Design, Modeling, and Construction of a New Tactile Sensor for Measuring Contact-Force

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ABSTRACT

This paper presents the design, modeling, and testing of a flexible tactile sensor and its applications. This sensor is made of polymer materials and can detect the 2D surface texture image and contact-force estimation. The sensing mechanism is based on the novel contact deflection effect of a membrane. We measure the deflection of the membrane with measuring the strain in the membrane with embedded strain gauge. It consists of a chamber and a membrane. Inner radius of the sensor element is 2 cm and its outer radius is 3cm. Furthermore, the size and shape can be easily tailored to the applications requirements. The proposed sensor with the potential for further miniaturization is suitable for using in medical applications, especially in minimally invasive surgery.

KEYWORDS

Tactile Sensor, Contact Force, Deflection, Membrane

1. INTRODUCTION

By touching an object, it is possible to measure contact properties such as contact force, torques, and contact position. From these, we can estimate object properties such as geometry, stiffness, and surface conditions. This information can then be used to control grasping or manipulation. So, tactile sense is a key to the advance robotic grasping and manipulation. In this regard, there are two different kinds of grasps; power grasps and precision grasps.

Power grasps are typically used for larger and rigid objects and in tasks that do not require more than simple manipulation of the object. Grasping a book to lift it, would be an example. More delicate and soft objects are typically held in a precision grasp. When lifting a fragile grasp or holding a soft tissue and in other precision tasks, primarily the fingertips are used for contact. The precision grasp has advantages such as enabling better control of contact force and motion of object, but it is also less stable than the power grasp [1]-[3].

In 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 [4]-[6]. Artificial palpation is another important application of tactile sensors. Additionally, tactile and visual sensing is of great importance in different types of surgeries. It has been demonstrated that automation technology can assist surgeons in minimal access treatment by enabling the benefits of steady tool motion through difficult access [7].

Minimal invasive surgery (MIS) is now being widely used as one of the most preferred choices for various types of operations [8]-[10]. MIS has many advantages, including reducing trauma, alleviating pain, requiring smaller incisions, faster recovery time, and reducing post-operation complications [11], [12]. Applications to the prostatectomy procedure and in the resection of tumors in neurosurgery [13] are excellent examples. Other examples of advanced mechatronic surgical tools to enhance skill levels are in the alignments of bone tissues in craniofacial surgery [14], the control of cutting tools in knee surgery [15], the precise control of tool motion in soft tissues in the ear [16], and needle penetration [17]. However, MIS decreases the tactile sensory perception of the surgeon. In this situation, 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 [18].

In order to take advantage of tactile sensing in

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applications that require imaging of force between two area surfaces, a versatile sensor technology is required. This technology should consider the spatial resolution, force sensing range, and sensing area size. A telesurgical system returns force sensation to the user corresponding with the reaction force of tissues to the tool action imposed. Fig.1 shows a function diagram of a master-slave system.

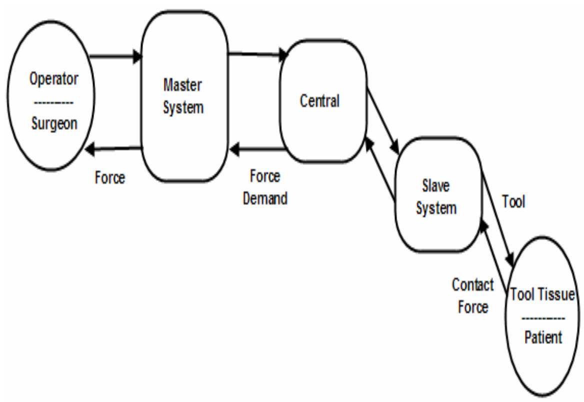

Figure 1: Function diagram of a master-slave system.

The user inputs a tool action. This is sensed at the master system and is transmitted as a control demand to the tool operated by the slave system. The tool responds automatically by following the user input demand signal. The reaction force of the tissue is sensed at the tool and is transmitted to the master system where actuators are applied to resist the motion of the input device. The control of tool motion in soft tissues requires careful monitoring of forces and responses. Such a system can be applied to scissors, forceps, needles, probes, and other tool actions.

This paper, for the first time, focuses on a novel technique for measuring tactile forces as an important component in force reflection systems. The sensor can measure, with reasonable accuracy, the magnitude and the position of an applied load on it. This sensor has good potential for use in surgical devices.

2. TACTILE FORCE SENSING TECHNIQUE

The term tactile sensing covers a range of applications from object recognition by touch to measurement of the magnitude of the force over the area in contact with an object. For measuring a distribution of contact forces, resistive transducers have been investigated [19]. These are positioned such that there is significant deformation when subjected to a change in applied force resulting in a change in conductivity. Capacitive transducers have been applied using a similar approach [20]. Piezoelectric sensors are an alternative and are most suited to measuring fast transients and not steady levels [21]. Magnetic, mechanical and optical transduction methods have also been used [22], [23].

