Design, Fabrication and Testing of a Piezoresistive-Based Tactile Sensor for Minimally Invasive Surgery

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***Abstract*—Minimally invasive robotic surgery (MIRS) has become a preferred method for patients for the last two decades, thanks to its crucial advantages over classical open surgeries. Although MIRS has some advantages, it has a few drawbacks. Since MIRS technology includes performing surgery through small incisions using long slender tools, one of the main drawbacks of MIRS becomes the loss of direct contact with the patient's body in the site of operation. Therefore, the surgeon loses the sense of touch during the operation which is one of the important tools to investigate the health condition of the treated tissue. This paper presents a novel piezoresistive-based multifunctional tactile sensor that is able to measure the contact force, relative hardness of the contact object, and the position of a hidden lump at the same time. A prototype of the designed sensor has been fabricated, simulated, analyzed, and tested numerically and experimentally. The experiments have been performed on hyperelastic materials, which are silicone rubber samples with different hardness values that resemble different biological tissues. The ability of the sensor to measure the contact force and relative hardness of the contact objects is tested with several experiments. A finite element (FE) model has been built in COMSOL Multiphysics (v3.4) environment to simulate both the mechanical behavior of the silicone rubber samples, and the interaction between the sensor and the silicone rubbers. Both numerical and experimental analysis proved the capability of the sensor to distinguish between different silicone rubber samples.**

***Index Terms*—Minimally invasive surgery/devices, Tactile sensor, Tissue modeling/contact, Hyperelastic material modeling, Piezoresistive films.**

I. Introduction

The sensory receptors in human skin instantly convert and transfer information about our tactile perception with surrounding objects, allowing us to differentiate between touched objects based on their size, temperature, shape, and texture. Having this information, one is able to identify the proper contact pressure needed to handle an object. Therefore, the loss of this sense crucially limits one's knowledge of the surroundings [1]. Tactile sensing could be defined as the assessment of vertical forces distribution over a specified sensory field [2]. It is a fundamental need for the improvement of many fields such as virtual systems, robotic manipulators, tele-operations and medical applications. Improvements in automation and micro-electro-mechanical systems enable researches to produce various tactile sensors [3].

Minimally Invasive and Robotic Surgery (MIRS) has attracted many researchers over the last twenty years by virtue of MIRS capability in replacing classical open surgeries in almost all surgical disciplines [4]. MIRS has become the most popular method for detection and treatment of cancer, which is predicted to be the single most reason of death in the world in the near future [5]. Also, it is reported that MIRS could give the same or improved results compared to the classical open surgery for heart valve replacement [6].

Despite many advantages, MIRS has some significant shortcomings. Since in MIRS the surgeon's hands are outside the site of operation, it is considered as a teleoperation application, where direct contact with the object is lost [7]. This main shortcoming is a limitation for expanding the MIRS applications. For example, palpation is one of the routine procedures in many surgeries, where the surgeon in open surgeries will pass his hand over the tissue to investigate whether there is any abnormality or tumor hidden in the tissue. Palpation is a common practice since it is known that tumors are harder than surrounding healthy tissues, and tissues composition usually changes under the effect of different illnesses [8]. Unfortunately, in MIRS the surgeon is not able to insert his hand through the small incisions to palpate the tissues. However, surgeons have found alternative palpation methods, such as inserting a metal rod through the incision to detect the abnormalities. Though, any deep tumors will be hardly detected with this method, and extensive training is needed to avoid mistakes and harms to the tissue [9]. Moreover, the metal rod method is not suitable for all MIRS surgeries especially those near sensitive tissues such as brain surgery.

All the problems associated with MIRS could be greatly reduced or even solved if a bio-comfortable, accurate, multifunctional, and in-vivo tactile sensor was used on the tips of MIRS tools such as endoscopic grasper. Such a tactile sensor will replace the sense of touch in the surgeon's hand and give a precise feedback to the surgeon, while performing surgery. This feedback provides the surgeon with important information about the contacted tissue regarding its relative hardness, contact force, and tumor existence. The tactile sensor

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improves MIRS efficiency and expands its applications. It is important to note that the property of soft tissues alters when detached from the body; therefore, the sensor should be able to work in-vivo.

