Real-time tactile sensory feedback during minimally invasive surgery

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REAL-TIME TACTILE SENSORY FEEDBACK DURING MINIMALLY INVASIVE SURGERY

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**SCHOOL OF MECHANICAL AND AEROSPACE
ENGINEERING**

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SUMMARY

Minimally Invasive Surgery (MIS) is now routinely used as the preferred choice for many operations. MIS, due to surgeons' indirect interaction with operation site, severely reduces the surgeon's visual and tactile sensory perception during manipulation. Though visual sensing for MIS is well-established, it still suffers from lack of a tactile sensing. The two most important aspects of a tactile sensing system for MIS application are force and stiffness information existing in the sensor-tissue interface. This information is necessary for safe handling and safe grasping of tissues.

In this study, a novel tactile sensing concept is proposed to measure both the tool-tissue interaction forces and tissue stiffness. The sensor is a square array of force sensing elements with sets of two different stiffness ranges and arranged alternately. A pair of two force sensing elements is referred to as a stiffness sensing unit. When sensor array comes in contact with an object, each group of force sensors would experience a different deflection. This difference in deflection is used for stiffness evaluation of the object. The sensor array is proposed to be a polymer based piezo-resistive sensing system.

A lumped model and FEM simulation was formulated at first, in order to investigate the design concept and its performance. Results of both approaches showed that higher stiffness difference between force sensing elements provides higher sensitivity in the evaluation of object stiffness. The design of MEMS force sensing elements was performed using ANSYS. Considering Polyimide and SU-8 polymer as structural base material, strain distribution across the sensor diaphragm and its thickness was first investigated along with the influence of mesa size in this distribution. The force-deflection relationship for the sensing diaphragm, with a span size in the range of 300~700 μm and diaphragm thickness in the range of 20~50 μm , was obtained. The results showed a linear relationship for the strain up to the yield strain of the polymers. For a given diaphragm thickness, the required stiffness of the sensor can be tuned by changing diaphragm's span size. Force capacity for the diaphragm can be supported by the above proposed dimensions.

For benchmarking of sensing principle, a macro tactile-sensor array was designed and tested. The results for the different stiffness combinations of sensing elements showed an inverse linear relationship between the stiffness of object and the output of stiffness sensing unit. This substantiated the proposed design concept for the tactile sensors for their capability in evaluating different stiffness of test objects.

Finally, the feasibility of fabricating a MEMS device based upon the proposed concept was conducted using existing MEMS facilities at NTU. Results indicated that the MEMS based tactile array using SU8 polymer as the diaphragm material, can be fabricated using the available fabrication techniques.

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List of Symbols

a_1, a_2	Measuring arms of the flexural mechanism (mm)
Δf	Shift in resonant frequency
ΔC	Capacitance change
C_0	Initial capacitance
Δr	Deflection in the <i>Reference</i> sensing element
Δm	Deflection in the <i>Measuring</i> sensing element
δr	Deflection in interface of the <i>Reference</i> element with tissue
δm	Deflection in interface of the <i>Measuring</i> element with tissue
K	Stiffness of the macro sensing element (N/mm)
K_m	Stiffness of the <i>Measuring</i> sensing element (N/mm)
K_r	Stiffness of the <i>Reference</i> sensing element (N/mm)
K_{obj}	Stiffness of the object (N/mm)
K_v	output signal change respect to applied deflection ($V/\mu m$)
R	Ratio of the <i>Measuring</i> element over the <i>Reference</i> element deflection
V_0	Initial signal value in the macro sensing element (V)
V_{drive}	Drive voltage

List of Abbreviations

2D	Two dimensional
3D	Three dimensional
DAQ	Data Acquisition Device
DC	Direct Current
DRIE	Deep Reactive Ion Etching
FEM	Finite Element Method
FFC	Flexible flat cables
MEMS	Micro Electro Mechanical Systems
MIS	Minimally Invasive Surgery
MOSFET	Metal Oxide Semiconductor Field Effect Transistor
MTS	Micro Tactile Sensor
<i>nm</i>	Nanometer
PDMS	Polydimethylsiloxasane
PVDF	Polyvinylidene Fluoride
PZT	Piezoelectric transducer
SC	Switched Capacitor
SU-8	High contrast epoxy based photoresist
μm	micrometer

Chapter 1 Introduction

Minimally Invasive Surgery (MIS) is now routinely used as the preferred choice for many operations. Despite MIS advantages, due to surgeons' indirect interaction with operation site, it severely reduces the surgeon's sensory perception during manipulation. Surgery is essentially a visual and tactile experience and any limitations on the surgeon's sensory abilities are most undesirable. MIS, besides the restricted manipulation mobility and lack of depth from 2D vision difficulties, suffers of almost complete lack of a sense of touch.

The high importance of tactile sensing in surgery is due to the fact that soft tissues can only be properly examined and identified by assessing their softness, viscosity and elasticity. The palpation of tissues and organs is an essential procedure that surgeons value highly. Surgeons have been known to insert their fingers through the access openings during MIS, simply to perform direct tactile exploration.

Tactile sensing can also play a key role in improving the navigation of medical equipment during MIS in which long slender tools such as endoscopes, catheters and guide wires are involved [1]. These endoscopic tools are long and flexible and have few

degrees of freedom. The surgeon's sensory (visual and tactile) perception needs to be enhanced during surgical manipulation.

For visual sensing and feedback, the state-of-the-art is well-established. But tactile sensing which is greatly needed in this area is still an open/potential research area. This study is an attempt to contribute in developing a novel tactile sensor for MIS applications.

In this study tactile sensing for MIS is categorized into two different applications as diagnostic and assistive. In diagnostic application the tactile sensing system is used to evaluate the material characteristics of tissues such as hardness/softness, stiffness, elasticity and distinguishing the abnormalities and it is mainly a non-force approach towards assessing tissues. A number of such sensors have been developed recently [2-7]. The assistive application of tactile sensing is aimed to safe handling and displacing of organs during MIS; as well as safe grasping of tissues like clipping of colon to prevent seepage. The grasping forces play a key role in obtaining such objectives and the measurement of forces can be valued the most important parameter. As the measurement of force is necessary, but is not sufficient to secure assistive tactile sensing requirements. Rather the sensing system has to be integrated with some of other sensing procedures used in diagnostic approach such as stiffness and temperature sensing.

The main focus of this study is on development of an assistive tactile sensing system to be used in MIS operations. Considering the importance of force sensing in MIS operations, it is necessary to have adequate understanding and quantitative assessment of the forces present in MIS operations.

1.1 Forces present at grasper/tissue interface

The forces present at the grasper/tissue interface from different aspects have been investigated in several studies so far. Frank T. *et al* [8] performed a study on prediction and measurement of sealing force during bowel surgery. During bowel surgery it is often necessary to occlude the bowel using clamps. They concluded that a bowel clamp must apply around 7 N to prevent seepage and the coefficient of friction between clamp and bowel should be between 0.6 and 0.9.

Herder J.L. *et al* [9] attempted to design a laparoscopic grasper with enhanced force transmission characteristics through which a constant force transmission function can be obtained. Apart from the constant force transmission function and low friction operating principle, the design criteria included the need to apply grasping forces of at least 10-20N. It was decided to maintain 30N as the objective. In their belief safely-attainable grasping force of 30 N was quite sufficient in normal uses.

Cartmill J.A. *et al* [10] studied the generation of high pressure at the tip of laparoscopic graspers. They used a pull force of 250 gf to investigate this issue. Based on their report, when the angle of incidence of the grasper in relation to the tissue increased from 0 to 135°, the localized pressure at grasper's tips too increased on average from 210 KPa to 650 KPa. They noted that using the same measuring equipment, the maximum pressure that could be generated between one of the author's thumbs and index fingers was of the order of 200 KPa. In another study carried out by the same authors, they studied how rounding the edge in graspers affects the pressure generation in the jaws' interface [11]. They again used the same 250 gf load as the pulling load at the above-mentioned incident angles.

Toledo L. *et al* [12] performed a total of 6,750 measurements for grasping, cutting,

dissecting, suturing and knotting action to investigate the sustained forces and working space of endoscopic instruments. The exerted forces on the instruments were between 0.5 and 12 *N*. Forces exerted on the head-point were between 0.4 and 10.5 *N*, whereas the friction forces were between 0.5 and 1.5 *N*. Their results can provide a basis for the development of endoscopic instruments.

Marucci D. *et al* [13] studied the effect of jaw design on grip security and tissue trauma. Seven pairs of aluminum jaws with a surface area of 8*15 mm² were used. The fresh rumens of three sheep were cut into 2.5*4 cm² strips. The jaws were closed with apposing pressures of 205, 410, and 615 *KPa* using a nitrogen regulator. Considering the jaw area, the pinch forces were calculated as 24.6, 49.2 and 73.8 *N* respectively. These pressures were considered as the approximate pressures generated by a surgeon's light, medium, and firm squeeze respectively. Increasing the apposing pressure resulted in an increase in the peak of pull load for each jaw type. Though high pull forces up to 50 *N* were applied, the forces required during surgery are much lower [14].

The most extensive study on the tissue-forceps interaction so far has been reported by Heijnsdijk E.A.M. and his colleagues [14-18]. Visser H. *et al* [15] aimed to find criteria for designing safer laparoscopic graspers for manipulation of delicate tissues. Four surgeons were asked to apply, via two clamps, enough force to a pig's colon to stretch the mesocolon for dissection. Directions, magnitudes, and application points of the forces were recorded. The surgeons applied an average of 2.5 *N* and maximum of 5 *N* to the colon. In their conclusion the 2.5 *N* force is the force that provides enough tension to the bowel and the maximal needed force can go as high as 5 *N*. From the grasper design point of view, a safe laparoscopic grasper should be able to transmit at least a 5 *N* pull force to the tissue (colon) without damaging it.

Heijnsdijk E.A.M. *et al* [14] studied the slip and damage properties of jaws when

the tissue was under a 5 N pull force. The study was based on a theoretical model shown in Fig 1.1. Based on the model, when the pull force on the tissue is increased, the pinch force should be large enough to prevent slip of tissue. While the pull force is high compared to the pinch force the tissue slips out of the forceps; whereas when both pull and pinch forces are high the tissue can be damaged. Even a high pinch force without pull force can cause damage.

Fig 1.1: A theoretical model of the combination of pinch force and pull force leading to slip, damage, or safe manipulation of tissue [7].

In their study, cecum of pig was clamped between 13 pairs of jaws in the form of 5 different series. Slip and damage forces for the plain jaws (without protrusions) consisted of three flat 4*8, 8*8, 16*8 mm² rectangles is shown in Fig 1.2. Pinch slip forces were 13, 17, 22 N and pinch damage forces were 15, 25, 37 N for the mentioned jaws respectively.

Fig 1.2: The slip and damage forces of the plain rectangular jaws of rect4 (4*8 mm²), rect8 (8*8 mm²) and rect16 (16*8 mm²) [7].

Fig 1.3 shows the views of the series d and e jaws respectively. Series d was a jaw with parallel triangular ribs while series e was jaws with diamond shaped profile protrusions. As Fig 1.4 shows, these profiles compared to the flat jaws significantly decreased the slip force from 22 N to approximately 3 N; while the damage force remained in the range of 24-30 N. For an 8*8 mm² jaw the maximum pinch force should be less than 30 N which is equivalent to an apparent pressure of 468 KPa.

Fig 1.3: Top and side views of the d and e series jaws used in the experiment [7].

Fig 1.4: The slip and damage forces for the rectangles with different profiles (left) and for various depths in the profiles (right) [7].

Heinsdijk E.A.M. *et al* also investigated influence of the fenestrations (holes) in the jaws of laparoscopic graspers during manipulation of delicate tissues [18]. The experiments were performed, on the small bowel of pigs, to determine the effect of a fenestration on the pinch force needed to prevent slip of tissue and on the pinch force leading to tissue damage. Pull forces of 0, 2.5, 4 and 5 N were used. Pinch force could be increased from 1 to 10 N. The jaw was 10x10 mm². Compared to the jaws without fenestration, a fenestration in the contact area decreased the damage line as shown in Fig 1.5. This figure is comparable with the model in Fig 1.1. For the fenestration of 6 mm in diameter the damage line could not be measured accurately as the tissue was damaged at pinch forces lower than 1 N. There was no difference in slip line. The jaws without fenestration resulted in the largest safe working range. Based on this experiment's results, pinch force was recommended to be equal to or less than 7 N.

Fig 1.5: The effect of having a fenestration with different diameter in the contact area on the slip force (three bottom lines) and the damage force (two top lines) [18].

Heinsdijk E.A.M. *et al* [16] in another study investigated the role of mechanical efficiency of laparoscopic grasper on different operative tasks. They again considered a 5 N force as pull force and carried out the experiments under a pinch force in range of 2-12 N as the common pinch force during a surgery.

The influence of force feedback and visual feedback in grasping of tissue too was studied by Heinsdijk E.A.M. *et al* [17]. A grasper with a force sensor in the jaws was developed. Subjects, with and without laparoscopic experience, were set to grasp and pull pig bowel with a pull force of 5 N. The applied mean pinch force was 6.8 N (± 0.5) and the force necessary to prevent slippage was 3.0 N (± 0.4).

Rentschler M.E. *et al* [19] have referred that during routine laparoscopic procedures grasping forces as high as 40 N have been recorded and the corresponding tissue pressure of was approximately 400 KPa.

Jayaraman S. *et al* [20] conducted a study on training the laparoscopists in using their force sensorized MIS instrument. They compared novice and expert

laparoscopists force and position profiles. It was found out that grasp force data have a Gaussian distribution. Therefore, mean grasp forces were compared. The experts exerted a grasp force in the range of 14.9-28.7 N while it was 7.9-22.3 N for the novices.

Callaghan D J *et al* [21] performed a study on sensorized medical scissor blades for application in MIS. The blades exploit the strain sensing capabilities of a single fiber Bragg grating (FBG) sensor bonded to the blade surface. Calibration of the sensorized scissor blade was carried out over the maximum available cutting range of the scissor blades (25° – 10.4°) and direct loads were applied normal to the blade cutting edge. The loads were in a range of 0–30 N load range to represent direct loading of the blade structure during its closing.

A study performed by Valdastrì P. *et al* [22] shows that the forces required to cut tissue range between 13 and 35 N, depending on the organ. It was also found by Rosen J. *et al* [23] that typical grasping and pulling forces when handling tissue in MIS can be as high as 17 N. Considering these information, Trejos A. L. *et al* [24] tried to identify the forces required to manipulate tissues during natural orifice transluminal endoscopic surgery (NOTES). Force sensors were placed as close as possible to the tip of the instruments, and measurements were taken during an *in vivo* NOTES procedure. Their results show that, for a pelvic procedure, the average forces required for the direct (transgastric) approach are significantly less than in the retroflexed (transperineal) approach (43% less), and that the maximum forces required are about 8 and 16 N in the direct and retroflexed approaches, respectively. Based on this study, it is recommended that NOTES instruments be designed to easily handle forces as high as 16 N.

Following the review, the discussion on the forces at grasper-tissue interface can be concluded. To occlude the bowel a clamp must apply around 7 N to prevent seepage

[8]. An average force of 2.5 N is the force that provides enough extension to the tissue (colon) [10, 11, 15], however this force may be increase to 5 N. A grasper should be able to transmit at least 5 N pull force to the tissue without damaging it [14-18]. For pinch force a wide range has been mentioned. Following [9, 20] a minimum 10-20 N and a maximum 30 N of grasping force looks preferable. The forces required to cut tissue range between 13 and 35 N, depending on the organ. Finally for grasping pressure a range of 200-400 KPa is preferable [13, 14, 19] in which 200 KPa is considered as the maximum pressure that a human subject can produce between one of his thumbs and index fingers [10]. In development of a force based tactile sensing system, the force requirements that should be met can be summarized as Table 1.1.

Table 1.1: Summarizing forces present at grasper/tissue interface.

Seepage preventing force	7 N
Pull Force	Average: 2.5 N Maximum: 5 N
Pinch Force	Minimum: 10-20 N Objective: 30 N
Pinch Pressure	200-400 kPa

1.2 Research Objective

To solve problems due to the lack of tactile sensing ability in MIS operations, sensing elements must be integrated into the surgical instruments. The placement of sensing elements itself is an important matter that could greatly influence the quality and extent of information which could be extracted from the tactile measurement. In cases where

sensors are placed far away from the point(s) of generation of forces, the effects of forces are transferred to the sensor position through mechanical linkages, friction, backlash, gravity and inertia. This could distort the actual value of these force and cause errors in the measurement. If high measurement accuracy is needed, forces should be sensed close to the place where they are generated to reduce any possible errors [25]. Secondly, in the systems where forces are measured remotely from the site of their generation, for instance, at/near the actuation mechanisms, they are lumped as a single value. However, in surgical environment, requiring highly resolved forces at multiple points, both for diagnostic and assistive applications, an array of sensing elements is desired.

This study focuses mainly on providing tactile sensory feedback for MIS manipulators, which are equipped with a grasper for holding and pulling the tissues during surgery as an essential part of any generic surgical protocol. This offers the possibility of integrating force sensors into the grasper jaws. The jaw is the place where forces acting at the surgical instrument tip can be measured most directly. Friction and other disturbance forces generated from any moving mechanisms do not affect the measurement of the tissue contact forces as much as that in the other cases [25].

For development of an assistive tactile sensing, force information is necessary but is not sufficient. The main objective of deploying a tactile sensing system in MIS is safe handling and safe grasping of tissues; while force information alone cannot guarantee this matter. The more a force sensor is pushed against a tissue, a higher force value would be observed and vice versa; thus it cannot provide a basis to judge on the security of tissues. It is also necessary to monitor tissue deflections in the sensor-tissue interface. Having both these information would provide information regarding the stiffness at the sensor-tissue interface. This way MIS operation can benefit of both the force and

stiffness information as two of the most important parameters in a tactile sensing system which exists in the sensor-tissue interface. Force information is necessary for both safe handling of tissues, like displacing organs during MIS, and safe grasping of tissues, like clipping of colon to prevent seepage. Stiffness information too can provide information regarding softness and hardness of tissues which leads not only to distinguish between soft and hard tissues but also distinguishing abnormal areas within a given tissue (provided tactile sensor is an array of sensing elements). Stiffness information is valuable for safe grasping of tissues. As in the first stages of grasping the tissues the stiffness is low while in firm grasping the stiffness tends to saturate at a higher level.

Thus, considering the placement of tactile sensing system in a MIS manipulator, the ideal sensing position for monitoring tissue deflections and stiffness at the sensor-tissue interface is achieved via integrating multiple tactile sensing elements into the grasper jaws. Hence grasper jaws were considered as the target point in positioning the tactile sensing elements. However, concerning the fact that the jaw is the place that exhibits most severe space limitations, it is proposed to develop a clinically feasible tactile sensing system based on MEMS fabrication procedures.

The tactile sensor itself was proposed to be an array of force sensors made up of two set of force sensors that are different in their stiffness. These force sensors are arranged alternately in a square array. So a pair of two force sensing elements with different stiffness is referred to as a stiffness sensing unit. When sensor array comes in contact with an object each group of force sensors would experience a different deflection. The difference in deflection was used for stiffness evaluation of tissue phantoms.

The basic principle for functioning of the force sensing elements, irrespective of

the dimensional limitations (micro or macro), was considered to be the same as in the commonly used force sensors, in which they undergo a mechanical deflection following the application of external forces. These force sensors behave very much like a spring, and produce the force signal upon application of the deflection to their mechanical structure. Therefore, they can be referred to as passive, spring-based force sensors. Integrating this force sensing principle in the proposed tactile sensing system implies that the stiffness sensing in this study is also a spring-based approach. Thus, a novel force and stiffness integrated approach is proposed and implemented with individual elements having adjoining and/or overlapping range so as to cover a wide range of forces occurring in the grasping and pulling actions during surgery. A time/frequency multiplexed approach is proposed and implemented in the read-out (measurement) circuitry for real-time observation and estimation of the observed forces at multiple points. The proposed concept is demonstrated in a macro prototype set-up developed by the author for validation.

The mechanical structure of MEMS force sensors could be either silicon or polymer base. Silicon is very fragile and cannot come in direct contact with the biological/surgical environment except by being protected using a protective layer, whereas polymer-based sensors could be more compliant and can come in direct contact with the tissue without being damaged. Sensors developed using polymers like SU-8 and polyimide can provide good linearity in their behavior [26, 27]. Hence, it is proposed to fabricate the sensor-array using SU-8 or Polyimide polymer. However, the objective and/or the scope of this work didn't include a demonstration of a fully-functional MEMS prototype since the concept needed to be proved prior to investing into an expensive MEMS facility. Later, the feasibility of fabrication of a micro-prototype of desired dimensions, using existing MEMS based fabrication

techniques, was also tested.

In short, the main objective of this study was on developing a new tactile sensing system which could measure both the force and stiffness at the sensor-tissue interface during MIS surgery. It was considered to be an array of force sensing elements with two different stiffness.

1.3 Scope and methodology

This study was carried out through the following major steps. Design concept of the sensor was proposed after reviewing the literature. FEM modeling was used to simulate the interaction between sensor and object. The structural modeling of sensing elements was performed for detailed study of force sensing elements by ANSYS. A macro tactile sensing setup was designed and fabricated to bench mark the sensing concept experimentally. Finally, the feasibility of fabricating the MEMS tactile sensor prototype was investigated. These steps can be more elaborated as following:

- **Literature review**

Literature review contains a survey on both force and non-force tactile sensors which could be applicable for tactile sensing in MIS.

- **Design concept and defining sensing requirements**

The design concept was proposed and sensor requirements were defined. A lumped model was used to formulate sensor-tissue interaction. FEM simulation was used for further validation of the expected performance of the sensor and for obtaining preferred stiffness ratios in the stiffness sensing units; through which a higher sensitivity in the evaluation of stiffness at sensor-tissue interface can be secured.

- **Bench marking of sensing concept based on a macro tactile setup**

To pave the way for the development of a functional sensor prototype, an experimental validation was conducted. This experimental validation included both force measurement and the stiffness evaluation in practice. For this purpose, a macro tactile sensing setup was designed, fabricated, calibrated and used for data gathering. The results were presented and analyzed and corresponding conclusions were drawn.

- **Structural design of the MEMS sensor and its fabrication results**

Detailed structural modeling of the proposed MEMS sensor was performed by ANSYS. The force-deflection relationship for the mechanical structure of the sensor and strain distribution throughout the model was obtained; regions with maximum strains were identified too. The supporting force capacity that can be provided by the proposed diaphragm dimensions and the corresponding diaphragm deflections were obtained. The drawings of lithography masks for the fabrication of the MEMS device and the results of the initial fabrication attempts were addressed. The feasibility of fabricating the MEMS tactile array through the available techniques, the SU8-2000 polymer and the design proposed in this study was discussed.

1.4 Organization of this Report

This report has been organized as the following.

Chapter 1 gives a background to the study and highlights importance of tactile sensing in MIS. It also reviews the forces present during surgery between grasper and

tissues. It describes research objective, scope and research methodology finally.

Chapter 2 provides a literature review on stiffness sensing of tissues and available technologies for force sensors. It also tries to highlight the existing gap in the tactile sensing.

Chapter 3 presents conceptual design of the proposed sensor. It presents a lumped modeling for the problem formulation; as well as FEM simulation for further validation of design concept.

Chapter 4 is on the detailed structural modeling and performance analysis of the proposed MEMS sensor by ANSYS.

Chapter 5 is on the validation of sensor design concept and it's functioning experimentally. It includes both force measurement and the stiffness evaluation following the fabrication of a macro tactile sensing setup.

Chapter 6 was dedicated to investigate the fabrication feasibility of the MEMS sensor prototype. In this chapter, the design of the MEMS device is addressed first and the outcome of the fabrication attempts comes next.

Chapter 7 concludes the study and describes future work.

Chapter 2 Literature Review

Tactile sensing has a wide range of potential application in both the industrial and medical fields. Besides force sensing which is the most common application, texture classification [28], pressure distribution [29] and shear stress detection [30-32] can be realized through tactile sensing. The tactile sensors mainly work based on sensing the mechanical deflection in the sensor structure; however, in some studies functioning based on fiber Bragg grating [33], fiber optic micro bending [34, 35] and optical waveguide [36-38] has been investigated too. Since this study is focused on augmenting MIS by tactile sensing technology, only the approaches that can be implemented in MIS are reviewed. Most of studies in the tactile sensing field were more attempted towards sensing of a single parameter by the proposed sensor. In some studies however multimodal tactile sensors have been hybridized too; in result of which different sensing elements were combined in a single sensing node [39, 40]. Tactile sensing procedures which are applicable on tissues can be classified as material characteristics sensing and the force sensing in contact area. Material characteristics sensing itself covers a broad range of senses among which stiffness, hardness,

temperature, roughness and conductivity could be named. In this study, stiffness sensing and force sensing were considered to have the highest value for MIS and the review concentrates only on them.

2.1 Stiffness Sensing

In this review stiffness sensing not only is considered synonymous to measure the stiffness itself but to evaluate the hardness, softness and elasticity also. Even though there might be some differences in definition of any of these parameters with other ones, it is possible to drive a correlation between each two of the desired parameters. Stiffness sensing can be carried out either through an active or passive approach. The active stiffness sensing is mainly performed through resonant vibrating sensor. The passive stiffness sensing, however, makes use of a spring based model which is constructed based on the equivalent stiffness of the sensor and object/tissue.

