DESIGN AND FABRICATIN OF TACTILE INTERFACE FOR PALPATION

Literature Review

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REVIEWED LITERATURE:

B. Li *et al.*, "Mechanical Imaging of Soft Tissues with a Highly Compliant Tactile Sensing Array," *IEEE Transaction on Biomedical Engineering J.*, vol. 65, no. 3, pp. 687–698, March. 2018.

INTRODUCTION AND BACKGROUND:

Tactile elasticity imaging, or tactile elastography, aims to reconstruct the composition of a volume of tissue from surface mechanical measurements, in the form of stress (normal pressure and shear traction) or strain (displacement) captured by a measurement instrument. It is a valuable alternative to other imaging modalities, because it is relatively non-invasive, and able to capture surface mechanics of tissues that reflect the elasticity of sub-surface tissues. Physicians during palpation often miss nodules or lumps, due to tissue inhomogeneities, perceptual limitations in what can be felt, or to incorrect technique, leading to diagnostic errors. Equally concerning, most physical examinations currently yield little quantitative information. This can make diagnosis and prognosis error prone, and can impede tracking of disease progression over time. Imaging techniques based on softer sensors, like that presented here, could facilitate imaging of sensitive, curved, hard, or heterogeneous tissues, and could yield wearable sensors that are able to capture quantitative data during existing clinical physical examination practices.

A diverse variety of sensing methods for tactile imaging or elastography have been studied for the detection of tissue lumps or other features. Prevailing technologies have been based on optically-based mechanical sensing, capacitive strain sensing, and rigid point-force sensors based on resistive, piezoelectric and magnetic principles. Devices involved in tactile imaging frequently employ sensors distributed over rigid substrates, and as a result cannot conform to heterogeneous, stiff, highly curved, or sensitive bodily tissues. Within the broader field of mechanical sensing, several strategies have recently been developed for realizing mechanical sensors from stretchable materials, or somewhat stiffer polydimethylsiloxane (PDMS), or from flexible substrates, or through other geometric specializations, such as meshing.

PROJECT OVERVIEW:

The aim of the project is to develop a tactile screening method for the detection of the breast lumps. The tactile imaging is performed using a Force- resistance sensor and the obtained output signal is process to determine the location of lumps and provide an image of the tissue strain due to the lump.

SOFT TACTILE SENSOR:

The operating principle of the sensor used in paper is based on the changes in mutual capacitance between arrays of micro-fluid electrodes arranged in two orthogonal planes. The change in the mutual capacitance is due to the mechanical loading and deformation of electrodes which reflects the deformation of the actual tissues. The capacitance strain thus reflects local strain and indirectly normal pressure applied to the sensor.

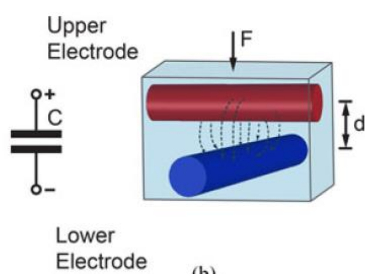

Fig 1. The sensing element consisting of upper and lower electrodes

The sensing array consists of a thin, composite membrane, comprised of three layers as shown in Fig 2. The electrodes are embedded into top and bottom polymer layers via microfluidic channels that are filled with a metal alloy that is liquid at room temperatures and above. Between these layers is a third layer, comprising a two-dimensional array of micropillars, which make it possible to tune the operating range of pressures and the electronic sensitivity of the device to application requirements. Due to the low elastic modulus of the soft polymer substrate, the sensing devices can recoverably deform to applied strains greater than 200% without mechanical or electronic damage.

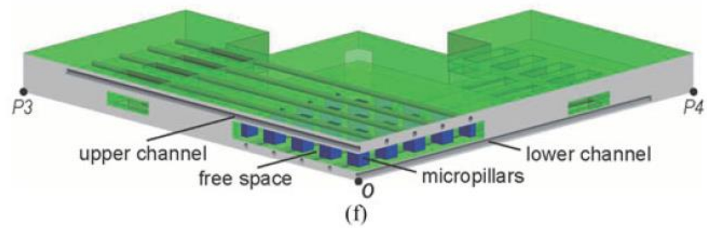

Fig 2. Configuration of micropillars and micro channels

EXPERIMENT:

The tactile sensing system used for the experiment was designed and fabricated using the method described above yielding in one embodiment 9 x 9 soft sensing array. An 8 x 8 micro pillar layer was used to provide enhanced sensitivity, linearity and operating range. The data acquisition system has an integrated custom circuit of 1:16 multiplexers with lower series resistance and analog-to-digital conversion unit to measure capacitance with femtofarad accuracy. Custom electronics, custom shielded cables and interconnections, active switched-grounding isolation method controlled by the measurement unit were used between sensing array and measurement unit because the capacitance measured is very small. Data transfer to the computer was via serial communication through Arduino. The overview of the setup is shown in Fig 3.