Numerous tactile sensors with different principles and designs have been presented for MIRS, many of which concentrate on contact force measurements [10]. For instance, Wisitsoraat et al. [10] proposed a piezoresistive-based micro-machined tactile sensor that can measure the contact force, only. On the other hand, some researchers have worked on evaluating different features of the contact object. For example, Sokhanvar et al. [11] designed a Polyvinylidene Fluoride (PVDF) piezoelectric-based micro-machined multifunctional tactile sensor for endoscopic grasper, which is able to measure the contact force, relative hardness of the contact object, and the position of any hidden lump inside the object. The sensor suffers from inaccuracy in evaluating the contact force because of the PVDF sensitivity to external noise. Furthermore, the sensor measures dynamic forces, only. Qasaimeh et al. [12] proposed another multi-purpose PVDF piezoelectric-based tactile sensor for endoscopic grasper, in which the functionality of Sokhanvar's [11] sensor is improved by modifying the endoscopic grasper so that the sensor is able to cover the entire grasped surface. Regardless of the reliable results of the sensor of Qasaimeh et al. [12], due to the PVDF properties, it measures dynamic forces only. Furthermore, the new proposed structure for the endoscopic grasper is brittle. Barmana et al. [13] presented a deformable force-stretch array (DFSA) tactile sensor that is able to detect nodules in palpation. The DFSA sensor consists of two main sensing elements: (i) the strain gauges attached to a piston-cylinder system harder than the palpated tissue; and, (ii) the stretch sensor elements that give an output proportional to their length. Despite the novel design of the DFSA sensor and its acceptable results, there is no proof that its complex structure is appropriate for micro-fabrication or MIRS applications. Shikida et al. [14] proposed a pneumatically actuated piezoresistive-based micro-fabricated and multifunctional tactile sensor which is able to measure the contact force and relative hardness of contacted objects. Even though the sensor is micro-fabricated and shows promising results, building and controlling an accurate pneumatic system for an array of this sensor is costly.

The literature review shows that the majority of tactile sensors are force sensors, and the other tactile sensors that measure several features of the contact object have some disadvantages. For instance, some of them have constraints in measurements by virtue of the properties of the materials used for either sensing mechanism of the sensor, or fabricating the sensor. Moreover, some sensors have a complicated structure that makes the micro-fabrication of the sensors a formidable task.

The presented sensor in this work—a piezoresistive-based multifunctional tactile sensor for MIRS applications—is sensitive for dynamic and static loading conditions, by virtue of the properties of piezoresistive sensing elements. The sensor is able to measure the contact force, relative hardness of the contact object, and the position of any inclusions inside the grasped object. Moreover, due to the combination of both hyperelastic and linear elastic materials used to build the sensor's composite structure, it is less fragile compared to other sensors [11, 12]. Although it is less breakable, the simplicity of its structure ensures the ability for micro-fabrication. In addition, the sensor can be integrated in a MIRS endoscopic grasper.

The first part of this paper describes the design of the proposed sensor and its working principle. Then, a brief background theory is presented about the sensing elements of the sensor. Next, the fabrication process is described. In the modeling section, the sensor is numerically modeled using a finite element modeling (FEM) software. Experiments section describes the tests that were carried out to model the silicone rubber samples, the calibration of piezoresistive sensing elements, and the functionality verification of the sensor. Finally, the results of both numerical analysis and experiments are compared and discussed on two different silicone rubber samples.

II. Design and Structure of the Sensor

Piezoresistive films are utilized as sensing element of the proposed sensor [15]. Among all known types of sensing elements, semi-conductive polymer composite film is used in this work due to its (1) quick response to both dynamic and static loading; (2) reduced sensitivity to external noise; (3) compliance with micro-fabrication techniques; and (4) cost effectiveness for mass production. In this section, the composition, mechanical properties, and working principle of this sensing material are presented.

