Determination of the Mechanical Properties of Soft Human Tissues through Aspiration Experiments

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Abstract. Mechanical models for soft human organs are necessary for a variety of medical applications, such as surgical planning, virtual reality surgery simulators, and for diagnostic purposes. An adequate quantitative description of the mechanical behaviour of human organs requires high quality experimental data to be acquired and analyzed. We present a novel technique for the acquisition of such data from soft tissues and its post processing to determine some parameters of the tissue's mechanical properties. A small tube is applied to the target organ and a weak vacuum is generated inside the tube according to a predefined pressure history. A video camera grabs images of the deformation profile of the aspirated tissue, and a pressure sensor measures the correspondent vacuum level. The images are processed and used to inform the fitting of uniaxial and continuum mechanics models. Whilst the aspiration test device has been designed to fulfill the requirements for in-vivo applications, for measurements obtained during open surgery, initial experiments performed on human cadaveric tissues demonstrate the ability to both differentiate between different organs and also between normal and diseased organs on the basis of the derived mechanical properties.

1 Introduction

The mechanical characterization of soft biological tissues is essential to a number of medical applications, such as surgery planning, surgical training deploying virtual reality based simulators, or diagnosis (see [1],[2],[3],[4]). Quantitative sets are available on the mechanical properties of soft tissues, however very limited data are available on the in-vivo behaviour of soft tissues associated with human organs ([5],[6],[7]). This deficiency is primarily due to the severe technical

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and ethical problems related to any investigations. Direct access to the internal organs is necessary for most techniques, leading to interference and disruption of the primary surgical procedure. The standard methods of material testing, such as tensile or bending experiments, are not appropriate for in-vivo application [8]. Procedures for quasi static tissue testing have been proposed, most of them based on indentation experiments ([9],[10],[11],[12]). Yet these techniques lack the necessary constraints to impose the well-defined boundary conditions required to allow accurate mechanical properties to be fitted to the acquired data.

We present an alternative technique based on tissue aspiration in which relatively large local tissue deformations are imposed and the time dependent behaviour of the tissue can be observed. The experiment is characterized by well defined kinematic and static boundary conditions and the time required for each measurement is very short (of order of 20 seconds). The aspiration device presented in this paper is an optimized version of the equipment previously presented by Vuskovic and Kauer ([5],[6]). The device that has been designed for the in-vivo characterization of the mechanical behaviour of human organs during open surgery with minimal disruption to, or prolonging the duration of, the operative procedure. Initial studies presented here are restricted to the testing of adult human organs (livers and kidneys) obtained from cadavers made available during post-mortem examinations. Future research efforts will focus on the application of this technique in-vivo during open surgery.

2 Aspiration Device and Investigations

2.1 Aspiration Device

The aspiration test device shown in figure 1 has been developed by V. Vuskovic [5]. The device has been designed for in-vivo applications addressing issues associated with: safety, sterilizability, space limitation and a short data acquisition cycle time. Several modifications in the acquisition and in the control system have been introduced in order to make the device suitable for the current research requirements.

The principle of working of the device is based upon the pipette aspiration technique [13]. The device consists of a tube in which the internal pressure can be controlled according to a desired pressure law. The investigation is performed by (i) gently pushing the tube against the tissue to ensure a good initial contact, (ii) creating a (time variable) vacuum inside the tube so that the tissue is sucked in through a smooth edged aspiration hole (diameter of 10 mm), see figure 2.

Assuming the tissue to be isotropic and homogeneous in the small portion under deformation, a complete description of the deformed tissue can be given by simply monitoring the side-view profile of the tissue during its deformation. An optic fiber connected to an external source of light provides the necessary illumination in the inner part of the tube.

The images of the side-view (figure 3) are reflected by a mirror and are captured at a frequency of 25 Hz by a digital camera mounted on the upper part