A suitable sensor for application in telesurgery and robot surgery should determine the force applied, and detect the force distribution and movement of surfaces by deformation. It should use operating media that do not present a risk to patients and it should be manufactured from biocompatible materials.

3. MATERIALS AND METHODS

Our proposed sensor is made of a thin polymer membrane that can detect the 2D surface texture image and contact-force estimation.

A. *Detection Principle*

The basis of detection is measuring the amount of membrane deflection due to force.

1) *2D Surface Texture Image Detection:*

This sensor can be used to sense a diverse range of stimuli ranging from detecting the presence or absence of a grasped object to a complete tactile image. Fig. 2 shows an array of two elements of our tactile sensor. Each sensor consists of a membrane with a mesa at the center. A salient feature of our tactile sensing is its ability to encode and decode the shape of objects.

When the tactile sensor array comes in contact with an object that has a bumpy surface, some of the mesa structures on the membrane push inwards and as a result of it. Hence, the system can detect the presence or absence of object above each of the elements. Therefore, we can have a 2D surface texture image of the object.

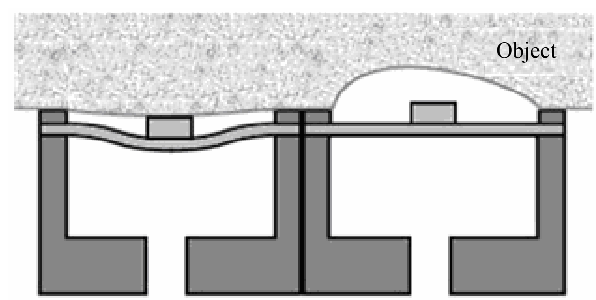

Figure 2: An array of two elements for detecting contact force distribution and surface texture image.

2) *Contact-Force Estimation:*

The structure of the estimating contact-force is illustrated in Fig. 2. When the mesa of membrane comes in contact with an object, the normal force or uniform pressure from it causes inward deformation of the membrane. Therefore, by determining the displacement at the center of the membrane and according to the mechanical properties of it, we can measure the amount of normal force or the uniform pressure actuating on it.

B. Sensor Design

The device has a cylindrical shape, and as a result, it simplifies the problem and reduces 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. The thickness of membrane is 100 µm and the radius of mesa is 0.5 cm with a thickness about 150 µm. Two series of theoretical and numerical analysis have been performed and the results of them are compared with the experimental results.

1) *Theoretical Analysis:*

The problem of axisymmetric large deformation of circular membrane is one with practical significance. For single-layer circular membranes under the concentrated force and by considering the large deformation theory of them, the solution for out-of-plane deflections (OPD) can be expressed as [24]:

$$(\frac{w_0}{h})^3 = [1-(\frac{1-3\nu}{4})^{1/3}]^3 \frac{4R^2}{(1+\nu)\pi E h^4} F \quad (2)$$

when $\nu = 1/3$,

$$(\frac{w_0}{h})^3 = \frac{3R^2}{\pi E h^4} F \quad (3)$$

In the above formulas w_0 is OPD (out-of-plane deflection) of membrane, R is radius of membrane, h is thickness of membrane, ν is Poisson's ratio, E is elastic modulus, and F is applied force at the central point.

The theoretical model of a single-layer circular membrane is shown in Fig. 3.

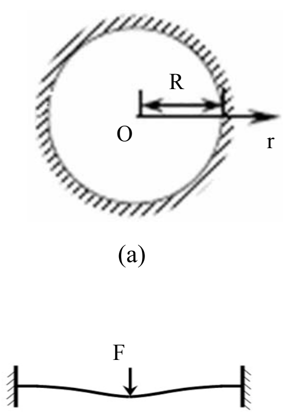

Figure 3: Theoretical model. (a) Top view (b) Front view.

2) *Numerical Analysis:*

The second series of tests were performed to simulate the mechanical responses of sensor numerically. The finite element modeling of sensor shown in Fig. 4 for which a commercial finite element analysis software package (ANSYS, version 10.0) was employed.

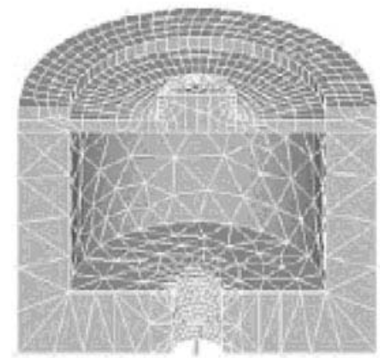

Figure 4: Finite element modeling of the sensor.

