FEM Analysis of the Interaction between a Piezoresistive Tactile Sensor and Biological Tissues

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Abstract—The present paper presents a finite element model and analysis for the interaction between a piezoresistive tactile sensor and biological tissues. The tactile sensor is proposed for use in minimally invasive surgery to deliver tactile information of biological tissues to surgeons. The proposed sensor measures the relative hardness of soft contact objects as well as the contact force. Silicone rubbers were used as the phantom of biological tissues. Finite element analysis of the silicone rubbers and the mechanical structure of the sensor were performed using COMSOL Multiphysics (v3.4) environment. The simulation results verify the capability of the sensor to be used to differentiate between different kinds of silicone rubber materials.

Keywords—finite element analysis, minimally invasive surgery, Neo-Hookean hyperelastic materials, tactile sensor.

I. INTRODUCTION

TACTILE sensing is the evaluation of spatial distribution of forces vertical to the identified sensory field and the consequent displacement of the touched material [1], [2]. The tactile sensing could have an important role in Minimally Invasive Surgeries (MIS). MIS technology is now one of the most recommended options for numerous types of surgeries [2], while offering many advantages for surgeons and patients in comparison with open surgeries. Nevertheless, MIS still has some drawbacks. For instance, surgeons do not have the tactile perception with biological tissues due to the use of MIS surgical tools [3]. The ability to accurately sense tactile information of tissues allows surgeons to distinguish any hidden abnormalities in biological tissues, such as tumours, which are generally harder than surrounding healthy tissues [4]. Therefore, presenting a tactile sensor that measures the relative hardness of tissues in MIS, helps surgeons to identify any abnormalities in the tissues.

In the last few years, several multipurpose tactile sensors have been presented, utilizing different designs and principles, to measure the tactile information in MIS. Most of these tactile sensors focus on measuring the contact force. For example, Wisitsoraat *et al.* [5] have proposed a piezoresistive-based micro-machined tactile sensor, which can be used only for force measurement. On the other hand, some research groups aimed to measure more properties of the contact object. For instance, Bonomo *et al.* [4] have presented a multipurpose tactile sensor that uses the ionic polymer-metal composite (IPMC) cantilever beams as the sensing element. The IPMC beams deform under the effect of electrical field and, once deformed, they generate electricity. Despite the numerous advantages of using IPMC technology, the range of hardness measurement is still limited to less than 1kPa in the sensor. This limitation is due to the properties of IPMC and the maximum force it can deliver. Engel *et al.* [6], [7] have presented a polyimide-based multimodal and micro-machined tactile sensory skin that measures several mechanical properties of the contact object including the relative hardness. Although their sensor shows reliable results, it cannot measure any hardness higher than the hardness of polyimide material; and a rough contact surface (object) causes inaccurate measurements of relative hardness. Dargahi *et al.* [8] have presented a Polyvinylidene Fluoride (PVDF) piezoelectric-based micro-machined tactile sensor for endoscopic grasper, which is able to measure the magnitude and position of the contact force. Despite the acceptable results of their sensor, it is complex to evaluate shear force from the sensor output. In addition, the sensor measures only dynamic loads by virtue of the PVDF properties. Sokhanvar *et al.* [9] have proposed a PVDF piezoelectric-based miniaturized multifunctional endoscopic tactile sensor. Their sensor measures the relative hardness of the contact object, the contact force, and the position of the concentrated force. However, it cannot measure static loads. Moreover, due to the PVDF sensitivity to external noise, the sensor suffers from some errors in evaluating the contact force.

This literature shows that most of the tactile sensors have been designed only for force measurement. However, those aimed to evaluate different properties of contact object suffer from limitations in measuring due to either the properties of the sensing elements or the properties of the materials used to fabricate the sensor. Furthermore, some of these multipurpose tactile sensors are hard to manufacture due to their complex structures.

