Application of Tactile Sensing in Determining Characteristics of Biological Tissues

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ABSTRACT

In this study, the design, fabrication, and testing of different piezoelectric and piezoresistive tactile sensors used in endoscopic tools for detecting compliance and softness of the biological tissues are being presented. In the comparison made between the experimental data and the results obtained from the finite element analysis of the systems, it was found that there is a reasonable correspondence between these two methods for the designed sensors. Also, we propose and test a novel method to investigate the effects of the tumor existence that appear on the surface of a biological tissue. Finite element analysis provided properties such as the shape, depth, and location of the tumor which are all important parameters for physicians to pinpoint the correct condition of the patients. We obtained good agreements between the numerical and experimental results received from various artificial tactile sensing systems.

KEYWORDS

Tactile sensing, finite element analysis, membrane, polyvinylidene fluoride (PVDF), grasper

1. INTRODUCTION

There are five main sensing modalities, i.e., sight, sound, smell, taste, and touch. It is important to briefly compare them to gain insight into the development of tactile sensing. Computer vision systems are now commercially available and widely used for industrial inspection, recognition, monitoring, and many other application areas. Similarly, for audio systems, many kinds of audio processing systems are available and there exists a long and substantial literature on the mathematics and technologies of signal processing. Human speech analysis and speech recognition are currently very active research areas and have identified a wide range of applications. Even the senses of smell and taste have their electronic analogies. Devices known as electronic noses are now readily available for the detection of a range of molecules, and chemical tests can be implemented to automatically analyze across particular spectrums in a simulation of taste [1].

We do have a history of research in tactile sensing, but it is nowhere near as large in volume or as well-developed as in the other sensory modalities. Drawing on the human analogy, we see some of the difficulties: no localized sensory organ, complex sensing, and difficult to imitate. Tactile sensing can be defined as a system that can measure a given property of an object or contact event, through physical contact between the system and the object [2]. Here, the sense of touch seems to have received less attention comparing to visual and auditory senses. A considerable number of research reports in tactile sensing exist, but it is simply not as much as the research in the other sensory modalities and this is both from the size and development point of view [3,4]. By referring to the tactile sensing abilities in human, a number of problems can be envisioned that may be related to the lack of sufficient progress in this important field of science [5].

Potential application areas for tactile sensing are robotic surgery through minimally invasive surgery (MIS), rehabilitation procedures, service robotics, MEMS research, agriculture, and food processing [6]. Tactile sensors are utilized to sense a wide range of stimuli in various biomedical engineering and medical robotics applications, such as detection of the presence or absence of a grasped tissue/object or even mapping of a complete tactile image [7]–[10].

From a biomedical engineering point of view, surgery is probably the most interesting and fastest-developing area of research in which the use of tactile and visual sensing is of critical importance [10,11]. In spite of the fact that minimally invasive surgery (MIS) is only a decade old, it is now being used routinely as one of the most preferred choices for various types of operations [12,13]. MIS is, in effect, both a visual and a tactile

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procedure, and hence any inhibitions on the surgeon's sensory abilities are highly undesirable [14]. Despite its many advantages, such as reduction of trauma, less pain, smaller injuries, faster recovery time, and reduction of post-operation complications, MIS decreases the tactile sensory perception of the surgeon during grasping or manipulation of biological tissues [15].

A surgeon should be able to feel the tissues and detect the presence of blood vessels and ducts during a procedure. This ability is especially important during controlled manipulation tasks such as grasping internal organs, gentle load transferring during lifting, removing tissues (e.g., gall bladder in laparoscopic surgery and loose bodies in knee arthroscopy), and suturing tissues together [16]. These capabilities, coupled with the ability to detect various tactile properties, demonstrate the importance of tactile sensing in MIS.

Tactile sensors could be incorporated into the surgical tools to detect and control the contact of the tools with delicate biological structures. Currently, commercially available endoscopic graspers employed in MIS do not have any integrated tactile sensors. This means that the surgeon normally relies only on his/her experience and visual abilities to perform the maneuvers mentioned earlier [17].

Following the above facts, it has become evident that in robotic surgery, the application of the sense of touch is of great importance. In this case, determining the stiffness/softness, surface texture of tissues and detecting the presence of any embedded object inside a tissue organ such as subdermic tumors or cancerous tumors of the human breast are interesting fields of study which have not been considered sufficiently. Therefore, in this paper, the most recent and novel applications of tactile sensing designed and tested by the author are being reported.