A. Sensor Design

The mechanical structure of the sensor, presented in our previous work [16], consists of an elastic beam sitting on two supports that are placed at both end of the beam. The gap between the elastic beam and the substrate is filled with a soft filler material, i.e. a soft silicone rubber, as it is shown in Figure 1. The piezoresistive semi conductive polymer sensing elements are placed at three positions in the sensor. As illustrated in Figure 1, two of the piezoresistive films are placed under the beam supports, one at the left side and one at the right side. The third film is in the middle, attached to the beam at the lower surface between the beam and the silicone rubber.

By using this configuration of the sensor structure and piezoresistive films, it would be possible to measure the contact force, relative hardness of the contact object. The piezoresistive films underneath the supports of the sensor measures the total contact force applied to the top of the sensor. On the other hand, the piezoresistive film attached to the beam measures the beam deflection resulted from the deformation of the contact object. Since piezoresistive films work with compressive forces only, the silicone rubber is used under the middle film to transfer enough pressure to the middle film. The presence of the silicone rubber would increase the sensor capability to stand larger forces without breakage. However, the hardness of the silicone rubber is an important factor in the design, since it would define the range of hardness values that

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the sensor can measure. Experiments showed that for a wider range of measurements the filler material should be softer. In other words, the filler material should be softer than the softest material to be tested by the sensor, in order to ensure that the beam is able to deflect under the deformation of the softest material. As a result, the contact force can be measured from the summation of the output of the piezoresistive films on both sides, since the supports are rigid and will transfer almost all the force to the piezoresistive films underneath. This has been also proved by FEM analysis in this work.

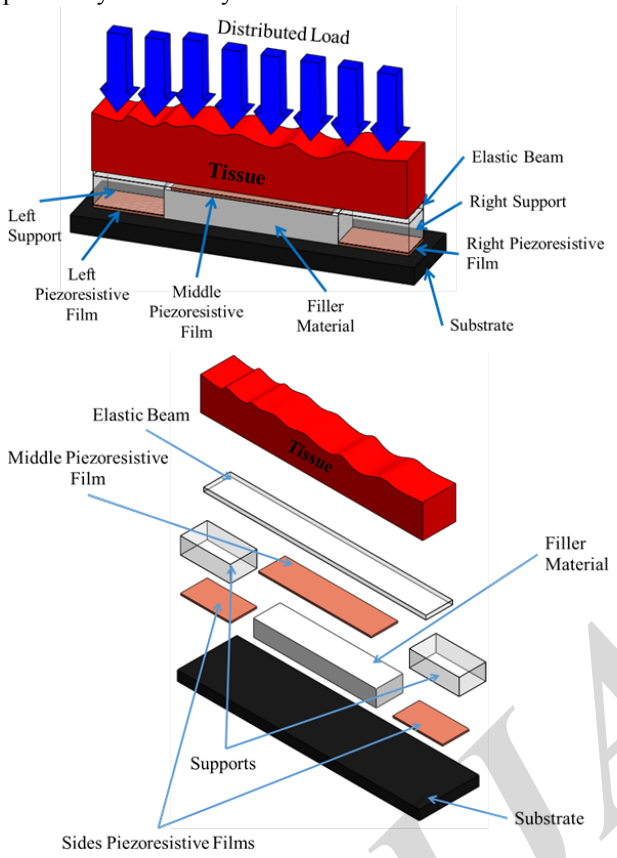


Figure 1. (a) Sensor design in contact with the tissue. (b) 3D exploded view of the sensor and the tissue.

B. Principle of the work

The sensor is aiming to measure the relative hardness as defined in [16] of the contact object depending on its behavior under different loading condition. Hard material will have less deformation, but as the material hardness decrease the deformation of the material would increase under the same loading condition. Hence, for a specific material, the deformation is proportional to the loading magnitude. Evaluating the relative hardness of a material requires measuring the contact force and the material deformation simultaneously. In the proposed sensor, it is assumed that the contact object is in full contact with the beam, and the applied load to the material by MIRS tool is distributed. Hence a soft object would have larger deformation under the same loading, generating larger deflection on the beam, which results in the higher output voltage of the middle piezoresistive film. Regardless of the hardness of the object, the piezoresistive films underneath the supports measures the contact force transferred to them through the piezoresistive films underneath the supports measures the contact force transferred to them through the sensor structure. Therefore, the relative hardness of the material can be evaluated. Figure 2 illustrates the measurement principle.

