract material parameters, some model of tissue behavior must be assumed, and the geometry of interaction must taken into account. For the large, solid porcine organs, a useful first approximation is that the tissue is isotropic and homogeneous (generally valid for liver or spleen), incompressible (approximately true for water-rich living tissue) and linear. This last assumption depends on the use of small deformations, relative to the size of the organ. Further, if the small deformation approximation is valid, then the organ can be treated as a semi-infinite body, which provides for closed form relationships between Young's modulus and the measured stiffness and the geometry of the indentation probe [20]. In (1, $k$ is the measured stiffness, $a$, the radius of the indenter, $K$, a geometric factor which is unity for a semi-infinite body, and greater than one for a thin material on a rigid substrate, and finally $E$ is Young's modulus.

$$E = K \frac{3k}{8a}$$

This expression can be extended to a frequency dependent form if $k$ and $E$ are replaced by functions of frequency, allowing viscous and inertial effects to be taken into account. An example of the results, showing $E$ as a function of frequency and preload is shown in Figure 3.

For the case when the organ is small compared with the motion of the instrument (e.g. rat kidney), or when the tissue model is more complex than the extremely simplified one described, finite element models can be employed, iteratively approaching the observed response by modifying the parameters given the measured geometry and stiffness. Analysis of the rat kidney and liver data is ongoing.

V Summary

Measuring the material properties of living tissue is a complex problem, not only because of the nature of the tissues, but also because of the limits imposed by minimizing or preventing damage to the test subject. As a result a wide variety of techniques have been and continue to be developed, including non-invasive ones, employing external stimulation and imaging technology, as well as invasive techniques which directly contact and deform the tissue to be studied. No one technique is suitable to capture the full range of behaviors, including non-linear stiffness, visco-elasticity, anisotropy and inhomogeneity. However each has its own advantages, and can be used to contribute data to and cross check the results of the others.
Figure 3. Frequency dependent and non-linear elasticity of porcine liver. Squares are results from sinusoidal tests, point clouds from chirp response.

Acknowledgement

This work was supported in part by the Department of the Army, under contract number DAMD17-99-2-9001. The views and opinions expressed do not necessarily reflect the position or the policy of the government, and no official endorsement should be inferred.