2. Design and Fabrication of Piezoelectric-Based Tactile Sensor for Detecting Compliance

In this part, we investigate on the proof of a conceptual and innovative design of an endoscopic piezoelectric tactile sensor. The sensor is capable of measuring the total applied force on the sensed object, as well as the compliance of the tissue/sensed object. The sensor has a rigid cylinder surrounded by a compliant cylinder. When an object/tissue is in contact with the sensor, the degree of transfer of the load from rigid element to that of compliant element is used to determine the compliance of the sensed object. One polyvinylidene fluoride (PVDF) film is placed between the rigid cylinder and the plate which measures the force applied on the rigid element. Another PVDF film is sandwiched between the two base plates measuring the total force applied on the sensor. The compliance of the sensed object is measured by recording the PVDF films response under different load sets. When an object is pressed against the sensor, the softer the object, the more the load is transferred to the compliant cylinder. If the object is rigid, then all the force is applied to the rigid cylinder only.

Both the experimental and theoretical aspects of PVDF-based tooth-like endoscopic grasper are presented in this study. Figure 1 shows the detailed dimensions of the constructed sensor.

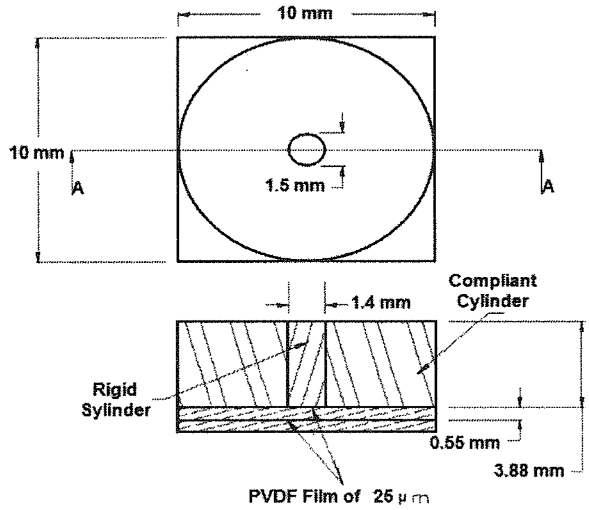

Figure 1: Dimensions of the designed sensor.

Figure 2 represents a comparison made between the FEM (finite element modeling) results and those obtained from experimental studies. As the modulus of elasticity increases, the sensed object becomes stiffer than the compliant cylinder of the sensor. The results show that the experimental and the FE approach agree with each other with a reasonable band.

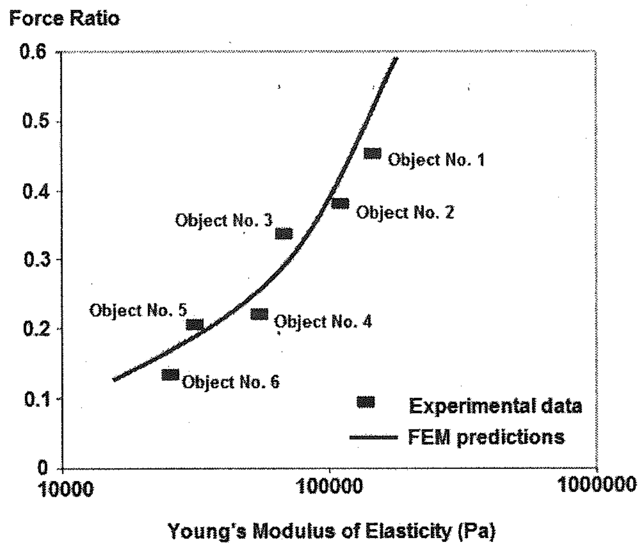

Figure 2: Comparison between experimental data and theoretical results.

3. Theoretical and Experimental Analysis of a Piezoelectric Tactile Sensor for Use in Endoscopic Surgery

Following the pervious study, we investigated the design, theoretical, and experimental analyses of a polyvinylidene fluoride (PVDF) tactile sensor, which could be integrated with an endoscopic grasper. This tactile sensor assembly consists of three main parts:

(1) endoscopic cylindrical tube,
(2) grasper jaws, and
(3) tactile sensors.

