

# *Design and Modeling of a New Type of Tactile Sensor Based on the Deformation of an Elastic Membrane*

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## **ABSTRACT**

This paper presents the design and modeling of a flexible tactile sensor, capable of detecting the 2D surface texture image, contact-force estimation and stiffness of the sensed object. The sensor is made of polymer materials. It consists of a cylindrical chamber for pneumatic actuation and a membrane with a mesa structure. The inner radius of the cylindrical chamber is 2cm and its outer radius is 3cm. The sensing mechanism of the sensor is based on the contact deformation of the membrane. Determination of the contact-force and stiffness of sensed object is based on the amount and variations of out of plane deflections at the center of a circular membrane. The amount of deflection depends on the force or pressure applied. Furthermore, the size and shape can be easily tailored to the applications' requirements. This versatility facilitates the use of the sensor in smart applications where tactile information is used to create system intelligence. The proposed sensor with the potential for further miniaturization is suitable for using in medical applications, especially in minimally invasive surgery (MIS).

## **KEYWORDS**

Tactile Sensor, Stiffness, Contact Force, Membrane.

## **1. INTRODUCTION**

Tactile sensing is an area of MEMS research that has the potential to impact a large number of industries and disciplines. The key feature among them is application to robotics in medicine and industrial automation [1]. In different biomedical engineering and medical robotics applications, tactile sensors can be used to sense a wide range of stimuli. This includes detecting the presence or absence of a grasped tissue/object or even mapping a complete tactile image [2]-[4].

Robust, reliable tactile feedback of forces and torques, contact shape and location, and dynamic slip sensing are required for dexterous, dynamic gripping and manipulation by robots and by humans through haptic interfaces [1]. Force and position signatures are the two factors that can provide a great deal of information about the state of gripping or manipulation of a biological tissue [5]. Lack of suitable commercial tactile sensors will limit development in robotic handling of soft, fragile or irregular objects. Several types of tactile sensors have already been proposed for handling objects in robotics and automation systems. They can handle soft and fragile materials with some difficulty [6].

Artificial palpation is an important application of tactile sensors. Normally, in order to improve the efficiency of these types of sensors, an array of sensors is utilized [7], [8]. Tactile and visual sensing is of great importance in different types of surgeries [9]. Minimally invasive surgery (MIS) is now being widely used as one of the most preferred choices for various types of operations [10]-[12]. In MIS, any inhibitions on the surgeon's sensory abilities might lead to undesirable results [13]. MIS has many advantages, including reducing trauma, alleviating pain, requiring smaller incisions, faster recovery time, and reducing postoperation complications [14], [15]. However, MIS decreases the tactile sensory perception of the surgeon. This effect is more pronounced during grasping or manipulation of biological tissues (i.e., veins, arteries, bones, etc.). In this regard, measuring the magnitude of the forces applied by the surgeon through the endoscopic graspers results in safer handling of biological tissues [16]. Controlled manipulation tasks are among the manoeuvres in which the ability to feel the tissues are very crucial [17]. The need to detect various tactile properties (such as stiffness, temperature, and surface texture) justifies the key role of tactile sensing which is currently missing in MIS [18], [19]. There are operation

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sites in human body that are otherwise difficult to see in order to examine or operate on. As a result, only relying on the visual tools is not sufficient to obtain satisfactory results [20]-[22].

We proposed a new type of tactile sensor that can detect both the contact force and the stiffness of an object. Our proposed sensor offers the following combination of characteristics:

- 1-Mechanical flexibility and robustness.
- 2-Relatively low processing temperature (<350° C).
- 3-Low fabrication complexity.
- 4-Improved strain transfer from membrane to strain gauges or other measurement devices.
- 5-Decrease in fabrication cost because of its simplicity.

