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Novel MEMS Stiffness Sensor for In-Vivo Tissue Characterization Measurement

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***Abstract*— This paper presents the design, mathematical model, fabrication and testing of a novel type of in-vivo stiffness sensor. The proposed sensor can measure both tissue stiffness and contact force. The sensing concept utilizes multiple membranes with varying stiffness and is particularly designed for integration with minimally invasive surgical (MIS) tools. In order to validate the new sensing concept, MEMS capacitive sensors are fabricated using surface micromachining with each fabricated sensor having a 1mm x 1mm active sensor area. Finally, the sensors are tested by touching polymers of different elastic stiffnesses. The results are promising and confirm the capability of the sensor for measuring both force and tissue compliance.**

I. INTRODUCTION

MINIMALLY Invasive Surgery (MIS) offers many advantages over open surgery including fewer complications, less discomfort after the operation, faster recovery times and lower health care costs. However, one significant existing difficulty of MIS is the loss of the intra-operative tactile feedback [1] which is readily obtained in open surgery by touching the tissue. In order to recover tactile sensing during MIS, measurements of contact force and stiffness are vital. The knowledge of tissue stiffness is also a valuable tool in providing real-time feedback on the type of tissue to the surgeon during MIS. In addition to MIS, there are many other biomedical applications such as ligament tension measurement during knee implant surgery, early detection of compartment syndrome, and cartilage hardness measurement where stiffness measurement could be extremely valuable.

Indentation type tissue stiffness tests have been performed on cancerous breast tissue by estimating elastic moduli from measured force displacement curves [2]. In telerobotics research, a laparoscopic grasper attached to a robot arm [3] has been designed to provide force and vision feedback. The stiffness of the testing object was tested by measuring the applied force and the angular displacement of the jaw. Another work on tissue based measurement employed piezoelectric cantilevers to investigate the force-deformation response of soft tissue [4]. As the active sensing alternative, an air-driven oscillating indenter was developed to detect bulk tissue compliance such as soft tissue compliance measurements in stumps of amputated lower limbs [5]. The pressure was controlled by a flexible rubber hose and the displacement was recorded by an electromagnetic sensor.

The existing technologies necessitate simultaneous measurement of both tissue deformation and applied force to estimate tissue stiffness. However, these methods are not suitable for direct in-vivo measurements and present difficulties for miniaturization. In another work, Dargahi et al. [6] proposed a softness sensor which consists of two coaxial cylinders with different stiffnesses. This method differs from the other approaches since no displacement information of the end-effector needs to be obtained during the sensing movement. However, the sensor needs the application of a dynamic load driven by a vibrating unit and utilizes rubber cylinders and PVDF films which render it extremely challenging to be miniaturized and fabricated by micromachining processes. In addition, since changing the stiffness of the rubber cylinder is cumbersome, the dynamic range (the range of Young's moduli the sensor is capable of measuring) will be limited for certain applications.

This paper presents a new sensor that employs an array of MEMS capacitive sensing membranes with different stiffnesses. This novel approach addresses the challenges regarding miniaturization for in in-vivo measurements and compatibility with existing micromachining processes. The method utilizes capacitive measurements from multiple nodes with different stiffnesses on the sensor and estimates tissue stiffness by manipulating the relative capacitive change data from these nodes. Further, the microfabrication of the sensors has been accomplished through a five-mask surface micromachining process. Finally, the MEMS sensors have been used to experimentally verify the developed sensing approach.

II. SENSING CONCEPT AND SENSOR DESIGN

Our sensor is designed to measure the stiffness k_T of the tissue under contact. In order to measure tissue stiffness, it is proposed to have two or more sensing elements with different spring constants (k_h and k_s for a hard and soft spring, respectively). The proposed mathematical model for tissue-sensor interaction analysis is shown in Fig. 1.

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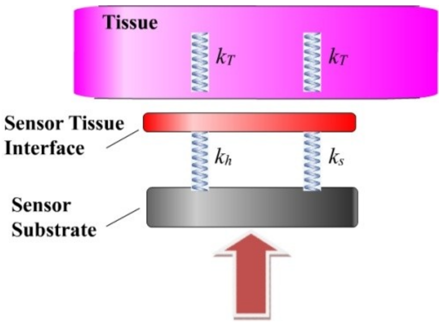
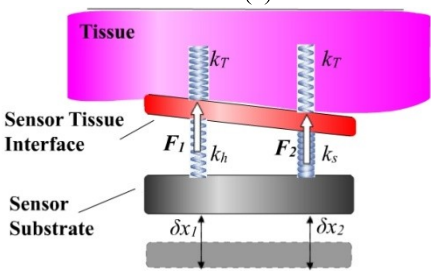


(a)



(b)