Figures 3 and 4 showthe detailed drawing of the tactile sensor's structure.

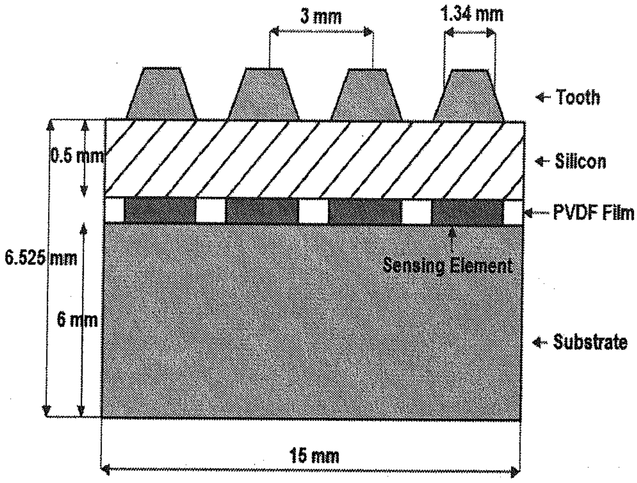

Figure 3: Lateral section of the sensing system (not to scale).

Applying forces to the top layer of the endoscopic grasper (the silicon part) leads to stress generation in the PVDF film. This stress, in turn, results in a polarization charge production at each electrode. There is a direct proportionality between the signal amplitude and the applied force magnitude. Similarly, the slope of the signals is an indication of any localized position of the exerted force.

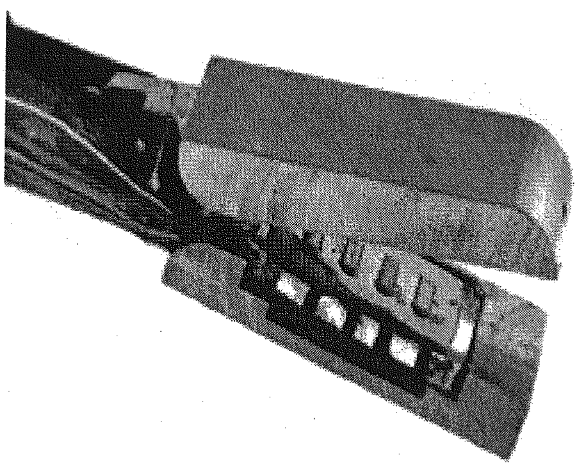

Figure 4: Photograph of the designed endoscopic grasper.

Although applying higher values of forces was feasible practically, the maximum loads were restricted to 2N to avoid possible damage to the sensors. In experiments, vertical loads were applied directly to the center of each tooth so that a linear relationship was obtained between the output charge from PVDF film (mVolts) and the reference load (Newtons). Finite element analysis was used to investigate the shear stress distribution and the deformation of the tactile sensor under various types of loading.

In the comparison made between the experimental data and the results obtained from the finite element analysis of the system, it was found that there is a reasonable correspondence between these two. The difference varied in the range of four to nine percent.

4. A SUPPORTED MEMBRANE TYPE SENSOR FOR MEDICAL TACTILE MAPPING

This study describes the design, fabrication, testing, and mathematical modeling of a supported membrane type polyvinylidene fluoride (PVDF) tactile sensor. Using the designed membrane type sensor (MTS), it is shown that the entire surface of the PVDF film can be employed as a means of detecting the force magnitude and its application point. This is accomplished by utilizing only three sensing elements. Unlike the array type tactile sensors, in which the regions between the neighboring sensing elements are not active, all the surface points of the sensor are practically active in this MTS. 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 (Figure 5).

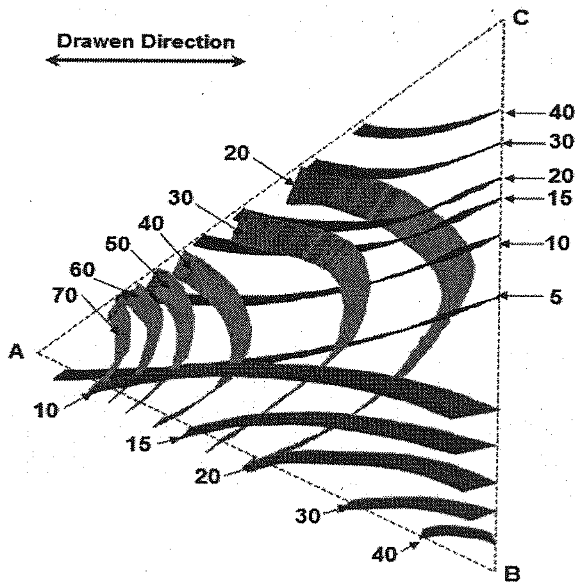

Figure 5: Three-dimensional map of the isocharge contours for the PVDF tactile sensor. The charge associated with each isocharge contour is measured in pico-Coulombs.

