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Liver Vessel Parameter Estimation from Tactile Imaging Information

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Abstract

Realistic tissue models require accurate representations of the properties of *in vivo* tissue. This study examines the potential for tactile imaging to measure tissue properties and geometric information about subsurface anatomical features such as large blood vessels. Realistic finite element models of a hollow vessel in a homogenous parenchyma are constructed in order to establish a relationship between tissue parameters and tactile imaging data. A linear algorithm is developed to relate the tactile data to linearized tissue parameters. The estimation algorithm shows low errors in estimating the model parameters. A preliminary study on two porcine livers results in errors on the order of 20% in estimating the liver geometry. This result is promising given the small sample size and parameter recording limitations of this preliminary study. Further work will reduce these sources of error and lead to *in vivo* testing with a minimally invasive tactile imaging scanhead.

1. Introduction

Realistic models for surgical simulation require accurate representations of the properties of *in vivo* tissue. Recent measurements of organs such as the liver have used specialized apparatus to characterize mechanical properties adjacent to the organ surface [1,2]. This study examines the potential for tactile imaging to measure tissue properties and geometric information about subsurface anatomical features such as large blood vessels. In addition to informing surgical simulation, this technique may provide a fast, simple, and noninvasive means of intraoperatively locating vessels and diagnosing diseases such as cirrhosis that are characterized by changes in mechanical properties [3].

Tactile Imaging uses an array of pressure sensors to map the surface pressures that result from indenting the tactile imager into the surface of a soft material [4]. This medical imaging modality quantifies the qualitative information provided by the human sense of touch through palpation. Tactile imaging evolved as a means of detecting and characterizing pathologies in the human breast that manifest as areas of increased stiffness [5,6] and has progressed to include similar pathologies in organs such as the prostate [7,8]. Previous work [9,10] has resulted in algorithms for estimating the parameters of stiff inclusions embedded in soft tissue, including lump diameter, depth, and modulus, as well as the modulus of the surrounding soft tissue. In vitro

mean absolute errors (MAE) for these algorithms are 5-16%, while clinical assessment shows 13% MAE for lump diameter.

These algorithms can be readily extended to the problem of estimation of liver properties. Liver anatomy shows a macroscopically homogenous parenchyma punctuated by large branches of the hepatic vein (Figure 1). These vessels leave a signature of lower pressure on the surface pressure data collected in tactile imaging, and this information can be used to estimate parameters of the vessels and the tissue in which they are embedded. Following the approach of previous work on tactile signal interpretation [4,10], we first characterize the forward relationship between tissue parameters and the tactile signal using mechanical models (Figure 2). We then develop a linear algorithm for inverting the relationship to estimate tissue parameters from tactile images (Figure 3). The algorithm is then tested on porcine livers.

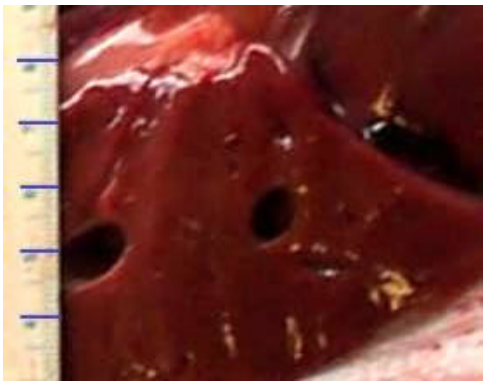

Figure 1. Cross-section through a porcine liver, showing large hepatic veins.

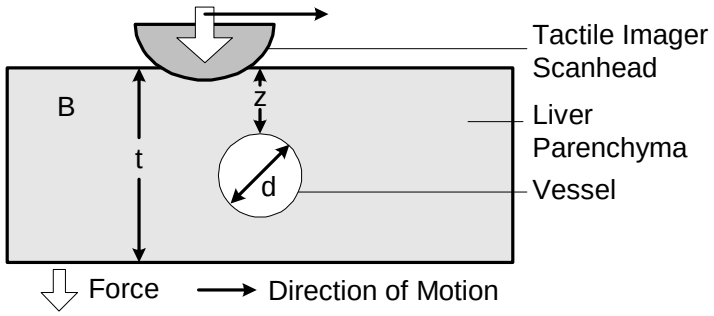

Figure 2. Model of tactile imager and liver, with a homogenous parenchyma and a single round vein.