Most tactile sensors can be categorized based on their sensing principles; some use IPMC [4], whereas others employ piezoresistive materials [5]–[7], piezoelectric materials

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[8]–[11], capacitive sensing principle [12], [13], optical fiber [14], or pneumatic-based principles [15]. Among them, the piezoresistive principle is preferred due to (1) its fast response to static and dynamic forces, (2) its compatibility with micro-machining process, (3) its low sensitivity to external noise, and (4) its availability and low price for batch production.

In the previous work [16], we proposed a novel tactile sensor based on piezoresistive sensing principle for MIS applications. The sensor is able to measure the relative hardness of soft objects as well as the contact force, with a simple design that ensures the capability for micro-fabrication.

In the present paper, the interaction of this tactile sensor with two different silicone rubbers, resembling biological tissues, is simulated and analyzed in finite element model (FEM) using the COMSOL Multi-physics (v3.4) software.

II. PROPOSED SENSOR

A. Design Principle

The scope of the current study is to measure the relative hardness of touched objects, based on their behaviour to applied force. Ideally, hard materials have low amount of deformation under loading, while, as the hardness of the material decreases, the deformation increases under the same loading condition. Hence, it would be possible to evaluate the relative hardness of a material by measuring both its contact force and its corresponding deformation with a tactile sensor.

Physically, the material hardness is related to its young modulus (E), as

$$E = \sigma/\varepsilon \qquad (1)$$

where

$$\sigma = F/A, \quad \varepsilon = \Delta L / l_0 \qquad (2)$$

where stress and strain are functions of applied force (F), and displacement (ΔL), along the axis of the applied load, respectively. For a given geometry, the surface area (A) and initial length (l_0) are fixed; therefore, the ratio of the applied force over displacement ($F/\Delta L$) can denote an approximation for the relative hardness of a material, as

$$E = C\left(F/\Delta L\right) \qquad (3)$$

where C is a constant. The ratio of the applied contact force to the deformation of the object can indicate a measure to compare the relative hardness of materials. Hence, the relative hardness $R.H$ can be approximated as

$$R.H \cong F/\Delta L \qquad (4)$$

Harder materials have higher $R.H$ values than soft ones. As the $R.H$ value increases the degree of hardness increases. Using this principle, different type of materials can be distinguished based on their relative hardness. In the proposed sensor design, the deformation of touched materials is proportional to the deflection (δ) of an elastic beam, which can be measured with a piezoresistive film. Therefore, the obtained relative hardness by the sensor is equal to the ratio of the contact force (F) to the beam deflection (δ).

B. Design and Fabrication

Fig. 1 illustrates the novel design of the proposed tactile sensor. The design of the proposed sensor is fully demonstrated in our previous work [16]. Briefly, the sensor is made up of two simply fixed supports that hold up an elastic beam from its ends. The sensor includes four piezoresistive films placed as shown in Fig. 1. A hyperelastic material, Silicone rubber, fills the cavity between the elastic beam and the substrate, to build enough pressure on the middle top film. A prototype of the proposed sensor was built and tested. Table I illustrates briefly the components and materials of the sensor mechanical structure.
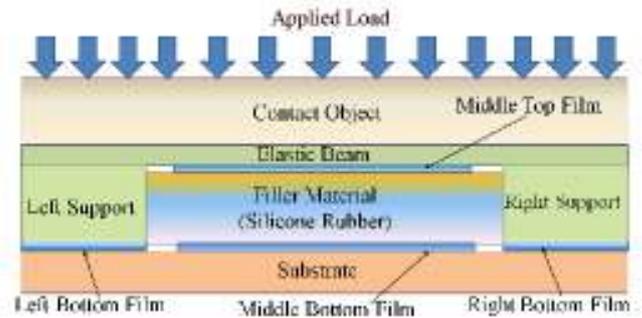

Fig. 1 Sensor design.