In order to form a criterion for the comparison between the experimental findings and the theoretical analysis data, and also to determine the magnitude of the stresses generated in the membrane, finite element modeling was utilized. The mathematical analysis was performed to predict how the PVDF-support assembly would behave theoretically when exposed to various loading conditions.

Figure 6 shows the finite element simulation results of the output charge for three typical angles vs. the distance from the sensing element of B. The experimental data are also shown as discrete points in the same figure. The finite element modeling results are in good agreement with those of the experimental findings.

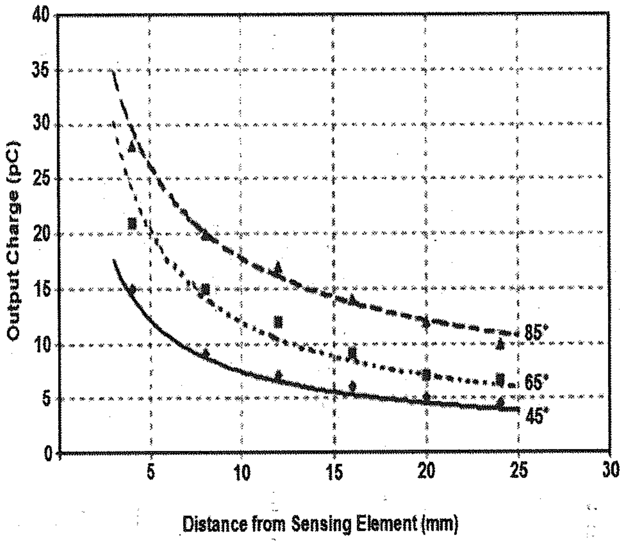

Figure 6: Comparison between the experimental output charges and the simulated results obtained by FEM (the solid and dashed-line curves are the finite element modeling outputs).

Potentially, the designed MTS can be incorporated into vaious medical probes for tactile imaging.

5. DESIGN AND MODELING A NEW TYPE OF TACTILE SENSOR BASED ON THE DEFORMATION OF AN ELASTIC MEMBRANE

In other study, we designed and modeled a flexible tactile sensor, capable of measuring contact-force and softness of a contact soft tissue/sensed object. The sensor was made of polymeric materials. This sensor can detect the 2D surface texture image, contact-force estimation, and softness of the sensed object. It consists of a chamber for pneumatic actuation and a membrane with a mesa structure as shown in Figure 7. The sensing mechanism is based on the novel contact deformation effect of a membrane. Determination of the contact-force and softness of sensed object is based on the amount and variations of out of plane deflection, at the center of a circular membrane as a result of applied force or pressure on it. The contact regions of the object are deformed according to the driving force of the mesa element and the softness of the object. Therefore, we can detect the softness of the object by measuring the relationship between the deformation of the membrane and the actuation force of it.

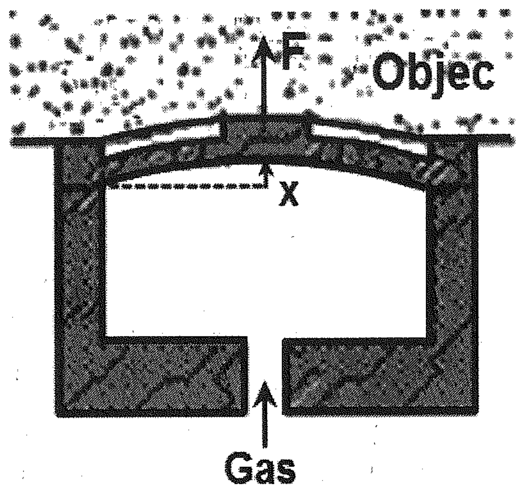

Figure 7: A schematic of the designed probe.