2.1.1 Resonant vibrating sensors

Resonant vibrating sensing, which is an active stiffness sensing approach, has been used by different investigators to explore the stiffness and characteristics of tissues [2-7]. Originally Omata S. *et al* [2] developed a sensor which was consisted of two piezoelectric elements, one as transducer and the other as vibration pickup, Fig 2.1. When an alternating voltage is applied across the electrodes of driving PZT element, it vibrates in the direction of its length. The pickup detects this vibration and an amplifier feeds the signal back into PZT transducer through a phase shift circuit. In result the sensor vibrates at its own resonance frequency. When the free end of the PZT element

is pressed against a surface the characteristics of the system changes which leads to a phase shift between input and output signals of driving PZT. This phase shift is found and compensated by the phase shift circuit such that the difference in the phase of pickup PZT is forced to zero again; as it was before coming in contact with object. But the change in resonance frequency of the vibrating system would reflect the contact between the vibration rod and the object and can be used to obtain the characteristics of the object.

Fig 2.1: Schematic of the PZT resonant vibrating sensor[4].

Fig 2.2 depicts a simple lumped parameter model that describes resonant vibrating sensor acting against a load. The mechanical behavior of the free vibrator is characterized by its modal mass, damping and stiffness. The electrical behavior is included in the model using electro-mechanical analogies. The electrical capacitance and resistance of the piezoelectric element are represented as a mechanical compliance and a mechanical damper respectively. The electro-mechanical coupling is represented by a rigid lever that converts electrical to mechanical quantities. The load

characteristics are depicted by an additional mass, damping and counteracting stiffness.

Fig 2.2: Mechanical model for resonant vibrating sensor and the load [5].

Simulation results for different load characteristics are depicted in Fig 2.3. Pure load stiffness (blue line) leads to an increase of resonance frequency and anti-resonance frequency of the vibrator. Counteracting to a load stiffness, load mass reduces resonance frequency and anti-resonance frequency (from blue line to green line). Load damping (red line) -as long as it is not too large- changes the frequency only slightly but has strong impact on amplitude and phase at resonance and anti-resonance. Large damping may even lead to loss of zero-phase crossing [5]. These results show that vibration characteristics of objects and their stiffness are closely correlated and magnitude of changes in vibration characteristics depend on the magnitude of the objects characteristics.

Fig 2.3: Simulation of electrical admittance characteristics for different sets of load parameters: black line is response without load; red line is influence of damping; blue line shows influence of stiffness; green is for combination of mass and stiffness [5].

Omata S. *et al* [2] have reported the changes in the resonant frequency of the sensor system when it comes in contact with different objects under a contact pressure about 20 gf, Fig 2.4. In result the softer tissues such as silicone gum, the forehead and the palm of the hand experienced a negative change in Δf , for hard samples such as a wooden plate or a nail, a positive change in Δf occurred. As it is shown in Fig 2.4 after applying the sensor to a specimen, the change in frequency shows a sharp downward or upward curve, and slowly changes with time thereafter. In another experiment, they reported a linear relationship between the changes in resonance frequency and the shear modulus of a 3 cm gelatin film. Based on this results, the hardness of gelatin can be measured as a function of the change in frequency.

Fig 2.4: Time chart of results obtained Omata's resonant vibrating sensor[2].

Omata S. *et al* [7] in another work modified the structure of the sensor in a way that, it could be embedded into fingers of a robotics hand. This time both PZT transducer and PZT pickup elements were 6mm in diameter and 0.5mm in thickness. To detect contact pressure, a strain gauge pressure sensor, 6mm in diameter and 0.5mm in thickness, was also included. A plastic hemisphere, 6mm in diameter, was fixed at the tip of the PZT transducer element as a contact element. They reported the shift in frequency of the sensor when: a) it comes in contact with silicone samples of different concentration; b) contact pressure varies for various samples; c) spring constant K (N/cm) of sample changes; d) Young's modulus of sample changes. For the case of changes in spring constant of samples they reported an approximate linear relationship between spring constant and frequency shift Δf .

Murayama Y. *et al* [3] exploited this approach to develop a Micro-Tactile Sensor (MTS) not only to measure the elasticity with high sensitivity but also to detect the instant of contact. Thanks to finding both the touch and mechanical characteristics of

objects, they developed an instrument to capture both the topological structure and materials characteristics image. They made use of a 30 mm long, 1 mm in outer diameter, 0.5mm in inner diameter glass needle as the vibrating rod, Fig 2.5. The 10 mm length from the tip of the glass needle was tapered to a spherical point with $100\ \mu\text{m}$ in diameter. They attempted to exploit both the longitudinal and rotational vibration mode. The longitudinal mode is the major mode while the rotational mode is the one with small amplitude. The rotational vibration mode exists in the non-loaded MTS but it disappears when the MTS makes contact with the sample. This fact is the principle behind detecting the touch with object and the frequency shift of longitudinal vibration helps to recognize material characteristics.

Fig 2.5: Micro Tactile Sensor (MTS) design[3].

They mounted the MTS probe on an XYZ platform in order to scan the sample's surface freely in both horizontal (X - Y) and vertical directions (Z), Fig 2.6. The XYZ platform system was controlled by a computer for movement in any direction and led to an instrument to capture both the topological and materials characteristics images.

Fig 2.6: Schematic diagram of the Tactile Mapping system [3].

Murayama Y. *et al* [4] in another study tried to highlight the idea that, the contact impedance is not only affected by the material characteristics but also is dependent on the vibration amplitude. By a change in vibration amplitude the extent of mass area deep into the object that contributes to the vibration would also change. That is, the sensor can obtain information regarding the stiffness of a large area deep inside the object. This point can be considered as the base for recognition of tumors inside original tissues. They developed a prototype in the form of an array of resonant vibrating sensors in order to examine the stiffness in the breast. Fig 2.7 shows this palpation probe which is a tactile sensor array with 64 sensing elements and enjoys a space resolution of 5.0 mm between the sensing elements. The contact pressure for each sensor element was controlled by a spring with stiffness of approximately 0.29 N/cm which was optimized from silicone model experiments.

Fig 2.7: The 64-channel tactile sensing probe for diagnosis of breast cancer [4].

The objective was to investigate whether the palpation probe could detect the hardness of the tumor when pushed against the surface of the breast cancer model. Since detection of a hard lump, such as a breast cancer lesion, through palpation is closely related to the hardness and depth of the tumor below the surface of the skin, they arrange an experimental setup as in Fig 2.8. Based on their report, this palpation probe can be used for detecting tumors located at an approximate depth of less than 20 mm.

Fig 2.8: The breast cancer phantom[4].

In conclusion resonant vibrating sensors can be used as a device for recognition of mechanical characteristics of objects in an active manner. These characteristics can be hardness/stiffness, elasticity, shear modulus and recognition of extraneous objects inside a host material. Despite all advantages it suffers from a number of drawbacks among which size limitation is highly crucial [4]. The resolution of the tactile arrays is limited to few millimeters which still are far from arranging it in the form of a high resolution tactile sensor array. This procedure does not reflect the force at sensor-tissue interface as well and it is more suitable for diagnostic application. As the focus in this study is more on the assistive applications, this principle cannot be used as the base for the stiffness sensing in this study.

2.1.2 Spring Based Stiffness Sensing

When two force sensing elements with two different stiffness factors come in contact with a compliant object they experience different deflections compared together. This difference in deflections can be formulated to obtain the stiffness of the object. Engel J. *et al* [39] presented a multimodal polymer-based sensor skin in which stiffness sensing mode takes benefit of the above mentioned principle. The stiffness sensing mode consisted of a membrane and a bulk element each of which has its own NiCr alloy strain-gauge for measuring contact force, Fig 2.9. The membrane represented *Measured Hardness* while bulk element represented *Reference Hardness*.

Fig 2.9: Schematics of a spring based stiffness sensor: (a) Cross-section of the membrane and the bulk element, (b) The deflection of sensing membrane during the contact with an object [39].

They assumed an equivalent spring model for the sensor-object interaction as in Fig 2.10. Based on their proposition, the ratio of pressure (P) at the interface of the reference and measured springs (K_r and K_m , respectively) with the contact object is a function of corresponding stiffness of the object. Eq. (2.1) expresses the proposed ratio of pressure in which K_{obj} is the stiffness of object. Based on their report, the hardness of the contact object was found to be empirically proportional to the ratio of apparent pressures explained by this equation.

$$P = \frac{q_m}{q_r} = \frac{K_m + K_{obj}}{K_r + K_{obj}} \quad (\text{Eq. 2.1})$$

Fig 2.10: Spring based model for the stiffness sensor[39].

Dargahi J. *et al.* [41] too used the same principle to tissue softness measurement but with a different sensing element. They made use of PVDF in their study instead of strain gauges as in the above case. In turn they had to measure dynamic loads instead of static ones because PVDF films do not have static response. They constructed a tactile probe consisting of a rigid base and two compressible cubes, Fig 2.11. In this probe each of the compressible cubes was considered as sensitive parts of the probe with different stiffness.

Fig 2.11: Schematic of the PVDF based tactile probe with its major constituting parts[41].

Two uniaxially poled PVDF films (i.e., PVDF1 and PVDF2), placed between the cubes and the bottom rigid base of the probe, made up the sensing part of the whole assembly. The working principle of the probe was based on applying a force to the test object in a way that it comes in contact with both the compressible cube 2 and the compressible cube 3. For a given setup, E_2 and E_3 as the elasticity of compressible cubes and T_2 and T_3 as their thickness are known; and F_1 and F_2 as the available forces in the probe-object interface can be measured. In result, it is possible to find E_1 which is the elasticity of object. However in this design, reading force values of the sensor has to be obtained based on the peak-to-peak values of a dynamic load.

In short, the spring based stiffness sensing can be used to measure both the contact force and object stiffness simultaneously. But since this principle is a passive approach, it is necessary to have an actuation in order to drive the sensor toward the object. The miniaturization of the sensor structure to an extent that could be exploited in MIS operation is required too.

2.2 Tactile Force Sensing

Force sensing has drawn the highest attention of the researchers when it comes to the tactile sensing field. As a natural choice, most of the tactile force sensors were favored to be developed as an array of sensing elements. Of course miniaturization of these sensors structure was a defining factor and MEMS technology was adopted to realize these sensors. Thanks to the MEMS technology many numbers of these tactile sensors have been fabricated so far. Their functioning is mainly based on either capacitive or piezoresistive principle. In this study, they have been review from

different aspects.

2.2.1 Capacitive Tactile Sensors

Capacitive tactile sensors have been one of the most favored sensors for the force sensing and image recognition. These sensors are about an order of magnitude more pressure sensitive for a given device size and more than an order of magnitude less sensitive to temperature compared to piezoresistive ones [42]. They can be fabricated as pressure (uniaxial force) or three-axial force sensing devices. The structural material of them is either silicon semiconductor or polymer.

Capacitive pressure/force tactile sensor comprises in general of a mechanical structure besides an electrical device. The mechanical structure, Fig 2.12, is either a bridge supported at two edges or a membrane fully supported all around its edges. The electrical device is a capacitor in which following the application of pressure/force the structure deflection causes the variation in capacitance.

Fig 2.12: Schematic design of a force/pressure variable capacitor[43].

The measurand parameter versus the pressure/force is considered to be capacitance change ΔC compared to initial capacitance C_0 . In practice there are two

series of capacitors in the tactile sensor, reference (insensitive) capacitors and sensory capacitors. Reference capacitors are few orders of magnitude less sensitive than sensory capacitors, for example over three orders of magnitude as in Suzuki K. *et al* work [43]. This is realized by providing thicker mechanical structure for the reference capacitors compared to sensory capacitors, Fig 2.13. Their capacitance is equal to zero force capacitance of sensory capacitors. But the parasitic capacitance and temperature effects are much the same.

Fig 2.13: Schematics of reference capacitor (top) and sensory capacitor (bottom)[44].

On the other hand the reference and sensory capacitors are arranged into an array of tactile cells which usually is an X-Y array. This arrangement can be accomplished in two different approaches. In first approach the reference capacitors are fabricated as one dummy row R_D beside sensory rows [45], Fig 2.14. In second method each tactile cell consists of both sensory and reference capacitors beside each other. Fig 2.15 illustrates an example in which two reference and two sensory capacitors are positioned

diagonally and compromise a single tactile cell [46].

In an array of force sensing elements, organized as an X-Y matrix with peripheral circuits, the voltage is fed as driving means on the row lines. In result the charge which is induced through the row-column cross point capacitors onto the columns is detected. The capacitance of these cross points is dependent on the local force on the elements. In order to minimize number of interconnections in read-out of large number of arrays, multiplexing of the output is necessary. This is despite the fact that multiplexing normally causes delays and readout errors.

Fig 2.14 shows a switched-capacitor (SC) readout scheme in which, as the selected row line is switched in voltage, the charge induced on the column is integrated. The force-independent dummy row line (or reference capacitor within each tactile cell) is switched using a voltage opposite to that of the selected row so that the net charge induced on the column is proportional to the difference between the selected cross point (the sensory) and the reference capacitances, allowing the zero-pressure capacitance to be subtracted out. By this approach as long as the integrator gain is high, the output voltage of the switched capacitor circuit would be independent of the parasitic column capacitance [45, 47].

Fig 2.14: Dummy row of reference capacitors, R_D , beside sensory rows in overall structure of sensor array and the read-out components [45].

Fig 2.15: Reference and sensory capacitors arranged as a single tactile cell[46].

Capacitive pressure/force sensors have been extensively used for uniaxial force measurement and there were only few attempts to extend this principle to measure

three-axial forces. Chu Z. *et al* [42] developed one such sensor which works based on the differential capacitive principle. Based on this principle each tactile pixel consists of a capacitor array of four identical quadratic capacitors, Fig 2.16. Once forced, the membrane (the top electrode) tilts and deflects; consequently every capacitance in the array varies in response to the applied force. The force components are decoupled based on the capacitance changes. Fig 2.17 illustrates the response of the sensor to calibration forces. The capacitance change is linearly proportional to the calibration force. The sensitivity to shear force differs from that to normal force such that the sensitivity to shear forces is 2.47 times more than to normal forces.

Fig 2.16: The structure of the three axial capacitive tactile force sensor[42].

Fig 2.17: The capacitance change in response to forces[42].

Silicon-based sensors have been the most attractive sensors because they provide higher resolutions by virtue of MEMS technology. Further more it is possible to integrate read-out circuits and sensing elements together. But they suffer from fragility and require proper packaging and protection for having successful operation in environments that are often harsh or destructive. Fig 2.18 shows two examples of such packaging. For the tactile sensor reported at Fig 2.18 (right), it is even more important to protect the boron-doped row lines from touching other objects as since they carry control signals; as well as to enclose the capacitive gap so that particulates cannot get in between the two planes. While for Fig 2.18 (left) the packaging is only a means to protect from mechanical damage and transfer of applied force to sensing cells. Chu Z. *et al* [42], in developing their three axial tactile sensor, used a hybridized approach for packaging the sensor. They first connected the sensor pixels on a polymer substrate to have a making large-area array and next a layer of silicon rubber was cast on the sensor array, Fig 2.16.

Fig 2.18: Possible packaging of sensors: left) Chun K., university of Michigan, 1985 [24], right) Suzuki K., university of Michigan, 1990[43].

Nevertheless packaging the tactile cells with elastic covers like silicon rubber introduces nonlinearity and hysteresis to the sensor performance which needs to put

into consideration. For example, Gray B. L. *et al* [48] covered their prototype with almost an uniform layer of silicon layer. Fig 2.19 shows the result of a test in which a force was applied via a 1 mm steel sphere under 2.5 mN load at time zero, and removed at time 60 seconds. The initial signal jumping to 90% of its peak, the slow leveling to peak value over 20 seconds and the extremely slow decay after probe removal was observed. The sensor took nearly a minute to decay to 10% of its peak value.

Fig 2.19: Temporal Response of a Rubber-Coated Sensor[48].

An alternative approach to offer flexibility and compliance to tactile sensor is to make use of polymers as structural materials. Lee H.K. *et al* developed polymer-based capacitive tactile sensors for both uniaxial force sensing [49] and three-axial force sensing purposes [50]. Polydimethylsiloxane (PDMS) elastomer was used as structural material. They were fabricated by using MEMS processes with 1 mm resolution for uniaxial tactile sensor and 2 mm resolution for the three-axial tactile sensor. Fig 2.20 shows the completed uniaxial tactile sensor module. Upper and lower electrodes are crossing each other to form a capacitive array of tactile cells and a bump is located on top of each crossing point. A cell is composed of upper and lower electrodes and an air gap encapsulated by polymer structure. The air gap is squeezed with applied force to change the cell capacitance. PDMS was cast on a substrate by spin

coating for reproducible thickness control and vulcanized on a highly leveled stage for better uniformity.

Fig 2.21 illustrates the schematic of the proposed structure for the three-axis tactile cell which consists of an array of four electrodes for both upper and lower portion of a cell and works based on differential capacitance. Furthermore the bump was shifted from crossing point of electrodes to the centroid of four electrodes area. The height of the bump should be large enough to generate a sufficient torque in order to measure shear force. They developed an 8×8 tactile sensor prototype with a resolution of 2 mm. The changes in the four capacitances normalized by their initial capacitances are displayed in Fig 2.22; in each respective direction of the applied force. The reported results are clearly nonlinear.

Fig 2.20: PDMS based uniaxial capacitive tactile sensor; (a) Fabricated tactile sensor module, (b) magnified view of cells, and (c) the bottom electrodes and spacer layer of four tactile cells[49].

Fig 2.21: Conceptual diagram of a unit cell of the proposed three-axis capacitive tactile sensor: (a) 3D schematic view, (b) top plane view, and (c) cross-section view along A-A'[50].

Fig 2.22: Measured capacitance change of a tactile cell as a function of applied force in respective directions: (a) single cell configuration ; (b) capacitance change following applied forces to the bump in the X, (c) the Y, and (d) the Z directions respectively[50].

Table 2.1 lists both the silicon-based and polymer-based capacitive force tactile sensors. Despite the fact that capacitive sensors provide a vast range of resolution, from 50 to 2200 μm , but they are more suitable for pressure (uniaxial force) sensing devices than three-axial force sensing devices. Quite few number of three-axial force sensors have been developed based on capacitive principle so far and the three-axial ones have not been proved to be as efficient as uniaxial sensors. The sensors with silicon-based structure provide high resolutions and integrated read-out circuit but suffer from fragility. The flexibility and compliance of the polymer-based tactile sensors provides a good damage protection and it is a good advantage for these sensors in dealing with objects. However, they show a nonlinear behavior and have a coarse resolution which is undesirable for accurate measurement and robust performance.

Table 2.1: Capacitive-based Force Tactile Sensors; the gray filled rows contain the polymer-based sensors and non-filled rows are silicon-based ones.

Description	Fabrication Technology	Read out method	Diaphragm Geometry (μm)	No of force components	Capacitive gap (μm)	Initial capacitance (pF)	Max. Operating force (mN/element)	Sensitivity	% $(C_{\text{final}}-C_0)/C_0$	Resolution (μm)
Chun, 1985 [45] (Array: 8x8)	Bulk Micromachining	SC charge integrator	1200x1200x10	1	4.0	1.62	100	60 mV/gr	-	2000
Suzuki, 1990 (32x32) [43, 47, 51]	Bulk Micromachining	SC charge integrator	414x386x12.5(2.5)	1	2.5	0.25	9.4	0.27 pF/gr	100	500
Wolffenbuttel M. R., 1990 [52-55]	Surface Micromachining	Charge amplifier	300x200x4	1	1.0	0.14	-	-	-	-
Dudaicevs, H., 1994 [44]	Surface Micromachining	SC, pulse width modulation	$\varnothing$ 100 x 1.5	1	0.9	-	7.8	-	-	-
Gray B. L., 1996 (8x8) [48]	Surface Micromachining	-	80x90x1.5	1	0.5	-	7.5	5000 (% $\Delta C/C_0$)/N	4	100
Chu Z., 1996 (3x3) [42]	Bulk Micromachining	Compensated charge amplifiers	x10	3	2.4	2.0	10	N: 13 pF/N S: 32 pF/N	-	2200
Leineweber M., 2000 (1x8) [46, 56]	Surface Micromachining	SC	$\varnothing$ 80x-	1	0.8	-	10	200-250 V/N	-	250
Sato N., 2003 (224x256) [57]	Micromachining	-	50x50x	1	1.0	-	-	-	-	50
Ko C.T., 2006 [58]	Micromachining	Frequency shift	$\varnothing$ 150x3.3	1	0.64	0.153	0.5	26.1 kHz/ μN	35	200
Lee H.K., 2006 (16x16) [49]	Polymer Based	Charge amplifier	600x600x-	1	12	0.171	40	-	230	1000
Lee H.K., 2007 (8x8) [50]	Polymer Based	-	1300x1300x-	3	-	0.180	30	-	175	2000

2.2.2 Piezoresistive Force Tactile Sensors

Piezoresistive force tactile sensors have been the most popular tactile force sensors. The basic structure of these sensors is generally a planar diaphragm/crossbeam upon which piezoresistive elements are implanted. The structure of the sensor is from either silicon or polymer but silicon-based sensors are more favored, accurate and fine in resolution. In the case of silicon-based sensors, piezoresistors are either boron doped or ion implanted. The piezoresistors are placed at the points of maximum stress to achieve high sensitivity. Pressure exertion results in resistance change in the piezoresistors which is measured through a read out circuit. The read-out circuit is usually either a Wheatstone bridge or a Half-bridge circuit. The reference resistors are fabricated at the periphery of the sensing diaphragm.

Sugiyama S. *et al* [59] reported one of the early prototypes for the silicon-based piezoresistive force tactile sensors. In their prototype the diaphragm and a cavity as the reference pressure chamber were formed by undercut-etching of the interface between the diaphragm and the silicon substrate. Fig 2.23 depicts both the correspondent anisotropic etched cavity for the sensor and the $5\ \mu\text{m}$ wide polysilicon piezoresistors which were photo-lithographically defined and formed by reactive ion etching. Fig 2.23(b) depicts the $80*80*1.4\ \mu\text{m}$ diaphragm, two polysilicon piezoresistors and four etch-holes with $8\ \mu\text{m}$ in diameter.

Because of piezoresistors positioning in Fig 2.23 the change in central resistor R1 is opposite in sign to the change in side resistor R2. Hence the read out circuit was arranged in the form of a half Poisson bridge, Fig 2.24. Since both of the piezoresistors are arranged in a single bridge, the sensor behaves as a pressure (uniaxial force) sensor. The bridge output voltage V_{out} as a function of pressure for the micro-diaphragm

pressure sensor is shown in Fig 2.24. A non-linearity of less than 0.5 % of the full scale is obtained.

Fig 2.23: Schematic of piezoresistive pressure sensor, a) anisotropically etched cavity, b) top view of microdiaphragm [59].

Fig 2.24: Output voltage as a function of pressure for the micro-diaphragm pressure sensor[59].

Sugiyama S. *et al* [60] in an extension to the above work fabricated a 1k element silicon pressure sensor array to obtain a tactile image detection. The micro-diaphragm size was $50 \times 50 \mu\text{m}^2$ and four polysilicon piezoresistors were arranged in a full bridge. The pressure sensor array consisted of a 32×32 elements and pressure-sensing cells had a resolution of $250 \mu\text{m}$. The array was addressed as a memory. The sensing signals from 1024 cells were taken out as a serial waveform to the analog output terminals [60].

Liu L. *et al* [61] too developed a 4×4 element array of pressure tactile sensor. In each tactile cell the sensing element was made of a single piezoresistor that was connected to the three reference resistors to build up a quarter Wheatstone bridge. The piezoresistor sensory element was made at the edge of the diaphragm. The area of the diaphragm was $600 \times 600 \mu\text{m}^2$ with thickness of about $50 \mu\text{m}$. There was a protuberance on the centre of each diaphragm. It transmits the external force to the diaphragm of each sensing unit.

Three-axis force tactile sensors originated readily out of the pressure (uniaxial force) tactile sensors following some necessary augmentations. Fig 2.25 shows a typical design and micro-photograph of a three-axis tactile force sensor in which a central plate was supported over an etched pit through four suspending bridges [62]. To obtain a three-axis force tactile sensor, enough number of piezoresistors should be fabricated upon diaphragm; through reading of which all three force components could be resolved. It can be seen from Fig 2.25(a) that enough number of read-out circuits for each tactile cell is necessary. The read-out circuit here is a Half-bridge circuit.

Fig 2.25: Three-axial force tactile sensor; a) Schematic of the sensor design, b) Micro-photograph of the sensor (Kane B.J., 1995-2000)[62].

There has been large number of piezoresistive three-axial force tactile sensors which were fabricated through MEMS technology. They can be categorized based on different aspects of their design. Each of these sensors can be identified based on their own mechanical structure of sensing element, positioning of piezoresistors, read-out circuits, protective and compliant layer, load transfer devices and overload protection devices from others.

Fig 2.25 illustrates indeed the work of Kane B.J. *et al* [62-64]; which is composed of a central shuttle plate suspended by four bridges over an undercut etched pit. The central plate is composed of several extra layers and it is stiff relative to the bridges. This relative stiffness ensures that the bulk of the structural deformation occurs in the bridge elements, thereby maximizing the sensing sensitivity. The sensing piezoresistors were placed on the bridges while the reference piezoresistors were located at the sensing element periphery. This silicon-based piezoresistive force sensor is one of the early works to measure forces in three axes. Each sensing element is almost

$150 \times 150 \mu\text{m}^2$ in size and offers a resolution less than $300 \mu\text{m}$.

Mei T. *et al* [65, 66] fabricated a prototype by silicon bulk-micromachining which consists of an E-shaped square membrane, Fig 2.26(a). Strain induced following the application of forces is sensed by three groups of integrated piezoresistors in the silicon membrane. The locations of the piezoresistors on the membrane are shown in Fig 2.26(a). Suitable positions for the piezoresistors were determined by FEM analysis of the E-shaped membrane to obtain optimal circuit output for F_x , F_y and F_z . Three half bridge circuits were deployed to read-out the output signals, one for each axis.

