Micromachined Tactile Sensor for Soft-Tissue Compliance Detection

Ahmed M. R. Fath El Bab, Koji Sugano, Toshiyuki Tsuchiya, Member, IEEE, Osamu Tabata, Senior Member, IEEE, Mohamed E. H. Eltaib, and Mohamed M. Sallam, Member, ASME

Abstract—Compliance detection becomes very essential in minimally invasive surgery (MIS). It can help in detection of cancerous lumps and/or for deciding on tissue healthiness. In this paper, a micromachined piezoresistive tactile sensor, with two serpentine springs and 500-μm cubic mesas, has been designed for detecting the compliance of soft tissue independent of the applied distance between the sensor and the tissue. The measuring range of the sensor is chosen to be associated with the soft-tissue properties. The sensor parameters are optimized to give high sensitivity and linearity of the sensor output. The design is simulated using ANSYS for checking the sensor performance. Then, the sensor is fabricated and tested by three types of specimens, namely, specimen chips with known stiffness, silicone rubber specimens, and chicken organs specimens (leg and heart). For the specimens chips fabricated and tested by three types of specimens, namely, specimen chips with known stiffness, silicone rubber specimens, and chicken organs specimens (leg and heart), the sensor distinguished between different stiffnesses independent of the applied displacement in the range of 50–200 μm. The sensor measured Young’s modulus up to 808 kPa with an average error of ±7.25%. For the chicken leg and heart, the sensor distinguished between them under the applied displacement from 100 to 200 μm, and they were calculated as 12 ± 1 kPa and 81 ± 8 kPa, respectively. [2011-0213]

Index Terms—Compliance detection, micromachining, soft tissue, tactile sensor.

I. INTRODUCTION

TODAY, MUCH attention is paid to tactile sensing in medical applications such as minimally invasive surgery (MIS) [1], [2]. In MIS, the surgeon operates through small openings (between 3 and 12 mm) in the abdominal wall of the patient. One of the openings is used to introduce a miniature camera (including a cold light source). Camera images are shown on a monitor; the camera is manipulated by an expert operator. A head physician and his assistants operate using a set of elongated slim rigid tools provided by an appropriate utensil on the tip. The tip is manipulated using the handle of the tools. The transmission of force and motion from the handle to the tip is actuated by means of levers. In such surgery, it is of concern that much of the tactile information available in open surgery will be lost. Compliance (reciprocal of stiffness) detection of soft tissue can help the surgeon in decision making in MIS. Moreover, it could be used for detecting cancerous lumps or for determining the health of a tissue [1].

For compliance detection, many concepts have been introduced. Carter et al. [3] have developed a probe for tissue characterization such as stiffness based on sensing the static force applied to the tissue and the corresponding tissue deformation/deflection. The tissue stiffness can be determined by simply dividing the applied force by the measured tissue deflection. Based on the same concept, Husegawa et al. [4] proposed a multifunctional tactile sensor device driven by a magnetic force. The device has the advantage of the ability to detect multiple physical values, such as contact force and the elastic and damping coefficients of an object. The same team previously used an external pneumatic system for actuating a diaphragm of microelectromechanical systems sensor [5].

The limitation of the previously referenced approaches is that the actuation force should be controlled to be constant in order to differentiate between different tissues by measuring the equivalent displacement. The extra device necessary to apply the constant force makes the sensor more expensive. Also, a reference point to measure the displacement may be needed in some designs, which makes a fixed frame essential for using it as a reference to measure the equivalent displacement. These make such a concept of limited use as a sensor attached to a probe freely held in a hand for medical applications.

Omata et al. [6]–[8] utilized an approach for the elasticity detection of soft tissue that involves the use of a piezoelectric ceramic as a transducer. The transducer applies vibration at its resonant frequency. When the free end of the probe touches a material, the resonant frequency shifts. The shift in resonant frequency depends on the elasticity of the material. The main limitation of this sensor is that the sensor output is varying with changing the contact pressure between the sensor and the tissue. This causes controversial results, as the sensor may give
different readings for the same tissue if the applied pressure changes between the sensor and the tissue.

Engel et al. [9] developed a micromachined polyimide stiffness sensor that does not rely on applying known actuation force and is based on signals from two sensors with considerably different stiffnesses. Dargahi et al. [10]–[12] used the same concept to develop piezoelectric sensors, capable of measuring the total applied force on the sensed object as well as its compliance. The sensors generally consist of rigid and compliant elements. Determination of the compliance of sensed objects is based on the ratio of force experienced by the rigid element to the total force applied to the sensor. Based on the same concept, Peng et al. [13] presented a polydimethylsiloxane capacitive tactile sensor for tissue elasticity measurement. Although the theoretical model implies that the reading of the sensor does not depend on the contact force/displacement between the sensor and the measured object, the experiment results showed that it is highly dependent on it. Although the sensor is designed for detecting the soft-tissue elasticity, the measurement range of soft-tissue elasticity was not taken into account in designing the aforementioned sensors.