6. A NEW TACTILE PROBE FOR MEASURING THE MODULUS OF ELASTICITY OF BIOLOGICAL SOFT TISSUE

In this section, the design, fabrication, communication, testing, and simulation of a new tactile probe called Elastirob used to measure the modulus of elasticity of biological soft tissues and soft materials are discussed (Figure 8). Elastirob determines the elasticity by drawing the stress-strain curve and then calculating its slope. The combination of the force sensing resistor (FSR), microcontroller, and stepper motor provides Elastirob with the ability to apply different rates of strain on testing specimens.

Elastirob is accompanied by a tactile display called TacPlay. This display (Figure 9) is a custom-designed user-friendly interface and is able to evaluate the elasticity in each part of the stress-strain curve.

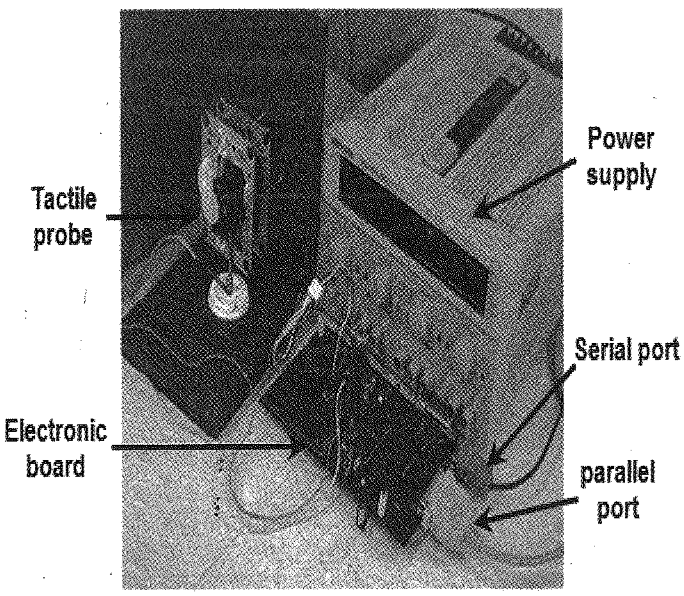

Figure 8: Elastirob and tactile data processing system.

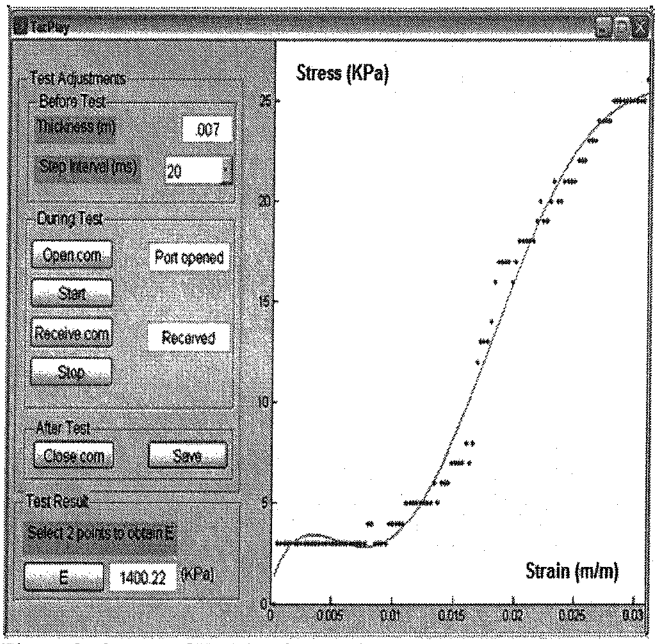

Figure 9: Output of TacPlay for specimen No.1.

The results of Elastirob applied on two specimens are reported and compared by two different methods; first, an industrial testing machine, and secondly, by a numerical approach of finite element method (FEM) used to simulate the compressing process. Acceptable validations of Elastirob obtained from the two comparison methods. The investigation of the effect of the strain rate on the results is also possible using Elastirob.