Fig. 1. Contact model for the proposed stiffness sensor (a) before contact (b) after contact

In its simplest embodiment, the model assumes a rigid substrate on which the sensing elements are attached. The sensing elements (k_h, k_s) and the tissue (k_T) are arranged in series. As the sensor assembly pushes towards the tissue, the sensor base remains parallel resulting in the same total displacement at the bottom of both sensing elements. This is given in (1).

$$\delta x_1 = \delta x_2$$
$$\Rightarrow \frac{F_1(k_h + k_t)}{k_h k_t} = \frac{F_2(k_s + k_t)}{k_s k_t} \tag{1}$$

which results in (2)

$$k_t = \frac{r_x k_h - k_s}{1 - r_x} \tag{2}$$

where r_x represents the relative displacement of two sensing elements $\delta x_h/\delta x_s$.

This result shows that the tissue stiffness k_T can be calculated with the knowledge of r_x. This eliminates the need for an additional measurement of the displacement of the tissue in contact to measure tissue stiffness. In other words, this method allows the operator to readout tissue stiffness value and contact force value simultaneously by simply pushing against the tissue under investigation.

The deformation of the sensing elements in the proposed tactile sensors will be measured by a membrane based capacitive sensor as shown in Fig. 2. In this simple configuration, a sensing membrane (diaphragm) is suspended over the sensor substrate. There are two electrodes with one on top of the membrane and the other on the substrate. The deflection of the membrane causes the capacitance of the sensor to change which can be measured and be converted back into a deflection readout. The stiffness of this membrane will be defined by the geometry (i.e. diameter) of the membrane such that several membranes with identical membrane thicknesses and gap heights will have different compliance under the same load.

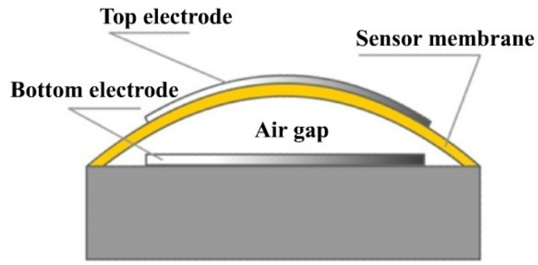


Fig.2. The conceptual diagram of the capacitive sensing membrane

III. Fabrication of The Prototype Sensor

The prototype MEMS sensors have been fabricated using a surface micromachining process. The fabrication process is shown schematically in Fig. 3 [7]. The sensing membrane is made of PECVD silicon nitride whereas the top and bottom electrodes are made of gold.

The fabrication process starts with a silicon wafer which is first covered with 6000A˚ of silicon nitride (SiN_x) layer for passivation. Plasma enhanced chemical vapor deposition (PECVD) is used to obtain the SiN_x passivation layer (Fig 3.a). A 2600A˚ Cr-Au metal layer is then electron-beam evaporated to form the bottom electrodes (Fig 3.b). An 8000-18000A˚ sacrificial aluminum layer is electron-beam deposited and patterned via wet etching (Fig 3.c). The sensing membrane is formed by depositing a 5000-10000A˚ PECVD SiN_x layer (Fig 3.d). Top electrodes are patterned by wet etching a 400A˚ Cr – 2000A˚ Au electron beam evaporated metallization layer (Fig 3.e). The etch holes are then patterned via dry plasma etch (Fig 3.f). The sacrificial Al layer is then etched in wet etching solution though the etch holes and the membranes are released in a critical point dryer (Fig 3.g).

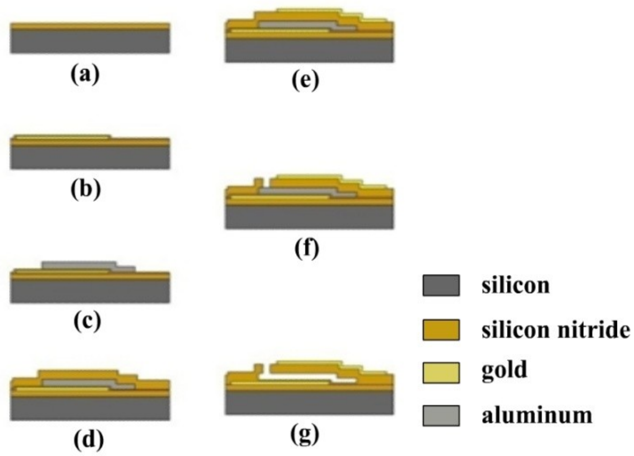


Fig. 3. The fabrication process for MEMS sensors

Fabricated sensors are shown in Fig. 4. The same-sized membrane capacitances form a parallel capacitor array and are added together to form the capacitance readout that will give us the deflection for that membrane size. Thus more numbers of smaller membranes can be utilized to make the capacitive readout comparable (equally sensitive) to those from larger membranes. A custom designed PCB equipped with three capacitance measurement chips (Model MS3110P, Irvine Technologies, CA) is utilized for capacitive readout. A single IC is dedicated for measuring the capacitance and consequently the deflection of a single array of capacitive membranes with identical diameter.