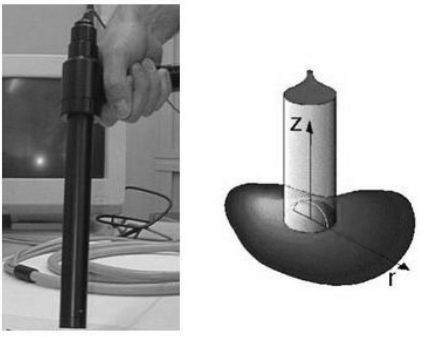

Fig. 1. Aspiration device and principle of working

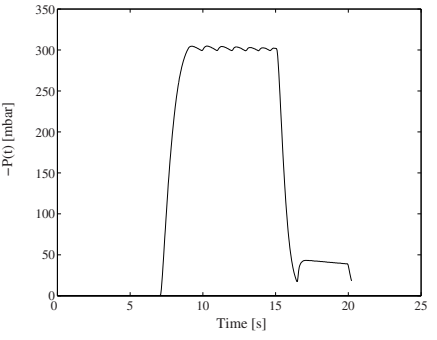

Fig. 2. A typical vacuum law imposed inside the tube

of the device. The grabbed images are processed off-line in order to extract the profiles of the deformed tissue (figure 4). A normal personal computer (running NI LabView Version 6.1) controls the pressure inside the device by means of a pump, an air reservoir and two valves. With such a control system, a predefined pressure law can be accurately realized and therefore the repeatability of the experiment is tightly bounded. The time of the experiment is about 20 seconds and the magnitude of the vacuum (maximum allowed 300 mbar) is deemed insufficient to cause tissue damage. Time histories of measured pressure and deformation profiles are the input data used to evaluate the mechanical properties and to determine the constitutive model.

2.2 Investigations

Investigations have been performed on organs obtained from adult cadavers during post-mortem examination. In the UK there are standard permissions to access post mortem examination rooms and materials for research and educational purposes. In addition the researchers obtained specific permissions from the Director of Laboratory Medicine, Central Manchester and Manchester Childrens' Hospital NHS Trust.

The cadavers were kept in refrigerated storage at +4 degrees centigrade (+/− 1 degree centigrade). The dissections took place in the post mortem examination room which is maintained at ambient temperature (c. 20 degrees centigrade) at least 30 minutes after the cadavers were removed from the cold storage facility.

The organs were removed from the cadaver before the testing and subsequently replaced back into the body cavity. Initial studies were limited to obtaining data from kidneys and livers. Two kidneys and three livers, all from different cadavers were tested. The applied tests were non-invasive and non-destructive. Tissues were not taken nor retained from test sites. There was no permanent deformation of these sites. For each organ multiple test were conducted both at

different locations on the organ's surface but also using different time-pressure profiles. In line with normal post-mortem procedures, the tested organs were subsequently dissected and where appropriate tissue blocks (not from the test sites) taken by the attending pathologist for diagnostic purposes.

One of the three livers tested was clearly different from the other two livers tested in terms of colour and size. Histological examination confirmed the presence of hepatic fibrosis/early cirrhosis and widespread hepatocytic steatosis most likely due to prolonged excessive alcohol intake.

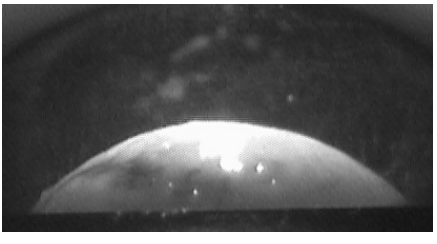

Fig. 3. Image grabbed by the digital camera

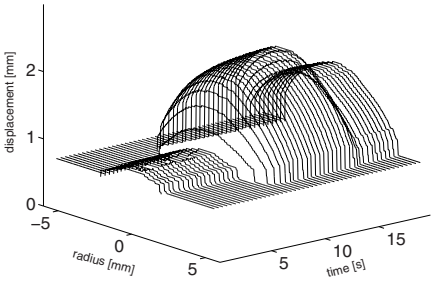

Fig. 4. Profiles of deformation extracted from the images

3 Mechanical Modelling

3.1 Uniaxial Model

A first evaluation of the obtained data can be performed by modelling the 3-D state of stress and deformation in the aspiration experiment by a simple 1-D problem, by correlating the displacement of the highest point of the aspirated tissue "bubble" to the instantaneous pressure inside the device. In the literature, there are a variety of different uniaxial models describing viscoelasticity ([14],[15]). We implemented different uniaxial models based on different combinations of lumped parameters elements (springs, dashpots and Coulomb elements). A simple model that closely fits the experimental pressure-displacement curves is represented in Fig. 5 and it is a composition of a linear spring with spring constant η, two linear dashpots with coefficient of viscosity μ_1 and μ_2, and, in order to obtain a nonlinear relation, a Coulomb element of dry friction (for which the displacement is null if the force applied (P) is less than a certain critical value (P_0) and it is set to an arbitrary constant if the force is higher than that value).