Based on the same concept of “applying two springs with considerably different stiffnesses to soft tissue for compliance detection,” Fath El-Bab et al. [14] developed a detailed design procedure, with taking into account the measurement range of soft tissue, for optimizing the sensor design parameters to give high sensitivity and linearity of the sensor output in order to get output reading independent of the pushing distance between the sensor and the tissue. The design assumed a relatively flat tissue surface relative to the size of the contact mesa. The separation distance between the two mesas is selected to be far enough, so the deformations at each mesa minimally affect the other one. A finite-element (FE) model is developed to investigate the sensor performance with the designed parameters. The results showed that the sensor can distinguish different tissue stiffnesses independent of the pushing distance of the sensor.

In this work, based on a previously proposed design procedure [14], a micromachined piezoresistive-type tactile sensor with serpentine spring structures is developed. The sensor mathematical model, sensor parameters, structure design, fabrication, and simulation using ANSYS for checking the sensor performance are covered in Section II. Finally, in Section III, the fabricated sensor is tested for monitoring the sensor ability of detecting object compliance, independently of the applied displacement between the sensor and the object.

II. SENSOR DESIGN

A. Sensor Model

The sensor is modeled as two springs, as shown in Fig. 1, namely, \( S_l \) (low-stiffness spring) and \( S_h \) (high-stiffness spring). The low- and high-stiffness springs have stiffness constants of \( k_l \) and \( k_h \), respectively. The soft tissue has stiffness of \( k_o \). When the two springs contact the soft tissue, two different forces are generated on the high- and low-stiffness sides, of values \( F_H \) and \( F_L \), respectively. By measuring \( F_H \) and \( F_L \), the soft-tissue stiffness \( k_o \) can be expressed as a function of \( Q \), as shown in the following [15]:

\[
k_o = \frac{k_h k_l (1 - Q)}{(Q k_l - k_h)}
\] (1)

where \( Q = \frac{F_H}{F_L} \) is a dimensionless variable. The sensor output is represented by \( Q \) as the tissue stiffness, and \( k_o \) can be estimated from \( Q \) using (1).

B. Sensor Parameter Selection

In this section, the sensor parameters are selected based on the design procedure [13]. As shown in Fig. 1, the sensor parameters are \( k_l \), \( k_h \), and \( S \), i.e., the low- and high-stiffness springs and the separation distance between them, respectively. Another parameter is \( Q_{\text{max}} \) (the \( Q \) value at the end of the measuring range), as shown in Fig. 2. Those parameters should be selected based on the sensor measurement range. Therefore, the first step of this procedure is to define the measurement range of the target object Young modulus. In the case of soft tissue, 1 MPa is selected as the upper limit of the soft-tissue Young modulus [14]. Since the sensor will measure the tissue stiffness (in newtons per meter) not the tissue Young modulus (in newtons per square meter), the sensor measurement range in terms of stiffness should be determined. Based on the fact that stiffness of tissue depends on not only the sensor mesa size but also the geometry of the tissue, Hayes et al. [16] presented a mathematical solution to the elasticity problem of indentation test as follows:

\[
E = \frac{(1 - \nu^2) k_o}{2a C_k}
\] (2)

where \( k_o \) and \( \nu \) are the stiffness and Poisson ratio of the tissue, respectively, and \( (2a) \) is the sensor mesa size (500 \( \mu \)m). \( C_k \) is a scaling factor depending on the Poisson ratio, deflection, and height of the tissue.

By assuming \( \nu = 0.5 \) (incompressible material) and setting \( C_k \) to one in the case of low tissue deflection [17], using (2), the
measurement range can be estimated as 666.66 N/m. Finally, the stiffness range from 0 to 800 N/m is chosen as a round figure.

The sensor parameter selection process is based on the following considerations.

1) The relationship between $Q$ and $k_o$ should be as linear as possible during the working range, in order to keep constant sensitivity along the measuring range. As shown in Fig. 2, the $Q-k_o$ relationship is originally nonlinear. By proper selection of $k_l$ and $k_h$, the nonlinearity ($n/N$), as defined in Fig. 2, can be minimized.

2) The sensitivity ($dQ/dk_o$), which is directly controlled by $Q_{\text{max}}$, should be as large as possible to realize high calculating accuracy when obtaining the object stiffness $k_o$.

3) The distance between the high- and low-stiffness springs $S$ should be as small as possible to increase the sensor spatial resolution.

4) The difference between tissue deflections at the two springs should not be very large to avoid the influence of one sensor mesa on the other due to the tissue deformation, which affects the reading of the object stiffness $k_o$.

$k_l$ and $k_h$ are chosen to be 170 and 5700 N/m, respectively, to give sensitivity ($Q_{\text{max}} = 5$) and a nonlinearity ratio of less than 0.05. Please note that, according to the design procedure [14], increasing the $Q_{\text{max}}$ to more than five (for increasing the sensitivity) and decreasing the $S$ distance to less than 1 mm (for increasing the spatial resolution) will dramatically increase the deflection difference between the two mesas and, subsequently, the sensor error.

After selecting the sensor parameters, the sensor structure will be designed to provide two linear springs with the aforementioned constant values.

C. Sensor Structure Design

In this section, the sensor shape and dimensions are selected to provide linear spring constant. Fig. 3 shows the FE model of the low-stiffness spring $S_l$. In this model, the $S_l$ is designed as serpentine structure with a thickness of 50 μm. The two ends of the serpentine are fixed as boundary condition. Pressure is applied to the top of the mesa. The force on the top of the mesa can be determined by multiplying the pressure by the mesa top area. The spring displacement is determined from the FE results.