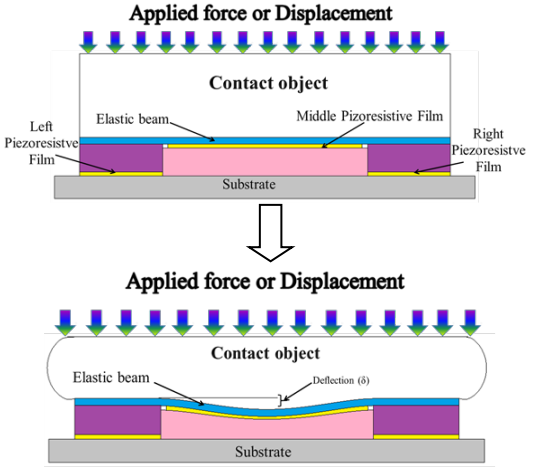


Figure 2. Working principle of the sensor (relative hardness measurements) (a) before applying force or displacement, (b) after applying force or displacement.

It is important to note that measuring only the relative hardness does not need more than two piezoresistive films. One film measures the contact force, which could be placed either on the base along with the sensor structure, or under one of the supports. The second film is placed in the same place as the current middle piezoresistive film.

The range of hardness measurements can be tuned to fit the required application, depending on many factors such as the hardness degree of filler material. In addition, the elastic beam dimensions and its modulus of elasticity play a crucial role in the sensor sensitivity, range of measurements, and the maximum applicable force before the sensor fails. For instance, thin beam increases the sensor sensitivity, and reduces the maximum applicable force. Same results would be obtained if the beam gets longer, narrower, or more elastic (lower modulus of elasticity).

The proposed design can be used in an array to be implemented in the endoscopic grasper jaws. In an endoscopic grasper, the number of sensors used will depend on the grasper dimensions. In order to cover the entire grasper jaw surface, each tooth and spacer would include two sensors, one in the tooth, and the other in the spacer. Each sensor contains the beam, the supports, three piezoresistive films, and the filler material. In order to give the toothed shape to the grasper and for staple grasping, the support of the tooth is higher than the support of the spacer.

III. Fabrication of the Sensor

Fabrication of the proposed prototype was carried out manually. Figure 3 shows the dimensions of the prototype. The elastic beam and the supports were cut out from an elastic Polystyrene sheet (Plaskolite Inc.) and an Acrylic sheet (Plaskolite Inc.), respectively. Then, the elastic beam and the

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supports were glued together. Next, the soft filler material was molded from a silicone liquid rubber called "Ecoflex 00-10", produced by SMOOTH-ON Inc. This type of silicone rubber was used since it is the softest material among other silicone rubbers, as shown in the experiment section of this paper. Therefore, a wider range of measurements and testing can be performed to investigate the functionality of the prototype. Table I shows the components of the mechanical structure of the prototype.

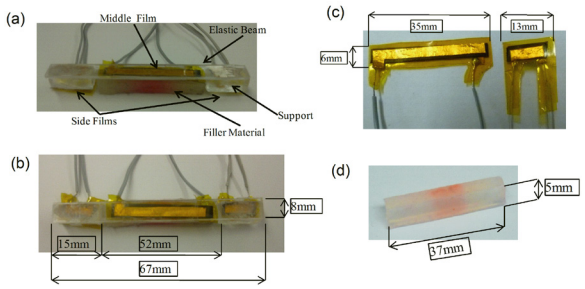


Figure 3. The fabricated prototype, its parts and dimensions: (a) Isometric view of the prototype. (b) Top view of the prototype. (c) Piezoresistive films used in the prototype. (d) The hyperelastic filler material.