Wang L. *et al* [67] used a monolithic diaphragm to realize their piezoresistive pressure sensor, Fig 2.26(b). It consisted of four piezoresistors oriented perpendicular to diaphragm edges and an additional force transmission structure (a square mesa) to convert an applied force into stress. Each piezoresistor serves as an independent sensing element to allow the sensor to resolve all three components of the applied force. Each of the four piezoresistors is connected into a separate Wheatstone quarter bridge.

Vazsonyi E. *et al* [68] and Kim K. *et al* [69, 70] too developed tactile sensors with a monolithic diaphragm as sensing element but with different geometrical structures. In the Vazsonyi E. sensor alkaline etching technique was used for the fabrication and piezoresistive elements were positioned on the bottom edge of the diaphragm. In the Kim K. prototype the sensing piezoresistors were placed on the top edge of the sensing diaphragm.

Fig 2.26: Monolithic diaphragm as sensing element; a) Mei T., (1999-2000)[65], b) Wang L., 2000 [67].

Adam M. *et al* [71-73] in developing their force tactile sensor used single crystalline membrane formed by combining porous Si bulk micromachining with a conventional CMOS process. Based on their report this approach offers the advantage of higher sensitivity piezoresistive sensing elements, compared to the similar size of a conventional surface micromachined reference. The single crystalline sensing elements, Fig 2.27, consist of a [100] central plate suspended by four bridges over an etched cavity. Each of the bridges includes a [110]-oriented p⁺ piezoresistor which act as an independent strain gauge to provide the signals for resolving the vector components of the load. The nominal value of piezoresistors is 2 kΩ. Since the central plate of the membrane is relatively rigid compared to the narrow suspensions, when the load is applied, deformation occurs only in the bridge elements. The center hole in the

membrane is provided for the insertion of an optional load transmitting element in order to enhance the shear sensitivity of the device.

Fig 2.27: Micro-photograph of Adam M. *et al* prototype (2004-2008) [71-73].

All silicon-based tactile sensors are fragile and need to be protected against mechanical shocks. Hence all of these sensors are covered with an elastic and compliant layer. Kane B.J. *et al* [62-64] and Mei T. *et al* [65, 66] stuck a 2 mm-thick rubber layer to the surface of their prototype for absorbing shock to the inner devices. Kim K. *et al* [69, 70] diced the wafer level sensor array into individual and attached on a flexible PCB. Next they covered the sensor array surface with two-component silicon rubber as a protective layer through which they obtained a hybrid structure for their tactile sensor. Natural Consequence to have these protective layers was that, the sensitivity of the individual sensing cells was reduced and the magnitude of this reduction was dependant on the thickness and the hardness of the layer.

The most comprehensive study on the influence of the protective layer over the response of the tactile sensor was performed by Adam M. *et al* [71-73]. Two sensors, with linear performance before creating protective layer, were covered by two elastic

layers of different thickness. The first sensor was covered with a 220 μm silicone rubber. This decreased the sensor's sensitivity highly but did not change its linear characteristics. The second sensor was covered with 500 μm rubber layer. This thicker rubber layer reduced the sensor's sensitivity even further and resulted in a nonlinear exponential behavior. They also reported that, the response of sensors with a protective elastic layer is extended from the loads applied to the center of bridges to the loads applied outside the sensor diaphragm area. The extension of reception field even leads in overlapping between these receptive fields [73].

An alternative approach for silicon-based piezoresistive force tactile sensors covered with rubbery protective layer is to develop polymer based piezoresistive force tactile sensors. Hwang E.S. *et al* [74, 75] developed such a tactile sensor array for measuring both normal and shear loads. Fig 2.28 shows the schematic view of the proposed tactile sensor. The general structure of the sensor consists of a sensing skin with a thickness of 70 μm , a 50 μm height bump deposited on top of the sensing skin and PDMS as substrate. The sensing skin itself consists of multilayer polyimide and strain gauges deposited out of Ni-Cr alloy functioned as sensing elements.

This sensor does not make use of a sensing diaphragm/crossbeam like a conventional silicon-based sensor; rather the strain gauges are embedded into bulk material and following the deformation of the substrate polymer they experience strain. Fig 2.29 illustrates the performance of the sensor for both the normal and shear loads. It is clear that this sensor does not provide an acceptable linear performance. Furthermore comparing the normal load response, Fig 2.29(a), with the shear load response, Fig 2.29(b), it is evident that the output voltage change due to shear load is much smaller than that due to normal load. In results this sensor cannot provide sufficient sensitivity when it comes to measure shear loading.

Fig 2.28: The schematic view of the polymer-based tactile sensor [76].

Fig 2.29: Characteristics of the polymer-based tactile sensor for a) normal load, b) shear load[74].

Kim J.H *et al* [26, 27] developed two polymer-based force tactile sensors. These sensors are benefiting from a square sensing diaphragm are similar in their structure; but are different in dimension and fabrication process. Fig 2.30(a) shows the schematic of one of the sensors containing four metal strain gauges implanted on the diaphragm edges. In this figure the sensing direction of stain gauges are perpendicular to edges

while in the second sensor, Fig 2.30(b), two of the strain gauges are perpendicular but the other two are parallel to their corresponding edges. They made use of polyimide polymer as the structural material for both of their sensors. The mesa in the second sensor was fabricated out of polyurethane. For the last prototype with a diaphragm thickness of $50\text{ }\mu\text{m}$, they reported a load capacity of 0.6 N which is quite considerable. As Fig 2.31 shows, the sensor shown in Fig 2.30 (b) showed a linearity under applied F_x , F_y force components, while the component, F_z , presents the second-order fitting curve [27].

Fig 2.30: Schematic of sensing diaphragm based polymer force tactile sensor design; a) [26], b) [27].

Fig 2.31: Applied load-Output signal relationship for the polymer-based force tactile sensor in Fig 2.30

(b): a) Application of F_z , b) Application of F_x , c) Application of F_y .

Table 2.2 summarizes the piezoresistive pressure/force tactile sensors and includes their technical specifications. Based on this table, sub-millimeter resolution for a force tactile array is achievable by existing MEMS fabrication techniques. These sensors have been more favorable for measuring three-axial forces than capacitive sensors and enjoy a good linear relationship between input force/pressure and their output signal. Piezoresistive sensors not only benefit from a simpler readout circuits, compared to capacitive ones, but also their electronic circuit can be separate from the tactile array itself. But the read-out circuit of the capacitive tactile cells should be close to them.

Table 2.2: Piezoresistive-based Force Tactile Sensors; the gray filled rows contain the polymer-based sensors and non-filled rows are silicon-based ones.

Description	Fabrication technology	Read out method	Diaphragm geometry (μm)	No of force components	Piezoresistive	Max. operating force (mN/element)	Sensitivity	Resolution (μm)
Sugiyama S., 1986 [59]	Silicon undercut-etching	Poisson half bridge	80x80x1.4	1	polysilicon	4.48	130 $\mu\text{V/V/bar}$	-
Sugiyama S., 1990 (32x32) [60]	Silicon undercut-etching	Wheatstone half bridge	50x50x-	1	polysilicon	-	5-10 mV/V/bar	250
Liu L., 1993 (4x4) [61, 77, 78]	Bulk micromachining	Wheatstone quarter bridge	600x600x50	1	Single crystal	7.2	320 mV/V/bar	-
Kane B.J., 1995-2000 [62-64]	Bulk micromachining	Half Bridge	150x150x2.85	3	Polysilicon	-	N: 51 S:12 mV/KPa	300
Jin W.L., 1998 [79]	Bulk micromachining	Wheatstone Full Bridge	300x300x4	3	Polysilicon	-	N: 20 S: 8 mV/mA	500
Wang L., 2000 [67]	Bulk micromachining	Wheatstone quarter Bridge	1900x1900x50	3	Single crystal	3000	-	-
Mei T., 2000 (4x8) [65, 66, 80]	Bulk Micromachining	Half Bridge	4000x4000x70	3	Single crystal	N: 50000 S: 10000	N: 13 mV/N S: 2.3 mV/N	4000
Vazsonyi E., 2005 [81-85]	Alkaline etching	Wheatstone half bridge	-x130	3	Single crystal	2000	3.68 mV/V/N	-
Kim K., 2006 (4x4) [69, 70]	Bulk Micromachining	-	1200x1200x-	3	Single crystal	2000	N: 2.1% dR/N S:0.5% dR/N	2400
Dao D.V., 2006 [86, 87]	Micromachining	Wheatstone bridge	1700x1700x400	4	Single crystal	1000	Refer to article	2000

Table 2.2: Piezoresistive-based Force Tactile Sensors; the gray filled rows contain the polymer-based sensors and non-filled rows are silicon-based ones. (Continued)

Description	Fabrication technology	Read out method	Diaphragm geometry (μm)	No of force components	Piezoresistive	Max. operating force (mN/element)	Sensitivity	Resolution (μm)
Dao D.V., 2002 [88, 89]	Micromachining	Wheatstone bridge	1700x1700x400	6	Single crystal	-	Refer to article	2000
Fujii I., 2006 [90]	Micromachining	Wheatstone half bridge	1200x1200x-	3	Single crystal	5000	N: 0.2 mV/mN	-
Wen C.C., 2008 [91]	Micromachining	Wheatstone bridge	3000x3000x~	3	Single crystal	200	-	5000
Adam M., 2008 (8x8) [81, 82]	Porous Si bulk micromachining	Half Bridge	300x300x8	3	Single crystal	70	4-6 mV/mN/V	700
Hwang E.S. [74-76] 2006 (8x8)	Polymer based	Wheatstone bridge	-x-x70	3	Metal layer	4000	0.8 mV/10 gF	-
Kim J.H., 2005 (10x10) [26]	Polymer based	-	260x260x14	3	Metal layer	300	-	1000
Kim J.H., 2006 (4x4) [27]	Polymer based	Wheatstone bridge	1500x1500x50	3	Metal layer	600	N: 0.52 V/N S: 0.23(0.28) V/N	2500

2.3 Embedding Tactile Sensors into Graspers

To solve problems due to the lack of tactile sensing ability in MIS operations, sensing elements must be embedded into the instruments. The placement of sensing elements itself is an important matter that could greatly influence the quality of tactile measurement. Basically, for force sensing alone, there are at least four locations where the sensing elements might be placed. Fig 2.32 shows the four possible positions to place sensing elements. The forces acting at the instrument tip can be measured by placing force sensor in either positions of (1), (2) and (3). In these placements, however, the effects of forces are transferred to the sensor position through mechanical linkages, friction, backlash, gravity and inertia. This could distort the force values and cause errors in the measurement. If high measurement accuracy is needed, forces should be sensed close to the place where they are generated to reduce any possible errors [25].

This study focuses mainly on providing tactile sensory feedback for MIS manipulators which are equipped with a grasper. This offers the possibility of integrating force sensors in the grasper. The jaw is the place where forces acting at the surgical instrument tip can be measured most directly. Friction and other disturbance forces generated from any moving mechanisms do not affect the measurement of the tissue-tool contact forces as much as that in the other cases [25]. This makes position (4) a favourable point in integrating the tactile sensor into the MIS manipulator.

Dargahi J. *et al* [92] integrated a PVDF film based tactile sensing in one of the grasper jaws, Fig 2.33. The structure of this sensing jaw consisted of three layers. The top layer is made of micromachined silicon, which has a rigid tooth-like structure. The bottom layer is a flat 6 mm thick Plexiglas substrate. Sandwiched between the Plexiglas

and silicon is a 25 μm -thick PVDF film. The two flat surfaces of the PVDF film contain patterned aluminum electrodes to provide electrical output to an amplifier and four output signals are derived from the sensing device. But this prototype has few major drawbacks. Not only its size, occupied a 22*45 mm² for a jaw, is not acceptable for a MIS grasper but also quit coarse resolution is provided by this sensor. On the other hand the integrated sensor suffers from lack of DC response due to the nature of PVDF piezoelectric film.

Fig 2.32: Possible locations for placing force sensing elements on a typical laparoscopic instrument: (1) near or at the actuation mechanism driving a joint; (2) on the instrument shaft outside the patient’s body; (3) on the instrument shaft inside the patient’s body; and (4) at the instrument tip [25].

Fig 2.33: Cross section of the grasper jaw equipped with tactile sensors[92].

Sokhanvar S. *et al.* [93] designed a tactile sensing system for MIS grasper. A row of sensing elements was integrated along grasper length. Each sensing unit comprises three PVDF films. One is attached to a flexible beam and two are located on both of its end supports. Fig 2.34 illustrates such a sensing configuration; the force applied on the beam can be measured by the PVDF films at its supports while the softness of an object being grasped can be quantified from the beam deflection detected by the PVDF film attached to the beam. This design enjoys a good sensing concept: the sensing jaw has a dimension of $18 \times 18 \text{ mm}^2$ and the length and width of each beam in a sensing unit are 7 and 2 mm, respectively. However, this design, besides using PVDF film as sensing elements, has a coarse resolution which makes it unfavorable in integrating into a commercial MIS grasper.

Fig 2.34: (a) The schematic of integrating tactile sensing array into the grasper jaw. (b) The cross section of a sensing unit comprising PVDF films, flexible beam and rigid silicon base [93].

Bonakdar A. *et al.*, [94] in their attempt for measuring the compliance of the objects between the grasper jaws, integrated an array of four individual sensors on both jaws of an endoscopic grasper. Figure 33 shows the grasper with the assembly of tactile array. Each sensor consists of four different parts, as shown in Fig 2.35, namely rigid

and compliant cylinders, a sensing element and a substrate. The rigid cylinder is machined from Plexiglas, 3mm in diameter and of 1mm thickness. The purpose of the rigid cylinder is twofold. It not only acts as one major sensing element, but also increases the friction between the grasper and the sensed object. The second cylinder is the compliant cylinder which surrounds the rigid cylinder. The compliant ring shaped cylinder is micromolded from liquid silicone rubber. Though the single sensor design benefits from a good design concept but, as its outer diameter is 6 mm, it is too not within the distance limitation to be integrated into a MIS grasper in the real case.

(a)

(b)

Fig 2.35: (a) Grasper jaws equipped with tactile sensors. (b) Cross-section view of the sensor unit [94].

S. Schostek *et al.* [95] developed a tactile laparoscopic grasper for surgical palpation. They exploited the resistance change following exertion of external force in the contact area between a spherical and a flat surface as the pressure sensing principle. The sensor, Fig 2.36(a), consisted of a substrate material, spherical electrodes and a layered-structure sensing sheet. The sensing sheet itself was built out of an electrical conductive polymer, a 10 μm copper foil and a silicone rubber encapsulation. Forces exerted on the silicone rubber layer leads to the change in the contact area between the conductive polymer sheet and the electrode surface; and thus in the change of the electrical resistance between the copper foil and the respective electrode. Fig 2.36(b) shows a photo of the prototype. This system, regardless of its robustness to higher forces, suffers from both low sensitivity and poor resolution. Furthermore such a system can measure only normal loads and cannot be used for sensing of pulling forces.

Fig 2.36: A tactile system for laparoscopic application; a) Schematic of structure, b) Prototype

photo[95].

Culjat M. C. *et al.*, [96] in their attempt in equipping the da Vinci commercial surgical robotic system with tactile feedback, integrated a piezoresistive force sensor into a modified Cadere grasper (with a contact area 5 mm*14 mm); as shown in Fig 2.37 (a). The force sensor was a commercially available piezoresistive force sensor (Tekscan FlexiForce) because of its thin-film profile (208 μm), small diameter (10 mm),

appropriate force range (0 to 110 N), high linearity, and good static and dynamic response. In their first attempt, they integrated a single force sensor into the grasper jaw. In another attempt, they developed a 3*2 tactile sensor array for mounting in the grasper [97]. As Fig 2.37 (b) shows, the sensor array was constructed by combining a 3*2 element array of gold electrode pads with the piezoresistive substrate of the above mentioned force sensor. The 3*2 sensing electrodes had a contact area of 1.75 mm*1.75 mm. Such a tactile system can only measure a single force value in the former case; or at the most, a coarse distribution of contact pressure by a small number of sensing elements in the latter case.

(a)

(b)

Fig 2.37: (a) Modified FlexiForce sensor mounted on a da Vinci robotic Cadiere grasper. (b) Tactile sensor array following combining electrodes pad and piezoresistive substrate [96].

Naish M. D. *et al* developed a palpation system for minimally invasive tumour localization [98, 99]. Their tactile sensing array incorporates a capacitive-based pressure sensor from Pressure Profile Systems (Los Angeles, CA). It is comprised of sixty 2 mm square elements arranged in a 4*15 grid, as shown in the Fig 2.38. The tactile array sensor was mounted at the distal end of a 10-mm diameter, 385-mm long rigid metal rod with a handle (surgical probe suitable for MIS). In this design, the sensing resolution is 2 mm, however, it can only measure the normal component of the force vector.

Fig 2.38: Tactile sensing instrument following incorporating the TactArray sensor into a surgical probe suitable for MIS [98, 99].

The available tactile sensing systems that are integrated into the surgery graspers so far, regardless of their robustness to higher forces, still suffer from poor resolution. Furthermore, these systems can measure only normal loads and cannot be used for sensing of pulling forces.

The endoscopic and laparoscopic instruments are inserted through incisions of about 3–12 mm diameter [25]; and the commercial MIS grasper modules have a

common jaws' length in the range of 20-40 mm [100]. These size limitations for the diameter and length of MIS graspers during the integration of tactile sensing array into the graspers should be met.

2.4 Summary

For assistive application of tactile sensing in MIS operation both stiffness sensing and force sensing are considered beneficial. Hence both stiffness sensing and force sensing approaches were reviewed.

Resonant vibrating sensors, used for stiffness sensing, do not reflect the force at the sensor-tissue interface. But spring based stiffness sensors can reflect both the contact stiffness and the contact force simultaneously. Hence the last approach is preferred to be exploited in the study.

Tactile force sensors can be developed based on capacitive or piezoresistive approaches. But piezoresistive sensors have been more favorable for measuring three-axial forces than capacitive sensors. On the other hand piezoresistive sensors benefit from a simpler readout circuits and provide the possibility of arranging their electronic circuits away from the tactile array. Hence it is proposed to develop force sensing elements based on piezoresistivity approach. From structural material aspect, polymer-based sensors can come in direct contact with tissue without being damaged. Hence it is preferred to construct the sensor out of SU-8 or Polyimide polymer.

Finally integration of tactile sensing systems into graspers have not been accomplished fully yet. Due to space limitation a high resolution sensing system is required. Furthermore for measuring both grasping (normal) and pulling forces, the force sensing elements need to have three-axial measurement capability.

Chapter 3 Design Concept and Formulation

Two of the most important parameters for an assistive tactile sensing system in MIS applications are force and stiffness information at the sensor-tissue interface. Force information is necessary for both safe handling of tissues, like displacing organs during MIS, and safe grasping of them, like clipping of colon to prevent seepage. Stiffness information too can provide information regarding softness and hardness of tissues which leads not only to distinguish between soft and hard tissues but also to distinguish abnormal areas within a given tissue (provided tactile sensor is an array of sensing elements). Similar to force sensing, Stiffness sensing is valuable for safe grasping of tissues. As in the first stages of grasping, the stiffness is low while in firm grasping the stiffness tends to saturate at a higher level. In this chapter, the design concept for including both force and stiffness sensing in a single device and its formulation is addressed.

3.1 Design Concept

Developing a tactile sensor which could measure both the interaction force and stiffness with an object provides the possibility of benefiting from the advantages of both force and stiffness sensing. For measuring stiffness of an object, it is necessary to have information regarding both force and deflection at the sensor-object interface. Through combining the force and deflection information together it would be possible to obtain force-deflection or stiffness information of the object.

Fig 3.1: Spring based model for the stiffness sensing [39].

The spring based stiffness sensing exploited by Engel J. *et al* [39] is considered again, Fig 3.1. A stiffness sensing unit consists of two force sensing elements but with different stiffness. The governing principle of Fig 3.1 not only provides the possibility of measuring tactile forces but the stiffness of the contact object as well. In this figure, K_{obj} is the stiffness of object; and K_r and K_m are the stiffness of force sensing elements. K_r is referred to as *Reference* stiffness and K_m is the *Measuring* stiffness. K_r is considered to have a higher stiffness compared K_m . When an object comes in contact with the *Measuring* (with K_m) and the *Reference* (with K_r) sensing elements within a

stiffness sensing unit, these sensing elements experience different deflections and different values for q_m and q_r are recorded. Considering K_r , K_m , it is proposed to use q_r , q_m readings to evaluate the stiffness of object K_{obj} .

A sensor prototype could be an array of force sensing elements with two different stiffness which are arranged alternately, as shown in Fig 3.2. In this figure different ‘+’ patterns represent either of *Measuring* and *Reference* force sensing elements; and a pair of one *Measuring* and one *Reference* sensing element is considered as a stiffness sensing unit. The force sensing elements are proposed to be piezoresistive-based sensors. The stiffness sensing sensitivity is expected to be a function of not only the stiffness of force sensing elements but also the stiffness of object itself. Hence in dealing with the sensitivity of the sensor, two different issues should be considered. Firstly a single sensor array cannot be effective to measure the whole stiffness range of objects. So for a specific range of stiffness, it is necessary to have a separate sensor. Furthermore to enhance the stiffness measurement, the whole span of stiffness can be covered by a set of sensor arrays in such a way that there is always an overlap between the working ranges of two individual sensors as shown in Fig 3.3. Secondly for each specific range of stiffness, the stiffness ratio of *Measuring* and *Reference* force sensing elements in the sensing array should be tuned precisely.

Each force sensing element is meant to be a three-axial force tactile sensor. Such a sensing system can provide both three components of force and contact stiffness and provides a four component vector as its output. For grasping the tissues only two components of the force vector are mainly required. These are the grasping (normal) and pulling forces. In result among four components of the output vector, only three out of them are expected to be valued the highest. These values are normal (grasping) force, normal (grasping) stiffness and pulling force during grasping. Normal (grasping) and

graspers’ pulling forces indicate values of interaction forces and normal stiffness indicates firmness of the grasping.

Fig 3.2: Schematic of the proposed force/stiffness tactile sensor array.

Fig 3.3: The concept of considering overlap between sensing ranges to enhance stiffness measurement.

3.2 Assistive Tactile Sensing

The proposed tactile sensing system is expected to perform an assistive function in the MIS. On the other hand, the proposed sensor is a passive sensor and its performance

is dependent on providing an external actuation as shown in Fig 3.4. The tactile sensor is expected to be integrated into one of the jaws of a laparoscopic grasper. When surgeons manipulate the grasper manually, the actuation necessary for the function of the sensor is provided by the surgeon's hand motion. Following the grasping of the tissue both the force and stiffness in the contact area can be measured and fed back into surgeon. In the case of a master-slave robotics system, actuation should be provided by slave part and the force/stiffness information is fed back to the master system.

The proposed sensor can also be adopted as a diagnostic tactile sensing system. In such case, for some reasons like space restrictions, not only it might not be possible to align the sensor array parallel to the tissue surface but the tissue itself may not have a relatively flat surface. Hence a semi-spherical external surface for the proposed tactile sensor array could help to overcome the problem to a large extent. In this way when the tactile sensor array is driven against a tissue, only a partial group of sensing elements would come in contact with the tissue. But still, this partial group of the engaged sensing elements in contact can measure both the force and stiffness.

Fig 3.4: Side view schematic for an assistive (or diagnostic) tactile sensor array equipped with actuation.

In the grasper-tissue interface, as mentioned earlier in Chapter 1, the common

pressure can be considered in 200-400 KPa range [13, 14, 19]. The sensor design should be based on having pressure values higher than this range. This issue can have two different implications. First, the stiffness sensing feature should not saturate even if the load exceeds beyond this range and it should be able to continue its function. Second, the sensor should be protected from overloading as well. Hence a grasping pressure of 600 KPa is considered as the maximum pressure limit below which sensing system would perform its function without stiffness saturation and beyond which the sensor would be protected against getting damaged. Following the table 2.2, a piezoresistive tactile sensor array capable of three-axial force measurement can be considered to have a resolution in the range of 250-1000 μm . Hence the expected support force by a single force sensing element can be tabulated as Table 3.1. So a minimum 37.5 mN and a maximum 600 mN force is expected to be supported by a single force sensing element at the extreme situation under 600 KPa of pressure.

Table 3.1: the expected supported force by a single force sensing element (mN).

		Pressure (KPa)			
		200	300	400	600
Resolution (μm)	250	12.5	18.75	25	37.5
	500	50	75	100	150
	750	112.5	168.75	225	337.5
	1000	200	300	400	600

3.3 Lumped Model Formulation of Design Concept

Fig 3.5 depicts a lumped model for the sensor and target object when they interact. The object is represented by two springs with K_{obj} stiffness and each of them is considered to engage with sensing elements with stiffness of K_r and K_m . K_r is the *Reference* stiffness and K_m is the *Measuring* stiffness such that $K_r \geq K_m$. When the sensor is given a displacement of Δ against the object, following the moment of contact, the interface of *Reference* and *Measuring* sensing elements with object would experience δ_r and δ_m displacements respectively. Based on force equilibrium in the vertical direction, it is possible to write:

$$\begin{aligned}
 K_{obj}\delta_m &= K_m(\Delta - \delta_m) \Rightarrow K_m\Delta = (K_{obj} + K_m)\delta_m \Rightarrow \delta_m = \frac{K_m}{K_{obj} + K_m}\Delta \\
 K_{obj}\delta_r &= K_r(\Delta - \delta_r) \Rightarrow K_r\Delta = (K_{obj} + K_r)\delta_r \Rightarrow \delta_r = \frac{K_r}{K_{obj} + K_r}\Delta \\
 \Rightarrow \Delta - \delta_m &= \Delta - \frac{K_m}{K_{obj} + K_m}\Delta = \frac{K_{obj}}{K_{obj} + K_m}\Delta \\
 \Rightarrow \Delta - \delta_r &= \Delta - \frac{K_r}{K_{obj} + K_r}\Delta = \frac{K_{obj}}{K_{obj} + K_r}\Delta \\
 \Rightarrow \boxed{R = \frac{\Delta - \delta_m}{\Delta - \delta_r} = \frac{K_r + K_{obj}}{K_m + K_{obj}}} & \quad \text{(Eq. 3.1)}
 \end{aligned}$$

Eq. 3.1 expresses the relationship between ratio of sensing elements reading, R , with object stiffness, K_{obj} and K_m and K_r as the stiffness of sensing elements. This equation can be rearranged to find K_{obj} through an explicit expression as:

$$\begin{aligned}
 (K_m + K_{obj})R &= K_r + K_{obj} \Rightarrow RK_m + RK_{obj} = K_r + K_{obj} \\
 \Rightarrow (R - 1)K_{obj} &= K_r - RK_m \Rightarrow \boxed{K_{obj} = \frac{K_r - RK_m}{R - 1}} \quad \text{(Eq. 3.2)}
 \end{aligned}$$

Fig 3.5: Schematic of a lumped model of the sensor and object interaction.