## 2. PREVIOUS WORK

In this section, the most recent advances in tactile sensors in different types of medical industries will be presented. Tactile sensors are divided into different categories: mechanical (binary touch mechanism), capacitive, magnetic, optical, piezoelectric (PZT), piezoresistive (strain gauges), and silicon-based (micro electromechanical). During the past decades, various types of tactile sensors have been tested in various robotic-related applications [6, 7].

One of the important parameters in determining the physical properties of living tissues is stiffness. Considerable biomedical attention has centered on the mechanical properties of living tissues at the single cell level [6]. The Young's modulus of zona pellucida of bovine ovum was calculated using both micro-tactile sensor fabricated and PZT material. The sensor consists of a needle shaped 20mm transduction point made using a microelectrode puller mounted on a micro-manipulator platform [23]. The stiffness of the cartilage of the human femoral condyles was measured via an ultrasonic tactile sensor under arthroscopic control. The tactile sensor was useful for determining the intraoperative stiffness of healthy and diseased human cartilage in all grades [24]. The development of milli-robotic tools for MIS is reported. It describes the limitations of current surgical practice and the technological and scientific issues involved in building a telesurgical workstation [25].

A tactile sensor system has been developed for accurate measurement of myocardial stiffness in situ [26]. Piezoresistive sensors applied to the fingertips of nonsensate fingers were used for the detection of touch and pressure in four patients with recent median nerve repairs, and in one patient using a myoelectric prosthesis [27]. The design, fabrication, testing, and mathematical modeling of a semiconductor micro strain gauge endoscopic tactile sensor have been investigated [13], [28].

A novel membrane-type PVDF tactile sensor has been reported, which is free of crosstalk and does not require an array of sensors. Three rectangular-shaped aluminum electrodes formed a triangle in the central region of the PVDF film. Unlike the array type tactile sensors, all the surface points of this membrane-type tactile sensor are active. Using a geometrical mapping process and by applying force at various points on the sensor surface, the loci of the isocharge contours for the three sensing elements are obtained [15].

The shortcomings of most of the current designs are mainly related to the complexity of the systems and different expensive accessories used. The major advantage of the system proposed in this paper is the simplicity and robustness of the design.

## 3. SENSOR PRINCIPLE AND DESIGN

### A. Device Specifications

The device has a cylindrical shape, to simplify the problem and reduce the amount of calculation. The radius of membrane is 2 cm and it is attached on a rigid cylinder which has a port for gas supply and exhaust.

A strain gauge will be embedded in the membrane exactly in peripheral and radial direction of membrane. The strain gauge is one of the most important tools of the electrical measurement techniques applied to the measurement of mechanical quantities. It has small physical size and low mass, and is highly sensitive. As the force is applied to the center of membrane, the embedded strain gauge measures the strain in the membrane due to membrane deflections caused by the applied force. As a result of membrane deflection and so on applying stress on strain gauge, resistive changes take place and result in a signal output, related to the stress value because of membrane deflections. So, the output signal of the strain gauge can be read as a force signal. Figure 1 shows the shape of sensor device and Table 1 shows typical specifications of the designed tactile sensor.

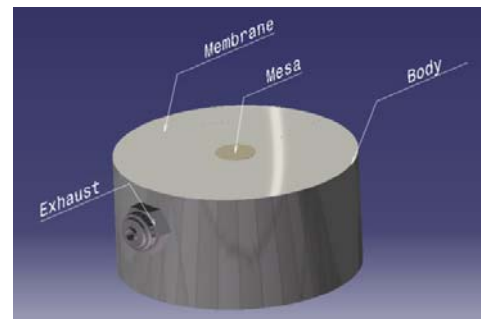


Figure 1: A general model of the sensing element.