Table 1 shows typical specifications of the sensor modeled in the finite element method.

TABLE 1
SPECIFICATIONS OF THE MODELED SENSOR

Device (cylindrical)	2 cm (inner radius) 3 cm (outer radius) 5 cm (height)
Membrane	2 cm (radius) 100 µm (thickness)
Mesa	0.5 cm (radius) 150 µm (thickness)
ν (Poisson's ratio)	0.33
E (Elastic modulus)	30 MPa

Figure 5 shows a typical sample of numerical analysis. In this sample, the applied force is 0.5N. It shows that the maximum amount of deflection occurs at the center of the membrane.

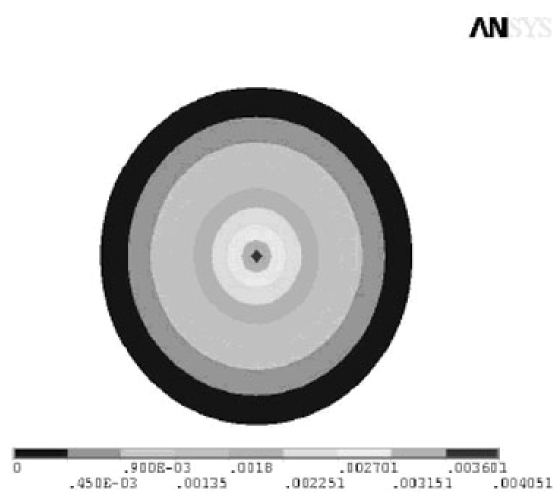

Figure 5: Deflection of the membrane due to the applied force.

C. Experimental Method

In order to evaluate the performance of proposed sensor, we have set one experiment for determining the contact-force between the sensor and an object and finding the relationship between the force and membrane deflection. In this test, a single tactile sensor was fabricated according to the theoretical model specifications. The material of membrane should yield sufficiently under low force, in an ideal elastic (no hystersis) and repeatable manner.

Several polymers were studied to identify one with suitable mechanical properties and more compatibility with the biological tissues. Materials with the high hysteresis were rejected. We chose a particular kind of silicon rubber as the membrane and we made the body of the sensor out of PVC. It should be noted that in tactile sensing, a force range of 0.1 N to 10N is considered to have practical applications in medical devices [25].

In this work, according to the hardness of membrane, the maximum magnitude of applied force was limited to 1 N. There is a possibility for determining the higher force with choosing a material with higher hardness for membrane. We embedded a strain gauge in the membrane exactly in peripheral and radial direction of membrane.

Axisymmetric shape of sensor is useful for simplifying and reducing the number of embedded strain gauges to only one. As we applied force to the center of membrane on mesa, the strain gauge measured the membrane stretching as a result of membrane deflection. So, the output signal of the strain gauge can be read as a force signal. We used a Wheatstone bridge circuit to measure the change of the resistance of the strain gauge, as shown in Fig. 6.

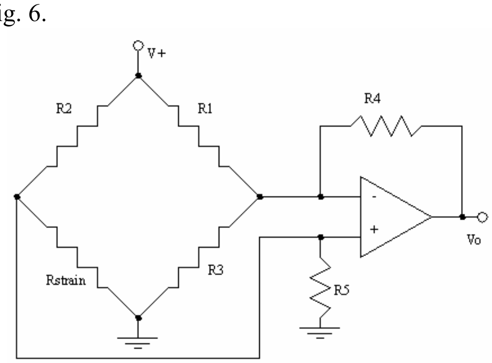

Figure 6: Wheatstone bridge circuit with amplifiers.

In this configuration, the change coming out via the strain gauge represented the force applied to the membrane. The change generated was amplified by a circuit and the output was measured by an oscilloscope. Static forces were applied to the sensor using steel weight and the output from the strain gauge was recorded. Loads were applied incrementally to nearly 1 N and then unloaded. The test setup is illustrated in Fig. 7.

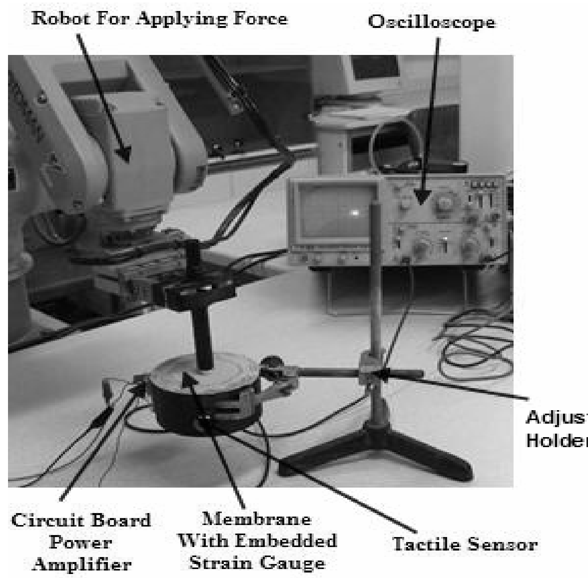

Figure 7: Photograph of the experimental setup.