Fig 3.6 visualizes Eq. 3.1 in which horizontal axis is the ratio of object stiffness to the stiffness of *Measuring* (softer) sensing element, K_{obj} / K_m , ranging from 0.1 to 10 and vertical axis represents the ratio of *Reference* (stiffer) element stiffness to the *measuring* one, K_r / K_m , ranging from 1 to 10. The contour lines are the representation of the parameter R which is the ratio of softer sensing element reading, $\Delta - \delta_m$, to the stiffer sensing element reading, $\Delta - \delta_r$.

A given sensor array would have a certain K_r / K_m ratio. hence its performance can be represented by a horizontal line drawn from the desired K_r / K_m value in Fig 3.6. In the figure by moving from lower K_r / K_m to higher ones, for example from 5 to 7 as shown, a wider range of R is experienced which is an indication of increase in the measurement sensitivity. Hence higher K_r / K_m provide higher sensitivity. For a specific sensor, with a given K_r / K_m ratio, as the stiffness of object increases the

sensor reading, R , tends to saturate. This means for having higher sensitivity, stiffness of the *Measuring* (softer) sensing element should not be smaller than the object stiffness. Rather, it is recommended that stiffness of the *Measuring* sensing element to be higher than the object stiffness.

Fig 3.6: Contour line representation of the ratio of sensing elements reading, R , based on Eq. 3.1.

Fig 3.7 is another representation of Eq. 3.1 for further explanation in which R is depicted against K_{obj}/K_m for the given values of K_r/K_m . Each graph in the figure represents a sensor with a specific K_r/K_m . It is clear that the larger the K_r/K_m value is the higher the sensitivity would be; especially when it comes to smaller K_{obj}/K_m values. Again in this figure for having higher sensitivity, stiffness of the *Measuring* (softer) sensing element should be higher than the object stiffness. However, firm quantitative judgment has to be postponed till experimental works are conducted.

On the other hand, having higher values for R is obtained at the cost of having small reading values for $(\Delta - \delta_r)$ in the denominator of Eq. 3.1. This may cause problem to get sufficient output out of the *Reference* elements. Hence, it is decided to make sure that K_r/K_m values in design stage are set in a way that R does not exceed beyond 10.

Fig 3.7: Expected Reading's ratio, R , versus object stiffness for sensors with different stiffness.

3.4 FEM Based Validation of the Design Concept

Since in real circumstance the object/tissue is a continuum medium rather than discrete elastic object, there is need for a more representative modeling than the lumped model. FEM simulation by ANSYS was chosen to consider this requirement in validating the sensing concept for the continuum medium. Fig 3.8 demonstrates both the 3D model (Fig 3.8 (a)) and 2D finite element model (Fig 3.8 (b)) of the sensor and object in their initial condition. The simulation condition was considered to be ax-symmetric. Fig 3.8

(b) shows the corresponding boundary condition for the meshed model. The sensor consists of a diaphragm constrained at its outer edges and a mesa protruded from its center downward. The mesa comes in contact with the object at its face.

Fig 3.8: Schematic of the sensor and object contact; a) 3D model, b) 2D FEM model.

When the sensor is driven against the object, both the sensor and object experience deflection. Of course the magnitude of the deflection in each of them depends on their stiffness. In this simulation, sensor diaphragm was considered to have constant geometry as 2.0 mm in diameter and 0.1 mm in thickness. Hence for resembling sensors with different stiffness, different elastic modulus was needed to be assigned. For the object too, a constant geometry as 2.0 mm for both its height and diameter was considered. This size was supposed to be large enough, compared to mesa diameter, for compensating deformation in the contact area before reaching the boundaries. This

means that, simulation results would depend on the magnitude of localized deformations and would not be affected by boundary conditions. Again, different elastic modulus was considered to simulate different object stiffness. As it can be seen from Fig 3.8 (b), for increasing simulation accuracy fine meshing was used in the contact area while in the remote areas coarse meshes were employed.

The contact area between sensor and object was simulated by contact elements. Since the mesa is expected to be much stiffer than object, TARGE169 element was used in meshing the mesa face and contact area on the object was meshed by CONTA175 element. To simulate the contact condition, the friction coefficients were examined in a range of 0.1-0.5 and the maximum shear stress for every simulation was considered to be 50 % of the object elastic modulus. For the different contact conditions, there was no clear difference in the results. This could be due to the fact that, the driving of sensor against object is more an indentation practice and no significant sliding phenomenon is present. Hence a friction factor of 0.5 and maximum shear stress equal to the 50 % of object elastic modulus was considered during all simulation cases.

Preliminary simulations were performed to observe behavior of the sensor stiffness and object stiffness with respect to different parameters. Results showed that the object stiffness in different indentation depths remains constant. This means that for a given elastic modulus, stiffness of tissue can be considered constant even in relatively deep indentation. Next the relationship between object elastic modulus and its stiffness was examined. Elastic modulus of the Silicone rubber was considered as the reference. While soft silicones, resembling a tissue similar to liver, have elastic modulus around 0.5 MPa, the range of 1.0-2.0 MPa can be considered as a common range for the elastic modulus of available silicone rubbers [101, 102]; on the other hand hard rubbers can have elastic modulus as high as 9.0 MPa [102]. Hence the elastic modulus of object was

set to vary in 1.0-7.0 MPa range. The object stiffness for three different mesa sizes was obtained as shown in Fig 3.9. The figure illustrates the object stiffness versus its elastic modulus for mesa diameters of 150, 250 and 350 μm . It shows that there is a linear relationship between the object stiffness and its elastic modulus. Any increases in the mesa size would increase the stiffness of.

Fig 3.9: Object stiffness variation respect to its elastic modulus.

In another attempt, variation of sensor stiffness respect to its deflection was monitored. Fig 3.10 illustrates the stiffness variation for sensors with two different elastic modulus. As in the figure, the overall stiffness of sensor shows a nonlinearity with respect to its deflection. The sensor stiffness for deflections up to 20 micron is quite constant and for the deflections of 60 micron only 8% of deviation respect to initial stiffness was observed. In result, the sensor during undergoing its deflection can provide a good constant stiffness.

Fig 3.10: Simulation result for the stiffness-deflection relationship of the sensor diaphragm.

But most importantly, the main objective of this simulation was to examine the performance of a given array of sensor that consists of two series of force sensing elements with different stiffness. Since in this simulation the interaction of only one sensor was considered, obtaining knowledge regarding the performance of different combinations of sensing elements was performed in two steps. In the first step, the deflection of sensors with different stiffness, when they come in contact with objects having different stiffness, was simulated. In the second step, these results were combined together. These two steps can be described as the following:

- **First step:** In order to have different combinations, sensor elastic modulus was considered to be 50, 100, 200, 300, 400 and 500 MPa once at a time. Next the deflection of a sensor, having one of these elastic moduli, was obtained when it comes in contact with an object that its elastic modulus varies in the range of 0.2-7.5 MPa. This means for every given object stiffness the deflections of sensors with different stiffness were obtained.
- **Second Step:** The sensor with elastic modulus equal to 50 MPa was considered as

the base *Measuring* element (with its stiffness as K_m); and the ratio of its deflections respect to stiffer sensors (K_r) were calculated. This means that *Reference* sensing element was considered to have 100, 200, 300, 400 and 500 MPa as its elastic modulus. This way a given sensor array consisted of two series of force sensing elements with different stiffness. The result was arranged as Fig 3.11. Each of the lines in this figure is the representative of a sensor array with a specific K_r/K_m and represents the performance of a sensor array which has a given K_r/K_m value. Horizontal axes is the ratio of object stiffness (K_{obj}) to *Measuring* (softer) force sensing element stiffness (K_m) and the vertical axes is the ratio of force sensing elements reading ($R = (\Delta - \delta_m)/(\Delta - \delta_r)$).

It is evident from Fig 3.11 that, sensor arrays with higher K_r/K_m ratios provide higher sensitivity in reading R . For larger values of K_m to K_{obj} higher reading in R is expected. This clearly indicates that, for having higher sensitivity in evaluation of object stiffness, the stiffness of *Measuring* (softer) sensing element should be higher than the object stiffness itself. These results are consistent with the results drawn from Fig 3.7 for the performance of the lumped model. Of course the reading values of these models are different.

Fig 3.11: Expected performance of different sensor arrays based on FEM analysis.

3.5 Summary

As information regarding force and stiffness in the tissue-grasper interface is valued highly in MIS, the sensor design concept for measuring both parameters was proposed. The design concept was described and the expected performance was discussed. It was shown that higher K_r/K_m value provides higher measurement sensitivity. At the same time, it is necessary to make sure that there is enough deflection and reading in the reference sensing elements. So there is need to find an optimized value for K_r/K_m ratio. Furthermore, larger K_m/K_{obj} values lead to further improvement in sensitivity and hence is recommended. It can be concluded that the design concept in this study can provide the desired functionality in measuring both tactile force and stiffness.

Chapter 4 FEM Simulation of Sensor Design

Following validation of sensor design concept from Chapter 3, it is necessary to analyze and simulate the force sensing elements in more details. This Chapter is dedicated for such analysis and simulation through which details of sensor performance can be understood. ANSYS is used to perform FEM simulations. This chapter starts with addressing structural design of sensor and follows strain distribution in the sensor structure, expected sensor performance, its stiffness, load capacity and extent of diaphragm deflection.

4.1 Structure of the sensor

Fig 4.1 illustrates the proposed design of a single force sensing element. Each sensing element consists of a square diaphragm with thickness ' t ' and span size of ' a '. A mesa with diameter ' D ' and height ' h ' is employed to transmit the external force to the diaphragm. Four independent piezoresistive elements are implanted on the diaphragm

edges. Following the application of external forces, the diaphragm would experience deflection; the deflection would appear as strain which leads to resistance change in the piezoresistive elements. It is understood that reference piezoresistive elements should be implanted besides the sensing elements but for the simplicity they are not shown in the figure.

As stated earlier in the report, the overall sensor would be an array of force sensing elements with two different stiffness values. Different stiffness can be obtained either by diaphragm thickness variation or by span variation. Considering constant thickness for diaphragm would require fewer fabrication steps and may facilitate sensor development. Hence it is preferred to tune the sensing elements stiffness by varying the diaphragm span size. As for the mesa diameter and height, in order to have simpler sensor design, it is also preferred to consider a fixed dimension size.

Following the literature review in Chapter 2 and Table 2.2, a resolution of 1000 μm was considered to be an achievable resolution for the proposed tactile sensing array, with polymer as its structural material. The sensing element diaphragm span was considered to vary within the range 300-700 μm . The sensing elements with the smallest span size are the *Reference* sensing elements and the ones with the largest span size are the *Measuring* sensing elements.

Fig 4.1: Schematic of a force sensing element with its geometric parameters.

4.2 Strain distribution in the sensor model

First simulation was the simulation of normal deflection in a quarter of the sensor model. Geometrically, the sensor was considered to have a diaphragm span size of $300\ \mu\text{m}$ and thickness of $20\ \mu\text{m}$ and mesa size of $75\ \mu\text{m}$ in diameter and $225\ \mu\text{m}$ in height. Two polymers with different material properties were considered; HD 4100 polyimide and SU8 2000. They are two of the most favorable polymers in the fabrication of polymer based MEMS devices. HD 4100 polyimide is the product of *HD Microsystems*

corporation and its elasticity modulus and Poisson ratio are 3.4 GPa and 0.33 respectively [27, 103]. As for SU 8 2000, a product from *Micro Chem* corporation, the elasticity modulus and poisson ratio are 4.02 GPa and 0.22 respectively [104]. The simulation results up to end of Section 4.4 are mainly general, so the results reflect the behavior for both of the polymers to a great extent. But for Section 5.5, SU-8 was used as structural materials to get mainly quantitative results.

For the boundary condition, nodes on all surfaces of the model (except the top surface i.e. diaphragm and mesa) were constrained from movement in the perpendicular direction to the surfaces. In order to apply external load all nodes on the top surface of mesa were coupled together first and the load was applied to one of them as a representative to all. Fig 4.2 shows the model meshed for this simulation.

Fig 4.2: Meshed model of a quarter of the sensor used in application of normal load.

Fig 4.3 illustrates equivalent *von Mises* strain distribution, within a model out of HD 4100 polyimide, when a -20 micron deflection was applied to the mesa in Z direction (downward). From strain contour on the bottom surface of the diaphragm and its thickness, it is evident that maximum strain distributions occur on the top and bottom surfaces of the diaphragm. Concerning the bottom surface of diaphragm only, it is clear that the maximum strain values occur at the areas of diaphragm which are close to the mesa and diaphragm edges. For a given mesa size strains, the strain near the mesa are greater than the edges.

The maximum equivalent *von Mises* strain in Fig 4.3 is above 6%. As the sensor is about to be developed based on a polymer as its structural material, the maximum possible strain in the sensor diaphragm should not be beyond the desired polymer yield strain. Dupont corporation has reported 3% strain as its criterion for determining the yield point for its Kapton HN polyimide films [105]. However, Micro Chem data sheet for SU-8 2000 has not reported the yield strain for this polymer and in return it has given value of 6.5 % as the elongation at break for it. The strain was decided to be confined at around 3 % during the simulations; thereby the maximum allowable normal deflection of the diaphragm has to be limited to 10 μm .

Strain distribution on bottom surface of the diaphragm (Fig 4.3) also shows that the strain distributions on the centerlines (symmetry planes) of diaphragm are higher than other positions. Hence implanting strain gauges in centerlines of the diaphragm can provide relatively higher sensitivity with respect to other locations. Positioning the strain gauges as close as possible to the symmetry lines is one of the necessary conditions for obtaining the highest possible sensitivity for the sensor.

Fig 4.3: von Mises strain distribution inside sensor model for -20 micron normal deflection.

In order to have clear demonstration of the strain distribution, its distribution on the symmetry plane of the sensor was extracted as shown in Fig 4.4. P0 and P4 are von Mises strain distribution on bottom and top line of symmetry plane respectively; While P1 and P3 illustrate the strain distribution $t/6$ deep inside diaphragm thickness with respect to the bottom and top surface respectively. P2 represents the equivalent strain distribution on the middle line of the thickness. This figure shows quantitatively that the maximum strain occurs on either top (P4) or bottom (P0) surface of the diaphragm thickness. Since obtaining maximum possible sensitivity is one of the major objectives in developing the sensor, it is necessary to implant the strain gauges as close as possible to the either surface of the diaphragm thickness.

Fig 4.4: Equivalent *von Mises* strain distribution on different thickness position in the symmetry plane of the diaphragm.

Fig 4.4 also demonstrates that the magnitude of strain in the nearby of mesa is larger than in the edges. Considering the fact that piezoresistive elements are supposed to be implanted on the diaphragm edges (Fig 4.1) it is favored to experience the highest strain on the edges. So it is required to investigate the conditions necessary to satisfy this preference.

4.3 Mesa size influence on strain distribution

The strain distributions are affected by varying dimensions in the sensor structure. For a given diaphragm span size, the mesa diameter is one of the parameters that could

significantly influence the strain distribution. In order to observe this influence the *von Mises* strain distribution on the bottom and top line of symmetry plane for the sensor diaphragm was investigated.

A sensor with 300 μm span size and 20 μm diaphragm thickness was considered. Von Mises strain distribution for mesa diameters from 75 to 150 μm was observed for a downward 10 μm deflection of the diaphragm. Fig 4.5 and Fig 4.6 illustrate the distribution on the bottom and top surfaces respectively. The horizontal axis starts from diaphragm edge and ends at sensor center. Based on Fig 4.5 and Fig 4.6, any increase in the mesa diameter leads to the increase in the strain values at the vicinity of diaphragm edge. This helps to have higher sensor sensitivity in the piezoresistive elements in the case of larger mesa diameters.

Fig 4.5: *von Mises* strain distribution on the BOTTOM center line of diaphragm for different mesa sizes.

Following the comparison of corresponding graphs in Fig 4.5 and Fig 4.6 for mesa

diameters of 125 and 150 μm , it is clear that maximum strain in the diaphragm edge nearby and the mesa nearby are almost the same. But for mesa diameter of 75 and 100 μm the maximum strain at mesa nearby is larger than the strain in the vicinity of diaphragm edge. As a result, making use of the relatively larger diameter for mesa can help to experience higher strains on the edges and improvement of the sensor sensitivity.

Fig 4.6: *von Mises* strain distribution on the TOP center line of diaphragm for different mesa sizes.

A fixed mesa diameter of 150 μm would be used in the case of all simulations hereafter. But the span size of diaphragm would be subject to change such that the 300 μm , which was considered so far, is going to be the smallest span size for sensing elements.

4.4 Force-deflection relationship

The force-deflection relationship during application of the normal force and lateral force to the mesa was investigated and the strain gauges' deflection was obtained. Finally performance of the sensor under both loadings was compared and discussed.

4.4.1 Normal Forces

The simulation was run for a model with 300 μm span size, 20 μm diaphragm thickness and mesa diameter 150 μm while it undergoes a maximum -7.5 μm deflection in Z direction. This magnitude of deflection was found to cause desired polymers to enter yield condition. The maximum force F_z supported at this deflection was 123.2 mN and diaphragm stiffness showed an average value of 16.838 N/mm . Fig 4.7 shows the relationship between mesa deflection and support force which exhibits a good linearity.

In practice, there is need to observe this linear relationship between mesa deflection and magnitude of resistance change in the strain gauges from. The magnitude of resistance change in a strain gauge element is proportional to the deflection observed in its length. To obtain this deflection in the length of a strain gauge, it is necessary to integrate the strain along its length. By increasing the measuring length of a strain gauge it observes larger resistance change when it undergoes a loading. Hence, serpentine shaped piezoresistive elements are used as it provides the possibility of increasing strain gauge length. It is proposed to implant these serpentine shaped strain gauges on positions with maximum possible strain. As shown earlier, the maximum possible strain occurs, at both bottom and top surfaces of diaphragm, on lines foremost close to symmetry planes perpendicular to diaphragm edges. Fig 4.8 illustrates the schematic of the serpentine shaped piezoresistive elements positioning on

one quarter of sensor structure. Considering $10\text{ }\mu\text{m}$ as the width of strain gauge [26], $30\text{ }\mu\text{m}$ offset distance was considered between AB and DC lines on both the bottom and top surfaces of the diaphragm.

Fig 4.7: Force-deflection relationship for diaphragm before reaching its yield point.

In order to determine appropriate magnitudes for the parameters ‘e’ and ‘f’ it is necessary to monitor the trend of strain distribution on lines with $15\text{ }\mu\text{m}$ and $45\text{ }\mu\text{m}$ distance away from symmetry planes. Fig 4.9 illustrates such a distribution for these lines on both bottom and top surface of diaphragm for the above model. The horizontal axis is representative of the horizontal position from point M to N in Fig 4.8 and the value 100 on the axis represents beginning of diaphragm span. Following the distribution of strain XX for the lines on top surface shows that the strain has sign change at a point $35\text{ }\mu\text{m}$ away from diaphragm edge. While for the same lines on bottom surface, this happens at $40\text{ }\mu\text{m}$ away from diaphragm edge. So for the lines on top and bottom surface the value of ‘e’ needs to be set as $35\text{ }\mu\text{m}$ and $44\text{ }\mu\text{m}$ respectively. The magnitude of

‘f’ for both the cases was considered to be $40\text{ }\mu\text{m}$.

Fig 4.8: schematic view of the sensor quarter and geometric parameters of its strain gauges.

In order to calculate the deflection in a strain gauge with the prescribed geometry, the strain component along each line segment was integrated. This means that for the lines AB and CD in Fig 4.8 integration was performed with respect to component XX of the strain but for the lines BC and DE it was necessary to carry out the integration with respect to component YY of the strain. As a result, the total magnitude of deflection between two ends of a serpentine strain gauge versus mesa deflection was obtained, Fig 4.10. It is evident that there is a good linear relationship between the magnitude of mesa deflection and the deflection in the length of the serpentine strain gauge when it is implanted on both the bottom and top surface of sensor diaphragm.

Fig 4.9: Strain XX distribution along two horizontal lines on both top and bottom surface of the diaphragm for mesa diameter of 150 micron.

Fig 4.10: Serpentine strain gauge deflection, implanted on top and bottom surface, versus sensor deflection for mesa diameter of 150 μm .

4.4.2 Lateral Forces

In an attempt to simulate expected performance of the sensor when lateral forces are applied, a half model of the sensor structure was modeled as it is symmetrical on other half. Fig 4.11 depicts top view of the model schematic with three of its strain gauges. A $-20\text{ }\mu\text{m}$ deflection in the direction of X was observed to be sufficient to reach the yield point of the desired polymers. Following the application of the above deflection, an average stiffness of 1.273 N/mm was observed. The corresponding supporting force for this deflection was 52.23 mN .

Fig 4.11: Schematic of the sensor half model with its serpentine strain gauges.

Similar to the case of normal deflection, the magnitude of deflections in each of the given serpentine strain gauges had to be obtained. For this purpose the value of strain along their consisting lines was integrated again. It was assumed that the sensing elements were implanted on the top surface of the diaphragm only. Table 4.1 lists the values of the different sensing elements deflection, shown in Fig 4.11, for different mesa deflection steps. As it is seen from the table, the path ABCDE experiences negative deflection while this is positive for the path FGHIJ. This is quite consistent with the fact that lateral deflection applied to the mesa was in the $-X$ direction. The

ratio of sensing elements deflection with respect to the mesa deflection for the lateral loading is -0.06828 and 0.08313 for the complete sensing elements of ABCDE and FGHIJ respectively (the values in the Table 4.1 were multiplied by 2 to obtain reading of complete element). As a much more important matter, the strain gauge implanted in the MNOPQRST path is expected not to be affected from the loading in the horizontal direction. Simulation results for the path MNOPQRST are quite consistent with the expectation and show almost a negligible reading. So it is evident that the sensing elements arranged in a perpendicular manner to each other can decouple the lateral load force components effectively.

Table 4.1: the observed deflections along the serpentine elements shown in Fig 4.11 (μm).

Mesa Deflection (μm)	1.0	2.0	3.5	5.75	9.12	14.19	20.0
ABCDE	-0.038	-0.0756	-0.1313	-0.2132	-0.3323	-0.5057	-0.6824
FGHIJ	0.0384	0.0772	0.1360	0.2258	0.3636	0.5771	0.8313
MNOPQRST	0.0001	0.0005	0.0018	0.0050	0.0130	0.0317	0.0629

Fig 4.12 is the pictorial representation of data in Table 4.1. Again data of ABCDE and FGHIJ paths had to be multiplied by 2 in order to obtain the data for the full path of these elements. It is clear that not only there is a good linear relationship between the lateral deflection of the mesa and readings of the horizontal sensing elements; but also the reading of the vertically aligned strain gauge is very close to zero.

Fig 4.12: Serpentine strain gauges deflection versus lateral deflection of mesa.

4.4.3 Performance comparison of normal and lateral loadings

First of all, the maximum allowable deflection following which the polymer goes beyond its yield point changes from $-7.5 \mu\text{m}$ in the normal loading to $-20 \mu\text{m}$ in the lateral loading. The same argument applies for the maximum supporting force in these cases. The maximum available force in the normal loading is 123.2 mN . This value is 2.36 times higher than the supporting force in the lateral loading which is 52.23 mN . Based on these points it is obvious to expect much lower stiffness for lateral loading than the normal one. Indeed the stiffness of the sensor for the lateral loading is more than 13 times lower than it for the normal loading. Even comparing the ratio of strain gauges deflection to the mesa deflection in both loading cases shows that the sensitivity for lateral forces is almost 10 times less than it for the normal forces.

4.5 Sensor Characteristics

In order to have further understanding of influence of diaphragm thickness and its span size on the sensor characteristics a series of simulations was carried out on a quarter model of the sensor when it undergoes normal loading. Both the material properties and the mesa size were considered to be constant. As fabrication attempts after the simulations showed that realizing tactile sensing device out of SU-8 polymer as its structural material is more straight forward than its fabricating out of HD 4100 polyimide; the simulation results in this section were modified to reflect the behavior of SU-8 2000 as structural polymer. Mesa diameter too was considered to be 150 micron for all cases. Stiffness variation, supporting force capacity and maximum allowable deflection were investigated.

4.5.1 Stiffness dependence on the diaphragm thickness and span size

As stated earlier, the proposed tactile sensor consists of two groups of force sensing elements with different stiffness. Hence understanding of stiffness dependency on the variation of sensor structure parameters is important. For this purpose variation of sensor stiffness respect to the changes in sensing diaphragm thickness and its diaphragm span size was investigated. Fig 4.13 illustrates the trend of stiffness changes for three different diaphragm thicknesses when span size of the diaphragm varies. The reported stiffness in the figure belongs to the normal deflection of the diaphragm and the maximum considered deflection was $7.5\ \mu\text{m}$.