Fig. 4 shows that the selected dimension of $S_l$ provided a linear force–displacement relationship with $k_l = 173$ N/m, which is close to the desired value of 170 N/m.

The same analysis is done with $S_h$. Fig. 5 shows that the selected dimension of $S_h$ provided a linear force–displacement relationship with $k_h = 5724$ N/m, which is close to the desired value of 5700 N/m.
Because this sensor will work based on the concept of piezoresistor, it is important to study the relationship between the spring displacement and the stress at the maximum stress location (location of the piezoresistor, as shown in Fig. 3).

Figs. 6 and 7 show that the stress–displacement relationships of $S_l$ and $S_h$, respectively, are linear. This means that, experimentally, the force–displacement relationship of each spring is expected to be linear too, as the force at each spring mesa will be detected by the stress induced in the spring at the piezoresistor location (shown in Fig. 3). Subsequently, the maximum deflection of $S_h$ is not more than 80 $\mu$m, because $S_h$ is much harder than $S_l$ and the expected force on $S_h$ is five times larger than that on $S_l$ ($Q = 5$). The stress–deflection relationship shows that, in the worst case, the stress in the sensor is limited to 3.5 GPa. For silicon, the fracture stress is about 7 GPa [18].

Finally, the sensing chip (5 mm $\times$ 6 mm), which is shown in Fig. 8, is consisting of low-stiffness spring $S_l$ and high-stiffness spring $S_h$ with the desired values of $k_l$ and $k_h$. The separation distance ($S$) between the two mesas is selected to be 3 mm to avoid any influence of one sensor mesa on the other due to the tissue deformation, which affects the reading of the object stiffness. Piezoresistors were placed in the maximum stress area to detect the spring deflection, as shown in Fig. 3. A half Wheatstone bridge measurement circuit is utilized for each piezoresistor to detect the force signal.

For the purpose of testing the sensor, a specimen chip shown in Fig. 9 was also prepared. It consists of two identical springs with known spring constant. Each spring’s shape and dimensions are similar to those of $S_l$ of the sensor.

D. Fabrication Process Design

The fabrication process shown in Fig. 10 is explained as follows.

1) The starting material is a (100)-oriented double-side-polished 4-in n-type silicon wafer. First, a silicon oxide ($SiO_2$) layer of 1-$\mu$m thickness was formed by thermal oxidation. $SiO_2$ was patterned to define the size and location of the piezoresistors (sensing element). A $SiO_2$ layer of 40-nm thickness was formed by thermal oxidation for protecting the silicon surface during boron ion implantation. Then, boron was implanted and diffused to form p-type silicon.

2) Contact holes were patterned, and $SiO_2$ was etched by hydrofluoric acid (HF). Then, an aluminum (Al) layer of 1-$\mu$m thickness was deposited and patterned by wet etching to form the electronic lead pattern on the chip.

3) An Al layer of 100 nm was deposited on the front side as masking material for the subsequent dry etching. Then, the Al and underlying $SiO_2$ layers were patterned to form the serpentine structure on the top surface (front side). The back-side $SiO_2$ was patterned to form the backside opening with protecting the front side by photoresist.

4) The back-side Si was etched by tetramethylammonium hydroxide 25% (80 $^\circ$C) while the front side was protected.
by wax. The structure thickness was adjusted to 50 μm by controlling the etching time.

5) A cubic mesa of size of 500 μm was glued to the top of the structure by epoxy glue. Then, the serpentine structure was generated in the Si layer by deep reactive-ion etching. Finally, the Al layer was removed by wet etching.

The fabrication of the specimen chips is the same as that of the sensor chip, as shown in Fig. 10, except that, after process number 4), the wafer is diced into chips. Then, the specimen chip thickness is adjusted one by one by controlling the etching time. Therefore, different specimens with different thicknesses can be prepared. Finally, fabrication process number 5) is carried out.

E. FE Analysis

In this section, a simulation of the sensor output using ANSYS is carried out to investigate the sensor performance with the designed \( k_l \) and \( k_h \) values. The soft tissue is assumed to be homogeneous, isotropic, and adhered to a rigid bony base. An axisymmetric FE model was established, as shown in Fig. 11, by meshing the tissue into solid eight-node elements. The tested material is assumed to be linearly elastic with Poisson’s ratio of 0.49, dimensions of \( 10 \times 10 \times 10 \) mm\(^3\), and Young’s moduli of 0.25, 0.5, 0.75, 1, and 1.25 MPa to represent tissue stiffnesses \( k_o \) of 166, 333, 500, 666, and 833 N/m, respectively. Large deformation was taken into consideration in the analysis by choosing nonlinear geometry and applying the load incrementally. The sensor is presented by two serpentine springs with a thickness of 50 μm. The sensor reading is the force ratio \( (Q) \) on the high- and the low-stiffness spring mesas.