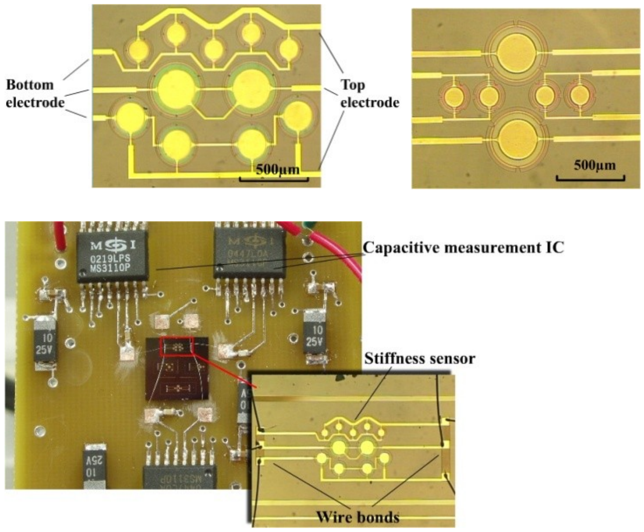


Fig.4. Fabricated MEMS stiffness sensors and readout circuitry

IV. Results and Discussion

Preliminary tests have been conducted by using the Micro-Mechanical Tester (MMT) located in the Characterization Facility of the University of Minnesota (Fig. 5). This equipment is employed with a modification wherein the diamond indenters are replaced with probes made of polymers with different stiffnesses. The probe is carefully driven towards the stiffness sensor by a stepper motor. The load values and capacitive readouts are recorded once the polymer probe touches the sensor.

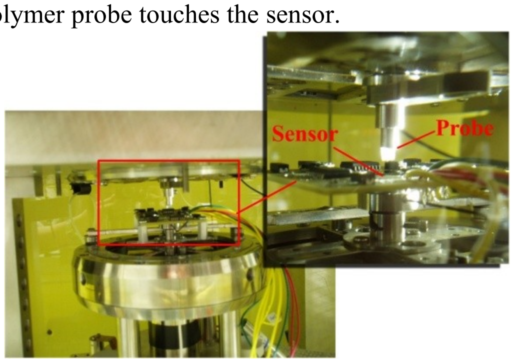


Fig. 5. MicroMechanical Tester and testing with probes attached with the polymer head

A. Force Sensing

The applied force and the corresponding capacitive readouts are given in Fig. 6 for the sensing arrays consisting of membranes with the same diameter of 300μm. The capacitive readout climbs with increasing load and holds the value when no additional loads are applied. For a load range from 0.35N to 0.55N, the change in voltage readings is around 1V which corresponds to a capacitance change of 1pF. The force sensing resolution is 0.2mN or 100Pa under uniform loading conditions on the membrane. It is also noticed that the capacitive response starts at an initial load value of 0.35N. This is due to fact that the contact area of the probe utilized in this experiment (Fig. 5) is larger than the sensing area defined by the fabricated capacitive membranes, and the probe surface cannot provide a perfect normal contact with sensor membranes. The contact at the probe-sensor interface therefore may not initiate at the sensing area.

The experimental results for 400μm, 300μm and 200μm membranes-arrays show deflections of the membranes to be in the elastic regime with negligible hysteresis. The experimentally determined compliances of these membranes (interpreted in units of pF/N) are listed in Table I. The compliance variation of the membranes agrees with the analytical membrane model.

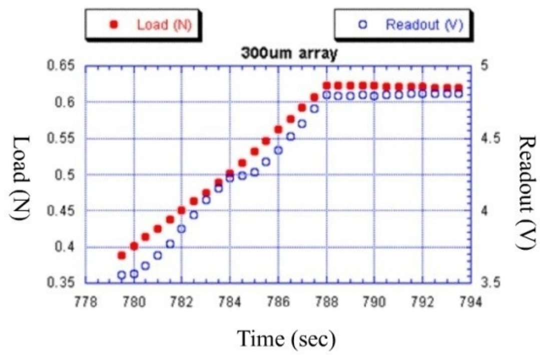


Fig. 6. Loads and capacitive readout (volts) for 300μm-membranes

TABLE I
Compliances of Sensing Arrays and Single Membrane

	400μm	300μm	200μm
Compliance of sensing array (pF/N)	15.34	5.42	2.36
Compliance of single membrane (pF/N)	7.67	1.36	0.47

B. Stiffness Sensing

As discussed in section II, the stiffness of the contacted tissue can be estimated by the relative deflection of two sensing elements (r_x). This value is approximately proportional to the ratio of capacitive change of a sensing pair (r_c). This result is shown in Fig. 7 where simultaneous readouts from two channels are taken and the ratio r_c is calculated. It is worth noting that our initial attempt at simultaneous multiple channel readout was not successful due to electrical cross-coupling between the membrane arrays. We then employed an alternative method in which a series of square waves generated by a DAQ board (National Instruments, SCB-68) alternately activated the two capacitive readout chips corresponding to the two membranes with different diameters at a frequency of 100Hz.