Since Fig 4.13 depicts a qualitative representation of stiffness variation with respect to the diaphragm thickness and span size, Table 4.2 was arranged to list the

same data in a tabular manner. For a sensor diaphragm thickness of 20 μm moving from 300 μm to 700 μm span size provides more than 10 times reduction in the sensor stiffness. This value for the diaphragm thickness 35 μm and 50 μm is 6.65 and 4.5 respectively. So it is evident that for a given diaphragm thickness, it is possible to tune diaphragm stiffness by varying its span size. The possibility of tuning stiffness independently from its thickness can play a great role in facilitating sensor development by reducing number of fabrication steps in practice.

Fig 4.13: Diaphragm stiffness variation with respect to its span size and thickness for SU-8 polymer.

Table 4.2: Diaphragm stiffness in the case of different thicknesses and span sizes for SU-8 (N/mm).

Diaphragm thickness (μm)	Diaphragm span size (μm)				
	300	400	500	600	700
20	21.52	9.08	4.94	3.03	2.08
35	56.75	30.13	18.17	12.10	8.53
50	89.97	57.25	38.35	27.09	20.05

4.5.2 Force capacity dependence on the diaphragm thickness and span size

Supporting force capacity is another important characteristic of a force sensing element. Table 4.3 lists the maximum force supported by a single force sensing element before undergoing SU-8 yield point. It is clear that for a given diaphragm thickness the force capacity decreases with the increase in the span size. Also any increase in diaphragm thickness provides higher supporting force capacity.

It is necessary to have an evaluation that whether the considered geometry parameters in this simulation can satisfy the load requirements during MIS process. Again from the attempts towards fabrication of MEMS device it was found out that, a tactile sensing array with diaphragm span sizes of 300 and 700 μm and a sensor resolution of 1000 μm can be realized. For this purpose, first the diaphragm thickness 50 μm is considered. For such an array, based on data from Table 4.3 the average value of 668.2 and 300.0 mN would be a rough estimate of its load supporting capacity. This average which is 484.1 mN should satisfy the expected values in the Table 3.1. The value is still larger than 400 mN which was expected to be secured for the maximum 400 KPa of pressure in the tissue-grasper interface. Even if a diaphragm thickness of 35 μm be considered, following the same discussion, the maximum supported force would be 303.4 mN . This capacity value is sufficient to secure only a 300 KPa pressure as reported in Table 3.1. In short the simulation results show that the MEMS device dimension values which are realizable in practice can secure grasping pressures of around 480 and 300 KPa for 50 and 35 μm thick diaphragms respectively.

Table 4.3: Supporting force capacity in the case of different thicknesses and span sizes (*mN*).

Diaphragm thickness (μm)	Diaphragm span size (μm)				
	300	400	500	600	700
20	157.5	133.1	117.9	106.2	100.7
35	417.2	325.4	253.4	249.7	189.6
50	668.2	508.4	416.5	403.4	300.0

4.5.3 Maximum allowable deflection

As it might be expected, while dealing with sensing diaphragms of different thickness and span size, the maximum allowable deflection before facing yielding varies from one case to another. Table 4.4 represents the maximum allowable deflections corresponding to the each of supporting forces in Table 4.3. Based on the table, for the 20 μm diaphragm thickness the deflection in 700 μm span size may be 4 times larger than it in 300 μm span size. But for the 50 μm thickness it would not be more than twice.

Table 4.4: Max. allowable deflection of the diaphragms for different thicknesses and span sizes (μm).

Diaphragm thickness (μm)	Diaphragm span size (μm)				
	300	400	500	600	700
20	7.5	14	20	25	30
35	7.5	11	14	20	21
50	7.5	9	11	15	15

4.6 Summary

A detailed study on the proposed force sensing elements was carried out through its simulation in ANSYS. The structural design of sensor, strain distribution within the sensor structure, sensor stiffness and load capacity were studied. For the proposed sensor structure, the maximum equivalent strain occurred on symmetry planes of the sensing diaphragm where they meet diaphragm at its surfaces. So for obtaining maximum possible sensitivity, it is necessary to implant the strain gauges as closer as possible to the either surface of the diaphragm thickness. Any increase in the mesa diameter intensified strain concentration on the diaphragm edges. This fact can be used for improving the sensor sensitivity to some extent. For both normal and shear loadings, a linear relationship was present between applied force and strain gauges deflection. It was found out that, for a given diaphragm thickness it is possible to tune diaphragm stiffness by varying its span size. The possibility of tuning stiffness independently from its thickness can play a great role in facilitating sensor development and may help to the reduction of necessary fabrication steps in practice. Simulation results also showed that, the dimensions of MEMS device, which could be realizable in practice are able to secure grasping pressures of around 480 and 300 KPa for 50 and 35 μm thick diaphragms respectively. Such pressures are quite consistent with the pressures present during grasping of tissues. Hence, the stiffness and force requirements for sensor performance can be met by choosing an appropriate polymer material and determining suitable size for both the span and thickness of sensing diaphragms.

Chapter 5 Macro Tactile Sensor Set up

Following the simulation of sensing concept, sensor design and discussion on its performance in previous chapters, it was decided to validate the sensing concept and it's functioning experimentally. This could help to pave the way in developing the MEMS sensor prototype in the later steps. The experimental validation includes both force measurement and stiffness evaluation. For this purpose, a macro tactile sensing setup was designed, fabricated, calibrated and used for data gathering. Finally the results were presented and analyzed and corresponding conclusions were drawn.

5.1 Design of Macro Tactile Sensor

The main element of this macro tactile sensing setup is a flexural mechanism as shown in Fig 5.1(a). In the design, a strain gauge was installed in the flexing point of each sensing element. Following the exertion of an upward force at the tip of sensing

element (contact area), the measure for the tip deflection appears as voltage change across the poles of strain gauge. By obtaining the deflection-voltage and force-voltage relationship for sensing elements the sensor can be calibrated. By having two force sensing elements next to each other but with different measuring arms ($a1 \neq a2$), as shown in Fig 5.1 (a), a stiffness sensing unit comprising two sensing elements with different stiffness can be realized. This design is to measure only one component of a general force vector (normal component) but can be used for validation of the stiffness sensing concept proposed in this study.

The bottom view of the macro tactile setup is shown in Fig 5.1(b) which is a 2*5 array of sensing elements with different stiffness. The different numbers beside each sensing element shows the corresponding length of sensing arm for that specific sensing element (dimension $a1$ and $a2$ in Fig 5.1 (a)). It worth noting that sensing elements with longer arms are softer compared to the ones with shorter arms that are stiffer.

Fig 5.1: Design of the Macro Tactile Sensor: (a) Side view of two sensing elements with different measuring arm length; (b) bottom view of macro tactile sensing array.

This design was analyzed in ANSYS. Fig 5.2 illustrates an example of meshed model of the sensing element which includes the boundary condition and applied force to the sensing element tip. A given distributed external force equal to 2 N was applied at the tip while the thickness in the flexing point was varying from 0.15 mm to 0.4 mm. The material was considered to be Aluminum with elastic modulus of 69000 MPa and poison ratio of 0.33. Plane stress condition was used for the simulation and model

thickness was assumed to be 3 *mm*. The flexing point arm length varied from 4-8 *mm*. In result the deflection at bottom right corner of the model and strain in the surface of flexing point was obtained. These results have been summarized as Table 5.1. At the cross point of a given flexing point thickness and its arm length in Table 5.1 three values have been reported. The top value belongs to the deflection and its unit is *mm*, the middle one is the percent of the maximum strain in the external surface of flexural point. The strain at external surface is expected to be monitored by the strain gauge later. The third one is the stiffness provided by the corresponding sensing element in *N/mm* which is the ratio of exerted force over the deflection in the sensor tip.

Fig 5.2: The meshed FEM model of the sensing element in ANSYS.

The sensing element with the flexing point of 0.25 mm in thickness, compared to the ones with 0.15 and 0.2 mm in flexing point thickness, could provide more mechanical strength; while it is handled in different stages of the fabrication, calibration and measurement processes. It is softer than the sensing element with 0.4 mm thickness in flexing point and this makes it favorable in dealing with soft objects. In result the flexing point of 0.25 mm in thickness was considered as the preferable thickness for the sensing elements in the fabrication stage. For such thickness in the flexing point the strain varies from 0.2711-0.6316 %, for different arm lengths. As strain gauges are very sensitive devices such a range of strain is a considerable value to be measured.

Table 5.1: Simulation results for the Macro Model Sensor following the exertion of 2 N vertical force. In each cross line top number represents tip deflection in mm, middle one is strain value in % and bottom one is sensor stiffness in N/mm.

			Flexing Point Arm Length (mm)				
			4	5	6	7	8
Thickness of flexing point (mm)	0.4	Def. (mm)	0.018747	0.033547	0.053243	0.077853	0.107547
		Strain (%)	0.107967	0.142567	0.1777	0.212033	0.2482
		K (N/mm)	106.6856	59.61844	37.56339	25.68933	18.59658
	0.25	Def. (mm)	0.051873	0.098263	0.161087	0.23899	0.332987
		Strain (%)	0.2711	0.360633	0.4518	0.5378	0.631567
		K (N/mm)	38.55546	20.35347	12.41568	8.368551	6.006246
	0.2	Def. (mm)	0.086523	0.167387	0.273793	0.4081	0.570933
		Strain (%)	0.4225	0.5602	0.695967	0.829933	0.976267
		K (N/mm)	23.11515	11.94838	7.30478	4.90076	3.503036
	0.15	Def. (mm)	0.172493	0.337467	0.559733	0.836167	1.1677
		Strain (%)	0.753267	1.0355	1.2611	1.510467	1.762833
		K (N/mm)	11.59465	5.926511	3.57313	2.391868	1.712769

5.2 Fabrication of the Macro Tactile Setup

The detailed drafting of the mechanical parts for the macro tactile setup was performed in the AutoCAD software. The macro tactile setup consisted of sensing elements, the holder for the sensing elements and the clamping part to fix sensing elements on the holder by screws. The sensing element itself, as shown in Fig 5.3, was fabricated by wire-cut process; because during wire-cut process almost no cutting force is present so cutting the flexing point of sensing element could be conducted easily. Fig 5.4 shows the whole macro tactile setup after its fabrication and assembly which includes the holder of sensing elements and its clamping parts. Of course the bonding of strain gauges and its wiring should be done before the setup could be put in function. Fig 5.4 also shows the translational axes of the setup which were used during the calibration of the sensing elements later.

Fig 5.3: The Macro Tactile Sensing Elements fabricated through wire-cut process.

Fig 5.4: The whole mechanical setup after its fabrication and assembly; including macro tactile setup and its carrying axes.

Fig 5.5 shows full setup of the macro tactile sensing system which includes all the mechanical, electrical and read-out parts. In the left hand side of Fig 5.5, the macro tactile setup following the bonding of strain gauges on sensing elements and soldering of the connection wires is visible. The right hand side of this figure illustrates the read-out and amplification circuits for the sensing elements signals.

Fig 5.5: Macro Tactile Sensor Setup after bonding of strain gauges on sensing elements and soldering of the connection wires; together with schematic of read-out circuit.

As Fig 5.5 shows, each sensing element out of 10 sensing elements in the tactile array was set up in the form of a quarter bridge configuration. A 120 Ω strain gauge, product of TML[106], was bonded on each sensing element and besides three other reference strain gauges formed a Wheatstone bridge. A single MOSFET device was used to isolate the connection between the zero line of the power source and the ground of all the 10 bridges. In this way current flow into strain gauges was avoided; and only during the read-out moment a signal coming from a digital output of DAQ device was activating the MOSFET and current could flow into the zero line. This measure protected strain gauges from overheating. The signals of quarter bridges were amplified 100 times by AD621 instrumentation amplifiers[107]; which benefits from internal resistors for gain setting. The amplified signal was acquired by a 16 bit DAQ device (NI USB 6258) through Lab View program. For reading the signals coming from all 10

sensing elements, 10 analogue input lines of DAQ device were used. In every measurement attempt, the program was set to take 10 samples of signals and provided their average as the representative value for that particular signal.

5.3 Calibration of Macro Tactile Sensors

The calibration of sensing elements was performed by making use of a mechanical setup that has been fabricated and assembled for this purpose. The horizontal axes setup for carrying the macro tactile array during calibration process was already shown in Fig 5.4. This setup was used to translate sensing elements in two X and Y directions as well as to rotate the macro tactile array around the Z (vertical) axis. This way each sensing element could be positioned accurately beneath the force sensor tip. The force sensor itself was installed on a Z axis and was pointing downward as shown in Fig 5.6. In order to have a high resolution for Z axis in displacing the force sensor, the original micro meter head of the linear axis has been replaced with a differential *Opto Sigma* micro meter head. This way the resolution of displacement in the Z axis was increased from 10 μm in original micro meter head to 0.5 μm in the case of *Opto Sigma* micro meter head. The commercial force sensor used in this study had a Max. 5 N loading capacity and senses the force along its vertical axis and benefits from a 0.001 N resolution in measuring the force.

The calibration process was aimed to get both the Force-Deflection and the Signal-Deflection relationships for each sensing element. Before applying any deflection, the initial signal or zero deflection signal of the sensing elements was collected first. Next the vertical axis was displaced 25 μm in every step and totally for 6 steps. So the displacement of vertical axis was totally 150 μm in the case of all sensing

elements. The commercial force sensor itself utilizes a stiffness based load cell and its stiffness was found to be around 57.044 N/mm . This means some of the applied $150\text{ }\mu\text{m}$ displacement was absorbed by the force sensor itself. Indeed the force sensor and the macro sensing element behave like two springs arranged in series. So to find the deflection of each macro sensing element the deflection in the force sensor was deducted from the $150\text{ }\mu\text{m}$ displacement.

Fig 5.6: Vertical linear axis of the calibration setup used in this study.

Fig 5.7 shows the performance graphs for one of the sensing elements. To obtain these graphs the above mentioned procedure was used. Every measurement was conducted at least 3 times for a given sensing element and then the collected data were averaged to have the representative graph. Fig 5.7(a) shows the applied force (N) versus sensing element deflection (μm). The slope of this graph represents the stiffness of the sensing element which accounts to $40.1 N/mm$ in this case. Fig 5.7(b) depicts the Signal (V) versus the applied deflection (μm). To have applied force (N) out of the sensing element Signal (V), the governing relationship in Fig 5.7(b) was combined with the one in Fig 5.7(a). So by obtaining these graphs, the relationship between Signal with both the force and the deflection was explicitly obtained. Fig 5.7 shows that there is a linear relationship between Force-Deflection and Signal-Deflection. This linear relationship for sensing elements was subject to an error in the range of 1.15-2.32 %, however.

Table 5.2 summarizes the calibration results for all sensing elements among which the sensing element 5 was not functioning. By having these data from Table 5.2, the deflection in each sensing element, Δ , can be calculated as:

$$\Delta = (1/K_v) * (V - V_0) \quad (\text{Eq. 4.1})$$

In which K_v is the slope of Fig 5.7(b) for each of the sensing elements. V_0 is the initial signal (zero deflection signal) and V is the current reading of signal (V). Next by having the value for Δ , the applied force to the sensing element can be obtained as:

$$F = K * \Delta \quad (\text{Eq. 4.2})$$

In which K is the stiffness (slope of Fig 5.7(a)) of the corresponding sensing

element. Hence by replacing equation Eq. 4.1 into Eq. 4.2 the force can be calculated as:

$$F = (K/K_v) * (V - V_0) \tag{Eq. 4.3}$$

(a)

(b)

Fig 5.7: An example of calibration graphs for sensing elements; (a) Force (N) versus Deflection (μm), (b) Voltage change (V) versus Deflection (μm).

Table 5.2: Calibration results for all sensing elements.

Sensor	K (N/mm)	Maximum Deflection (μm)	Maximum Support Force (N)	Maximum Signal Change (V)	Kv (V/μm)	Kv/K (V/N)	V0 (V)	Flexing Arm Length (mm)
1	40.1193	74.26	2.96	0.547	-0.0073	-0.18226	-1.0076	4
2	11.5667	118.92	1.41	0.588	-0.0049	-0.42483	0.0816	7
3	14.9060	103.69	1.55	0.367	-0.0036	-0.23890	-0.0857	6
4	5.7808	143.11	0.85	0.534	-0.0038	-0.65161	-2.1409	8
5*	xxx	xxx	xxx	xxx	xxx	xxx	xxx	xxx
6	7.3523	135.19	1.04	0.626	-0.0046	-0.62264	0.1002	8
7	15.2523	108.73	1.78	0.403	-0.0037	-0.24099	-1.4692	5
8	80.3840	50.63	4.33	0.471	-0.0090	-0.11175	0.1959	4
9	15.0530	105.3	1.64	0.635	-0.0060	-0.40075	0.0528	5
10	9.0479	123.55	1.20	0.455	-0.0037	-0.40769	-0.3263	7

* Sensor 5 was blind (not available!)

In Table 5.2, the second column lists the stiffness of the sensing elements and the last column contains the flexing arm length for the sensing elements. These two columns show that, simply by having sensing elements with different flexing arm in length, a stiffness ratio of more than 10 can be achieved. The stiffness ratio of the stiffest sensing element, sensing element 8, to the softest sensing element, sensing element 4, in Table 5.2 is 80.384/5.7808 which is equal to 13.9 times; while the flexing arm of the sensing elements varies only from 4 mm in the case of the stiffest one to the 8 mm in the case of the softest one. Three out of nine functioning sensing elements (the sensing elements 3, 7 and 9) have very close stiffness values. Hence in order to avoid redundancy, the results of only one of them are put into account. This way there would be totally seven different sensing elements as it comes to their stiffness.

The maximum deflection column represents the highest deflection that a specific

sensing element went through during its calibration. All of the sensing elements except the two stiffest ones have experienced deflections more than $100\ \mu m$. It is assumed that such level of deflection is sufficient in calibrating the sensing elements. The values in this column is the deflection absorbed in sensing elements following a total $150\ \mu m$ displacement which was introduced by the vertical axis of the calibration setup. This means that the remainder of this value has been absorbed by the commercial force sensor itself.

The maximum signal changes for all the sensing elements, irrespective from the applied deflection and force, shows readings of the same order. This is a positive indication that it is possible to have considerable and comparable signal reading for all the sensing elements; either the stiffest one(s) or the softest one(s). In other word, there may not be any concern on the availability of sufficient signal change in the design of tactile sensor arrays.

Sensitivity of sensing elements is one of the key factors in their performance. In this study two different sensitivity parameters are present. First one is the sensitivity of signal respect to the applied deflection. The column K_v in Table 5.2 contains this sensitivity parameter which has been presented as $V/\mu m$. The highest sensitivity of signal to deflection belongs to the stiffest element(s) and the lowest signal sensitivity to deflection belongs to the softest element(s).

During grasping of tissues, the pressure in the tissue-grasper interface would be directly translated as the force applied to the sensing elements. So the sensitivity of signal respect to the applied force is crucial for the performance of the tactile sensors. Hence this sensitivity is considered to have higher importance than the signal-deflection sensitivity in designing the tactile array. The column K_v/K in Table 5.2 represents this sensitivity parameter. Comparing K_v/K of different sensing elements,

it is obvious that softer sensing elements have higher signal to force sensitivity than stiffer ones. For example for the softest sensing element this value is -0.65161 V/N while for the stiffest sensing element it is -0.11175 V/N . The increasing trend in the K_v/K sensitivity parameter, while moving from stiffest sensing element to softest one, is in opposition to the trend in the case of K_v sensitivity parameter. As stated earlier, in the case of K_v sensitivity parameter the stiffer the sensing element the higher the K_v sensitivity parameter.

5.4 Rubber Sample as Artificial Tissue

Liu H. *et al* [108] in their study on identifying tissue abnormality by a rolling indentation probe have provided the measurement results for live tissues of heart and kidney as well as RTV6166 silicone gel. Their results show a stiffness of some 0.4 N/mm for heart tissue and 0.2 N/mm for kidney tissue and the silicone gel. In our study, however, it was required to have a number of different tissue phantoms in which a considerable range of stiffness could be present. A series of rubber samples was considered for such tissue phantoms. The rubber samples found to provide an acceptable range of stiffness. However, the magnitude of this stiffness, is clearly higher than live tissues.

For measurement of rubber samples stiffness a mechanical setup was designed and fabricated. Fig 5.8 shows the stiffness measurement arrangement and the consisting components of the mechanical setup separately. As Fig 5.8 (a) shows, the rubber sample is positioned in the clamping area of a wheeled cart. Making use of the wheeled cart is to make sure that only vertical forces are applied in the force sensor and rubber sample interface. Next a vacuum pump – through the black hose- starts to pump out the air inside the air suction channel. This way the rubber sample is held firmly in place. Then

the force sensor is displaced toward the rubber sample in an orderly manner. This was to obtain the force-deflection relationship for the rubber samples. Fig 5.8 (b) shows the contact apparatus in this measurement. The contact area of the apparatus was in square shape of 2 mm by 3 mm. This area is the same as the contact area considered for the tactile sensing elements. Hence it is realistic to expect the consistency in the stiffness measurement through both this setup and the tactile sensing elements later.

Fig 5.8: (a) the mechanical setup used in measuring the stiffness of rubber samples; (b) the contact apparatus to come in contact with the rubber samples; (c) Vacuum wheeled clamping cart.

In the measurement stage, 15 different rubber samples were investigated to see

whether they provide the necessary stiffness range for the study. Each rubber sample's stiffness was obtained by measuring its force-deflection relationship. In each measurement the force sensor was displaced 300 μm towards each rubber sample; in steps of 50 μm . This amount of deflection is considered to be a significant magnitude. Each measurement was performed 3 times for each rubber sample and the readings were averaged. Fig 5.9 shows one of the force-deflection graphs; and in the case of all rubber samples the force-deflection relationship found to be linear; and its linearity was subject to an error in the range of 0.72-1.9 %. However as it was noted that hysteresis could be a shortcoming for such rubber samples, a static loading procedure during obtaining rubber samples stiffness was adopted. This was aimed to avoid the hysteresis characteristics of the samples as much as possible.

Fig 5.9: Force-Deflection relationship for one of the rubber samples resembling the tissue.

Eight out of 15 rubber samples were found to have distinct different stiffness. Fig 5.10 shows a photo of the series of 8 rubber samples. The measurement results for these rubber samples are summarized in Table 5.3. The rubber samples stiffness varied in the

range of 2.34 ~ 10.01 N/mm and this means that the stiffness ratio of stiffest rubber sample to the softest one is some 4.28. This ratio is assumed be an acceptable wide range in having rubber samples with different stiffness.

Fig 5.10: The series of final 8 rubber samples in providing different stiffness values.

Comparing the data of Table 5.3 with the one in Table 5.2 provides the ground to discuss on the available stiffness combination between sensing elements and the rubber samples. Based on Table 5.2, the stiffness of sensing elements present in the tactile sensing array varies from 5.78 N/mm to 80.384 N/mm . The stiffness 80.384 N/mm represents the stiffness of *Reference* sensing element K_r ; and the range 5.78-40.1193 N/mm is the range that stiffness of the *Measuring* sensing elements, K_m , varies within. Comparing the range of *Measuring* sensing elements stiffness with the one provided by rubber samples in Table 5.3, the ratio of stiffness for the stiffest *Measuring* sensing element, $K_m=40.1193\ N/mm$, coming in contact with softest rubber sample is $40.119/2.34=17.145$. While the same ratio for the softest sensing element to the stiffest

rubber sample is $5.78/10.01= 0.577$. Considering the ratios range 0.577-17.145, the available combinations of the sensing elements and rubber samples, as far as their stiffness is concerned, cover a wide enough range. This can provide the possibility of drawing general conclusion in verifying the validity of the sensing concept; which was proposed in this study.

Table 5.3: Measurement results for force-deflection relationship of rubber samples.

<i>Row</i>	<i>Rubber Sample No.</i>	<i>Max Deflection (μm)</i>	<i>Max Support Force (N)</i>	<i>Stiffness (N/mm)</i>
1	1	274	2.757	10.01
2	3	293	1.857	6.28
3	4	312	1.43	4.51
4	5	310	1.185	3.85
5	7	357	0.842	2.34
6	9	319	1.725	5.41
7	13	344	1.24	3.57
8	15	302	2.188	7.26

5.5 The evaluation of rubber samples’s stiffness by macro sensing elements

Following the calibration of sensing elements and obtaining rubber samples’ stiffness, the stiffness of rubber samples was evaluated by the macro tactile array and the results were compared with the corresponding stiffness values obtained by the force sensor earlier. For this purpose another mechanical setup was designed and fabricated. Fig

5.11 shows this setup in which the macro tactile array was fixed on a linear axis; to be able to displace vertically. The rubber sample is fixed on the wheeled cart. The cart itself sits on the base plate through four ball bearings; back to Fig 5.8. In this way no horizontal force component can be created in the interface between macro tactile array and rubber sample. The rubber sample itself is held firmly on the mobile cart by providing vacuum clamping which is created by a vacuum pump through a pneumatic hose. Standard weights of 100 g were used to implement the normal (grasping) force in the macro tactile array and rubber sample interface. Of course before using the weight gauges, they were calibrated by the commercial force sensor.

Fig 5.11: The mechanical setup used for the grasping of the rubber samples by the macro tactile sensor array.

To collect the data for the rubber samples' grasping, each of the 8 rubber samples was picked one by one. The chosen rubber sample was located on the central part of the cart's top surface where the air suction channel for the vacuum clamping has been created; then the vacuum pump started to work and clamped the rubber sample in its position firmly. Next the macro tactile probe, installed on the vertical axis, was brought to contact with the rubber sample gently. The first value of the normal force between the macro sensor array and the rubber sample was equal to the weight of the

vertical axis itself. This weight included the weight of the linear carrier, the weight of the vertical bar and the weight of the macro tactile array itself which all together added up to 3.235 *N*. In order to increase the normal force, 100 *gr* weights were added to the previous loading values.