TABLE I  
SPECIFICATIONS OF THE MODELED SENSOR

Device (cylindrical)	2 cm (inner radius) 3 cm (outer radius) 5 cm (height)
Membrane	2 cm (radius) 100 $\mu\text{m}$ (thickness)
Mesa	0.5 cm (radius) 150 $\mu\text{m}$ (thickness)
$\nu$ (Poisson's ratio)	0.33
$E$ (Elastic modulus)	30 MPa

### B. Analytical Analysis

We assumed that the membrane and contacted object are elastic materials. The relationship between the deflection  $w$  (displacement at the center of the membrane or mesa) and the applied force  $F$  is given by:

$$F = (k_m + k_o) w \quad (1)$$

where  $k_m$  and  $k_o$  are the elastic constants of the membrane and object, respectively [29].

To obtain  $k_o$ , we used a contact model in which the surface profile of the object changes according to the pushing depth of the mesa toward the object. The theoretical model of a single-layer circular membrane is shown in Figure 2.

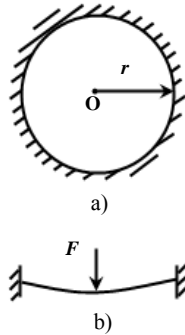


Figure 2: Theoretical model. a) Front view, b) Top view.

For single-layer circular membranes under the concentrated force and by considering this related large deformation theory, the solution for out-of-plane deflections (OPD) at the center of the membrane can be expressed as [30]:

$$\left(\frac{d}{h}\right)^3 = \left[1 - \left(\frac{1-3\nu}{4}\right)^{1/3}\right]^3 \frac{4r^2}{(1+\nu)\pi E h^4} F \quad (2)$$

$$\text{if } \nu=1/3, \quad (3)$$

$$\left(\frac{d}{h}\right)^3 = \frac{3r^2}{\pi E h^4} F$$

where  $d$  is out-of-plane deflection at the center of the membrane,  $r$  is radius of membrane,  $h$  is thickness of membrane,  $\nu$  is Poisson's ratio,  $E$  is elastic modulus, and  $F$  is the applied force at central point.

### C. Numerical Analysis

The second series of tests were performed to simulate the mechanical responses of sensor numerically. Figure 3 shows the finite element modeling of the sensor. A commercial finite element analysis software package (ANSYS, version 11.0) was employed. Membrane of the sensor is considered to have elastic and large deformation behavior.

In this numerical analysis, element type shell 181 is used to model the elastic and large deformation behavior of the membrane of the designed tactile sensor and element type solid 92 is used to model the behavior of rigid cylindrical body of the designed tactile sensor.

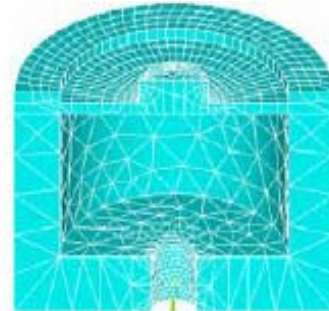


Figure 3: Finite element modeling of the sensor.

## 4. MODES OF OPERATION

### A. 2D Surface Texture Image Detection

Figure 4 shows an array of two elements of our tactile sensor. Each sensor consists of a membrane with a mesa at the center and a chamber for pneumatic actuation. The tactile sensor can encode and decode the shape of objects, (see Figure 4). When the sensor array comes in contact with an object that has a bumpy surface, the mesa structure on the membrane push inwards and as a result of it, the system can detect the presence or absence of the object. So, we can save a 2D surface texture image of the object.

### B. Contact-Force Estimation

The structure of the estimating contact-force is shown in Figure 4. When the mesa of membrane comes in contact with an object, the normal force or uniform pressure from the object causes inward deformation in the membrane. Therefore, by determining the displacement at the center of the membrane and according to its mechanical properties, we can measure the amount of normal force or the uniform pressure actuating on it.

### C. Stiffness Detection

In this mode, the contacting mesa elements are pneumatically driven against the object (Figure 5). Gas is used for applying uniform pressure on the membrane. The contact regions of the object are deformed according to the applied force of the mesa and the stiffness of the object. Therefore, we can detect the stiffness of the object

by measuring the relationship between the deformation of the membrane and its actuation force.