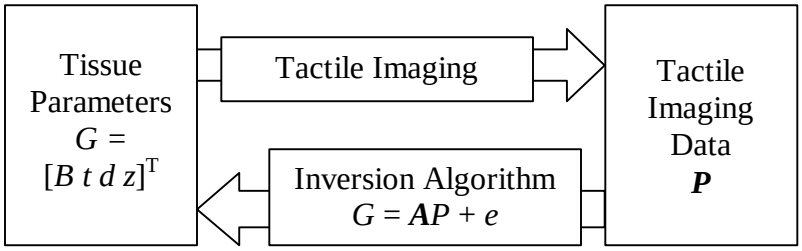

Figure 3. Imaging and signal interpretation approach.

2. Algorithm Development

A closed-form relationship between the parameters of interest and the tactile signal is not feasible due to the large strains and contact interactions of the imaging process, so finite element method (FEM) models are used to generate the tactile images for ranges of the tissue parameters of interest. A typical cross-section through a liver is shown in figure 1. This main features of this cross-section are captured by the two-dimensional model shown in figure 2; a plane strain model is used to minimize computational efforts; the plane approximates the situation along the centerline of the cylindrical tactile scanhead.

2.1. Mechanical Modeling

We constructed finite element models based on figure 2, modeling the tactile scanhead on the device used in the experimental validation, a 16 x 16 array of capacitive sensors spaced 2 mm on center mounted on a section of a cylinder with a 38 mm radius. The location and orientation of the scanhead in 6 DOF was determined by a magnetic tracker (miniBird, Ascension Technologies, Burlington VT). The model represented the tissue as incompressible, isotropic, and linearly elastic. The interaction between the scanhead and the liver was assumed well-lubricated (i.e. frictionless) and the bottom of the liver was fixed to the substrate.

The tissue parameters of interest are the background modulus, B, the tissue thickness, t, the vessel diameter, d, the vessel depth z, and the vessel pressure, V. The range for the modulus B of the tissue encompassed average moduli ranging from human to porcine livers [11]. The ranges for the geometry parameters were based on expected human anatomy [12]. The values of the parameters used in the thirty-six models created are specified in table 1. The vessels were set in the middle of the tissue so that in the models, $z = (t - d)/2$. Preliminary studies indicated that the vessel pressure had insignificant effect on the tactile data for physiological ranges, and so the pressure was set to zero for the models used here. Tactile data was calculated for every 2.5 mm displacement of the scanhead for 40 mm to either side of the vessel, approximating experimental data collection.

Table 1. Parameters for the finite element models constructed.

Parameter	Value
B	10, 12.5, 15 kPa
T	40, 50, 60 mm
d	5, 6.5, 8, 10 mm

2.2. Inversion Algorithm

For each model (i.e. combination of tissue parameters), the calculated tactile pressure data at each 2.5 mm displacement was concatenated into a single row vector P. Similarly, the tissue parameters were assembled into a column vector $G = [B\ t\ d\ z]^T$. We then characterize the problem as a linear inversion, and seek the transformation matrix $\boldsymbol{A}$ that minimizes the error e in $G = \boldsymbol{A}P + e$. The relationship between the pressure in P and the parameters $[B\ t\ d\ z]$, however, is not directly linear. Rather use the tissue parameters directly in G, we therefore look for functions of the parameters from which the parameters $[B, t, d, z]$ can be calculated, but which are more linearly related to the tactile information.

For example, if we approximate the tissue far from the vessel as a linear spring, the scanhead force is related to the equivalent spring constant B/t. This suggests that B and the function $1/t$ are approximately linearly related to the surface pressure data. Following similar arguments for the other parameters, we then use the input parameters $G = [B\ 1/t\ d/z\ 1/z]^T$.