First data value was collected while there was no loading between macro tactile array and the rubber sample. This was necessary to have the signal values of sensing elements under zero normal force. But the next data collections were conducted following every corresponding application of loading. That means right after the macro tactile array came in contact with the rubber sample and also after every 100 *gr* increase in the amount of the loading. As for the data acquisition, the program in the Lab View software was designed to collect 10 data sample in every measuring attempt; and calculate and save their average as the representative value for every measurement attempt.

Fig 5.12 is an example of a set of eight graphs which illustrate the collected data during the grasping of rubber samples. This figure belongs to the data obtained during the grasping of rubber sample 5 in Table 5.3 with the stiffness of 3.85 *N/mm*. To obtain this graph, the readings in deflection of all sensing elements, excluding the stiffest one, were divided over the deflection in the stiffest sensing element. As mentioned earlier, K_r is used to represent the stiffness of the stiffest element (the *Reference* element) and K_m is used to represent the stiffness of the rest of elements (the *Measuring* ones). So Fig 5.12 is the illustration of the ratio of deflection in the *Measuring* elements (with the stiffness K_m) to the deflection in the *Reference* element (with the stiffness of K_r). The horizontal axis is the applied grasping (normal) force; and the vertical axis represents the deflection ratio of $R = (\Delta\delta_m) / (\Delta\delta_r)$. $(\Delta\delta_m)$ is the deflection occurred in the *Measuring* elements; and $(\Delta\delta_r)$ is the deflection of the *Reference* element. As it is clear

from the legend, Fig 5.12 shows the R values for different quantities of K_r/K_m . Since $K_r > K_m$ hence the R is always bigger than 1.0.

Fig 5.12: The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=3.85\text{ N/mm}$.

The most important conclusion that can be drawn from the graph (and other graphs of the same series) is that, the R ratio for different values of grasping force remains greatly constant. This is true for of all K_r/K_m combinations as shown in Fig 5.12. This means that a set of tactile elements which consists of a fixed value for both K_r and K_m as the stiffness of sensing elements would provide a constant reading for the deflection ratio, R , even if the grasping force changes. Hence, the deflection ratio of the tactile sensor array, R , is independent of the change in the grasping force present at the interface of macro tactile sensor and the rubber sample.

The information available as the set of eight graphs, similar to Fig 5.12, can be

used to get more conclusive information and data in the form of a table or another graph. Such table and graph extracted from the experimental data could be derived and presented in the same way as the simulation results on the sensor performance was presented in Chapter 3 (Fig 3.11). This way the validation of the sensing principle can be investigated clearly. This means, the table and the graph resulted from it should illustrate the changes in R as the stiffness of the rubber sample increase from the lowest available value to the highest one for different K_r/K_m .

As R for different values of grasping force remains greatly constant, each of the eight graphs can be summarized as six average values of R for different K_r/K_m . By sorting R values starting from the softest rubber sample to the stiffest one, the whole trend for the change of R for each K_r/K_m can be obtained. Every graph of R for a given K_r/K_m consists of eight data points; and there would be six graphs for six different K_r/K_m .

Table 5.4 contains the data that have been obtained and arranged in the above mentioned manner. Based on Table 5.2, the stiffest sensing element in the macro sensor array has a stiffness of 80.38 N/mm . This sensing element was considered as the *Reference* sensor and the other eight sensing elements softer than it were considered as the *Measuring* sensors. As stated earlier, the stiffness of the *Reference* sensor was shown by K_r ($=80.38$) and the stiffness of the *Measuring* elements by K_m . The values of K_m were listed as the first column in Table 5.4 and the second column shows the all available combination of the *Reference* sensor respect to the *Measuring* ones in the form of K_r/K_m . This way the K_r/K_m was arranged in an ascending order. It was expected to have eight possible combinations for K_r and K_m as there are nine functioning sensors in the sensing array. But since as mentioned before the sensing elements 3, 7 and 9 have almost the same stiffness in Table 5.2, one of these sensing elements, sensing element 7,

combined with the *Reference* element and total number of K_r/K_m ratios was reduced to six.

Table 5.4: The readings of Macro tactile sensors in grasping the rubber samples with different stiffness

($K_r=80.38\text{ N/mm}$).

K_m (N/mm)	K_r/K_m	K_{obj} (N/mm)							
		10.01	7.26	6.28	5.41	4.51	3.85	3.57	2.34
40.12	2.004	K_m/K_{obj}	4.083	5.541	6.470	7.412	9.008	10.610	11.226
		R	0.977	0.840	0.830	0.822	1.002	1.004	0.942
15.25	5.270	K_m/K_{obj}	1.552	2.107	2.460	2.818	3.425	4.033	4.268
		R	1.729	1.816	2.001	2.002	2.174	2.283	2.186
11.57	6.950	K_m/K_{obj}	1.177	1.597	1.865	2.137	2.597	3.059	3.236
		R	2.134	2.402	2.410	2.320	2.787	2.757	2.852
9.05	8.884	K_m/K_{obj}	0.921	1.249	1.459	1.672	2.0312	2.393	2.532
		R	2.938	3.002	2.993	3.253	3.601	3.536	3.828
7.35	10.933	K_m/K_{obj}	0.748	1.015	1.186	1.358	1.651	1.944	2.057
		R	2.827	3.119	3.257	3.407	4.056	4.064	4.225
5.78	13.905	K_m/K_{obj}	0.588	0.798	0.932	1.068	1.298	1.529	1.617
		R	3.142	3.324	3.480	3.534	3.994	4.037	4.162

K_{obj} represents the stiffness of rubber samples from Table 5.3 which was ordered in a descending manner. This way the K_m/K_{obj} ratios appeared in an ascending order. The focal part of Table 5.4 is the sub-tables which have listed the R values versus K_m/K_{obj} variation for a given K_r/K_m . In result, this table contains the experimental data for the performance of the macro tactile array.

Table 5.4 can be represented in a pictorial format as Fig 5.13. The horizontal axis represents K_{obj}/K_m and the vertical axis is $R=(\Delta-\delta_m)/(\Delta-\delta_r)$. Each of the six graphs

belongs to one of the K_r/K_m values in Table 5.4. This figure illustrates the outcome of experimental results for the macro tactile sensor array; and was represented in the same way as Fig 3.11. However the horizontal axis only covers values up to $K_{obj}/K_m=1.33$. This means that in most cases the *Measuring* elements stiffness, K_m , is higher than object stiffness, K_{obj} . This point was meant to provide higher sensitivity in sensing object stiffness (as it was proposed in Chapter 3). Fig 5.13 shows by moving towards data points of higher K_m/K_{obj} (lower K_{obj}/K_m), higher sensitivity for R is achieved.

To ease the comparison between Fig 5.13 and Fig 3.11, their horizontal axis (instead of K_{obj}/K_m) can be recast based on K_m/K_{obj} . This helps to cover a wider range in the horizontal axis. Fig 5.14 (a) is the result of applying such change in horizontal axis on Fig 5.13. Fig 5.14 (b) is the replication of Fig 3.11, FEM simulation results from Chapter 3 for K_m/K_{obj} . Comparing Fig 5.14 (a) with Fig 5.14 (b), experimental results follow the same pattern as the simulation results. Though, R values of the macro tactile array are lower than the ones from FEM simulation.

Based on Fig 5.14 (a), any increase in K_m/K_{obj} (moving from stiffer rubber samples to the softer ones) leads to the increase in R for all the K_r/K_m combinations in the tactile array. Considering a given K_m for a tactile sensor array, the softer the object the higher the reading signal of the *Measuring* element compared to the *Reference* one. This shows the capability of the proposed design concept in evaluating the stiffness of objects.

Fig 5.13: Performance of Macro Tactile Sensor in grasping rubber samples with different stiffness.

As Fig 5.14 (a) shows, there is almost no sensitivity in $K_r/K_m=2.0$ combination in evaluating the object stiffness. In the case of $K_r/K_m=5.27$, there is a clear sensitivity in evaluating the object's stiffness (there is an acceptable slope for the trend line). But it is from the $K_r/K_m=6.95$ combination onward that a good sensitivity in the evaluation of the object stiffness is observable; and even the highest sensitivities are belonging to the $K_r/K_m=8.9$, 10.9 and 13.9 . In short, higher K_r/K_m combinations provide better sensitivity in reading of R . So to improve the sensing sensitivity in evaluating the objects stiffness, the K_r/K_m combination for the tactile sensing array should be close to 10 and above.

(a)

(b)

Fig 5.14: Re-presentation of the tactile Sensor performance; a) experimental results, b) Simulation results from Fig 3.11.

The *Reference* element being the stiffest sensing element undergoes the lowest deflection among all the sensing elements. In evaluating the performance of macro sensor array, it was needed to make sure that the *Reference* element shows sufficient magnitude of deflection, $(\Delta\delta_r)$ during its performance. This is because $(\Delta\delta_r)$ appears at the denominator of the R ratio and a trivial reading for the denominator should be avoided. Otherwise it would lead to misestimate the object stiffness. As another point, the *Reference* element shows the lowest value for K_v/K ratio in Table 5.2; which is the sensitivity of signal respect to the applied force on the sensing element. Hence both the deflection, $(\Delta\delta_r)$, and signal reading in the *Reference* element was monitored.

The maximum signal change in the *Reference* element during grasping the rubber samples was in a range of $0.06 \sim 0.14 V$; and such a signal change is indicating of a $28.6 \sim 37.2 \mu m$ deflection range in the *Reference* element. This was while a signal change of up to $0.61 V$ and a deflection of $141.9 \mu m$ were experienced in some of softer elements. So comparing the signal reading in the *Reference* element with the ones in the *Measuring* elements emphasizes the above mentioned concern.

This means that as it is favorable to have high K_r/K_m to improve the sensitivity in the stiffness evaluation; there is need to have sufficient signal reading and deflection in the *Reference* sensing elements too. Otherwise the results regarding the evaluation of objects stiffness might be misleading.

5.6 Summary

A macro sensor array consisting of 10 sensing elements with different stiffness was designed, fabricated and calibrated. All the sensing elements showed a linear force-deflection relationship. It was found that sensing elements with different stiffness

can be realized simply by providing different measurement arms for them. This way a stiffness ratio of 13.9 between the stiffest sensing element, 80.384 N/mm , and the softest one, 5.7808 N/mm , was achieved. Rubber samples were found to provide an acceptable range of stiffness to resemble tissues during the experiments. The stiffness range of rubber samples was in the range of 2.34 ~ 10.01 N/mm . In result a wide range of stiffness combinations for sensor- rubber were available during the experiments. Such a range provided the possibility of drawing general conclusion in verifying the validity of sensing concept.

The results showed, when a specific K_r/K_m combination in the tactile sensor array comes in contact with a given rubber sample, given K_m/K_{obj} , the ratio of R for different values of grasping forces remains greatly constant. In other words, the deflection ratio of the tactile sensor array, R , is independent from the change in grasping force. But as the same K_r/K_m comes in contact with rubber samples with different stiffness, K_m/K_{obj} is subject to change, the R ratio varies almost linearly. This shows the ability of proposed design concept in evaluating the stiffness of objects.

K_r/K_m combinations as low as 2.0 have to be avoided; because there would be almost no sensing sensitivity. Indeed the results suggests, for having good sensing sensitivity in evaluating the objects stiffness, the K_r/K_m combination for the tactile sensing array should be as high as possible; provided sufficient signal reading and deflection in the *Reference* sensing elements is secured. So the ratio of K_r/K_m specially in moving into fabrication of the MEMS devices should be optimized. Finally a K_r/K_m combination as high as 10.0 can provide a high stiffness evaluation sensitivity while the possibility of trivial readings in the *Reference* element is avoided.

Chapter 6 On the MEMS Sensor Development

The FEM simulation of the sensor design in the previous chapter showed that, stiffness and force requirements for sensor performance can be met by choosing an appropriate polymer material and determining suitable size for both the span and thickness of sensing diaphragms.

6.1 Design of sensor array

The sensor array design, addressed in this section, mainly consists of the masks' designs for the lithography processes during MEMS fabrication. The masks' design presented in this section has included the final changes and amendments necessary to be made in the earlier design attempts. The changes and amendments came as further MEMS fabrication experiences were gained. In result the design proposed in this section can be a good step stone in the fabrication of the final MEMS prototype.

Considering size limitations, the endoscopic or laparoscopic instruments are inserted through incisions of about 3–12 mm diameter [25] and the jaws' length commonly varies in range of 20–40 mm [100]. Following the literature review in Chapter 2 and Table 2.2, a resolution of 1000 μm was considered to be an achievable resolution for the proposed tactile sensing array with polymer as its structural material.

For the general design of sensing unit presented in Chapter 4 (Fig 4.1), the lithography masks were designed to provide a sensor resolution of 1000 μm and the sensor array was considered to be a 9*18 array of sensing elements. Such array would occupy a sensing area of 9*18 mm² in which the 9 mm edge sits on grasper's diameter and the 18 mm edge aligns along the grasper's length. It worth noting that in the first design for the MEMS device, the resolution of tactile sensor array was considered to be 750 μm ; but observing some fabrication constraints, the 1000 μm was found to be an achievable resolution for the proposed polymer-based tactile sensing array. These constraints would be clarified later in this chapter.

Two different combinations of diaphragm sizes were considered. In first combination, diaphragm sizes were set to be 300 μm and 700 μm . In the second combination, diaphragms were decided to have four different span sizes as 300, 400, 600 and 700 μm . By this means, it could be possible to investigate the variation of sensor stiffness with respect to its diaphragm size; and also compare the results obtained from the experiments with the simulation results.

Following the literature, the strain gauges are expected to be sputtered out of nichrome alloy (Ni-Cr 80/20 wt. %) [26, 27, 109, 110]. It is due to the fact that nichrome provides high resistivity and low coefficient of thermal resistance (TCR) [26]. Indeed nichrome's large resistivity helps to minimize the effect of contact and interconnect resistances on sensor performance [26]. The width of the strain gauges

have been reported to be $10\ \mu\text{m}$ in the literature [26, 109]. However in this study, it was decided to consider mainly a width of 20 and $30\ \mu\text{m}$ for the strain gauges in designing their patterns. This way the performance of strain gauge, its sensitivity and resistance change following the load exertion could be evaluated for the two different widths. It worth noting that, in the early designs strain gauges with $10\ \mu\text{m}$ in width was considered to be investigated too. But fabrication attempts showed that obtaining a feature as fine as $10\ \mu\text{m}$ takes unnecessarily more effort while the strain gauges as narrow as $20\ \mu\text{m}$ and higher in width can be obtained easily. So it was decided to go ahead with these two widths for strain gauges. Fig 5.1 shows the strain gauge patterns considered in designing its lithography mask. The patterns on the left and right of this figure are $30\ \mu\text{m}$ in width but different in the shape. The pattern in the center, however, is the strain gauge with $20\ \mu\text{m}$ in width. Fig 5.1 also contains insight into the exact arrangement of the strain gauges on the diaphragm edges and geometrical shape of the strain gauges in the proposed design.

Fig 6.1: Design of the serpentine strain gauges and their arrangement on the diaphragm edges; left and right) $30\ \mu\text{m}$ wide strain gauge, center) $20\ \mu\text{m}$ wide strain gauge.

In the same way as the simulation of sensor performance in Chapter 4, the mesa diameter for all diaphragm span sizes was set to be $150\ \mu\text{m}$. This way the stiffest diaphragms -the ones with 300 and $400\ \mu\text{m}$ span sizes- would have 75 and $125\ \mu\text{m}$

flexing arm between the edge of the sensing diaphragm and its mesa respectively. While this arm for the 600 and 700 μm diaphragm span sizes would be 150 and 200 μm respectively, Fig 6.2. Of course in the early design, the mesa diameter for the 600 and 700 μm span sizes diaphragms were set to 300 μm . But following the fabrication attempts it was found out that, with such a mesa size the stiffness ratio of the 300 μm diaphragm to the 700 μm diaphragm cannot be large enough. This indeed means that the sensing sensitivity in evaluating the tissue stiffness would be lost. So at the final design the mesa was considered back to be 150 μm in diameter for all the diaphragms again.

Fig 6.2: A top view of positioning the mesas on the sensing diaphragms center point.

Since the sensor would be in direct contact with soft tissues, such tissues can touch the base surface from which the mesa extrudes upward. In result some force could be induced between the tissue and the base surface. This issue could result in misrepresentation of the force in tissue-mesa interface. In order to overcome this short

coming, mesa was designed in a way that it is equipped with a crown, Fig 6.3. In mean time, in the design of the masks for lithography process few of the mesas were considered not to have this crown. This way, the mesa with crown can serve to prevent the tissue from touching the base surface. The mesas without crown were meant to facilitate the calibration of the sensing elements; especially when they come under lateral loading. At the same time, the comparison of the performance of sensing elements without crown with the ones with crown can be conducted too. This way a better understanding of the influence of the existence of the crown on sensor performance can be obtained.

Fig 6.3: Side view of the sensing elements and their mesa; Top) mesa without crown, Bottom) mesa with crown.

For the reading of the sensor signals, multiplexing electronics has to be used to scan the array of sensors. Though different electrical functioning can be adopted in the read-out circuit, for example a constant current source circuit [109]; but voltage dividing scheme during the multiplexing of signals was found to be more useful, Fig 6.4 [80]. In this read-out procedure, a control circuit sequentially connects each driving column to a drive voltage V_{drive} and ground while the rest of the lines are

equipotential to V_{out} through feedback lines. The signals of each sensor cell on the driving column is multiplexed to an operational amplifier circuit, then to an accurate (let say 16 bit), high speed, analog to digital converter, whose output is feed into a computer for further data analysis.

Fig 6.4: The read-out scheme applicable for the scanning of the sensor array signals [80].

Based on the read-out scheme presented in Fig 6.4, every sensing unit consists of 2 strain gauges connected in series between the V_{drive} and ground line; hence it is the voltage value in the middle point of every two strain gauges which is acquired by DAQ device as the sensor output signal. Indeed every sensing unit functions as a voltage divider. As it comes to the geometrical design of sensing units, Fig 6.5 shows the arrangement of both the sensing strain gauges in the edges of diaphragms and the

reference strain gauges in the space between the diaphragms (the later ones had been omitted in the previous figures for the sake of keeping these figures simple). In this figure too, it is clear that every two strain gauges build up a voltage divider setup. The blue connecting lines in Fig 6.5 are connecting the strain gauges vertically to form a column of the sensing array. So the V_{drive} and ground lines need to be connected to the blue column lines -and whole set of array columns- through the multiplexing devices. The figure also shows how middle point in every strain gauges pair can be connected into the DAQ device to provide the output signal. For Fig 6.5, there are only 2 connecting lines between the midpoint of stain gauges and the multiplexer device but in the real case -as there are 8 pairs of strain gauges in this figure- there should be 8 input lines into the multiplexer device on the right hand side of the figure. These lines should consist another series of the connecting lines which connect the middle points to the output terminal of the sensor array. In the desired sensor array, there was need for 18 columns and 9 rows of sensing diaphragms in the geometrical layout. Considering 2 vertical connecting lines for each column in Fig 6.5, there has to be 36 vertical lines for the whole columns of the array (18×2). As for the horizontal lines, since 9 sensing elements exist in a column and every sensing element itself consists of 4 sensing pairs; the total number of the horizontal connecting lines for the array is again 36 (9×4). So the read-out circuit has to deal with 36 lines of columns and 36 lines of the rows.

Fig 6.5: Detailed view of the wiring for the vertical lines together with strain gauges in order to read-out the outputs.

The Connecting lines ending to signal terminals were considered to have $75\ \mu\text{m}$ width and were spaced $50\ \mu\text{m}$ from each other. The terminals themselves were designed to be compatible with FFC connectors. One of the terminals was designed horizontally and the other one was considered to be vertical. Since both the terminals only need 36 connecting lines, they were designed to have 40 connecting pads with $0.5\ \text{mm}$ pitch (shown in Fig 6.7). This means that 4 lines of the 40 lines are extra.

Final layout resulting from the superposition of the masks designed for the lithography process on a $4''$ silicon wafer is shown in Fig 6.6. Indeed every sub layout in this figure represents a top view of a whole tactile sensor array. It shows how a whole sensor array is expected to look like following completion of the fabrication processes. It is clear that through such a layout, 4 sets of sensor arrays can be arranged on a $4''$

silicon wafer. Fig 6.7 shows relatively a closer view of one of the sensor arrays while Fig 6.8 provides a localized view of strain gauge and connecting lines configuration. Of course the horizontal and vertical connecting lines in Fig 6.8 are designed to be fabricated on separate layers so the apparent crossing of these connecting lines does not happen in the real case.

Fig 6.6: General top view of sensor arrays arrangement on a 4" silicon wafer following the superposition of the lithography masks.

Fig 6.7: General top view for a whole sensor array following the superposition of the lithography masks; including horizontal and vertical FFC signal terminals and their corresponding connecting lines.

Fig 6.8: Localized detail view of the strain gauges and their connecting lines inside a sensor array.

6.2 Attempts to fabricate the tactile sensing array

From the fabrication point of view, the tactile sensor array consists from two specific parts. First part is so called the sensing skin and second part is the mechanical structure which is needed to host and support the sensing skin on itself, Fig 6.9. The sensing skin consists of the polymeric sensing diaphragm, the strain gauges and the connecting lines. The mechanical structure on the other hand can be fabricated out of silicon wafer.

6.2.1 Etching of Diaphragm Cavity

The silicon wafer needs to be etched for creating square shaped cavities in it. This way a sensing diaphragm can be created by releasing it from beneath. In the first fabrication attempts, work started by selecting 500 μm thick silicon wafer as the substrate material. The lithography masks too were designed for the sensing array resolution of 750 μm . Such a resolution and the fact that sensing diaphragms were consisting of 300 and 700 μm means that, the sensing elements in the tactile array were designed densely next to each other. So the natural choice for creating the etch cavities was to go for a cavity with vertical walls. Such cavities can be created through DRIE (Deep Reaction Ion Etching) process. Hence a polymeric dry etch stop mask out of AZ9260 positive photoresist was spin coated, soft baked, exposed to UV light, developed, rinsed and speed dried on the silicon wafer. Next the masked wafer, with the designed square patterns, was subjected to the DRIE process. Fig 6.10 shows the cavities created on the silicon wafer. Fig 6.10 (left) shows the view of cavities from the wafer side that etch stop mask was present. Indeed this view belong for the side that dry etch has started from. Fig 6.10 (right), on the other hand, belongs to the opposite side of the wafer or side that etching has ended. Indeed the wafer has been etched through its thickness to

obtain through square holes. Both the top side view, the side from which dry etching has started, and bottom side view, the side that etching ended there, show that; DRIE process can provide good pattern definition as long as the forms are concerned. But dimension wise, as top side followed the exact dimension defined by the mask layer; the bottom side showed deviation from the desired dimension for the design. Indeed the cavity while was going through the wafer thickness has got widen. This means that the walls of the cavity during DRIE did not stay vertical as it is expected in the sensor design. Technically it means that strain gauges on the sensing element would be left out of the real edges of the sensing diaphragm; in result the sensitivity of the sensor performance can be lost in this way. As another matter of fact, through etching of the silicon substrate has to done following finishing of the fabrication of sensing skin on the wafer. This means that wafer has to be etched from beneath till the sensing diaphragm is released, Fig 6.9. Hence the possibility of damaging sensing skin during etching of cavities should be put into consideration too. As for the DRIE process, it was thought that the process can damage the sensing skin when it reaches to the opposite side of the silicon substrate. That is why it was decided to consider the wet etching process for creating the cavities.

Fig 6.9: Schematic of the sensing skin and the mechanical structure in the tactile sensing array design.

Fig 6.10: The views of square cavities on silicon wafer created by DRIE process; Left) starting (top) side of wafer etch, Right) Ending (bottom) side of wafer etch.

On the other hand, wet etching inherently results in inclined cavity walls because of the orientation selective etching in the silicon crystal. As for a commonly used (100) silicon wafer there would be a 54.7° angle between the walls of the etched cavities and the wafer plane. This means that for having a desired span size on the sensing skin side of the wafer, there is need to start with bigger window size in the back side of the silicon wafer. Indeed for the cavities with span sizes of 300 and 700 μm on the sensing skin side of the wafer there is need for 1008 and 1408 μm window sizes on the opposite side of a 500 μm wafer respectively. This means that for the 750 and 1000 μm sensor array resolutions, no supporting structure in the sensing array area can exist if the wafers with 500 μm in thickness were used as the mechanical structure. In considering 300 μm thick wafers as the support structure, for the 750 μm resolution too there would be no appropriate supporting structure for the sensing array; while this structure for the 1000 μm resolution of tactile array was found to be sufficient. Fig 6.11 illustrates the side view of sensing elements for the resolution of 1000 μm while the substrate wafer was

chosen to be $300\text{ }\mu\text{m}$ in its thickness. This figure also shows that, etch window sizes for 300 and $700\text{ }\mu\text{m}$ span sizes have to be 725 and $1125\text{ }\mu\text{m}$ respectively. Also the minimum available width for the supporting surface of the silicon substrate was found to be $75\text{ }\mu\text{m}$. Hence it is assumed that in this design the loading, applied to the sensing elements, can be supported appropriately in the contact area between the sensor's mechanical structure and the ground. In result it was decided to focus on the fabrication of the tactile array on top of the $300\text{ }\mu\text{m}$ thick silicon wafers; and perform the wet etching process in KOH solution to create the sensing skin cavities from the behind.

Fig 6.11: Schematic of the tactile array fabricated on the $300\text{ }\mu\text{m}$ thick wafer with the resolution of $1000\text{ }\mu\text{m}$, with cavities created through wet etching process.

