Hand-held stiffness measurement device for tissue analysis

*T. A. Zodage[1], A. N. Chaudhury[2], R. A. Srivatsan[2], N. Zevallos[2] , H. Choset[2]

[1] Mechanical Engineering, BITS Pilani,

[2]Robotics Institute, Carnegie Mellon University
*tzodge@gmail.com

INTRODUCTION

We develop a novel means of detecting tumors to better assist in early detection of cancer. Several conventional methods such as MRI or CT scans detect tumors, but these methods are time consuming -- taking around 20-90 minutes, need expertise to handle the machines and are less affordable in lower income settings [3]. Due to the cost and time involved in these methods, surgeons use palpation techniques for preliminary tests on superficial tumors on the breast, tongue or the prostate. Physicians typically palpate the tissue manually to get a qualitative estimate of stiffness [4]. As these palpation techniques are qualitative, the diagnosis depend on the expertise of the doctors investigating a particular case. The qualitative nature of the estimation makes it difficult to keep records of the tissue stiffness and analyze it over time.

To quantitatively interpret palpation, researchers have worked towards developing devices for tissue stiffness measurement. Fischer [4] was one of the first to devise tissue compliance meter to quantitatively investigate soft tissue consistency. Arokoski et al. [5] and Oflaz [6] measured the tissue's resistance to a deforming force as a measure of stiffness for a constant, preset indentation depth. The use of a constant length probe in [5] and [6] limits the range of stiffness which can be measured. Jalkanen et al. [7] suggested measurement of resonance frequency piezo crystal in contact with the tissue to estimate the stiffness of the underlying tissue. However, this only gives information about the local stiffness, but not the location of the point of investigation. Nguyen et al. [3] suggested the use of tactile photoelastic films to quantify the location and shape of the tumours. These techniques involve complex setup and specialized hardware.

Taking these issues into account, this work presents a stiffness measuring device which uses a simple setup using a monocular camera [9] along with a commercial force sensor. In addition to measuring the stiffness at a point, our device also generates a map of the stiffness distribution over the surface of the tissue. This makes it easier to visualize the shape and size of the tumor (refer to Fig. 1) compared to [4], [5], [6]. Since we do not need to use a constant length indenter as [5] our device gives a wider range of operation. In order to test our device, we perform experiments on silicone tissue phantoms embedded with stiff inclusions.

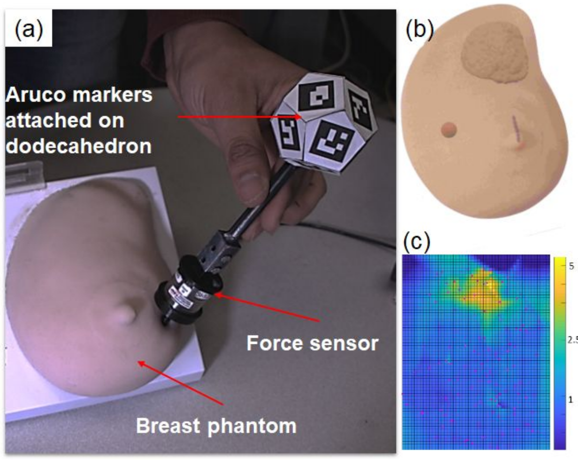

Fig. 1 (a) Our device with position sensing at the top and a force sensor at the tip. (b) Approximate location and shape of the tumors in breast phantom (c) Stiffness map generated using our device shows a stuff region in the top of the breast.

We also provide qualitative (stiffness map images) and quantitative (RMS error in stiffness) comparison results of our method to a ground truth generated using an industrial robotic arm.

MATERIALS AND METHODS

The position sensor is inspired from a recent work by Wu et al. [9] on a 6 degrees of freedom (DoF) tracking pipeline for a passive stylus for mixed reality applications. The method uses a 3D printed fiducial of known geometry (a dodecahedron of 2 cm side length in this case) with binary square markers (we used ArUCo markers [9]) glued to the faces, as shown in Fig. 1(a). The 6 DoF pose of the stylus is inferred by imaging the fiducial with a monocular machine vision camera (we used a Point Grey Chameleon3 camera with a Fujinon 2.8mm-8mm/f1.3" lens) and refining the pose estimate by performing two unconstrained minimization steps. The first step minimizes the reprojection error of the marker corners for the pose estimate and the second step minimizes the appearance distance (difference between observed and expected image intensities) to further refine the estimate on the pose. In our experiments, we achieved a worst case position error of 0.4 mm.