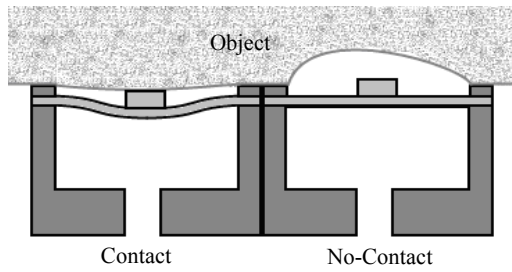


Figure 4: An array of two elements of the tactile sensor (Detecting contact-force distribution and 2D surface texture image).

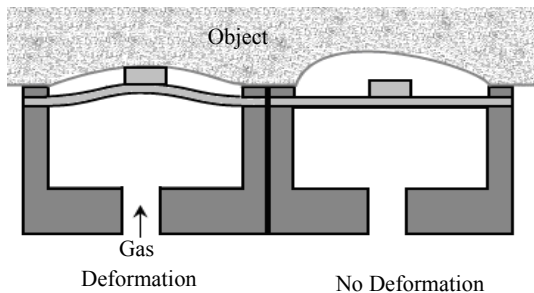


Figure 5: An array of two elements of the tactile sensor (Detecting stiffness distribution)

## 5. RESULTS AND DISCUSSION

The analytical and numerical results have been in good agreement. We investigated the deformation at the center of mesa changes with variations of thickness of the membrane and applied force. Figure 6 shows the variations of OPD (out-of-plane deflections) of membrane with the constant radius and unique force at different thicknesses.

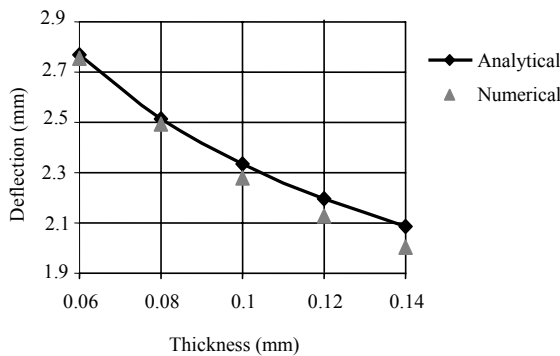


Figure 6: Variations of out-of-plane deflections vs. different thicknesses.

As a result of applied force at the center of membrane, we have an out-of-plane deflection on it. Figure 7 demonstrates the deformation or out-of-plane deflections of membrane according to the variations of the applied force.

Figure 8 shows a typical sample of numerical analysis. It shows that the maximum amount of deflection occurs at the center of the membrane.

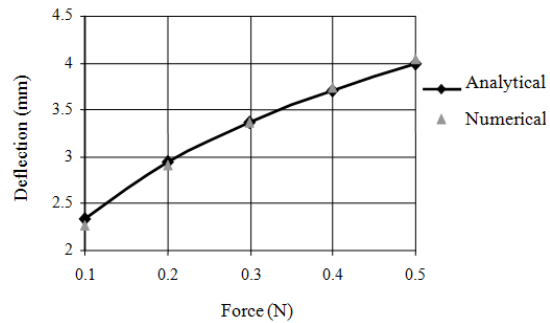


Figure 7: Variations of out-of-plane deflections vs. different applied forces.

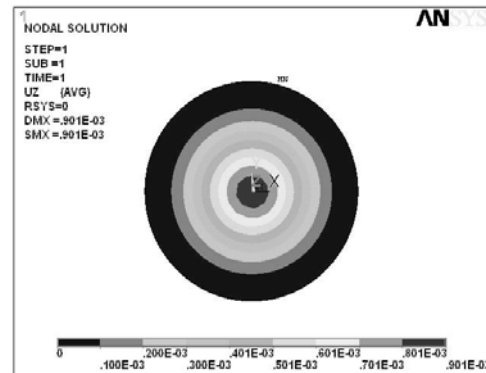


Figure 8: Deflection of the membrane due to the applied force.