2.3. Finite Element Model Parameter Estimation Results

For each of the model data sets, the other 35 models were used to generate the transformation matrix used in the estimation algorithm using the pseudo inverse $\boldsymbol{A} = G(P^{T}P)^{-1}P^{T}$. The results of estimating the model tissue and vessel parameters are summarized in table 2.

Table 2. Mean Absolute Error in estimating the underlying parameters of liver finite element models.

Parameter	Mean Absolute Error in Estimation
Liver Modulus *B*	0.8%
Liver Thickness *t*	9.3%
Vessel Diameter *d*	8.1%
Vessel Depth *z*	7.2%

3. Preliminary Physical Testing

3.1. Physical Data Collection

The inversion algorithm was validated on two porcine livers from healthy 40 kg pigs harvested within one hour of sacrifice. The livers were immediately flushed with Heparin to minimize clotting, and perfusion with physiological saline solution at 36°C commenced approximately one hour later. Resulting tactile images are shown in figure 4.

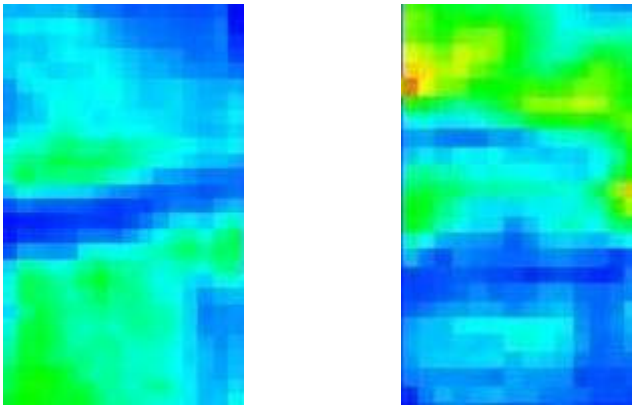

Figure 4. Tactile maps (averages of the spatially registered image sequences) of sections of porcine liver lobe, showing decreased pressure over vessels. The left image shows one vessel spanning the width of the image, indicating the presence of a large vessel beneath the surface. The image at right shows two vessels running from left to right, with the upper one leaving a smaller impression in the tactile image (due to smaller size or greater depth). The images shown are approximately 80 mm x 40 mm.

Eight vessels were found in the two porcine livers, and multiple tactile images were made of several of the vessels, resulting in 14 usable maps for testing the inversion algorithm. For each set of tactile image data, tactile frames were collected every 2.0 mm for a 40 mm linear region centered on the vessel. It was noted that the thin porcine liver lobes vary considerably in

thickness, and even over a 40 mm section the thickness varied up to 15 mm. Since the images were obtained using a hand-held sensor the pressure applied varied for each frame. The average pressure across all the frames of interest was 18.2 Pa with a standard deviation of 10.3 Pa.

For each set of data in turn, the transformation matrix was found using the other 13 sets of data and the parameter estimation tested on the set in question. The actual parameters were recorded after tactile imaging was complete by dissecting the liver lobes and measuring the vessel diameter, depth from surface, and total tissue thickness.

3.2. Porcine Liver Parameter Estimation Results

Due to the small sample size, the background modulus of the livers studied was assumed to be the same across all samples. The results of estimating the geometry parameters using our inversion algorithm are summarized in table 3.

Table 3. Results of estimating the underlying parameters of porcine livers with large embedded veins.

Parameter	Mean Absolute Error in Estimation
Liver Thickness *t*	20.0%
Vessel Diameter *d*	25.6%
Vessel Depth *z*	13.6%

4. Discussion

The algorithm tests on the finite element models showed excellent accuracy, with all mean errors under 10% across the range tested. Estimation results from physical livers resulted in errors approximately twice those of the finite element models. This is not surprising, given the unconstrained data collection in the laboratory setup and the nonlinearities inherent in the tissue properties that are not captured by a linear algorithm. A key difference between finite element analysis and physical data collection is the large range of input pressures observed during physical data collection. This variable is controlled to better than 1% in the finite element analysis, but was observed to vary more than 50% in the physical data collection, due to human operation of the tactile imaging system.