Figure 4 shows a sensing element. The piezoresistive films (sensing elements) were fabricated in the lab from a semi conductive polymer composite sheet. A double layer of semi conductive sheet was inserted freely between two copper electrodes, which were soldered to electrical wires. Then, two layers of nonconductive silicone sheets were used to cover the electrode and create the initial pressure that was enough to hold all the sheets together. The resolution of this fabricated sensor is 0.1 N. The sensors then are connected to DAQ and to a LabView code for displaying results.

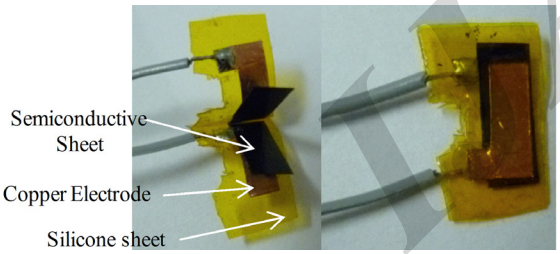


Figure 4. A fabricated sensing element and its components.

IV. Finite Element Model (FEM) of the Sensor

A finite element (FE) model for the proposed prototype is built to investigate the sensor response and functionality regarding the measurement of the contact force and relative hardness of the contact object. The silicone rubbers, which are hyperelastic materials, were used in the sensor structure. They were also used as testing samples that interact with the proposed sensor in the experiments, since the biological tissues can be simulated as silicone rubbers. Figure 5 shows both FEA model of the sensor with a silicone rubber sample.

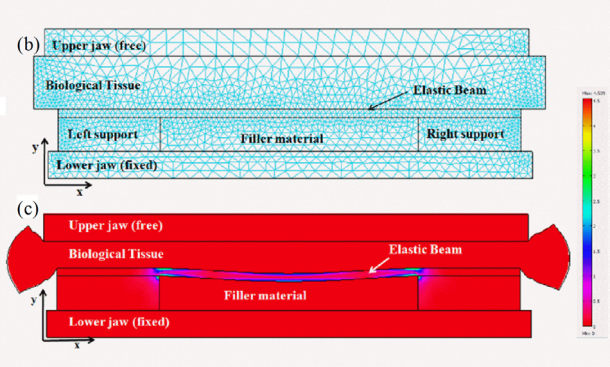


Figure 5. The biological tissue and sensor structure after applying displacement to the upper jaw.

V. Experiments

Several experiments were carried out to calibrate and to validate the proposed prototype in measuring the relative hardness of the contact object and contact force.

A. Calibration

In order to use the proposed prototype in measuring the relative hardness and contact force, all sensing elements (piezoresistive films) should be calibrated experimentally. With a simple analysis of the electrical voltage divider circuit the following equation can be obtained

$$R_{film} = R_{const} \times \left[\frac{V_i - V_{out}}{V_{out}} \right] \qquad (1)$$

where R_{film} is the resistance of the piezoresistive film, R_{const} is the constant resistance (9.4 kΩ), V_i is the DC input voltage to the voltage divider circuit, and V_{out} is the output voltage taken from the voltage divider circuit.

Once Equation (1) is implemented in LabView, the readings of the output voltage can be processed to calculate the resistance of the piezoresistive film corresponding to each output voltage value. The output voltage entered in LabView was filtered with a third-order Butterworth low-pass filter with a cut off frequency of 6 Hz to eliminate the noise from the voltage signal. An Electroforce 3200 Bose device (BOSE Co.) was used in order to calibrate the sensing elements. The accuracy of this device in measuring displacement and force is 0.01mm and 0.01N, respectively.

The Elecroforce 3200 Bose was used to apply different distributed forces discretely on each of the side piezoresistive films. The resistance of the film was recorded from LabView one second after applying the force. The applied force was increased each time with a step of 1N up to a maximum force of 9N and the resistance of the film was recorded. Figures 6.a. and 6.b. show the obtained results for this calibration. The experiments showed a linear relationship between the film conductance and the applied compressive force.