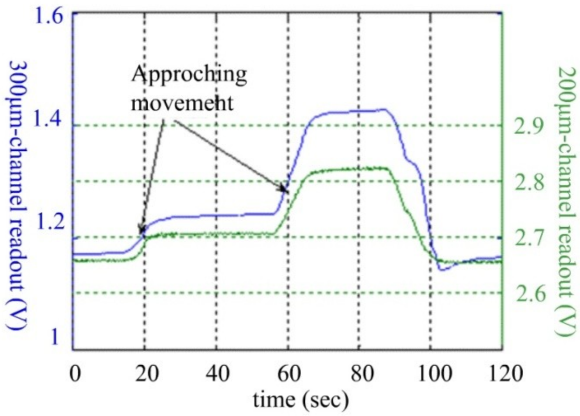


Fig. 7. Capacitive readouts of 300μm and 200μm–membrane during the contact between the polymer probe and the stiffness sensor

Experiments were conducted with probe materials of different elasticity and thickness in order to evaluate the ability of the sensors to estimate their elasticity. In Fig. 8, different polymer materials attached on the probes are shown.

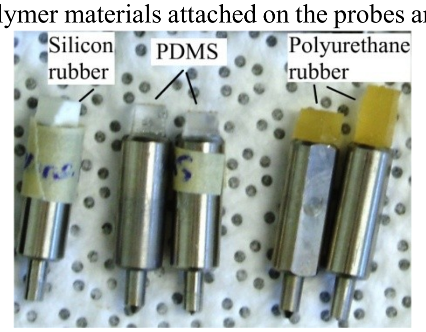


Fig. 8. Probes of different polymer materials and different thickness values

The ratios of changes in capacitances between the 300μm (diameter) and 200μm membranes are shown in Fig. 9 for different probe materials. The test on each probe is conducted for three times. The results show a clear variation in the ratio of capacitance changes with elasticity of the probe material. The values of Young's moduli of the testing polymers are obtained from an IRHS tester (Model IRHD micro compact II, Bareiss®, Germany).

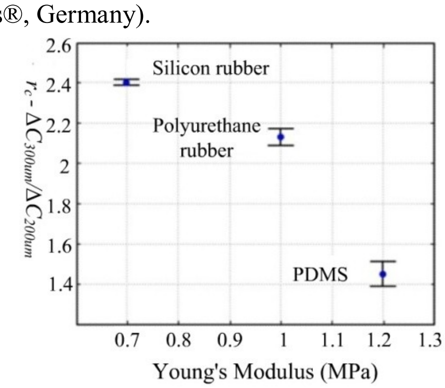


Fig. 9. Estimated r_c values from sensor versus Young's Moduli of different specimens (MPa)

It can be seen that the ratio of capacitance changes is a unique function of the Young's modulus of the material used in the experiments. Further, the ratio of capacitance changes is independent of the thickness of the material used in the sample. This result can be attributed to the fact that the sizes of the sensing membranes are much smaller than the polymer specimens. The elastic moduli of the contact materials can be therefore measured regardless of the thickness of the specimens. To verify this contact scheme, the contact model shown in Fig. 1 can be modified by modeling tissue as an elastic semi-space object. Another important note is that under small deflections such as those obtained during our experiments, the capacitive change of the sensing membranes is proportional to the membranes deflection and therefore is proportional to the stresses applied on the contact area. This observation ensures the feasibility of our estimation by using the relative capacitive change (r_c) instead of the relative deflection of sensing elements (r_x).

V. Conclusions

In this paper, we have proposed a novel method of stiffness measurement by employing sensing elements with different stiffnesses. This method can be good candidate for in-vivo characterization of tissue properties in Minimally Invasive Surgery. A prototype MEMS stiffness sensor has been successfully fabricated and evaluated. Preliminary testing on polymers with different compliances has been completed to validate the sensing method. The sensing resolution can reach 0.2mN for force sensing and at least 0.2MPa for stiffness sensing. The experimental results demonstrate the sensor capability of being both a stiffness sensor and a force sensor.

Acknowledgment

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