As shown in Fig. 12, the sensor reading possesses stability in reading during pushing against the tissue. The relationship between the tissue stiffness and the sensor output is shown in Fig. 13, which is linear as it is previously designed for.
III. SENSOR CHARACTERIZATION AND DISCUSSIONS

Since the sensor output \((Q)\) is calculated from the force ratio on \(S_l\) and \(S_h\) when the sensor contacts an object to be measured, \(S_l\) and \(S_h\) should be calibrated separately as two force sensors.

The setup for carrying out this calibration is shown in Fig. 14. The sensor is placed on a sensitive digital weight scale \((\text{resolution}=0.0001 \text{ g})\) for measuring the applied force to the sensor. The force is applied to the mesa top by a needle attached to the end of a lever. The lever motion is controlled by an \(x\)-\(y\)-\(z\)-axis precise table (resolutions are equal to \(10 \mu\text{m}\) in the \(x\) and \(y\)-directions and \(1 \mu\text{m}\) in the \(z\)-direction). Forces are applied to the spring tip (mesa) by moving the lever in the \(z\)-direction. The force value was determined by the weight scale reading. Then, the corresponding output voltage is determined by a digital multimeter (Agilent-34410A).

Figs. 15 and 16 show the output voltages of \(S_l\) and \(S_h\), respectively. From Figs. 15 and 16, the sensitivities of \(S_l\) and \(S_h\) were determined to be 51.67 and 9.743 mV/gf, respectively.

Figs. 15 and 16 show also that the sensor’s springs are possessing a linear load–voltage relationship which implies that the sensor’s springs \((S_l\) and \(S_h\) are linear springs. This point is very important for keeping the sensor mathematical model, as described in Fig. 1 and (1), as the model basically depends on the fact that the sensor’s springs \((S_l\) and \(S_h)\) are linear.

When object stiffness is measured, the sensor touches the object; then, \(S_l\) and \(S_h\) will deflect according to the object stiffness, and the deflection is detected as the output from the Wheatstone bridge circuit. By dividing the output voltage by the sensitivity, which is previously determined from Figs. 15 and 16, the generated force on \(S_l\) and \(S_h\) can be determined, and subsequently, the force ratio \((Q)\) was obtained, which is proportional to the measured object stiffness [15].

After calibrating \(S_l\) and \(S_h\) as force sensors, the sensor performance as a compliance sensor was examined using three types of specimens.

A. Specimen Chips

The sensor was tested using specimen chips, as shown in Fig. 9. By controlling the specimen spring thickness, different stiffnesses were prepared. The setup for carrying out this process is shown in Fig. 17. In this setup, the specimen was laid on a fixed table. The sensor was stuck to a glass slide and put upside down to touch the specimen for detecting its stiffness. The glass slide can move up and down in the \(z\)-axis (vertical) direction by a movable table. The forces generated on \(S_l\) and \(S_h\) were measured, and subsequently, the sensor output \(Q\) was calculated.

Figs. 18–20 show the force-versus-displacement characteristics on the low-stiffness springs \((F_L)\), the high-stiffness springs \((F_H)\), and the corresponding sensor output \((Q = F_H/F_L)\), when detecting specimen chips of thicknesses of 41, 44, and 51 \(\mu\text{m}\), respectively.

Although the contact forces \(F_L\) and \(F_H\) are increasing with increasing the applied displacement between the sensor and the specimen, the sensor output \(Q\) (force ratio) remains constant with the applied displacement (independent of the applied displacement).
The reason why the sensor output $Q$ shows dependence on the applied displacement in the range from 0 to 50 $\mu$m is clarified here. In the beginning of contact, one of the sensor mesas (tip) may touch the specimen chip’s mesa before the other one. This happens because the heights of the epoxy glue layer are not the same in the low- and high-stiffness sides of the sensor, as the mesas were glued one by one. This difference in the height is measured by a light microscope with a micrometer to be in the order of a few micrometers. Because the step of applied displacement is 10 $\mu$m, any difference in the mesa height between $S_l$ and $S_h$ smaller than 10 $\mu$m cannot be detected during the applied displacement process. In this paper, the force necessary to deflect the spring of the higher mesa to make it in the same height of the other one is termed as the error force. The magnitude of this error force is proportional to the difference in the mesa height and also to the stiffness of the specimen. Please note that, if the sensor chip is not parallel to the object surface and/or the object surface is not flat, this error force is induced.

In the range from 0 to 50 $\mu$m, $F_H$ and $F_L$ are relatively small compared to the error force. Thus, calculating $Q$ from $F_H$ and $F_L$ shows a large error. Due to increasing of the applied displacement from 50 to 200 $\mu$m, $F_H$ and $F_L$ increased and become high enough to overcome the error force, and the $Q$ value gets independent of the applied displacement.

The sensor output for specimens of thicknesses of 41, 44, and 51 $\mu$m is shown in Fig. 21. This figure shows not only that the sensor gives output reading independent of the applied displacement but also that it can distinguish between different specimen stiffnesses. The specimens of thicknesses of 41, 44, and 51 $\mu$m are equivalent to spring stiffnesses of 94, 116, and 180 N/m, respectively, which were evaluated by FE analysis. The sensor output provided mean readings of $1.89 \pm 0.094$, $2.55 \pm 0.126$, and $3.53 \pm 0.147$ for the specimen chips with stiffnesses of 94, 116, and 180 N/m, respectively. The error is determined here based on the maximum upper and lower deviations in the sensor reading. This error is not exceeding 4.97% in the worst case.
The sensor average output versus the specimen chip stiffness is shown in Fig. 22. The figure shows that the sensor has good linearity as it is designed for. Four output readings are recorded in Fig. 22; three of them are experimental (which are mentioned previously), and one is theoretical. The theoretical output reading \( Q \) for the zero-stiffness specimen is one as calculated from (1).