Following the above concept, the biomechanical properties of the skin and the subcutaneous tissues influence the transmission of mechanical vibration at different frequencies when touched by the designed sensor. As a result, the knowledge of the behavior of the skin under different loads is important in understanding the processes which involve tactile sensing such as minimally invasive surgery and designing tactile displays. As a typical system, we have proposed a finite element model to predict the response of fingertip subjected to a tactile stimulus. As most of the tactile displays are composed of arrays of vibrating pins, we have modeled the process of indentation of a typical pin as it interacts with the skin. A 2D finite element (FE) model of the fingertip is used for the analysis, as shown in Figure 10. The dimensions of the fingertip are assumed to be representative of the index finger of a male subject. The fingertip is assumed to be composed of a skin layer (representing epidermis and dermis), subcutaneous tissue, bone, and nail. The skin is assumed to have a thickness of 0.8 mm. The Young's module of the bone and nail were assumed to be 17.0 GPa and 170.0 MPa, respectively; while Poisson's ratio was assumed to be 0.30 for both. The skin and subcutaneous tissue were assumed to be hyperelastic and viscoelastic. The nail was fixed at its three upper nodes as in most experiments which fix the finger on a table for indentation test. A rectangular steel bar was modeled to deform the skin surface of the fingertip.

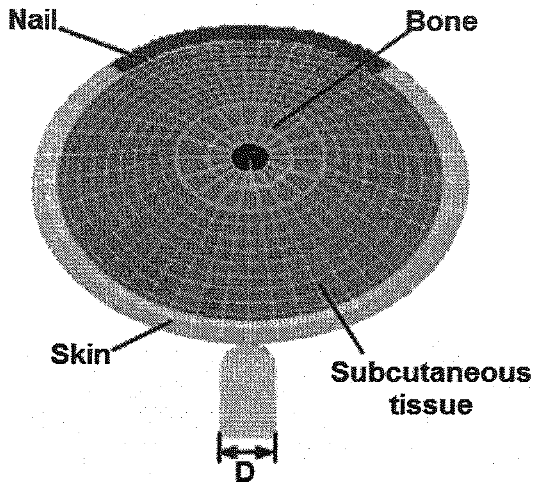

Figure 10: Finite element model of fingertip showing essential anatomical structures of a finger. The indenter diameter D was varied (0.1, 0.2, 2, and 4 mm) and its effect on stress and strain fields was studied.

Two sets of finite element analyses were performed to study the mechanics of tactile sensation of the fingertip. In the first set, the indenter was given a displacement of $h = 1mm$ in a time of $t = 1$s. If the skin on the fingerpad formed an infinite flat surface, then it would be sufficient to characterize the responses of afferents terminating in the region of the stimulus. However, the finger is a closed viscoelastic body with a curved surface. The results obtained from our model are plotted in Figure 11 which shows that the areas that are away from the indenter are also experiencing considerable amount of shear strain. There is a good correspondence between our results and the results of other research.

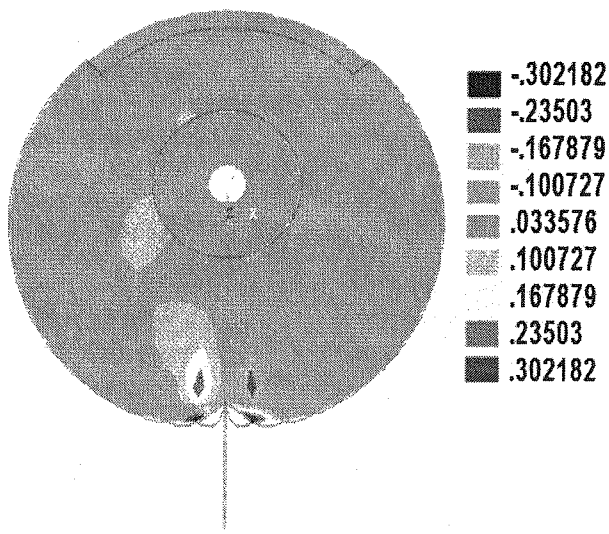

Figure 11: xy shear strain (e_{12}) at t = 1s with an indenter diameter of 0.1 mm.

It is known that the mechanoreceptors in the human finger sense the vibrations in the range 5-250 Hz. To verify this case, a second set of finite element analysis was performed in which the displacement of the indenter reached a level of 1mm at time 0.1s, remained in the same place for 0.4s and decreased to zero in 0.1s. Figure 12 shows the variation of strain energy density at different depths from the skin surface as a function of time in an indentation test. It is seen that when the indenter is stationary during t =0.1 to t =0.5, the strain energy density decreases.