This variable input pressure range affects the data in two ways. First, due to the nonlinearity of the tissue modulus, we inadvertently probe the tissue at different effective moduli. This precluded simple normalization of the frame information by the difference in the total applied force. This change in the effective background modulus will result in an incorrect measure of the tissue thickness, as we had assumed a constant background modulus. The wide range of input pressures also affects the way the tissue is probed in that as the tactile imager is indented further into the tissue, the surface pressure effectively senses deeper tissues [13], which results in information that our simple parameter system does not characterize. This problem of a wide range of applied forces can be alleviated by implementing bounds on the tactile data, using only

data in a fixed range of total force, and signaling the user when they are operating in the acceptable range.

The preliminary experimental evaluation did not include estimation of the background modulus because only two livers were available for testing. We note that previous studies that estimated tissue modulus for in vitro models were successful, with 5.4% MAE [10]. Because modulus was not estimated in the present study, the livers were assumed to have a constant background modulus. This assumption is reasonable as the subject animals were both healthy, the same age, and raised together. Any deviations of the actual modulus will result in errors similar to those mentioned above for errors caused by a changing input force.

With only eight vessels found in the two livers studied, the parameter range was sampled only sparsely. Obtaining extra maps on some vessels allowed for estimation of all map parameters, since we were not extrapolating for any one variable. Maps taken on the same vessel were not identical, and so represented different maps taken on vessels with similar parameters. Since these double maps were not identical, however, they may have negatively affected the parameter estimation, by providing a different pressure signature for the same parameters. The duplicate maps were well spaced over the tissue thickness and vessel diameter, but were biased towards larger diameter vessels since the smaller vessels were difficult to image repeatedly, most likely due to temporary collapse. Therefore, although the total parameter range was spanned, the estimation of the vessel diameter was most likely adversely affected since the range spanned by the majority of the data was narrow, with more than one pressure profile representing the same diameters.

The actual liver parameters were recorded after all maps were taken, by cutting the lobe perpendicular to the vessel along the line of data recording. Recording the parameters this way is the most direct and readily available method, although it may have contributed to inaccuracies in vessel parameter information. Since the cutting and data recording were done by hand, the planes of tactile data and dissection may be offset by a few millimeters. In this range, the vessel diameter and tissue thickness may vary as well. The vessel diameter may vary by up to a millimeter and the tissue thickness by twice that. The liver parenchyma also was prone to swelling in the cut plane. This is due to the natural tension that is present in the liver, maintained partly by the perfusion under which the data was recorded. Perfusion was necessary, however, in order to maintain mechanical viability of the liver, so that despite the above sources of error, subsequent maps recorded on the same vessel record approximately the same conditions. These sources of errors largely affect the input parameters, and may adversely affect the apparent estimation by presenting incorrect information for the creation of the transformation matrix. These errors can contribute a relatively large error to the parameters in question, and so the above work should only be considered a proof of concept for the use of tactile scanning to record liver vessel parameters. Within these constraints, the algorithm performed remarkably well in estimating the underlying parameters.

Further experimental work should be conducted on livers after the main causes of error noted above are addressed. A method for regulating the force to a near-constant level should be implemented. This can be as simple as generating a specific sound for when the data collected is in a narrow range around the ideal input pressure [9]. An improved method for measuring the

geometry parameters without cutting the tissue should be employed. For further *ex vivo* tests using an accurate method such as MRI or CT scanning should be employed. A method to measure the parenchymal stiffness independently should also be utilized, so that the ability of tactile imaging as a means of recoding the background stiffness can be assessed. Ideally, human *ex vivo* livers will be tested before moving ahead to an *in vivo* setting using a smaller tactile imager in minimally invasive data recording.

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