The sensor sensitivity \( \frac{dQ}{dk_o} \) determined from Fig. 22 is about 0.014. This is better than the simulation sensitivity (0.005), as shown in Fig. 13. This is because the sensor thickness may be reduced to less than 50 \( \mu \)m during the fabrication process, which increases its sensitivity. Also, there is uncertainty in the measurement of the silicon chip specimen thickness as well due to the accuracy of the etching process at which the thickness is controlled by time. Then, the thickness is checked by an optical microscope with an ocular micrometer of 1-\( \mu \)m resolution.

Fig. 22 is considered as a calibration curve of the proposed stiffness sensor. In other words, by using Fig. 22, the stiffness of a specimen can be determined by knowing the sensor output \( Q \) of it.

### B. Silicone Rubber Specimens

Next, the sensor is tested using silicone rubber (TSE 3331K EX) specimens. The specimen stiffness was controlled by changing the mixing ratio of its two components A and B. Since the B component is the hardener, the higher the B component, the harder the specimen. According to the silicone rubber (TSE 3331K EX) data sheet, the mixing ratio of A:B = 1:1 is resulting in a rubber of Young’s modulus equal to 3 MPa. Four silicone rubber stiffnesses were created with mixing ratios A:B of 1:0.75, 1:0.5, 1:0.25, and 1:0.1, to get rubber softness near to soft-tissue softness. The rubber was cut into approximately cubic shape of 1-cm size and tested by the same setup shown in Fig. 17, except that the specimen chip is replaced by the rubber cube, as shown in Fig. 23.

Figs. 24 and 25 show the sensor output for the rubber specimens. The sensor can distinguish the stiffness difference between the four specimens and shows reading independent of the applied displacement in the range from 50 to 200 \( \mu \)m.

In the range of applied displacement from 0 to 50 \( \mu \)m, an error force exists due to the difference in the mesa height of the sensor as well as the nonflatness of the silicone rubber surface.

| Mixing Ratio | Equivalent Stiffness | Sensor Output |
|--------------|----------------------|---------------|
| A:B=1:0.1    | 58 ±4.3              |
| A:B=1:0.25   | 360 ±15              |
| A:B=1:0.5    | 505 ±35              |
| A:B=1:0.75   | 538 ±39              |

In the range of applied displacement from 50 to 200 \( \mu \)m, \( F_H \) and \( F_L \) are high enough to overcome the error force and got sensor mean readings of 1.75 ±0.13, 5.66 ±0.234, 7.6 ±0.522, and 8 ±0.577 for specimens of mixing ratios A: B of 1:0.1, 1:0.25, 1:0.5, and 1:0.75, respectively, as shown in Fig. 15. The stiffness of the four silicone rubber specimens was calculated by using Fig. 22 as a calibration curve. The equivalent stiffness of each rubber specimen is shown in Table I.
C. Soft-Tissue Specimens

Finally, the sensor is tested using two tissue specimens with different stiffnesses, namely, chicken leg and heart. They were cut into approximately cubic shape of 1-cm size and tested by the same setup shown in Fig. 17, except that the specimen chip is replaced by one of the aforementioned soft-tissue cubes (similar to what have been done with the silicone rubber specimen, as shown in Fig. 23).

Figs. 26 and 27 show the sensor output for the chicken leg and heart specimens, respectively. In the case of the chicken heart specimen, the sensor showed big error in the range from 0 to 100 μm. Then, from 100 to 200 μm, the output starts to be steady with a mean value of 1.17 ± 0.126.

For the chicken leg, the sensor showed steady output with a mean value of 1.1 ± 0.093 from the applied displacement of 10 μm.

The reason why the chicken heart did not give steady reading (independent of the applied displacement) is also attributed to the same reason described before. The error force in the beginning of contact is large due to the corrugated surface of the tissue, which causes one of the sensor mesas to touch the tissue before the other one (it is very difficult to cut a tissue in 1-cm cube with flat surfaces). Due to relatively low stiffness of the soft tissues (the sensor output readings are 1.7 for the heart and as high as 8 for the silicone rubber), the generated \( F_H \) and \( F_L \) are low and, subsequently, cannot overcome the error force.

In the case of the chicken leg specimen, the results show stable reading from the beginning of the applied distance from 10 μm until the end of applied displacement at 200 μm, which is dissimilar to the behavior of the rest of specimens’ results. This happens because the two mesas of \( S_l \) and \( S_h \) touched the object surface simultaneously by chance. Therefore, the error force was near zero, and the output is proportional to the specimen stiffness only.

The stiffness of the chicken leg and heart was determined by using Fig. 22. The equivalent stiffness of the chicken leg and heart is shown in Table II.

After determining the stiffness of the silicone rubber and the soft-tissue specimens, the Young modulus \( (E) \) was calculated by (2). Table III shows the Young modulus of the rubber and the soft-tissue specimens.