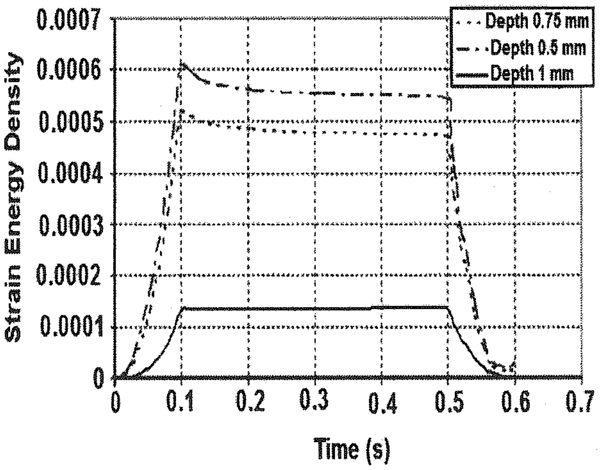

Figure 12: Strain energy density as a function of time for points located at depths of 0.5, 0.75, and 1 mm from the skin.

Since the mechanoreceptors in the human finger cannot sense the strain energy below a certain value, the frequency of indenter displacement cannot be reduced greatly, otherwise the mechanoreceptors will be adapted. This adaptation causes the output of these receptors to become zero and the stimuli could not be felt.

7. Application of Tactile Sensing to Determine Embedded Object Characteristics

Determining the existence of an alien object in the patient's body with the aid of tactile sensing is of great importance since it does not involve any invasive operation or penetration into the body. We propose a model shown in Figure 13 to investigate the effects of the tumor existence that appear on the surface of the tissue when compression is applied on the surface.

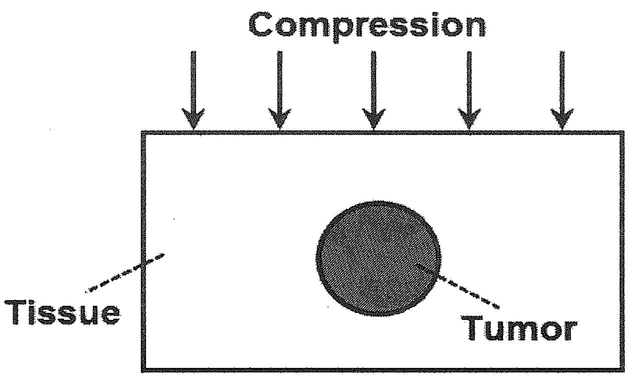

Figure 13: Transversal section (2D) of the 3D model.

Finite element analysis provides properties such as the shape and location of the tumor which are important parameters for physicians to distinguish the correct condition of the patients. Several different cases were created and solved by the ANSYS software (Release 10.0) and the following results were obtained:

- Appearing of the effects of an embedded object on the tissue surface.
- Determining the tumor location and shape with respect to the stress distribution produced on the tissue surface.

The appearance of the effects of the tumor on the surface of the tissue is the most fundamental result that confirms the accuracy and reliability of the artificial tactile method. Figure 14 shows the stress distribution when the solution is done and demonstrates that applying compression on the tissue, which has an embedded object, will cause a non-uniform stress distribution to be produced at the contact surface.

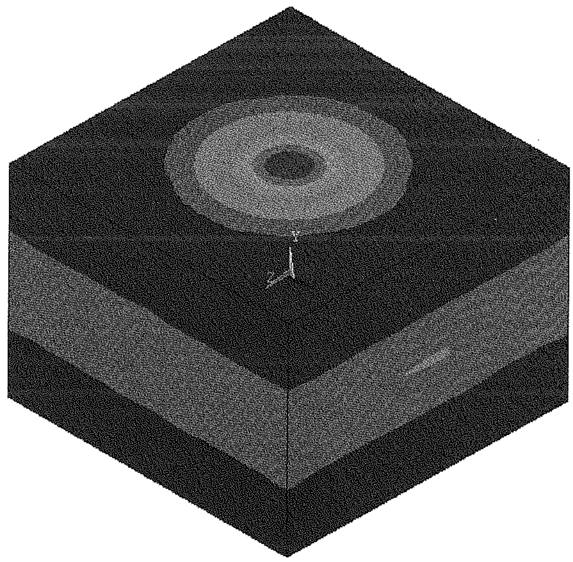

Figure 14: Stress distribution in the model.