TABLE I
THE COMPONENTS AND MATERIALS OF THE SENSOR'S MECHANICAL STRUCTURE

Structure Component	Material	Dimensions (mm)
Supports	Plexiglas	$15 \times 8 \times 5$
Filler Material	Silicone rubber (Ecoflex 00-10)	$37 \times 8 \times 5$
Elastic Beam	Polystyrene	$67 \times 8 \times 1.2$

III. FINITE ELEMENT ANALYSIS

The purpose of the FEA is to simulate the proposed tactile sensor and test it when touching biological tissues, which are simulated as silicone rubbers [17]. Since silicone rubbers, in this work, act as both a filler material in the sensor structure and as biological tissues in experiments, an accurate model for silicon rubbers was developed by several experimental tests.

A. Mathematical Model

Silicone rubbers, as any other rubberlike materials, are isotropic and incompressible materials which undergo large elastic deformations [18]-[20]. Consequently, silicone rubbers can be modeled as isotropic incompressible hyperelastic materials using the non-linear elastic theory. A hyperelastic material (Green elastic material) is an elastic material and a special case of Cauchy materials that is defined by its strain energy function (W), i.e. a scalar objective function [19], [20]. For an isotropic hyperelastic material, the strain energy function, W, can be defined using principle invariants (I_1, I_2, I_3) of the left Cauchy-Green deformation tensor [20]:

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$$W = W (I_1, I_2, I_3) \qquad (5)$$

for an incompressible material, where the rate of change of the volume is zero, I_3 is equal to one; therefore, W depends on I_1, and I_2 only. So, (5) becomes [20]:

$$W = W (I_1, I_2) \qquad (6)$$

for such materials the stress tensor (S) is given by [20]:

$$S = \partial W(A)/\partial A \qquad (7)$$

where A is the deformation gradient tensor.

Many mathematical models have been established to represent the mechanical behavior of hyperelastic materials. The Neo-Hookean model is one of the most important models that are suitable for modeling isotropic incompressible hayperelastic materials [19], [20]. The Neo-Hookean model obtains the strain energy function for an incompressible hyperelastic material using the following formula [21]:

$$W (I_1, I_2) = \frac{\mu}{2} (I_1 - 3) + \frac{1}{d} (J - 1)^2 \qquad (8)$$

where μ is the initial shear modulus, d is the material incompressibility parameter, and J is the ratio of current to reference volume (J=det(A)). The initial shear modulus and the incompressibility parameter are obtained using [21]:

$$\mu = 2(c_{10} + c_{01}) \qquad (9)$$

$$\kappa = \frac{2 (c_{10} + c_{01})}{1 - (2\nu)} = \frac{2}{d} \qquad (10)$$

where c_{10}, and c_{01} are the Mooney-Rivlin constants for the material, κ is the initial bulk modulus, and ν is the Poisson's ratio of the material ($\nu \simeq 0.5$, for an incompressible material).

Mooeny-Rivlin constants of a hyperelastic material can be determined from the experimental data of any stress–strain test, such as compression test [19], [18]. In this work, three different types of silicone rubbers with different hardness values were molded and used, which are Ecoflex® 00-10, Ecoflex® 00-30, and Dragon Skin® 20, all from SMOOTH-ON Inc. (Pennsylvania, USA). A compression test was carried out for specimens from all these silicone rubbers using the BOSE ElectroForce 3200 device (BOSE Corporation, Minnesota, USA). The device accuracy is 0.01 N in load measuring, and 0.01 mm in displacement measuring. A Matlab code [19] was used to obtain the Mooney-Rivlin constants from the experimental data. Accordingly, equations (8) and (9) are used to calculate the mechanical properties of the silicone rubber to implement them in FEA.