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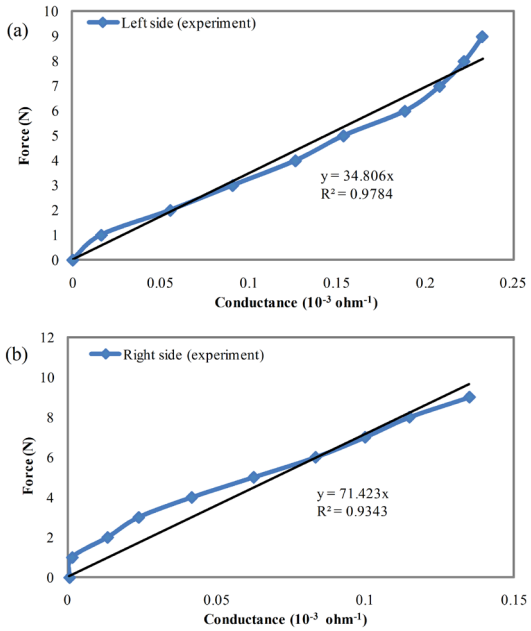


Figure 6. Calibration results used to obtain the mathematical relationships between the piezoresistive films conductance and either force or deflection. (a) Calibration of the left side piezoresistive film. (b) Calibration of the right side piezoresistive film.

Using interpolation for the collected data from the calibration tests, the following equations can be obtained to calculate the contact force and the beam deflection

$$F_{cont} = F_{ls} + F_{rs} \quad (2)$$

$$F_{ls} = 34.806\, X_{ls} \quad (3)$$

$$F_{rs} = 71.423\, X_{rs} \quad (4)$$

$$\delta_{beam} = \sqrt{(35.15\, X_{beam} - 0.0432)} \quad (5)$$

where F_{cont} is the contact force (N), F_{ls} and F_{rs} are the reaction force at the left and right support (N), respectively, X_{ls} and X_{rs} are the conductance of the piezoresistive film underneath the left and right support, ($k\Omega^{-1}$), respectively , δ_{beam} is the deflection of the elastic beam (mm), and, X_{beam} is the conductance of the middle piezoresistive film ($k\Omega^{-1}$).

VI. Prototype Testing

The same experimental setup as the one used in the calibration test is conducted to test and validate the functionality of the proposed prototype. The prototype was fixed to a base on the stationary jaw of the Electroforce device. The specimens were individually placed on top of the prototype. Then, using the movable jaw of the Electroforce, a distributed compressive load was applied to the specimen as a reference square force function to simulate the grasping process in MIRS. The reference square force function was repeated five times with a frequency of 0.2 Hz and a magnitude of 0 to 5 N. The applied force function tests the repeatability of the sensor and its ability to work with both dynamic and static forces, and shows the quick response of the sensor.

A. Force sensing

Figure 7 shows the contact force measurement when different types of silicone rubber are placed on top of the sensor. It is shown in the figure that the sensor is able to evaluate the contact force, precisely, for all hardness values of the contact object.

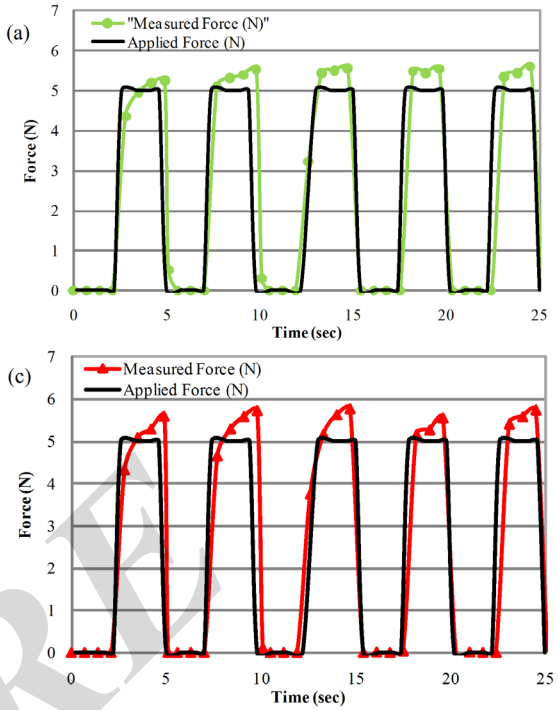


Figure 7. The Measured contact force by the prototype using different silicone rubbers. (a) Measured contact force when Ecoflex 00-50 is tested. (b) Measured contact force when Dragon Skin 20 is tested.