6.2.2 Etch-Stop Masks for Wet Etching

During the wet etching of the $300\text{ }\mu\text{m}$ thick wafers in KOH solution two different etch stop masks (sacrificial layer in Fig 6.11) were examined. First one was SiO_2 sacrificial layer with a thickness of 1200 nm and second one was Si_3N_4 sacrificial layer with a thickness of 200 nm . As for the SiO_2 , it was found out that 1200 nm thick etch stop mask is sufficient only to an etch depth around $200\text{ }\mu\text{m}$ of the wafer thickness.

Since it is needed to have through holes (cavities) in the 300 μm thick wafer, the etching process continued despite total removal of the sacrificial layer beyond the 200 μm etch depth. This way the etching of through holes was secured but at the price of the weakening of the supporting structure for the sensing diaphragm. Indeed the etching was happening all over the wafer surface following total removal of the sacrificial layer. This means that such a sacrificial layer cannot be used for the through etching of 300 μm thick wafers.

The Si₃N₄ etch stop mask, on the other hand, was very strong against the KOH etchant solution as it was expected. So the mask pattern lasted for the through etch of the cavities in the wafer. But as the KOH solution reached to the other side of the wafer where only the Si₃N₄ on the opposite side and the polymer diaphragm were left (no more silicon bulk), the Si₃N₄ on the opposite could not last. In result, following the damage in the Si₃N₄ on the opposite side, the polymer diaphragm was damaged fully too. It is believed that this happened because of thinness of the Si₃N₄ as it could not resist against the gentle turbulence in the KOH solution which was present during the etching process. So considering both the 1200 nm thick SiO₂ sacrificial layer and the 200 nm thick Si₃N₄ layer examined in the fabrication stage, it is proposed to use a Si₃N₄ sacrificial layer as the etch stop mask during etching process. But as it comes to the thickness of sacrificial layer, it is recommended to be as high as 800 nm.

6.2.3 Realization of Sensing Diaphragms

Despite the above fact on the 1200 nm thick SiO₂ sacrificial layer, it was managed to get acceptable sensing diaphragm for both the HD4100 polyimide and the SU8-2000 polymer. The sensing diaphragm obtained in this way was including a mesa pattern as thick as the diaphragm itself. The HD4100 diaphragm thickness after its baking in the

oven was around $15\ \mu\text{m}$. The thickness of the SU8-2000 diaphragm too was measured to be around $35\ \mu\text{m}$ after it was hard baked. Fig 6.12 shows views of both of the sensing diaphragms fabricated in this study. As it is visible in Fig 6.12, the sensing diaphragms show some deviation from the expected square shape for them in the sensor design. Two different reasons can be named for this issue. Firstly, some image deviation is resulted from the remaining of the sacrificial layer beneath the sensing diaphragms. So further cleaning of these residues could help to have better looking for the diaphragms appearance. Secondly some misalignment was observed following the fabrication between the edge of the silicon wafer and the patterns of the etching windows. Indeed it is because of the direction dependent etch rate of the silicon wafer that such a misalignment could result in the distortion of etched cavities from the desired shapes. So the more special care in avoiding such misalignments the squarer looking of the fabricated sensing diaphragm.

Fig 6.12: View of the polymeric sensing diaphragm with short mesas fabricated in the study; Left) Sensing diaphragm out of SU8-2000 with a thickness of $35\ \mu\text{m}$, Right) Sensing diaphragm out of HD4100 with a thickness of $15\ \mu\text{m}$.

The mechanical properties of these two diaphragms were measured almost by the same mechanical setup that was mentioned in chapter 4. Of course the load sensor was equipped with a needle size apparatus to have the possibility of applying force to the mesa. A microscope device too was used for the accurate positioning of needle apparatus. The measurement were aimed to measure the force-deflection relationship and hence the stiffness of diaphragms. As for the HD4100 polyimide diaphragm, the maximum diaphragm deflection was found to be around 100 μm beyond which the diaphragm tore down. Even though undergoing such level of the deflection without diaphragm's failure is significant; but the observed stiffness for the diaphragm span sizes of 300 and 700 μm was only 5.43 and 3.86 N/mm respectively. Such values for the diaphragm stiffness not only is much lower than the stiffness values required for the sensing elements; but also the stiffness ratio of the 300 and 700 μm wide diaphragms is lower than 2.0. Based on the experimental results for a macro tactile array- study on the macro tactile array was performed in parallel to the MEMS fabrication attempts; but as its results are more conclusive of this chapter it was reported after this chapter- for such a stiffness ratio sensing sensitivity would be lost. The poor mechanical properties found for the HD4100 diaphragm was despite the fact that this polymer was considered as one of polymers of choice in the fabrication of the tactile sensor array in this study. Indeed all this is opposite to the mechanical properties reported for this polymer in its data sheet. It is believed that, this inconsistency could be due to the oxidation happened during the baking of the polyimide in the oven. As despite the recommendation of HD4100 data sheet for baking the polyimide in an oven with nitrogen ambient, such an oven was not available during this study while the baking temperature was as high as 375° C.

In the case of the SU8-2000 diaphragm however promising results were found.

For the SU8-2000 polymer, the maximum deflection for the 300 μm span size was 50 μm beyond which the diaphragm tore down; as the 700 μm span size diaphragm went through 85 μm of deflection without experiencing the fracture. Fig 6.13 shows the force-deflection relationship for both the 300 and 700 μm span sizes of the diaphragm. It is clear that, a good linearity between the diaphragm deflection and the applied force exists. The maximum force supported by 700 μm diaphragm was 0.944 N for the 85 μm of deflection. This force for the 300 μm diaphragm was 1.2 N which was observed under a deflection of 50 μm . Translating these forces into the ongoing grasping pressure in the sensor-tissue interface for the sensor resolution of 1000 μm , the sensing element with 700 μm diaphragm span size can bear up to a pressure of 944 KPa and the 300 μm one to a pressure of 1200 KPa . This pressure is much beyond the 400 KPa pressure present in the grasper-tissue interface in the laparoscopy operations; which was reported earlier in the Introduction chapter of this report. Such supporting force is even beyond the needs for having the grasping pressure of 600 KPa as design criteria for the sensing elements of tactile array; and also this supporting force is beyond the obtained force values in the chapter 5 in simulating the MEMS sensing elements. This means that the diaphragm out of SU8-2000 polymer can provide good supporting force features; hence SU8-2000 can be considered as the favorable polymer in the fabrication of the tactile sensor array in this study.

As it comes to the stiffness ratio between 300 and 700 μm span size diaphragms, the stiffness ratio is 2.35. Definitely such a ratio cannot provide acceptable sensing sensitivity for the evaluation of tissue stiffness (Referring to the results reported in Chapter 6). Of course it should be pointed that the mesa size for the 700 μm span size diaphragm had a diameter of 300 μm . Indeed this mesa size is other than the mesa size considered for the simulation of the sensor design in Chapter 5. In order to obtain

higher orders of stiffness ratios, it is need to go back to mesas with $150\text{ }\mu\text{m}$ in diameter in the case of the $700\text{ }\mu\text{m}$ span size diaphragms. Though to what extent the stiffness ratio would increase needs to be seen in fabrication again. It worth noting that, these results are for the SU8-2000 diaphragm of the thickness around $35\text{ }\mu\text{m}$. Hence the force-deflection relationship can be tuned further by changing the diaphragm thickness to find the optimum thickness for the design needs.

Fig 6.13: The force-deflection relationship of the SU8-2000 diaphragm with a thickness of $35\text{ }\mu\text{m}$ and for the 300 and $700\text{ }\mu\text{m}$ span sizes.

6.2.4 Fabrication of Strain Gauges

The fabrication of strain gauges out of Nichrome alloy was conducted through e-beam evaporation process. Fig 6.14 shows a partial view of the fabricated strain gauges and one of the connecting lines. The connecting lines, in this figure, were fabricated out of the copper and through sputtering process. In this figure three different strain gauges can be seen. They are the same as the strain gauges that were discussed

about in the beginning of this chapter and their design was illustrated in Fig 6.1. In Fig 6.14, similar to Fig 6.1, the strain gauges on the left and right of this figure are $30\ \mu\text{m}$ in width but different in the shape. The strain gauges in the center, however, are the one with $20\ \mu\text{m}$ in width. The requested thickness was $200\ \text{nm}$ for the connecting line and $40\ \text{nm}$ for the strain gauges. As a matter of fact the resistance of the connecting line and strain gauges in Fig 6.14 was measured. The resistance on the connecting line from one of the FFC terminals to the furthest point of it was around $100\ \Omega$. The resistance of the strain gauges on the right hand side and left hand side of the figure were found to be 1.7 and $2.1\ \text{K}\Omega$ respectively. The resistance of the strain gauges in the center of figure, the $20\ \mu\text{m}$ wide ones, was around $4.8\ \text{K}\Omega$. These results show that fabricated strain gauges have high electrical resistivity. This fact can lead to the creation of high temperatures when the strain gauges are connected to power source. So this issue should be considered in the design of the final read-out circuit.

6.2.5 Fabrication of Sensing Skin

Fabrication of the sensing skin inclusive of all the strain gauges and the both the horizontal and vertical connecting lines was conducted on a $500\ \mu\text{m}$ thick silicon wafer. Fig 6.15 shows one out of the four tactile arrays on the silicon wafer. The polymer that was used to fabricate this prototype was HD4100 polyimide. The strain gauges and the connecting lines were fabricated through sputtering process. Since the fabrication was conducted on a $500\ \mu\text{m}$ thick wafer, there was not the possibility of the etching from the back side to come up with a functioning tactile array. Even though the outcome was the fabrication of whole sensing devices on top of a solid silicon wafer, without the desired functionality for it; but at least the possibility of the fabricating the sensing skin on the silicon substrate was examined and it was a strong prove that such tactile array can be

realized in practice.

Fig 6.14: A partial view of the fabricated strain gauges and one of the connecting lines. The strain gauges on the left and right side are 30 μm in width but different in the shape. The strain gauges in the center are 20 μm in width.

In Fig 6.15 both the horizontal and vertical connecting lines ending to their corresponding FFC terminal are seen. The vertical connecting lines are from gold and the horizontal ones were sputtered out of copper. Though copper is inexpensive and it's choosing might be preferred; but as at the end of the fabrication the polyimide had to be baked, at 200° C for 30 min followed by at 375° C for one hour; hence the copper FFC terminal, which was exposed to the oven ambient, got totally oxidized. This means that such FFC terminal cannot function as connecting terminal in the real prototype and it should be avoided. Hence it is strongly recommended to fabricate the connecting lines

and their terminal only through the sputtering of the gold. Indeed even if the connecting lines don't need to go through high temperature recipe; still there is the possibility of oxidation for metals like copper. This may lead to the poor electrical connection in the terminals and high interconnection electrical resistance that is not favorable in the case of sensing devices.

Though this fabrication was performed for the sensor design with $750\ \mu\text{m}$ resolution, but it is evident that the fabrication procedures for either of the sensor design with 750 and $1000\ \mu\text{m}$ resolution remain the same.

Fig 6.15: The view of the sensing skin inclusive of all the strain gauges and both the horizontal and vertical connecting lines fabricated on the silicon wafer.

Despite $500\ \mu\text{m}$ thick wafers, in working with $300\ \mu\text{m}$ wafers; it was observed that spin coating of polymer on the wafer causes it to distort and it gets into a concave shape and hence it is not possible to vacuum clamp it on the lithography machine. This issue

creates a lot of problems in handling the device which is under fabrication. So it is crucial to consider this point into account in the stage of designing the lithography masks. It is proposed to design the masks of polymer layers in a way that, polymer layer is slotted in the form of separate islands. This way tension caused by the continuity of polymer layer would be eased and serve for securing wafer flatness.

6.3 Summary

The detailed design of lithography masks for the MEMS tactile sensor array was conducted and the results of the initial fabrication attempts were obtained. The masks were aimed to cover a sensing area of $9 \times 18 \text{ mm}^2$ by considering the $1000 \mu\text{m}$ resolution for the 9×18 array size. Diaphragm span sizes of 300, 400, 600 and $700 \mu\text{m}$ were included in the masks design. Diaphragm mesa, based on fabrication results, was set to have a diameter of $150 \mu\text{m}$ for all the diaphragm span sizes. To avoid misrepresentation of the grasping force, crowns were considered in mesas' design. Connecting lines were designed to end at two vertical and horizontal FFC terminals. The sensing arrays were arranged in a way that on every 4" silicon wafer 4 of the tactile arrays can be fabricated.

The initial fabrication attempts showed that, wet etching is the preferable choice in creating the diaphragm cavities and silicon wafers need to be $300 \mu\text{m}$ thick wafers. As for the choosing the polymer, the SU8-2000 is preferred over the HD4100 polyimide. As SU8-2000 can be hard baked at lower temperatures and does not have oxidation problem. Moreover, SU8-2000 provides good force supporting characteristic; which is beyond the present force in the tissue-grasper interface during MIS operations. This fact can provide a large maneuverability in optimizing the sensing diaphragm parameters; as it comes to its thickness and the diameter of the mesa. This way the

diaphragm stiffness can be optimized. To increase the stiffness ratio of the 300 μm wide diaphragms over the 700 μm wide ones, the mesa diameter should be reduced to 150 μm . Finally it was shown that the MEMS tactile array can be fabricated through the available techniques, the SU8-2000 polymer and the design proposed in this study. Though, the final development of the tactile array prototype needs an independent and focused fabrication attempt.

Chapter 7 Conclusion and Future Work Plan

This chapter summarizes the study conclusions and highlights the contributions. The future plans for this study as the next steps to the study have been identified.

7.1 Conclusion

- Minimally Invasive Surgery (MIS), as a preferred choice for many operations, severely reduces the surgeon's sensory perception during organ manipulation. Though visual sensing for MIS is well-established but it still suffers from lack of a tactile sensing.
- Tactile sensing have two different applications in MIS; diagnostic and assistive. The diagnostic application is used to evaluate the material characteristics of tissues and is mainly a non-force approach to assess tissues. Assistive application is aimed to safe handling and safe grasping of tissues. The grasping force plays a key role in obtaining these objectives and is

considered to be the most valued parameter. But force information is not sufficient to secure assistive tactile sensing requirements. Rather the tactile sensing system has to include the stiffness sensing procedures too.

- The main focus of this study was on development of an assistive tactile sensing system by measuring both force and stiffness information existing in the grasper-tissue interface. When two force sensing elements with different stiffness come in contact with a soft object, like tissue, they experience different deflections. This difference in deflections can be formulated to obtain the stiffness of the object. This approach, the spring based stiffness sensing, can be adopted to measure both the force and stiffness during MIS.
- The sensor was proposed to be an array of two series of force sensors with different stiffness. These force sensors are arranged alternately in a square array. A pair of two force sensing elements with different stiffness is referred to as a stiffness sensing unit.
- A lumped model was used to describe the sensor performance. The stiffness of stiffer (*Reference*) sensing element, softer (*Measuring*) sensing element and object was referred to as K_r , K_m and K_{obj} respectively. The ratio of deflection in the *Measuring* sensing element to the deflection in the *Reference* sensing element was defined as the output for stiffness sensing unit, R . The lumped model correlated R to K_r , K_m and K_{obj} . For validation of the model, ANSYS was used to simulate sensor's performance. The results of simulation found to be in a good agreement with the results of the lumped model. Both results showed, higher K_r/K_m provides higher sensitivity in the evaluation of K_{obj} . Of course too high K_r/K_m ratios may be synonymous to have trivial reading in the *Reference* sensing element and should be avoided. Furthermore, larger K_m/K_{obj}

values can help to improve stiffness sensing sensitivity and is recommended.

- Force sensing elements can be developed based on capacitive or piezoresistive approaches. But piezoresistive sensors are more favorable in measuring three-axial forces and benefit from a simpler readout circuits; and it is possible to arrange their electronic circuit away from the tactile array. Hence it is advised to develop force sensing elements based on piezoresistivity approach. Polymer-based sensors can come in direct contact with tissue without being damaged. Hence it was preferred to consider SU-8 or polyimide polymer as structural material for the sensor.
- Simulation of MEMS scale force sensor was performed in ANSYS. For higher sensing sensitivity, strain gauges need to be implanted on diaphragm symmetry lines in either surface of diaphragm thickness. Any increase in the mesa diameter can be used in improving the sensor sensitivity too. For both normal and shear loadings, a linear force-deflection relationship was present. For a given diaphragm thickness, it is possible to tune diaphragm stiffness by varying its span size. This can help in reducing number of MEMS fabrication steps. Feasible dimensions for the MEMS device can provide grasping pressures of around 480 and 300 KPa for 50 and 35 μm thick diaphragms respectively. Such pressures are consistent with the pressures present during grasping of tissues. Hence, the stiffness and force requirements for sensor performance can be met by choosing an appropriate polymer material and determining suitable size for both the span and thickness of sensing diaphragms.
- Following the fabrication of a macro tactile array consisting of 10 sensing elements, sensing elements with different stiffness were realized by providing

different arm length for the flexural mechanism. A stiffness ratio of 13.9 between the stiffest sensing element, 80.384 N/mm , and the softest one, 5.7808 N/mm , was achieved. By using rubber samples to resemble tissues a stiffness range of 2.34 ~ 10.01 N/mm was obtained. In result a wide range of sensor-rubber stiffness combinations were available during the experiments. Such range made it possible to draw general conclusion in validating the sensing concept. The results showed, when a specific K_r/K_m combination in the tactile array comes in contact with a given rubber sample, given K_m/K_{obj} , the R ratio for different values of grasping forces remains greatly constant. In other words, the deflection ratio of the tactile sensor array, R , is independent from the change in the grasping force. But as the same K_r/K_m comes in contact with rubber samples with different stiffness, K_m/K_{obj} is subject to change, the R varies almost linearly. This shows the proposed design concept is capable of evaluating the object stiffness. K_r/K_m combinations as low as 2.0 have to be avoided; because there would be almost no sensing sensitivity. Indeed the results suggest that for having good sensing sensitivity for object stiffness, the K_r/K_m ratio for the tactile sensing array should be as high as possible; provided sufficient signal reading and deflection in the *Reference* sensing elements is secured. So the ratio of K_r/K_m specially in moving towards fabrication of the MEMS device should be optimized. Finally a K_r/K_m combination as high as 10.0 can provide a high stiffness evaluation sensitivity while the possibility of trivial readings in the *Reference* element is avoided.

- By available MEMS fabrication techniques, on a 10 mm diameter MIS grasper a tactile sensor array with 1000 μm resolution can be realized. Hence, the lithography masks were designed to cover a sensing area of 9*18 mm² by

considering 9*18 array sizes.

- The MEMS fabrication attempts showed that, by choosing wet etching for creating the diaphragm cavities silicon substrate needs to be a 300 μm thick wafer. SU8-2000 can be hard baked at lower temperatures and does not have oxidation problem. Moreover, SU8-2000 provides good force supporting characteristic; which is beyond the force present in the tissue-grasper interface during MIS operations. Hence in developing the polymer based sensor SU8-2000 is preferred over HD4100 polyimide. It can be concluded that the MEMS tactile array can be fabricated through the available techniques, the SU8-2000 polymer and the design proposed in this study. Though, the final development of the tactile array prototype needs an independent and focused fabrication attempt.

7.2 Contributions

Tactile sensing application in MIS is identified as diagnostic and assistive. Diagnostic application is used for evaluating the material characteristics of tissues and is mainly a non-force approach to assess tissues. Assistive application of tactile sensing is for safe handling/grasping of tissues and needs to include both force and stiffness sensing procedures. In this thesis, tactile sensing in assistive application was targeted. The contributions made by the author in terms of originality of the work are as follows:

1. Force with Stiffness as a Multimodal Sensor for MIS applications:

The basic principle for functioning of the force sensing elements was adapted to be the same as in the commonly used force sensors, in which they undergo a

mechanical deflection following the application of external forces. These force sensors behave very much like a spring, and produce the force signal upon application of the deflection to their mechanical structure. Therefore, they can be referred to as passive, spring-based force sensors. Integrating this force sensing principle in the proposed tactile sensing system implies that the stiffness sensing in this study is also a spring-based approach. Thus, a novel force and stiffness integrated multi-modal approach is proposed and implemented with individual elements having adjoining and/or overlapping range so as to cover a wide range of forces occurring in the grasping and pulling actions during surgery. Such a hybrid approach for MIS applications, to the best of knowledge of the author, is not available yet. A lumped model and FEM simulation was formulated at first, in order to investigate the design concept and its performance. The proposed concept was demonstrated in a macro prototype set-up developed by the author for validation of the sensing concept.

2. Multi-point Array Tactile Sensor and Feedback during MIS:

This study focuses mainly on providing tactile sensory feedback for MIS manipulators, which are equipped with a grasper for holding and pulling the tissues during surgery as an essential part of any generic surgical protocol. For better accuracy and amount of information required, the sensing elements must be integrated directly into the surgical instruments. Most of the MIS manipulators based upon force-feedback protocol, measure the forces at/near the actuators by translating the information from tool-tip using mechanical linkages, which involve friction, backlash, gravity and inertia. This could distort the actual value of these forces and cause errors in the measurement. Secondly, in these systems the

interaction force values at any given time are normally measured as lumped single value. However, in surgical environment involving soft tissues, either for diagnostic or assistive application, it is desirable to resolve the forces at multiple points within a very small area of tool-tissue contact. Thus, an array of sensing elements is proposed and demonstrated in this work with a time/frequency multiplexed approach implemented in the read-out (measurement) circuitry for direct observation and estimation of the observed forces at multiple points.

3. Sensor Placement and MEMS based Design:

An ideal sensing position for monitoring tissue deflections and stiffness at the sensor-tissue interface is achieved via integrating multiple tactile sensing elements into the grasper jaws. The jaw is the place where forces acting at the surgical instrument tip can be measured most directly. Friction and other disturbance forces generated from any moving mechanisms do not affect the measurement of the tissue contact forces as much as that in the other cases explained above. However, concerning the fact that the jaw is the place that exhibits most severe space limitations, it is proposed to develop a clinically feasible tactile sensing system based on MEMS fabrication procedures.

The design of MEMS force sensing elements was performed using ANSYS. Considering Polyimide and SU-8 polymer as structural base material, strain distribution across the sensor diaphragm and its thickness was first investigated along with the influence of mesa size in this distribution. The force-deflection relationship for the sensing diaphragm, with a span size in the range of 300~700 μm and diaphragm thickness in the range of 20~50 μm , was obtained. The results showed a linear relationship for the strain up to the yield strain of the polymers.

For a given diaphragm thickness, the required stiffness of the sensor can be tuned by changing diaphragm's span size. Force capacity for the diaphragm can be supported by the proposed dimensions. Finally, the feasibility of fabricating a MEMS device based upon the proposed concept was conducted using existing MEMS facilities at NTU. Results indicated that the MEMS based tactile array using SU8 polymer as the diaphragm material, can be fabricated using the available fabrication techniques. Developing a clinically feasible MEMS prototype, however, was not included in the scope of this work. Some recent work on multiple force sensors embedded into an artificial skin (tissue engineering) is in progress, however, it doesn't include stiffness measurement.

4. Benchmarking and Validation of the Sensing Concept:

The sensor design concept and its functioning were investigated experimentally by fabricating a macro sensor array. It was consisting of 10 sensing elements with the stiffest and softest sensing element with $K=80.384$ and 5.781 N/mm respectively. Various commercially available rubber samples were used as test-samples during the experiments with their stiffness ranging between $2.34 \sim 10.01$ N/mm. As a result, a wide range of sensor-rubber stiffness combinations were available. The results showed to be in agreement with simulation results and it was shown that the proposed design concept for the tactile sensors is capable of evaluating the stiffness of objects. When a specific K_r/K_m combination in the tactile sensor array comes in contact with a given rubber sample the ratio of sensor reading, R , for different values of grasping forces remains greatly constant. However, when the sensor array comes in contact with rubber samples with different stiffness the sensor reading, R , varies almost linearly.

7.3 Future Work

The future plan can be organized as two groups of plans. First plan is to develop MEMS tactile array in order to integrate it into a commercial size MIS grasper. Second plan is aimed to use the tactile sensing during MIS; which can be realized by providing stiffness feedback into the surgeon or the master robot during surgery. This plan should aim in visualization and reproduction of ongoing reality in the sensor-tissue interface.

The following is further elaboration of the future works:

- **Development of MEMS Tactile Array** is the first straight step following this study. Following the fabrication attempts this study, this objective is achievable. But there is still the need for further investigation of technical features of such tactile array. An experimental study is needed to find out the dependency of the sensing diaphragm stiffness respect to different mesa diameter, diaphragm thickness and span sizes; as well as for different combination of these parameters. There is also need to study the performance of strain gauges with different width and shapes. This way the optimum design for the strain gauges can be obtained and the appropriate sensitivity can be obtained. Following these investigations the final prototype with an optimum design can be realized. The sensor prototype has to go through a calibration stage finally. For this purpose, a mechanical test stage quiet similar for the one used in this study can be suitable. The external force needs to be applied in both the normal and lateral directions.

- **Signals Read-out Scheme** can be established based on the approach which was elaborated in Chapter 6 of this report. But extracting

small sensing signals with high accuracy is practically a challenging task that needs to be dealt with.

- **Integration of tactile sensor array into a laboratory grasper** is needed to pave the way for using the tactile array in grasping the tissues. The clamping agent of the tactile array on the grasper can be vacuum force. This way installing and removing of the tactile array can be facilitated to a good extent.