Through an optimization routine, we determined the constants of such model that best fit the experimental curves obtained (see figure 6). Figure 7 shows the obtained values for the spring constant η. This clearly demonstrated that it is possible to distinguish using this simple model, between kidneys and livers, and

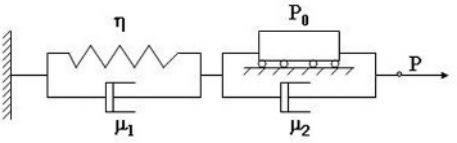

Fig. 5. Uniaxial model

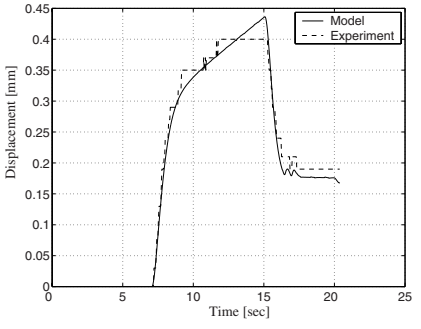

Fig. 6. Measured (dashed line) and calculated (continuous line) vertical displacements

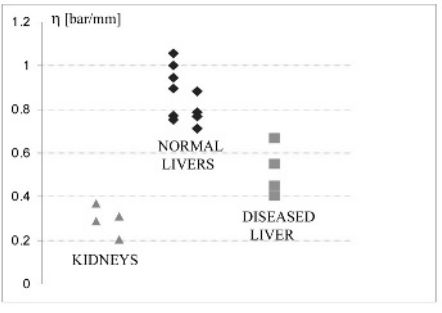

Fig. 7. Values of the spring constant η for the normal livers, the diseased liver and the kidneys

between normal livers and a diseased liver. In particular the two kidneys are softer (smaller value of the constant η) than the three livers, and the diseased liver under consideration was softer than the normal ones.

3.2 Continuum Mechanics Constitutive Model

A more accurate model for the tissue under consideration can be provided by a 3-D model based upon a continuum mechanics approach. Such models can be applied to simulate the deformation of the material under any loading condition. We consider the material as an homogeneous and isotropic continuum, showing a quasi-linear viscoelastic behaviour [14] and due to the high water content, as nearly incompressible. Hyperelastic materials are usually described in terms of a "strain potential energy" U which defines the energy stored in the material as a a function of the strain. We used the so called "reduced polynomial form" [16].

$$U = \sum_{n=1}^{N} C_{n0} \left(\overline{I}_1 - 3\right)^n + \frac{1}{D}\left(J - 1\right) \tag{1}$$

where J is the total volume change, $J = \lambda_1\lambda_2\lambda_3$ (λ_i: principal stretches), $\overline{I}_1 = \overline{\lambda_1^2} + \overline{\lambda_2^2} + \overline{\lambda_3^2}$ is the first strain invariant $\left(\overline{\lambda_i} = J^{-\frac{1}{3}}\lambda_i\right)$, and D is a material constant that takes into account the small material compressibility.

Viscoelasticity is taken into account by applying time dependent (relaxation) coefficients to the constants that define the energy function:

$$C_{n0}(t) = C_{n0}^{\infty}(t)\frac{\left(1 - \sum_{k=1}^{K} \overline{g}_k^P \left(1 - e^{-\frac{t}{\tau_k}}\right)\right)}{\sum_{k=1}^{K} \overline{g}_k^P} \tag{2}$$

where $C_n^{\infty}0$ (long term elastic module), $\overline{g}_k^P$ and τ_k are the material parameters to be determined experimentally. In the current implementation we chose to stop at the fifth order of the series expansion for the strain potential energy ($N = 5$) and at the fourth order for the Prony series ($K = 4$).

The experiment is simulated by an axisymmetric finite element (FE) model. The FE program Abaqus 6.2 has been used for this purpose [17]. The starting geometry of the tissue is the one grabbed by the camera at the beginning of the experiment, at atmospheric pressure inside the tube. The force by which the tube is pressed against the soft tissue leads to a non-zero initial deformation.

The FE calculation proceeds by applying the measured pressure to the free tissue surface (D-X, Fig.8). Axisymmetric hybrid triangles are used with linear pressure and quadratic displacement formulation. The overall dimensions of the model are selected in order to minimize the influence of the boundary conditions at the bottom (A-B) and at the side (B-C) on the displacement of the aspirated tissue "bubble". The contact between the tissue and the device (C-D) is modelled as rigid-deformable contact with sliding.