As shown in Table III, the sensor can measure objects of elasticity up to 808 kPa. The error percentage is around 7.25% (the highest and the lowest values are removed during calculating the average). This error is limiting the resolution of the sensor. Generally, the resolution of a sensor can be defined as the smallest change in the measured quantity that can be detected by the sensor. Because the sensor should measure the object elasticity independent of the applied displacement, in this paper, the sensor resolution may be determined by summing the ±errors during pushing the sensor against the object. Thus, the sensor resolution can be 14.5% of the sensor reading. Therefore, it is very important to keep the reading as much as possible independent of the pushing distance between the sensor and the tissue to get higher resolution.

### IV. Conclusion

Based on the concept of applying two springs, with considerably different stiffnesses to soft tissue for compliance detection, a micromachined piezoresistive-type tactile sensor has been designed. The sensor parameters are optimized to give high sensitivity and linearity of the sensor output. The measuring range of the sensor is chosen to be associated with the soft-tissue properties \( (E \text{ up to } 1 \text{ MPa}) \). The design shows that, for 500-μm mesa size, the best \( k_l \) and \( k_h \) values are 170 and 5700 N/m, respectively. Then, a FE model is developed to
investigate the sensor performance with the designed $k_1$ and $k_h$ values using serpentine structure as spring. The soft tissue is simulated as linearly elastic material with Poisson’s ratio of 0.49 and Young’s modulus changing from 0.25 to 1.25 MPa to represent tissue stiffnesses $k_0$, 166, 333, 500, 666, and 833 N/m, respectively. The simulation results showed that the sensor can distinguish between different stiffnesses with stable reading during the pushing of the sensor into the tissue.

Then, the sensor is fabricated and tested. The testing process was carried out by three different ways.

1) By subjecting the sensing chip to another chip with known spring stiffness. The results showed that the sensor can distinguish different stiffnesses independent of the applied displacement in the range of 50–200 μm and gives mean output readings of 1.89, 2.54, and 3.53 for specimen stiffnesses of 94, 116, and 180 N/m, respectively.

2) By subjecting the sensing chip to silicone rubber (TSE 3331K EX) specimens. The stiffness of a silicone rubber specimen was controlled by changing the mixing ratio of its two components. The sensor can distinguish four different silicone rubber stiffnesses independent of the applied displacement in the range of 50–200 μm and gives mean output readings of 1.75, 5.66, 7.6, and 8 for specimens of mixing ratios A:B of 1:0.1, 1:0.25, 1:0.5, and 1:0.75, respectively. Based on Hayes’s formula, the sensor output was equivalent to Young’s moduli of 87, 538, 762, and 808 kPa for the silicone rubber specimens of mixing ratios A:B of 1:0.1, 1:0.25, 1:0.5, and 1:0.75, respectively.

3) Finally, the sensor is tested by two chicken organ specimens (leg and heart), for detecting their stiffnesses. The results show high dependence on the applied displacement in the range from 0 to 100 μm. Then, from 100 to 200 μm, the output starts to be steady with mean values of 1.1 and 1.7 for the chicken leg and heart, respectively. Based on Hayes’s formula, the sensor output was equivalent to Young’s moduli of 12 and 81 kPa for the chicken leg and heart, respectively. The sensor can measure tissue elasticity up to 808 kPa with an error percentage around ±7.25% of the measured elasticity.

The future design should take into account the following.

1) The sensor should be packaged to avoid wire failure and electric shock to a patient.
2) The sensor sensitivity should be increased for detecting soft-tissue stiffness with higher accuracy.
3) The sensor design should take into account the effect of the inclination angle between the sensor and the tissue for eliminating the effect of the corrugated surface of soft tissue to reduce the error and to increase the sensor resolution.

**ACKNOWLEDGMENT**

The authors would like to thank S. Washio, Technical Officer of Cybernet System Company, Ltd., for his assistance in using ANSYS.

**REFERENCES**

[1] M. E. H. Eltaib and J. R. Hewit, “Tactile sensing technology for minimal access surgery—A review,” *Mechatronics*, vol. 13, no. 10, pp. 1163–1177, Dec. 2003.

[2] J. Tegin and J. Wikander, “Tactile sensing in intelligent robotic manipulation—A review,” *Ind. Robot—Int. J.*, vol. 32, no. 1, pp. 64–70, 2005.

[3] J. Carter, T. G. Frank, P. J. Davies, D. MacLean, and A. Cuscheri, “Biomechanical testing of intra-abdominal soft tissues,” *Med. Image Anal.*, vol. 5, pp. 231–236, 2001.

[4] Y. Hasegawa, H. Sasaki, T. Ando, M. Shikida, K. Sato, and K. Itoigawa, “Multifunctional active tactile sensor using magnetic micro actuator,” in *Proc. 18th IEEE Int. Conf. MEMS*, 2005, pp. 275–278.

[5] M. Shikida, T. Shimizu, K. Sato, and K. Itoigawa, “Active tactile sensor for detecting contact force and hardness of an object,” *Sens. Actuators A, Phys.*, vol. 103, no. 1, pp. 18–22, 2003.