For the force sensor, we use a commercial 6 DoF force torque sensor (ATI nano 25E). We fabricated a tissue phantom by embedding a hard plastic of unknown shape in a silicone specimen (Smooth-on EcoFLEX-20) of dimensions 10cm x 10cm x 2cm.

EXPERIMENTS AND RESULTS

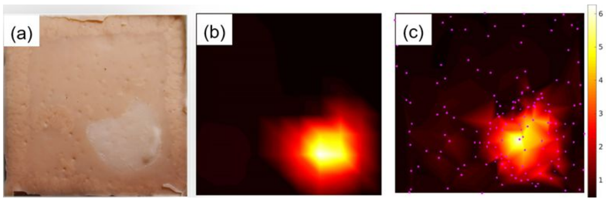

Fig. 2 (a) Flat tissue phantom with embedded plastic tumor. (b) Ground truth stiffness map generated using industrial robot arm (c) Stiffness map generated using our device. (Pink points shows probing locations)

The tissue phantom was manually palpated at 150 points at approximately uniform intervals and the visual and the force data were logged in real time. A threshold force was estimated by placing the probe in contact with the organ. The readings of position and force sensor were recorded. The stiffness at a point was calculated when the force measured by the force sensor was above the threshold force. For each indentation, we calculated the slope of force vs displacement of the indentation point, to obtain the local stiffness. This data was used to interpolate the stiffness information of the surface in between two indentation points [8].

To generate the ground truth stiffness map, we repeated the probing experiment using an industrial robot arm attached with the same force sensor. The tissue sample was probed at 100 points in a raster scan pattern. The stiffness map generated using our device was found to be similar to that of the ground truth (see Fig. 2 (b) , (c)).

In order to calculate the error in the stiffness, We selected a grid of 20x20 points at approximately same position on the ground truth and the stiffness map generated using our device. The root mean square error in stiffness was 0.86 N/mm. Work is currently underway to improve the bounds on the measurement errors.

As an edge usage, we demonstrate our device to obtain surface geometry of organs. Since generating point clouds for tissues is challenging using vision-based sensors, we use our probe to perform contact based geometry estimation. We show a 3D surface map of a kidney phantom, generated using our device in Fig. 3.

DISCUSSION

This work demonstrates the development of a hand held device for quantitative and qualitative estimation of stiffness over the surface of a tissue sample. In addition to finding the stiffness map, we also demonstrate the use of this device to find the geometry of an organ surface. Since our device uses a combination of position and force sensing, it does not require prior knowledge of stiffness range of the

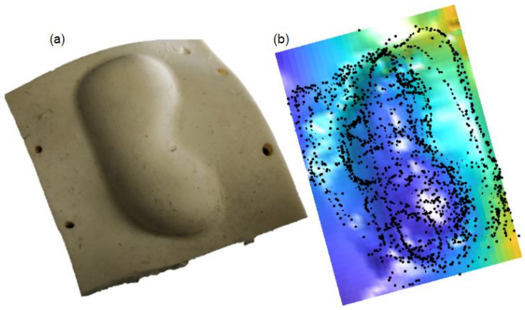

Fig. 3 (a) Kidney phantom. (b) 3D map of kidney phantom generated using our device. (Black points show tip contact locations)

organ, unlike most prior devices. As a result, our device can be used for soft organs such as the breast as well as stiffer organs such as the kidney (Fig. 1, Fig. 3).

An important benefit of our device is the low cost. However, the use of an industrial force sensor slightly offsets this benefit. As future work, we plan to integrate an inexpensive and light-weight, miniature force sensor developed in our previous work [10]. We also plan to perform tests on ex-vivo and live tissues to obtain better understanding of the problems that might arise in real world situations. Currently we process the data offline, however, future work will address this by using a C++ implementation of the pipeline for realtime tip tracking.

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