The material parameters are determined by minimizing the error function E

$$E = \sum_i \left(z_i - \widetilde{z}_i\right)^2 z_i \tag{3}$$

where z_i and $\widetilde{z}_i$ are the measured and the calculated displacement of the point X respectively . A proper evaluation of the material constants requires with our current configuration a time of about 60 hours for about 400 iterations for each experiment (with a Pentium4-2GHz). The implemented optimization procedure is the Nelder-Mead simplex (direct search) method [18] and the material constants obtained by this procedure provide the constitutive model of the tissue. A "physical" description of the mechanical behaviour described by that equation is hereby given by applying the general constitutive model, valid for any condition of loading, to uniaxial tension of the tissue. In figure 10 we plot the stress-stretch relation we would obtain by applying an instantaneous uniaxial stress T and measuring the stretch λ. In order to analyze the time dependance of the mechanical properties, we linearize the constitutive model for small deformations and plot the tensile relaxation modulus $E(t)$ (figure 11). Figures 10 and 11 clearly show how the three livers are stiffer than the two kidneys and how the two normal livers are stiffer than the diseased liver. The multiple traces in these figures are the result of acquiring data at different points on the organs' surface.

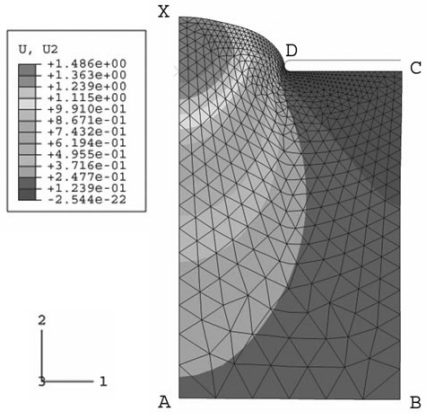

Fig. 8. Finite element model

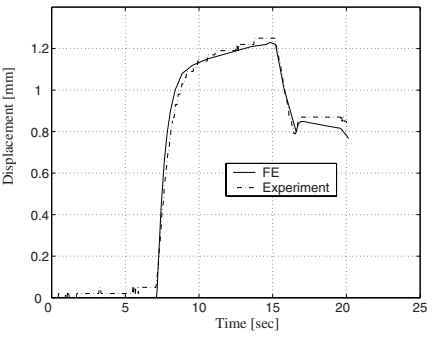

Fig. 9. Measured (dashed line) and calculated (continuous line) vertical displacements

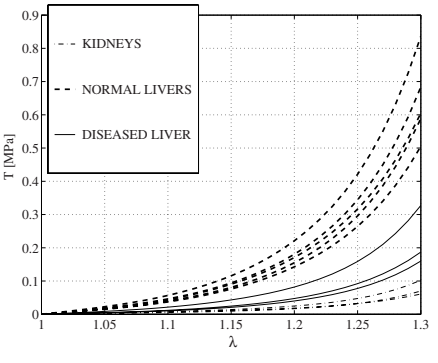

Fig. 10. Tensile curves

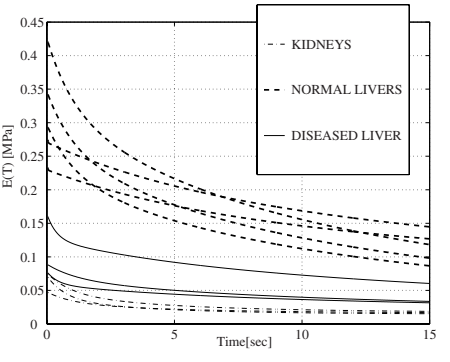

Fig. 11. Relaxation modulus $E(t)$

4 Conclusions and Outlook

A novel technique for human soft tissues testing has been proposed and tested on adult human organs obtained from cadavers during post mortem examination. Two different procedures for the fitting of the data, one based on uniaxial modelling and the other based on a 3-D continuum mechanics approach, have been implemented and the results presented. A biologically plausible constitutive model for the tissue is presented. The device demonstrated its capability to distinguish between two different types of organs (liver and kidney) and in the case of the liver between the normal ones and a diseased organ.

Future work will focus on in-vivo testing during open surgery and the creation of a database of the mechanical properties of different human organs.

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