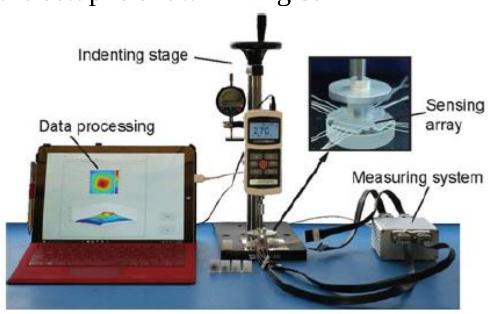

Fig 3. Over of experimental setup

In total, 24 distinct simulated tissue samples were fabricated from soft polymer (Ecoflex 00-30 M100 modulus of 7 x 10E4 Pa) of dimensions 15mm (width) x 15mm (length) x 20mm (height). Lumps were represented in four diameters(d): d = 4,6,8,10mm and were placed at different heights: h = 0, 1, 2, 3, 4, 5 mm.

The soft tactile sensor was applied to the simulated tissues under mechanical or human control using a range of displacements. A high-resolution force test stand was used to apply the vertical indentation displacement to the tissue. Three sets of experiments were designed to evaluate the performance. The first two used samples with varied depth and diameter and the force applied was strain controlled mechanically. In the third set of experiment, loading was provided by a human finger.

The captured data had a spatial resolution of 2mm and was smoothed by taking the moving average over 10 frames. The data is then interpolated via linear interpolation to form high resolution images. Enhanced images were obtained by background subtraction. The background signals were obtained by recording blank tactile images $f(x, y; z)$ for each load value z. These signals are separated from the measured signals $f(x, y; z)$ at the same load value yielding enhanced images $I(x, y; z)$.

$$I(x, y; z) = \frac{f(x, y; z)}{\max_{x,y} f(x, y; z)} - \frac{\hat{f}(x, y; z)}{\max_{x,y} \hat{f}(x, y; z)}$$

To get a quantitative estimate of lump size from the image data, the lump image is modelled via a two-dimension Gaussian function, whose parameters fit to data. The fit function is given by

$$f(x,y) = f_0 + a\exp\left(-\frac{(x-x_c)^2}{2w_1^2} - \frac{(y-y_c)^2}{2w_2^2}\right)$$

The image area A corresponding to the lump is evaluated via the portion of pixels that lie above the threshold. The diameter is then calculated as

$$A = \pi d_e^2/4 \quad \Rightarrow \quad d_e = \sqrt{4A/\pi}$$

In the final set of experiments, the force was applied by a human finger on samples with lump depth h = 0mm, lump diameter d= 10mm and indentations 0.5 to 2.0 mm.

RESULT:

Tactile images were captured for tissues of four lump diameter values (*d* = 4, 6, 8, 10 mm) as indenting displacement and depth were varied. Depth was measured from the top of the lump to the surface of the tissue (*h* = 0, 1, 2, 3, 4, 5 mm). Indenting displacement *p* is shown over a range from 0 mm to 2.5 mm in steps of 0.5 mm. For each lump size, the highlighted signal in the capacitance change map that reflects the subsurface lump becomes more prominent as the depth *h* decreases or the indentation increases. At equal values of depth and displacement, a larger lump size yields a higher capacitance value, and wider extent of the area in which this increase is observed.