B. Relative hardness testing

Figure 8 shows the beam deflection for different types of silicone rubber. The two tested samples are Ecoflex 00-50, and Dragon Skin 20. As Figure 8 illustrates, the Dragon Skin 20 (the hardest) has the minimum deflection, and the Ecoflex 00-50 (the softest) has the maximum deflection. Figure 8 shows that as the material hardness increases, its deformation and the corresponding beam deflection decrease. Therefore, the sensor is able to distinguish between different materials based on their hardness.

VII. Discussion and Conclusion

A novel multifunctional piezoresistive-based tactile sensor for MIRS applications is presented in this paper. The sensor measures the contact force and relative hardness of the contact object. In order to test the proposed sensor for MIRS application, silicone rubber samples were chosen to resemble different samples of biological tissues with different hardness values in both the FE models and experiments.

The FE models for the silicone rubbers were built in Comsol Multiphysics software. FE models are validated using the mechanical compression tests. Furthermore, the proposed sensor was modeled in Comsol to investigate its interaction with silicone rubber samples. The analysis of the FE model of

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the sensor ensures the validity of the sensor principle to measure the contact force and relative hardness of the silicone rubber samples. The FE model of the sensor simulates the grasping process in different scenarios.

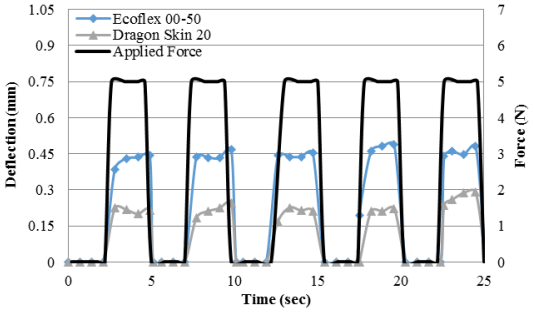


Figure 8. The deflection of the elastic beam measured by the middle piezoresistive film for three different silicone rubbers under a square force function. Ecoflex 00-50 is the softest material, while Dragon Skin 20 is the hardest material.

A miniaturized prototype of the proposed sensor was built and tested to investigate the sensor functionality and to validate the FEA results. The experiments tested the ability of the sensor to measure the contact force and the relative hardness of the contact object to differentiate the materials depending on their hardness. The Square force function applied in the experiments shows the fast response of the sensor, the output stability, and the sensor repeatability. The experimental results show some acceptable variation in the outputs value that can be eliminated by using a suitable filter, advanced calibration devices, or an advanced fabrication process such as using an accurate commercial pressure sensor.

Both simulation and experimental results ensure the validity of the sensor principle to measure both contact force and relative hardness of the contact object. The results show the same order of hardness degree for the three used silicone rubbers. In addition, the experiments show the sensor ability to work with both static and dynamic loading condition. It is important to note that the loading in MIRS applications is not dynamic all the time. In fact, grasping in MIRS is often considered as a static loading process than dynamic loading. This feature makes the proposed sensor more suitable for MIRS applications.

Although, in this study, the range and sensitivity of the sensor are reliable, the range of the hardness values considered in MIRS applications is wide depending on the targeted tissues in the surgery. However, the range of hardness measurements and the sensor sensitivity can be tuned by changing the design parameters such as the hardness of the filler material that should be softer than the hardness of the softest targeted tissue, or by changing other design parameters such as the dimensions and mechanical properties of the elastic beam.

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