- **Visualization and Reproduction of Ongoing Reality** in the sensor-tissue interface is an independent study to conduct; in shadow of which the tactile sensing cycle starting from the development of tactile array and representing its output signals can be complemented by the stiffness feedback into the surgeon or the master robot during MIS. This objective can be obtained in two different ways. The simplest way is to visualize the stiffness evaluation in the sensor-tissue interface in the form of a single graph or a 2D (3D) counter. This approach can be useful for the surgeons that manipulate the surgery graspers manually. But more advanced approach is to implement an impedance control over the surgical manipulator in a master-slave robotics surgery system. Hence there would be need for a stiffness feedback to the controller of the robotic system. As the tactile array provides a stiffness matrix in the contact area, rather than a single value, such impedance control could be a challenging task to deal with.

References

1. Tanaka, M., et al., *Development of a 3D tactile sensor*. Journal of Materials Processing Technology, 2007. **181**(1-3): p. 286-290.
2. Omata, S. and Y. Terunuma, *New tactile sensor like the human hand and its applications*. Sensors and Actuators A: Physical, 1992. **35**(1): p. 9-15.
3. Murayama, Y., C.E. Constantinou, and S. Omata, *Development of Tactile Mapping system for the stiffness characterization of tissue slice using novel tactile sensing technology*. Sensors and Actuators A: Physical, 2005. **120**(2): p. 543-549.
4. Murayama, Y., et al., *Development of a new instrument for examination of stiffness in the breast using haptic sensor technology*. Sensors and Actuators A: Physical, 2008. **143**(2): p. 430-438.
5. Hemsell, T., et al., *Resonant vibrating sensors for tactile tissue differentiation*. Journal of Sound and Vibration, 2007. **308**(3-5): p. 441-446.
6. Sakai, N., et al., *Diagnosis of the extent of gastric cancers by a new endoscopic ultrasonic tactile sensor*. Gastrointestinal Endoscopy, 2000. **51**(1): p. 69-73.
7. Omata, S., Y. Murayama, and C.E. Constantinou, *Real time robotic tactile sensor system for the determination of the physical properties of biomaterials*. Sensors and Actuators A: Physical, 2004. **112**(2-3): p. 278-285.
8. Frank, T., et al., *Clamping the small intestine during surgery: predicted and measured sealing forces*. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 1995. **209**(2): p. 111-115.
9. Herder, J.L., M.J. Horward, and W. Sjoerdsma, *A laparoscopic grasper with force perception*. Minimally Invasive Therapy and Allied Technologies, 1997. **6**(4): p. 279-286.
10. Cartmill, J.A., et al., *High pressures are generated at the tip of laparoscopic graspers*. Australian and New Zealand Journal of Surgery, 1999. **69**(2): p. 127-130.
11. Shakeshaft, A.J., et al., *A curved edge moderates high pressure generated by a laparoscopic grasper*. Surgical Endoscopy, 2001.
12. Toledo, L., et al., *A study of sustained forces and of the working space of endoscopic instruments*. Etude des forces subies et de l'espace de travail des instruments de chirurgie endoscopique, 1999. **53**(7): p. 587-597.
13. Marucci, D.D., et al., *Patterns of failure at the instrument-tissue interface*. Journal of Surgical Research, 2000. **93**(1): p. 16-20.
14. Heijnsdijk, E.A.M., et al., *Slip and damage properties of jaws of laparoscopic graspers*. Surgical Endoscopy, 2004. **18**(6): p. 974-979.
15. Visser, H., et al., *Forces and displacements in colon surgery*. Surgical Endoscopy, 2002. **16**(10): p. 1426-1430.
16. Heijnsdijk, E.A.M., et al., *The optimal mechanical efficiency of laparoscopic forceps*. Surgical Endoscopy, 2004. **18**(12): p. 1766-1770.
17. Heijnsdijk, E.A.M., et al., *The influence of force feedback and visual feedback in grasping tissue laparoscopically*. Surgical Endoscopy, 2004. **18**(6): p. 980-985.
18. Heijnsdijk, E.A.M., et al., *Fenestrations in the jaws of laparoscopic graspers*. Minimally Invasive Therapy and Allied Technologies, 2005. **14**(1): p. 45-48.
19. Rentschler, M.E., et al., *Modeling, analysis, and experimental study of in vivo*

- wheeled robotic mobility*. IEEE Transactions on Robotics, 2006. **22**(2): p. 308-321.
20. Jayaraman S., T.A.L., Naish M. D., Lyle A., Patel R. V., Schlachta C. M., *Toward construct validity for a novel sensorized instrument-based minimally invasive surgery simulation system*. Surgical Endoscopy, 2011. **25**: p. 7.
 21. Callaghan D. J., R.G., McGrath M. M., Coyle E., Semenova Y., Farrell G., *Investigation and experimental measurement of scissor blade cutting forces using fiber Bragg grating sensors*. Smart Materials and Structures, 2011. **20**: p. 1-9.
 22. Valdastrì P., T.S., Menciassi A., Dario P., *A scalable platform for biomechanical studies of tissue cutting forces*. Measurement Science and Technology, 2009. **20**(4): p. 1-11.
 23. Rosen J., M.M., Richards C., Hannaford B., Sinanan M. *Surgeon-tool force/torque signatures—evaluation of surgical skills in minimally invasive surgery*. in *Proceedings of Medicine Meets Virtual Reality*. 1999. San Francisco, CA.
 24. Trejos A. L., J.S., Patel R. V., Naish M. D., Schlachta C. M., *Force sensing in natural orifice transluminal endoscopic surgery*. Surgical Endoscopy, 2011. **25**: p. 186-192.
 25. Puangmali, P., et al., *State-of-the-art in force and tactile sensing for minimally invasive surgery*. IEEE Sensors Journal, 2008. **8**(4): p. 371-380.
 26. Kim, J.-H., et al., *Design of Flexible Tactile Sensor Based on Three-Component Force and Its Fabrication*. 2005.
 27. Kim, J.-H., et al., *Development of tactile sensor with functions of contact force and thermal sensing for attachment to intelligent robot finger tip*. 2006.
 28. Sung-Hoon Kim, J.E., Chang Liu, Douglas L Jones, *Texture classification using a polymer-based MEMS tactile sensor*. JOURNAL OF MICROMECHANICS AND MICROENGINEERING, 2005. **15**: p. 912-920.
 29. Umemori, M., et al., *A pressure-distribution sensor (PDS) for evaluation of lip functions*. American Journal of Orthodontics and Dentofacial Orthopedics, 1996. **109**(5): p. 473-480.
 30. Noda, K., et al., *A shear stress sensor for tactile sensing with the piezoresistive cantilever standing in elastic material*. Sensors and Actuators A: Physical, 2006. **127**(2): p. 295-301.
 31. Maeno, T., T. Kawai, and K. Kobayashi. *Analysis and design of a tactile sensor detecting strain distribution inside an elastic finger*. in *IEEE International Conference on Intelligent Robots and Systems*. 1998.
 32. Sohgalwal, M., et al., *FABRICATION AND CHARACTERIZATION OF SILICON-POLYMER BEAM STRUCTURES FOR CANTILEVER-TYPE TACTILE SENSORS*, in *IEEE*. 2007.
 33. Heo, J.-S., J.-H. Chung, and J.-J. Lee, *Tactile sensor arrays using fiber Bragg grating sensors*. Sensors and Actuators A: Physical, 2006. **126**(2): p. 312-327.
 34. Jenstrom, D.T. and C.-L. Chen, *A fiber optic microbend tactile sensor array*. Sensors and Actuators, 1989. **20**(3): p. 239-248.
 35. Heo, J.-S., C.-H. Han, and J.-J. Lee, *System Design and Evaluation of the Robot Tactile Sensor Using the Microbending Fiber Optic Sensors*, in *IEEE International Conference on Robot & Human Interactive Communication*. 2007.
 36. Ohka, M., et al. *Three-axis optical tactile sensor (FEM contact analyses and*

- sensing experiments using a large-sized tactile sensor*). in *Proceedings - IEEE International Conference on Robotics and Automation*. 1995.
37. Ohka, M., H. Kobayashi, and Y. Mitsuya. *Sensing Characteristics of an Optical Three-axis Tactile Sensor Mounted on a*. 2005.
 38. Ohka, M., et al., *Sensing Precision of an Optical Three-axis Tactile Sensor for a Robotic Finger*, in *IEEE International Symposium on Robot and Human Interactive Communication*. 2006.
 39. Engel, J., et al., *Polymer micromachined multimodal tactile sensors*. *Sensors and Actuators A: Physical*, 2005. **117**(1): p. 50-61.
 40. Dario, P., *Tactile sensing: Technology and applications*. *Sensors and Actuators A: Physical*, 1991. **26**(1-3): p. 251-256.
 41. Dargahi, J., S. Najarian, and B. Liu, *Sensitivity analysis of a novel tactile probe for measurement of tissue softness with applications in biomedical robotics*. *Journal of Materials Processing Technology*, 2007. **183**(2-3): p. 176-182.
 42. Chu, Z., P.M. Sarro, and S. Middelhoek, *Silicon three-axial tactile sensor*. *Sensors and Actuators A: Physical*, 1996. **54**(1-3): p. 505-510.
 43. Suzuki, K., K. Najafi, and K.D. Wise, *A 1024-element high-performance silicon tactile imager*. *IEEE Transactions on Electron Devices*, 1990. **37**(8): p. 1852-1860.
 44. Dudaicevs, H., et al., *Surface micromachined pressure sensors with integrated CMOS read-out electronics*. *Sensors and Actuators, A: Physical*, 1994. **43**(1-3): p. 157-163.
 45. Chun, K. and K.D. Wise, *HIGH-PERFORMANCE SILICON TACTILE IMAGER BASED ON A CAPACITIVE CELL*. *IEEE Transactions on Electron Devices*, 1985. **ED-32**(7): p. 1196-1201.
 46. Leineweber, M., et al., *New tactile sensor chip with silicone rubber cover*. *Sensors and Actuators A: Physical*, 2000. **84**(3): p. 236-245.
 47. Suzuki, K., K. Najafi, and K.D. Wise, *Process alternatives and scaling limits for high-density silicon tactile imagers*. *Sensors and Actuators, A: Physical*, 1990. **23**(1-3): p. 915-918.
 48. Gray, B.L. and R.S. Fearing, *A Surface Micromachined Microtactile Sensor Array*. 1996.
 49. Lee, H.K., S.I. Chang, and E. Yoon, *A flexible polymer tactile sensor: Fabrication and modular expandability for large area deployment*. *Journal of Microelectromechanical Systems*, 2006. **15**(6): p. 1681-1686.
 50. Lee, H.K., et al. *Polymer tactile sensing array with a unit cell of multiple capacitors for three-axis contact force image construction*. in *Proceedings of Future Generation Communication and Networking, FGCN 2007*. 2007.
 51. Suzuki, K., K. Najafi, and K.D. Wise. *1024-element high-performance silicon tactile imager*. in *Technical Digest - International Electron Devices Meeting*. 1988.
 52. Wolffenbuttel, M.R. and P.P.L. Regtien, *Polysilicon bridges for the realization of tactile sensors*. *Sensors and Actuators A: Physical*, 1991. **26**(1-3): p. 257-264.
 53. Wolffenbuttel, M.R. and P.P.L. Regtien, *Design considerations for a silicon capacitive tactile cell*. *Sensors and Actuators A: Physical*, 1990. **24**(3): p. 187-190.
 54. Wolffenbuttel, M.R., R.F. Wolffenbuttel, and P.P.L. Regtien, *An integrated charge amplifier for a smart tactile sensor*. *Sensors and Actuators A: Physical*,

1992. **31**(1-3): p. 101-109.
55. Wolffenbuttel, R.F. and P.P.L. Regtien, *Integrated tactile imager with an intrinsic contour detection option*. Sensors and Actuators, 1989. **16**(1-2): p. 141-153.
56. Paschen, U., et al., *A novel tactile sensor system for heavy-load applications based on an integrated capacitive pressure sensor*. Sensors and Actuators A: Physical, 1998. **68**(1-3): p. 294-298.
57. Sato, N., et al., *MEMS fingerprint sensor immune to various finger surface conditions*. IEEE Transactions on Electron Devices, 2003. **50**(4): p. 1109-1116.
58. Ko, C.T., et al. *A highly sensitive CMOS-MEMS capacitive tactile sensor*. in *Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS)*. 2006.
59. Sugiyama, S., et al. *MICRO-DIAPHRAGM PRESSURE SENSOR*. in *Technical Digest - International Electron Devices Meeting*. 1986.
60. Sugiyama, S., et al., *Tactile image detection using a 1k-element silicon pressure sensor array*. Sensors and Actuators A: Physical, 1990. **22**(1-3): p. 397-400.
61. Liu, L., X. Zheng, and Z. Li, *An array tactile sensor with piezoresistive single-crystal silicon diaphragm*. Sensors and Actuators A: Physical, 1993. **35**(3): p. 193-196.
62. Kane, B.J. and G.T.A. Kovacs. *A CMOS compatible traction stress sensing element for use in high resolution tactile imaging*. in *International Conference on Solid-State Sensors and Actuators, and Eurosensors IX, Proceedings*. 1995.
63. Kane, B.J., M.R. Cutkosky, and G.T.A. Kovacs, *A Traction stress sensor array for use in high-resolution robotic tactile imaging*. Journal of Microelectromechanical Systems, 2000. **9**(4): p. 425-434.
64. Kane, B.J., M.R. Cutkosky, and G.T.A. Kovacs, *CMOS-compatible traction stress sensor for use in high-resolution tactile imaging*. Sensors and Actuators A: Physical, 1996. **54**(1-3): p. 511-516.
65. Mei, T., et al., *An integrated MEMS three-dimensional tactile sensor with large force range*. Sensors and Actuators A: Physical, 2000. **80**(2): p. 155-162.
66. Mei, T., et al. *Design and fabrication of an integrated three-dimensional tactile sensor for space robotic applications*. in *Proceedings of the IEEE Micro Electro Mechanical Systems (MEMS)*. 1999.
67. Wang, L. and D.J. Beebe, *Silicon-based shear force sensor: development and characterization*. Sensors and Actuators, A: Physical, 2000. **84**(1): p. 33-44.
68. Vazsonyi, E., et al., *Three-dimensional force sensor by novel alkaline etching technique*. Sensors and Actuators A: Physical, 2005. **123-124**: p. 620-626.
69. Kim, K., et al. *3-Axes flexible tactile sensor fabricated by SI micromachining and packaging technology*. in *Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS)*. 2006.
70. Kim, K., et al., *A silicon-based flexible tactile sensor for ubiquitous robot companion applications*. Journal of Physics: Conference Series, 2006. **34**(1): p. 399-403.
71. Adam, M., et al. *Three dimensional single crystalline force sensor by porous Si micromachining*. in *Proceedings of IEEE Sensors*. 2004.
72. Adam, M., et al., *CMOS integrated tactile sensor array by porous Si bulk micromachining*. Sensors and Actuators A: Physical, 2008. **142**(1): p. 192-195.
73. Vasarhelyi, G., et al., *Characterization of an integrable single-crystalline 3-D tactile sensor*. IEEE Sensors Journal, 2006. **6**(4): p. 928-934.

74. Hwang, E.S., J.H. Seo, and Y.J. Kim, *A polymer-based flexible tactile sensor for both normal and shear load detections and its application for robotics*. Journal of Microelectromechanical Systems, 2007. **16**(3): p. 556-563.
75. Hwang, E.S., J.H. Seo, and Y.J. Kim. *A polymer-based flexible tactile sensor for normal and shear load detection*. in *Proceedings of the IEEE International Conference on Micro Electro Mechanical Systems (MEMS)*. 2006.
76. Hwang, E.-S. and Y.-J. Kim, *A Polymer-based Flexible Tactile Sensor and Its Application to Robotics*. 2007.
77. Wang, Y., et al., *Novel pressure sensor structure for integrated sensors*. Sensors and Actuators, A: Physical, 1990. **21**(1-3): p. 62-64.
78. Wang, Y., et al., *A novel structure of pressure sensors*. IEEE Transactions on Electron Devices, 1991. **38**(8): p. 1797-1802.
79. Jin, W.L. and C.D. Mote Jr, *Development and calibration of a sub-millimeter three-component force sensor*. Sensors and Actuators, A: Physical, 1998. **65**(1): p. 89-94.
80. Shan, J.H., et al., *The design and fabrication of a flexible three-dimensional force sensor skin*.
81. Adam, M., et al., *CMOS integrated tactile sensor array by porous Si bulk micromachining*. Sensors and Actuators A: Physical, 2008. **142**(1): p. 192-195.
82. Adam, M., et al. *Three dimensional single crystalline force sensor by porous Si micromachining*. in *Proceedings of IEEE Sensors*. 2004.
83. Vasarhelyi, G., et al., *Effects of the elastic cover on tactile sensor arrays*. Sensors and Actuators A: Physical, 2006. **132**(1): p. 245-251.
84. Vasarhelyi, G., et al., *Characterization of an integrable single-crystalline 3-D tactile sensor*. IEEE Sensors Journal, 2006. **6**(4): p. 928-934.
85. Vazsonyi, E., et al., *Three-dimensional force sensor by novel alkaline etching technique*. Sensors and Actuators A: Physical, 2005. **123-124**: p. 620-626.
86. Dao, D.V. and S. Sugiyama, *Development of 4-DOF Soft-Contact Tactile Sensor and Application to Gripping Operation of Robotics Fingers*, in *IEEE sensors*. 2006.
87. Dao, D.V., Q. Wang, and S. Sugiyama, *Fabrication and characterization of 3-DOF soft-contact tactile sensor utilizing 3-DOF micro force moment sensor*. IEEJ Transactions on Sensors and Micromachines, 2007. **127**(3): p. 177-181+13.
88. Dao, D.V., et al., *a multi-axis force-moment micro sensor for application in fluid mechanics*, in *International Symposium on Micromechatronics and Human Science*. 2002.
89. Dao, D.V., et al. *Development of Six-Degree of Freedom Micro Force Sensor for Application in Geophysics*. in *IEEE*. 2002.
90. Fujii, I., et al., *Tactile Perception using Micro Force/Moment Sensor Embedded in Soft Fingertip*, in *IEEE SENSORS*. 2006: Korea.
91. Wen, C.-C. and W. Fang, *Tuning the sensing range and sensitivity of three axes tactile sensors using the polymer composite membrane*. Sensors and Actuators A: Physical, 2008. **145-146**: p. 14-22.
92. Dargahi, J., M. Parameswaran, and S. Payandeh, *Micromachined piezoelectric tactile sensor for an endoscopic grasper - theory, fabrication and experiments*. Journal of Microelectromechanical Systems, 2000. **9**(3): p. 329-335.
93. Sokhanvar S., P.M., Dargahi J., *MEMS Endoscopic Tactile Sensor: Toward In-Situ and In-Vivo Tissue Softness Characterization*. IEEE Sensors Journal,

2009. 9(12): p. 9.
94. Bonakdar A., N.N., *Determination of tissue properties using microfabricated piezoelectric tactile sensor during minimally invasive surgery*. Sensor Review, 2010. 30(3): p. 9.
95. Schostek, S., et al., *Artificial tactile sensing in minimally invasive surgery - A new technical approach*. Minimally Invasive Therapy and Allied Technologies, 2006. 15(5): p. 296-304.
96. Culjat M. O., K.C.H., Franco M. L., Lewis C. E., Bisley J. W., Dutson E. P., Grundfest W. S., *A Tactile Feedback System for Robotic Surgery*, in *30th Annual International IEEE EMBS Conference*. 2008: Vancouver, British Columbia, Canada.
97. King C. H., C.M.O., Franco M. L., Bisley J. W., Carman G. P., Dutson E. P., Grundfest W. S., *A Multielement Tactile Feedback System for Robot-Assisted Minimally Invasive Surgery*. IEEE TRANSACTIONS ON HAPTICS, 2009. 2(1).
98. Trejos A. L., J.J., Perri M. T., Naish M. D., Patel R. V., Malthaner R. A. *Experimental Evaluation of Robot-Assisted Tactile Sensing for Minimally Invasive Surgery*. in *The 2nd Biennial IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics*. 2008. Scottsdale, AZ, USA.
99. Perri M. T., T.A.L., Naish M. D., Patel R. V., Malthaner R. A., *Initial Evaluation of a Tactile/Kinesthetic Force Feedback System for Minimally Invasive Tumor Localization*. IEEE/ASME TRANSACTIONS ON MECHATRONICS, 2010. 15(6).
100. Endoscopy, A.S.T.-. *Laparoscopic Instruments - Product Catalog*. 2010.
101. Allison M. Okamura, C.S., Mark D. O'Leary. *Force Modeling for Needle Insertion Into Soft Tissue*. in *IEEE TRANSACTIONS ON BIOMEDICAL ENGINEERING*. October 2004.
102. Hongbin Liu, D.P.N., Kaspar Althoefer, Lakmal D. Seneviratne, *Rolling Mechanical Imaging: A Novel Approach for Soft Tissue Modeling and Identification during Minimally Invasive Surgery*, in *IEEE International Conference on Robotics and Automation*. 2008: Pasadena, CA, USA.
103. *Product Bulletin HD-4100 Series Polyimide*. 2008, HD Microsystem.
104. <http://memscyclopedia.org/su8.html>. [cited.
105. Dupont, *Dupont Kapton HN Polyimide Film*, Dupont.
106. <http://www.tml.jp/e/download/catalogdownload.html>.
107. <http://www.analog.com/en/specialty-amplifiers/instrumentation-amplifiers/ad621/products/product.html>. [cited.
108. Liu H., L.J., Song X., Seneviratne L. D., Althoefer K., *Rolling Indentation Probe for Tissue Abnormality Identification During Minimally Invasive Surgery*. IEEE TRANSACTIONS ON ROBOTICS, 2011. 27(3): p. 11.
109. Engel, J., J. Chen, and C. Liu, *Development of polyimide flexible tactile sensor skin*. Journal of Micromechanics and Microengineering, 2003. 13(3): p. 359-366.
110. Kazi, I.H., et al., *Characterization of sputtered nichrome (Ni-Cr 80/20 wt.%) films for strain gauges applications*. Thin Solid Films, 2006.

List of Publications

The following publications have been generated from this research work:

- 1- S. Pour Abbas, S. Chauhan, S.J. Phee, G.K. Lau, “*Real-time, non-vision sensory feedback during minimally invasive surgery*”, 2009 IEEE Region 10 Conference, TENCON 2009, Nov. 2009.
- 2- S. Pour Abbas, S. Chauhan, S.J. Phee, G.K. Lau, “*Tactile Sensory feedback during minimally invasive surgery*”, AOTULE 2009, National Taiwan University, Dec. 2009.

The following paper has been submitted:

- 3- S. Pour Abbas, S. Chauhan, “*Spring-based Tactile Array for Assistive Robotic Surgical Applications*”, submitted to IEEE Sensors Journal.

Appendix A

The macro tactile array reading for different rubber samples are as followings:

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=2.34 \text{ N/mm}$.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=3.57 \text{ N/mm}$.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=3.85$ N/mm.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=4.51$ N/mm.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=5.41$ N/mm.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=6.28$ N/mm.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=7.26$ N/mm.

The macro tactile array reading, $R = (\Delta\delta_m)/(\Delta\delta_r)$, for different K_r/K_m in grasping the rubber sample with $K=10.01$ N/mm.

Performance of the Macro Tactile Sensor in grasping rubber samples with different stiffness.

Appendix B

Strain gauges used for fabrication of the macro tactile array all were strain gauge type FLA-2-23 from Tokyo Sokki Kenkyujo Co.

FOIL STRAIN GAUGE series "F"

Compatible adhesive & Operational temperature
CN : -20~+80℃
P-2 : -20~+80℃

EB-2 : -20~+80℃

Operational temperature -20~+80℃
Temperature compensation range +10~+80℃

GENERAL USE

Gauge pattern		Type	Gauge size L W		Backing L W		Resistance in Ω
<p>This gauge employs alloy foils which are 0.003 to 0.007 mm thick. Its gauge backing is made of epoxy resin with thickness of 0.03 mm which exhibits excellent electrical insulation performance. The backing is color coded for distinction of object specimen material for self temperature compensation.</p>			L : length W : width (Unit : mm)				
<p>■Single-element (G.F. 2.1 approx.)</p>							
 FLG-02	 (X3)	Single-element	FLG-02-11 -17 -23	0.2 1.4	3.5 2.5		120
 FLG-1	 (X3)		FLG-1-11 -17 -23	1 1.1	6.5 2.5		120
 FLA-03	 (X3)		FLA-03-11 -17 -23	0.3 1.4	3.0 2.0		120
 FLA-05	 (X3)		FLA-05-11 -17 -23	0.5 1.2	5.0 2.2		120
 FLA-1	 (X3)		FLA-1-11 -17 -23	1 1.3	5.0 2.5		120
 FLA-2	 (X3)		FLA-2-11 -17 -23	2 1.5	6.5 3.0		120
 FLA-3	 (X3)		FLA-3-11 -17 -23	3 1.7	8.8 3.5		120
 FLA-3-60	 (X3)		FLA-3-60-11 -17 -23	3 1.2	8.0 3.0		60
 FLA-5	 (X3)		FLA-5-11 -17 -23	5 1.5	10.0 3.0		120
 FLA-6	 (X3)		FLA-6-11 -17 -23	6 2.2	12.5 4.3		120
 FLA-1-350-11	 (X3)	350 Ω	FLA-1-350-11 FLA-1-350-17 FLA-1-350-23	1 2.0	5.0 4.0		350
 FLA-2-350-11	 (X3)		FLA-2-350-11 FLA-2-350-17 FLA-2-350-23	2 1.9	6.1 3.5		350
 FLA-3-350-11	 (X3)		FLA-3-350-11 FLA-3-350-17 FLA-3-350-23	3 3.2	8.5 5.0		350
 FLA-6-350-11	 (X3)		FLA-6-350-11 FLA-6-350-17 FLA-6-350-23	6 2.6	12.5 4.5		350

Each package contains 10 gauges.