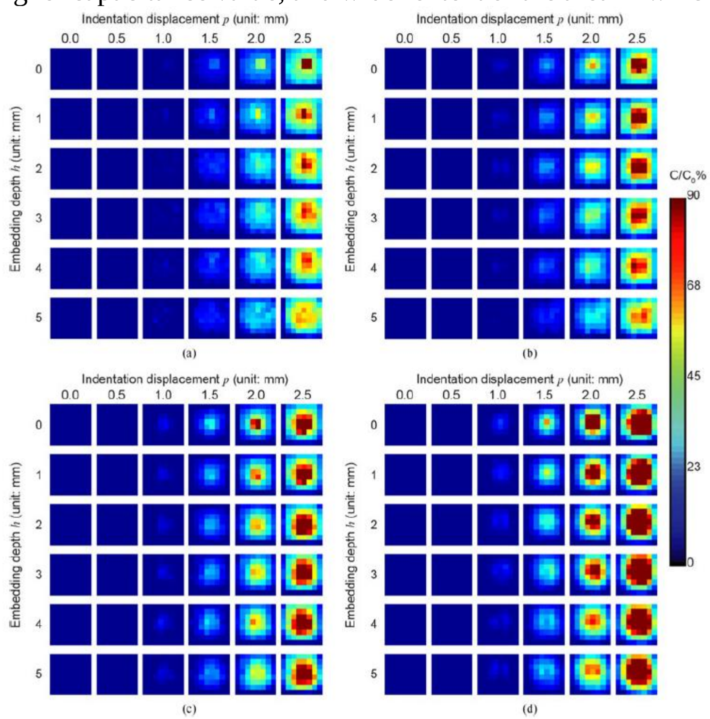

Fig 4. Sensor readings at different depths and lump diameters

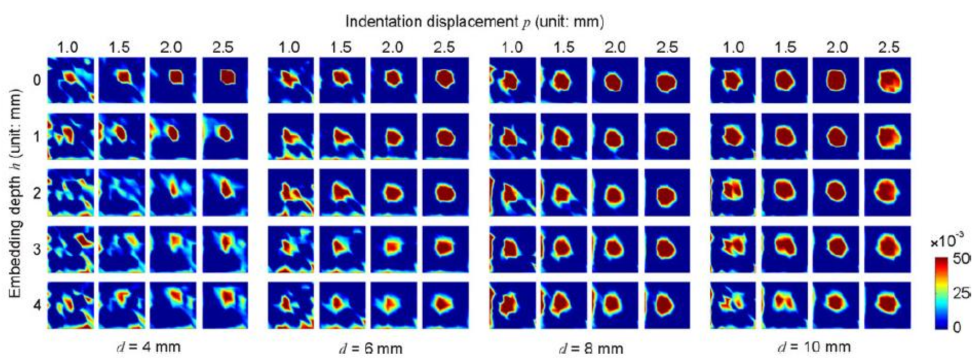

Fig 5. Enhanced image at varied lumps depths and diameters

The authors had compared the results with and without lumps. When present, the lump was prominent up to an indentation of at least 1.5 mm. For larger or shallower lumps their presence was releveled much sooner. Beyond 1.5mm, there was a broad increase in signal in the central area where strain was large, but the spatial changes were slowly varying. For lump containing tissues, there was an increase in capacitance in the lump area and the increase grew with decreased embedded depth, yielding quality images. The results indicate that the estimates predict the true lump size with an accuracy of a few percent, irrespective of the embedding depth, lump size, or indentation. The standard deviation across all conditions was 12%, indicating that this size estimate was robust to the lump parameters and palpation method.

LIMITATIONS:

The authors have mentioned several limitations that were not addressed in the paper.

- In human palpation, shear stress also plays an important sensory role. The sensor described is adapt for capturing normal stress, but shear stress aspect needs to be explored.
- The experiments conducted include spherical lumps while most lumps are irregular in geometry. Further experiments should be warranted to understand how this could affect performance of sensor.
- The electronic system described uses high signal to noise ratio and high sensitivity and translation of technology to the hospitals is not possible until the sensitivity issues have been resolved.

ASSESSMENT AND RELEVANCE:

The main advantages of the paper are:

- The hardware design aspects were detailed along with the reasoning for the choice of design. This helps other researchers in debugging electronics and provides an idea of related problems in customized design.
- The images after each software stack is included. This helps in better understanding of the processes and their outcomes.
- The samples designed had the same stiffness and characteristics as the biological tissues. This warrants that the sensor is suitable for clinical applications.

The main disadvantages of the paper are:

- The results of the experiments were not analyzed in detail. The success rate is discussed as "accuracy of few percentage" and have concluded that the method is robust for palpation applications which questions the experimental implementation and success of the tactile imaging process.
- The normalized image is included in the figures, but the experimental method does not explain the specific stage when the normalization was performed.
- Estimation of area through gaussian modelling is not an idle solution as the sensor data in practice is not capable of determining the depth of the lump.
- The cycle time and response time of sensor and the overall process time for data processing is not mentioned.

The paper discussed was chosen due to relevant to the current project "Design and Fabrication of Tactile interface for Imaging". The key aspects of the paper that was used in the project are:

- The experimental setup where an intending stage was used to image at different depths and the visual representation of the imaged data.
- The background detection and subtraction methods were adapted from the paper though not used in similar ways. The background subtraction was used in the project to eliminate the effect of sensor construction and obtain a better image representation of the images area.

CONCLUSION:

The study investigated the utility of soft sensors for imaging lumps in simulated tissues. A method for imaging including the integrated electronic sensing system to achieve high levels of sensitivity was discussed in the paper. Further, the authors discussed an algorithm for the detection of lumps on the subsurface. Although the study has limitation discussed above, it successfully demonstrated the feasibility of tactile imaging of tissues.

REFERENCE:

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