[6] S. Omata and Y. Terunuma, “New tactile sensor like human hand and its applications,” *Sens. Actuators A, Phys.*, vol. 35, no. 1, pp. 9–15, Oct. 1992.

[7] Y. Murayama and S. Omata, “Fabrication of micro tactile sensor for the measurement of micro-scale local elasticity,” *Sens. Actuators A, Phys.*, vol. 109, no. 3, pp. 202–207, Jan. 2004.

[8] Y. Murayama and S. Omata, “Considerations in the design and sensitivity optimization of the micro tactile sensor,” *IEEE Trans. Ultrason., Ferroelect., Freq. Control*, vol. 52, no. 3, pp. 434–438, Mar. 2005.

[9] J. Engel, J. Chen, Z. Fan, and C. Liu, “Polymer micromachined multi-modal tactile sensors,” *Sens. Actuators A, Phys.*, vol. 117, no. 1, pp. 50–61, 2005.

[10] R. Sedaghati, J. Dargahi, and H. Singh, “Design and modeling of an endoscopic piezoelectric tactile sensor,” *Int. J. Solids Struct.*, vol. 42, no. 21-22, pp. 5872–5886, Oct. 2005.

[11] J. Dargahi, S. Najarian, and B. Liu, “Sensitivity analysis of a novel tactile probe for measurement of tissue softness with applications in biomedical robotics,” *Mater. Process. Technol.*, vol. 183, no. 2/3, pp. 176–182, Mar. 2007.

[12] S. Sokhanvar, M. Packirisanmy, and J. Dargahi, “A multifunctional PVDF-based tactile sensor for minimally invasive surgery,” *Smart Mater. Struct.*, vol. 16, no. 4, pp. 989–998, Aug. 2007.

[13] P. Peng, R. Rajaman, and A. G. Erdman, “Flexible tactile sensor for tissue elasticity measurements,” *IEEE Trans. Ultrason.*, vol. 28, no. 1, pp. 102–107, 2001.

[14] A. M. R. Fath El Bab, T. Tamura, K. Sugano, T. Tsuchiya, O. Tabata, M. E. H. Eltaib, and M. M. Sallam, “Design and simulation of a tactile sensor for soft-tissue compliance detection,” *IEEE Trans. Sens. Microfluid.* vol. 16, no. 5, pp. 1262–1233, Dec. 2009.

[15] A. M. R. Fath El Bab, M. E. H. Eltaib, M. M. Sallam, and O. Tabata, “Tactile sensor for compliance detection,” *Sens. Mater.*, vol. 19, no. 3, pp. 165–177, 2007.

[16] W. C. Hayes, L. M. Keer, G. Herrmann, and L. F. Mockros, “A mathematical analysis for indentation test of articular cartilage,” *J. Biomech.*, vol. 1, no. 2, pp. 541–551, Sep. 1972.

[17] M. Zhang, Y. P. Zheng, and A. F. T. Mak, “Estimating the effective Young’s modulus of soft tissues from indentation tests–nonlinear finite element analysis of effects of friction and large deformation,” *Med. Eng. Phys.*, vol. 19, no. 6, pp. 512–517, Sep. 1997.

[18] N. Maluf and K. Williams, *An Introduction to Microelectromechanical Systems Engineering*. Norwood, MA: Artech House, 2004.

Ahmed M. R. Fath El Bab was born in Cairo, Egypt, in 1974. He received the M.Sc. and Ph.D. degrees from Assiut University, Assiut, Egypt, in 2002 and 2008, respectively. His Ph.D. work was in the field of micromachined tactile sensors for robotics and medical applications.

From October 2007 to October 2008, he was a Visiting Researcher at the Tabata Laboratory, Kyoto University, Kyoto, Japan. During this period, he gain experience in microfabrication experimentally, photomask design, and MEMS simulation using IntelliSuite software. Since January 2009, he has been a Lecturer (Assistant Professor) in the Department of Mechanical Engineering, Faculty of Engineering, Assiut University. His current interests include microsensors (principle, simulation, design, and fabrication), micromachining and its application in MEMS, tactile sensing systems (tactile sensing and display), micro energy harvesting devices, and kinematics and dynamics of robotic mechanisms.

Dr. Fath El Bab was a recipient of the Best Ph.D. Thesis Prize in Engineering Sciences from Assiut University in 2010.
Koji Sugano received the M.S. and Ph.D. degrees from Ritsumeikan University, Kusatsu, Japan, in 1998 and 2003, respectively. He studied silicon dry etching techniques during his dissertation work. Since 2003, he has been with the Department of Mechanical Engineering, Kyoto University, Kyoto, Japan, where he is currently an Assistant Professor. His current interests include micro-/nano-electromechanical systems, microfluidic systems, micro-/nanoparticle synthesis and assembly, and biosensor and chemical sensor systems. Dr. Sugano is a member of the Institute of Electrical Engineers of Japan.