B. FEA in COMSOL

Each silicone rubber was modeled in COMSOL Multiphysics software (v3.4) separately, using the dimensions of the tested specimens and the calculated mechanical properties. Table II presents the mechanical properties and number of meshing elements for each model. The FE models were formulated using the structural mechanics module in a plane strain application with a parametric analysis in 2D space. The structural mechanics module was suitable for hyperelastic modeling, because it allows large deformations

[19]. The solution was obtained using Lagrangian-quadratic elements, parametric analysis, and the ideal constrains in the software. Each model represents the compression test applied to a specimen. Therefore, the models include three main sub-domains; two plates sandwiched with the tested specimen. The lower and upper plates represent the lower (fixed) and upper (free) jaws of the ElectroForce 3200 device, respectively. In a simulation of the experimental compression test, the upper jaw moves in steps a total displacement of 1.5mm downward, each step is 0.05mm. The simulation results were compared to the experimental data and confirmed acceptable accuracy of the FE models for the three types of silicone rubbers. The experiments show the order of the silicone rubbers hardness values: Ecoflex 00-10 < Ecoflex 00-30< Dragon Skin 20.

TABLE II
MECHANICAL PROPERTIES, DIMENSIONS, AND MESHING ELEMENTS FOR EACH SILICONE RUBBER MODEL

Silicone rubber	Initial shear modulus (μ) (MPa)	Initial bulk modulus (κ) (MPa)	Dimensions (mm)	Meshing elements
Ecoflex 00-10	0.002743	0.030477	$37.11 \times 7.92 \times 5$	708
Ecoflex 00-30	0.009542	0.106017	$37.47 \times 8.12 \times 5.3$	752
Dragon Skin 20	0.063297	0.703228	$37.49 \times 7.98 \times 5.4$	712

After developing the FE models for the silicone rubbers, the mechanical behavior of the proposed tactile sensor in contact with a simulated biological tissue was analyzed and modelled. The new FE model, Fig. 2, in this stage includes the lower jaw, upper jaw, supports, filler material, elastic beam, and the
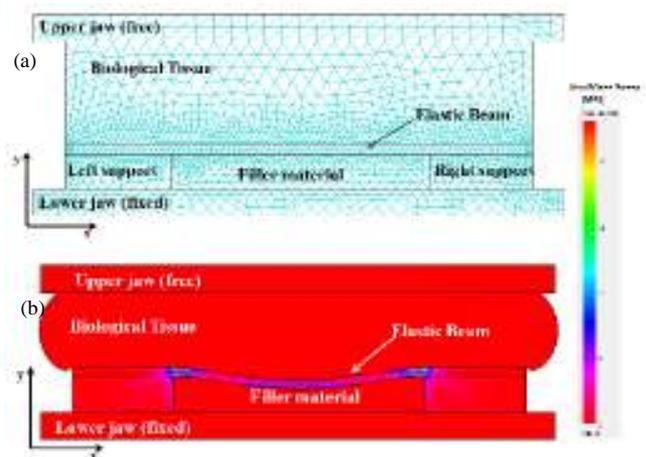

Fig. 2 The finite element model of the tactile sensor and its interaction with a biological tissue. The model simulating a practical test of this interaction using the Electroforce 3200 device. (a) Meshed structure. (b) The biological tissue and sensor structure after applying displacement to the upper jaw.
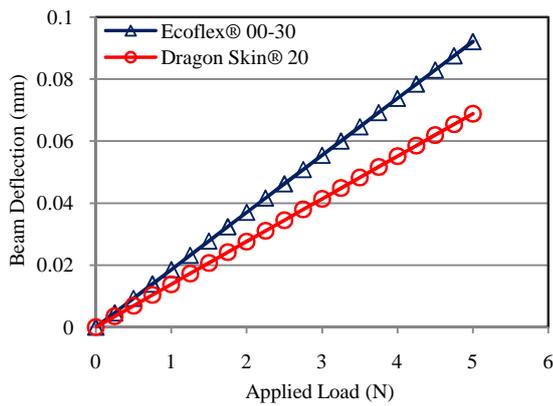
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Fig. 3 Deflection of the elastic beam under a specific load for both materials: Ecoflex 00-30 and Dragon Skin 20

Fig. 4 The measured contact force vs. the measured deflection of the elastic beam. (a) For Ecoflex 00-30, and (b) For Dragon Skin 20.

biological tissue in contact. The properties of the FE model are the same as silicone rubber models. The mechanical properties of the material used in the sensor in the FE model are estimated from properties of the materials used to build the experimental prototype.