The contact regions of the object are deformed according to the driving force of the mesa element and the stiffness of the object (see Figure 9). Therefore, we can detect the stiffness distribution of the object by measuring the relationship between the deformation of the membrane and the actuation force of the elements.

According to Figure 9, the amount of deflection of the membrane in contact with the stiffer objects at unique driving force is less than that of the others. So, with changing the stiffness of object in contact with membrane, the OPD of membrane changes and hence, we can estimate the variations of object's stiffness.

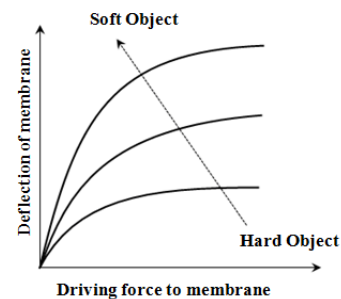


Figure 9: Relationship between the deflection of the membrane and driving forces.

We investigated how the deformation of the membrane changes with  $k_o$  and we obtained the following results:

a)  $k_o \ll k_m$ : In this region, the elastic constant of the object is much smaller than that of the membrane. Therefore, the membrane always deforms by the same amount under a constant force even if the stiffness of the object changes. As a result, in this region, the sensor cannot detect the changes in stiffness of the object.

b)  $k_o \gg k_m$ : The elastic constant of the object is too large compared to the membrane. Therefore, the deflection of the membrane is very small and with increasing the stiffness of the object, the amount of deflection of the membrane declines to zero. In this region, the sensor cannot detect the changes in stiffness of the object.

c)  $k_o \approx k_m$ : The elastic constant of the object is in the same range as that of the membrane. Therefore, the deformation of the membrane changes with the stiffness of the touched object. In this region, the sensor can detect the changes in stiffness of the object.

We conclude that in order to detect a change in the stiffness of the touched object, the elastic constant of the membrane should be almost the same as that of the touched object. Figure 10 shows how the deformation of the object changes with elastic constant of object ( $k_o$ ).

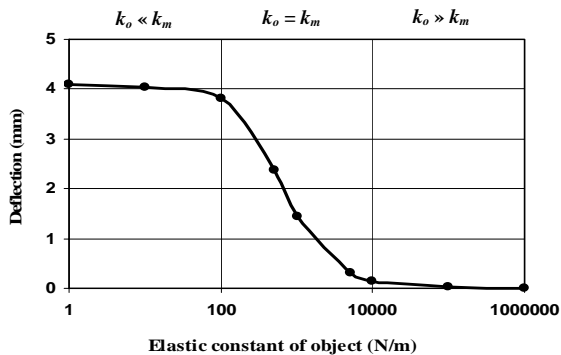


Figure 10: Membrane deflection vs. elastic constant of object.

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When the gas pressure is applied to the membrane of sensor in contact with a tissue, the membrane deflection and hence output voltage of the electronic circuit of the sensor will be changed. The amount of deflection of the membrane in contact with the stiffer objects at unique pressure is less than that of the other. Hence, with moving the sensor on an identified area of a tissue at a constant pressure, the changing of the output voltage of sensor describes the changing of surface stiffness of that area.

## 6. CONCLUSION

The demonstrated polymer-based tactile sensor is a major step towards realizing sensors that can provide robots with direct tactile feedback similar to the biological sense of touch. We proposed a new type of tactile sensor that can detect both the contact force and stiffness of an object. We analyzed theoretically and numerically the operation of the tactile sensor. This device can be made from robust, flexible polymers used to directly touch objects, and serve as a skin on robotic actuators. Providing such information to robotics opens new areas to development and exploration. A major advantage of the designed system is that it can be easily miniaturized and micromachined. As a result, it could be mass-produced at low cost and even be disposable. The designed sensor has two main applications, one in MIS and one in artificial palpation.

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