In addition to predicting the existence of the embedded object, stress distribution produced on the surface can locate the tumor. The position of the maximum stress demonstrates the location of the tumor. Moreover, circularly distributed stress contours in Figure 14 states that the tumor is spherical.

In order to measure the stiffness of the embedded objects, we describe a new method. A tailor-made tactile probe equipped with a polyvinylidene fluoride-based (PVDF-based) piezoelectric sensor is used in the experimental tests. The structure of the probe shown in Figure 15 is such that it deforms in a specific way when pressed against a large object.

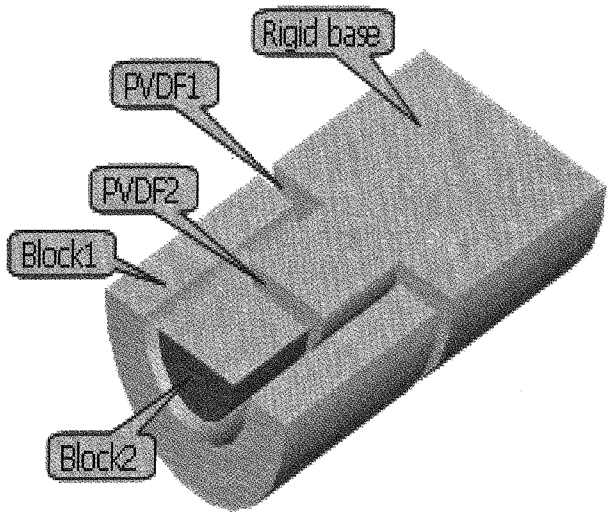

Figure 15: Schematic representation of sensor.

Two different elastic materials, in the form of two concentric cylinders, with different moduli of elasticity compose the major structure of the sensor assembly. Young's modulus of elasticity for the hidden object located inside a block is determined experimentally when the probe is applied to the outside of a rubber like matrix. This matrix simulates the human organs (such as breast). Figure 16 shows the assembly of the probe and tissue containing two different tumors.

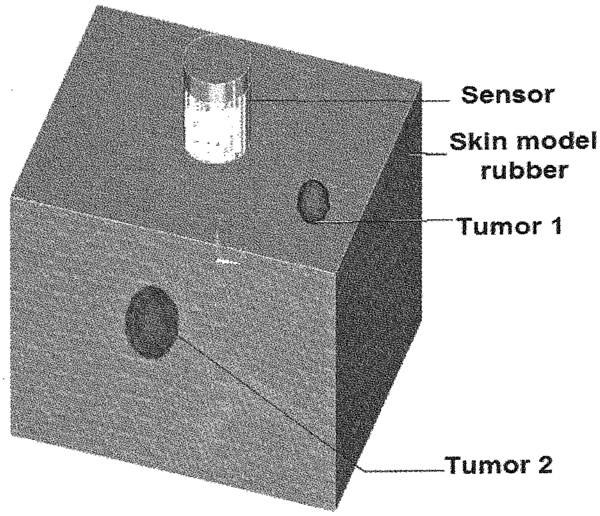

Figure 16: Model of block-probe assembly generated in ANSYS.

The probe, when scanning the surface of the tissue, receives deformation relating to the stiffness of the tissue. In the locations where there is a tumor beneath the surface, the probe senses different amount of deformation so that it can predict the stiffness of the tumor.

8. Conclusion

The present paper proposes different applications of tactile sensing which are useful in minimally invasive and robotic surgery. Design and fabrication of different artificial tactile sensor were suggested which can measure the compliance of the sensed object based on piezoelectric or piezoresistive characteristics. In one case, a typical polyvinylidene fluoride film was put between two different cylinders that transferred different amount of deformation to the PVDF film. In another sensor, a PVDF film was used as a supported membrane which can produce a three-dimensional map of the isocharge contours. The charge associated with each isocharge contour is measured in pico-coulombs. Applications of these sensors on surgical graspers showed the reliability of the artificial tactile sensing technology.

In addition to measuring the stiffness and compliance of the sensed objects, the accuracy of the proposed methods was demonstrated by detecting the presence and measuring the stiffness of the alien objects inside the simulated biological tissues.

9. Acknowledgments

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