4. RESULTS AND DISCUSSION

The theoretical, numerical and experimental results have been in good agreement. We investigated the deformation at the center of mesa and found that it changes with variations of applied force and the thickness of membrane. Fig. 8 shows the variations of OPD of membrane with the constant radius and unique force (0.1N) at different thicknesses.

As a result of applied force at the center of membrane, we have an out-of-plane deflection on it. Fig. 9 demonstrates the deformation or out-of-plane deflections of membrane according to the variations of applied force. At the range of 0.2-0.5 N, output of strain gauge was nearly linear and there was a good agreement with theoretical and numerical results.

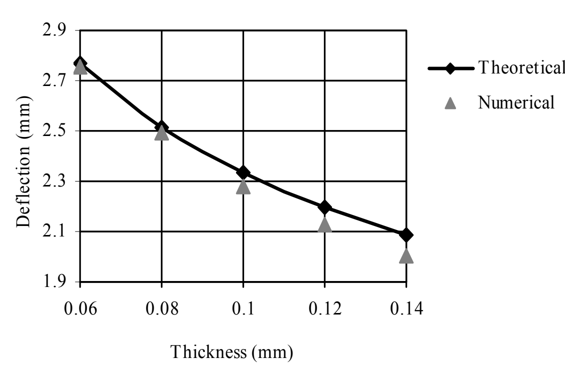

Figure 8: Variations of out-of-plane deflections vs. different thicknesses.

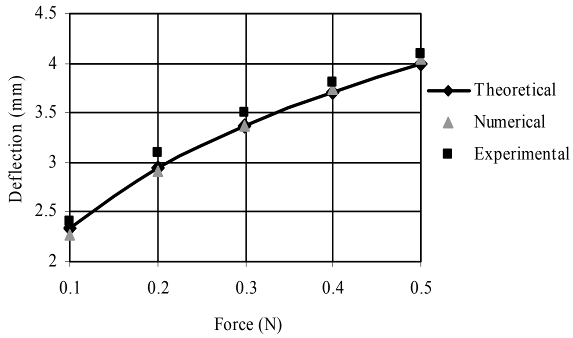

Figure 9: Variations of out-of-plane deflections vs. different applied forces.

Table 2 shows the values of numerical analysis and theoretical analysis along with the experimental results. As can be seen, there is good agreement between these results.

TABLE 2
COMPARISON OF NUMERICAL ANALYSIS, THEORETICAL ANALYSIS AND EXPERIMENTAL RESULTS

F (N)	Theoretical w_0 (mm)	Numerical w_0 (mm)	Experimental w_0 (mm)
0.1	2.335	2.277	2.4
0.2	2.942	2.910	3.1
0.3	3.368	3.363	3.5
0.4	3.707	3.732	3.8
0.5	3.993	4.051	4.1

As mentioned, the change measuring via the strain gauge represented the force applied to the membrane. Static forces were applied to the sensor using steel weight and the output from the strain gauge was recorded. Loads were applied incrementally to nearly 1N and then unloaded.

Figure 10 shows the measured output voltage of the strain gauge versus the applied force to the membrane.

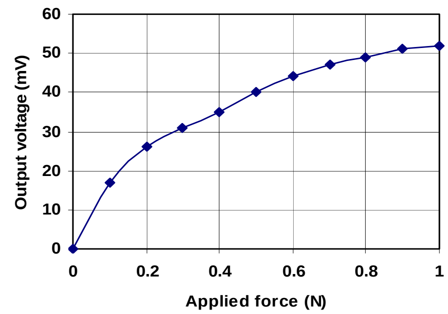

Figure 10: Output voltage vs. the applied force.

5. CONCLUSION

Although a considerable number of sensor technologies and strong theoretical models have been developed, there is still much left to be done in intelligent grasping and manipulation. Also, there is a gap in applying the theoretical models in association with tactile sensing. We proposed a new type of tactile sensor that can detect the contact force. First, we analyzed theoretically and numerically the operation of sensor and secondly we constructed the sensor according to these analyses.

This sensor is made from robust and flexible biocompatible polymers that can be used to directly touch biological objects. A major advantage of the designed system is that it can be easily miniaturized and micromachined. As a result, it can be mass-produced at low cost and even be disposable. Because of its biological compatibility, the designed sensor has two main applications, one in MIS and one in artificial palpation.

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