In order to analyze the interaction between an MIS tool and a biological tissue, it was assumed that the surgeon applies either displacement or load to the tissue. Both scenarios could be simulated in this model. Simulating the displacement scenario is possible through a negative displacement in y-axis that is applied to the free jaw by the use of parametric analysis. The load scenario can be set through the use of parametric analysis to apply an incremental distributed load in the negative direction of y-axis to the upper jaw. Both simulations are applied to the silicone rubbers Ecoflex 00-30 (soft material) and Dragon Skin 20 (hard material).

IV. SIMULATION RESULTS

Fig. 3 shows the results of applying a load to both Ecoflex 00-30 and Dragon Skin 20. The maximum load was 5 N with increment of 0.25 N. From Fig. 3, it is clear that the Ecoflex 00-30 had larger beam deflection than Dragon Skin 20 for the same amount of applied force. Therefore, Ecoflex 00-30 is the softer material.

Fig. 4 presents the results of applying displacement to the upper jaw for both Ecoflex 00-30 and Dragon Skin 20. The total displacement was 3 mm, reached by an incremental step of 0.1 mm. The total thickness of the simulated biological tissue was assumed to be 15 mm. The slope in Fig. 4 (a), and (b) is F/δ, which is the relative hardness measured by the sensor. From the equations of the linear trend line for each silicone rubber, it was found that the slope of Ecoflex 00-30 and Dragon Skin 20 is 55.218 N/mm, and 79.204 N/mm, respectively. Here, again, the simulation shows that Dragon Skin 20 is the harder material.
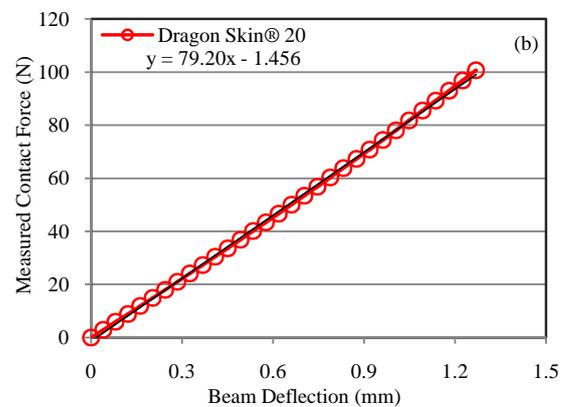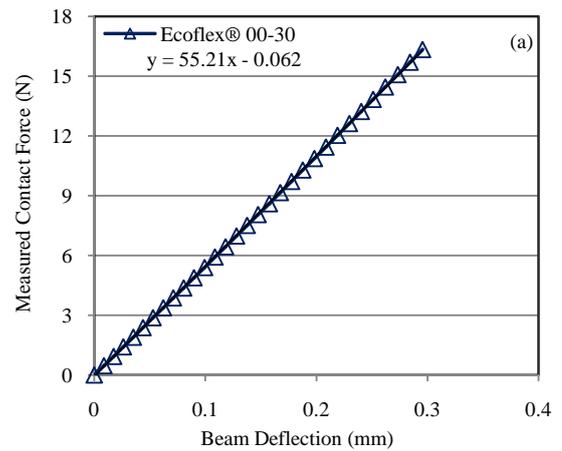
V. DISCUSSION AND CONCLUSION

The simulation results ensure the validity of the sensor principle to measure the relative hardness of two different biological tissues represented by Silicone rubber materials. The model was able to simulate different scenarios of practical application. Although the range and the sensitivity of the sensor are reliable, it is possible to improve them by changing the design parameters of the sensor, such as the hardness of the filler material.

As a future work, the simulation of interaction between the sensor and silicone rubber would be carried out experimentally, and then the experimental data will be compared to the numerical simulation results. Furthermore, FEA could be performed to verify the ability of the sensor to find the position of concentrated loads, such as abnormalities in tissues.

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