Toshiyuki Tsuchiya (M’96) received the M.S. degree from The University of Tokyo, Tokyo, Japan, in 1993, and the Ph.D. degree from Nagoya University, Nagoya, Japan, in 2004. From 1993 to 2004, he was with the Toyota Central Research and Development Laboratories, Inc. In 2004, he joined Kyoto University, Kyoto, Japan, as an Associate Professor, where he is currently in the Department of Micro Engineering, Graduate School of Engineering. He is currently engaged in research on silicon surface micromachining, its application in MEMS, the mechanical property evaluation of micromaterials, and the reliability of MEMS devices. Dr. Tsuchiya is a member of the Materials Research Society, the Institute of Electrical Engineers of Japan, and The Japan Society of Mechanical Engineers.

Osamu Tabata (M’89–SM’01) was born in 1956. He received the M.S. and Ph.D. degrees from Nagoya Institute of Technology, Nagoya, Japan, in 1981 and 1993, respectively. In 1981, he joined the Toyota Central Research and Development Laboratory, Inc., Nagakute, Japan. In 1996, he joined the Department of Mechanical Engineering, Ritsumeikan University, Kusatsu, Japan. From September to December 2000, he was a Guest Professor at the Institute of Microsystem Technology, University of Freiburg, Freiburg, Germany. From January to March 2001, he was a Guest Professor at the Swiss Federal Institute of Technology (ETH) Zurich, Zurich, Switzerland. Since May 2010, he has been a Senior Research Fellow at the Freiburg Institute for Advanced Studies University of Freiburg. Since 2010, he had a visiting professorship for senior international scientists of the Chinese Academy of Sciences, Shenyang, China. From July 2011 to July 2014, he is a Guest Professor at Huazhong University of Science and Technology, Wuhan, China. In 2003, he joined the Department of Mechanical Engineering, Kyoto University, Kyoto, Japan, where he has been a Professor in the Department of Micro Engineering, Graduate School of Engineering, since April 2005. He is currently engaged in research on micro-/nano-process MEMS, and micro-/nanosystem synthetic engineering. At his new position, he started the research to realize a unique and novel nanosystem by assembling the various functional components such as a microchip, a particle, a microcapsule, deoxyribonucleic acid origami, and a cell, with sizes ranging from the nanometer to micrometer scale on a MEMS substrate of a few square millimeters. This technology is termed synthetic engineering for nanosystems (SENS), and experimental and theoretical research on the establishment of SENS is pursued. He is an Editorial Board Member of Sensors and Actuators. Dr. Tabata is a member of The Japan Society of Mechanical Engineers and Japan Institute of Electronics Packaging and a senior member of the Institute of Electrical Engineers of Japan. He is an Associate Editor of the JOURNAL OF MICROELECTROMECHANICAL SYSTEMS. Also, he is a program committee member of many international conferences. He was the recipient of the Science News Award for research in “Monolithic pressure-flow sensor” in 1987, the Presentation Paper Award for research in “Anisotropic etching of silicon in TMAH solutions” in 1992, the R&D 100 Award for research in “Thin film Young’s modulus measurement apparatus” in 1993 and the R&D 100 Award for research in “Thin film tensile tester” in 1998, the Best Poster Award of the 19th Sensor Symposium on Sensors, Micromachines, and Applied Systems for “Determination of optimal mask movement pattern for moving mask deep X-ray lithography” in 2002, and the Best Patent Award from Ritsumeikan University for “Material processing and its apparatus using X-ray lithography” in 2004.

Mohamed E. H. Eltaib was born in 1963. He received the M.Sc. degree from Assiut University, Assiut, Egypt, in 1987, and the Ph.D. degree in the area of tactile sensing for robotics and medical applications from the University of Dundee, Dundee, U.K., in 2001. From 1996 to 2001, he was a Visiting Scholar in the Department of Mechanical Engineering and Mechatronics, University of Dundee. Since December 2001, he has been an Assistant Professor in the Department of Mechanical Engineering, Faculty of Engineering, Assiut University. Since February 2008, he has been on sabbatical leave from Assiut University and is currently with the Mechanical Engineering Department, Qassim University, Qassim, Saudi Arabia. He is currently engaged in research in the fields of mechatronics, micro tactile sensors for robotics and medical applications, haptic displays for virtual simulators, adaptive neuro-fuzzy inference systems, smart piezoelectric actuators, and piezoelectric-based energy harvesting.

Mohamed M. Sallam was born in 1943. He received the M.Sc. degree from Assiut University, Assiut, Egypt, in 1968, and the M.S. and Ph.D. degrees from Kansas State University, Manhattan, in 1969 and 1972, respectively. In 1973, he joined the Department of Mechanical Engineering, Assiut University, as an Assistant Professor, where he was an Associate Professor from 1977 to 1981, was the Head of the department from August 1991 to August 1997, and has been a Professor since 1982. During the period of September 1980–May 1984, he was on a leave of absence from Assiut University and was with King Faisal University, Al-Ahsa, Saudi Arabia. His fields of research include system dynamics, kinematics and synthesis of mechanisms, mechanical vibrations, stress analysis, optimization, machine design, and computer-aided design and manufacturing. Dr. Sallam is a member of the National Committee of Theoretical and Applied Mechanics, Academy of Science and Technology, Cairo, Egypt, and of the Scientific National Committee for Promotion of Professors in the Fields of Mechanical Design, Production Engineering, System Dynamics, and Industrial Engineering. He is a member of Sigma Xi, Pi Mu Epsilon, and the American Society of Mechanical Engineers.