Microelectromechanical force and tactile sensors for minimally invasive surgery

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**NANYANG
TECHNOLOGICAL
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**MICROELECTROMECHANICAL FORCE AND
TACTILE SENSORS FOR MINIMALLY
INVASIVE SURGERY**

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**SCHOOL OF MECHANICAL AND AEROSPACE
ENGINEERING**

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Microelectromechanical Force and Tactile Sensors for Minimally Invasive Surgery

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List of Abbreviations

MIS	Minimally Invasive Surgery
MEMS	Micro-electro-mechanical System
NEMS	Nano-electro-mechanical System
SiNW	Silicon Nanowire
Si	Silicon
NW	Nano Wire
CMOS	Complementary Metal Oxide Semiconductor
FEA	Finite Element Analysis
FEM	Finite Element Modelling
MRI	Magnetic Resonance Imaging
IVUS	Intravascular Ultrasound
PZT	Lead Zirconate Titanate
IMPC	Ionic Polymer Metal Composite
ASIC	Application-Specific Integrated Circuit
SOI	Silicon on Insulator
SOI-MUMPS	Silicon on Insulator Multi-user MEMS Process
BOX	Buried Oxide
CD	Critical Dimension

DRIE	Deep Reactive Ion Etching
PECVD	Plasma Enhanced Chemical Vapor Deposition
PCB	Printed Circuit Board
MC	Meissner Corpuscle
CMC	Carbon Micro-coil
RA	Rapid-adapting
SA	Slow-adapting
SAC	Sn/Ag/Cu Solder Compositions
PDMS	Polydimethylsiloxane
SEM	Scanning Electron Microscopy
SiO₂	Silicon Oxide
PMD	Premetal Dielectric
BAW	Bulk Acoustic Wave
SE	Square Extensional
Q-factor	Quality-factor
PSG	Phosphosilicate Glass
dB	Decibel
DC	Direct Current
AC	Alternating Current
PEMA	Poly (ethylmethacrylate)

Abstract

Sense of touch plays an important role in surgery. Minimally invasive surgical (MIS) procedures provide great benefits to patients; however, the surgeon's ability of perceiving force and tactile sensation from tissues is severely impaired. Performing surgery without such sensory information could lead to increase of tissue trauma and vital organic tissue damage. This dissertation presents the development of miniaturized force and tactile sensors that are able to measure the contact force and relative hardness of contact objects, with the purpose of enhancing the surgeon's touch sensibility during minimally invasive surgery. The force and tactile sensors developed in this dissertation are based on microelectromechanical systems (MEMS) technology utilizing the transduction principles of silicon nanowire (SiNW)-based piezoresistance and bulk acoustic resonance.

A miniaturized force sensor that can be integrated on the distal tip of a commercial guidewire is firstly developed. The sensor has a sensory area of $200\text{ }\mu\text{m} \times 200\text{ }\mu\text{m}$, and utilizes SiNWs as piezoresistive sensing element by taking advantage of its ultra-small dimension and high sensitivity. Together with the movable core wire, the sensor can detect contact forces while maintaining the original welded tip which is an essential part of the standard guidewire assembly. A finite element (FE) model is built in COMSOL Multiphysics[®] v4.2, in the aim of finding the maximum stress location in order to locate the SiNWs at the high stress regions and simulating the mechanical behaviour of the sensor-guidewire interaction as well. The results from simulation have been verified by the experimental measurement results. By taking advantages of the high sensitivity of SiNWs, the fabricated sensor is capable of detecting small displacement in nano-meter scale with a sensitivity of $13.4 \times 10^{-3}\text{ }\mu\text{m}^{-1}$ in the z-direction. It has been shown from the characterization results that the sensor has high linearity ($> 99.9\%$) to the applied load without obvious hysteresis.

To enhance the force sensitivity of the ring-shaped sensor that has a flat surface, a biomimetic tactile sensor is then designed and characterized. The Meissner corpuscle inspired sensor is designed with a rigid spherical ball attached on top of the suspended ring structure mimicking the biological structure of the Meissner corpuscles. The sensor is developed with a simplified fabrication and packaging process by dropping spherical balls on the SiNW-based ring structure. Finite element analysis (FEA) has been used to predict the mechanical behaviours of the tactile sensor with rigid solder ball and with elastic bump covering of rigid solder ball underneath. The performance of the SiNWs on the bio-inspired sensor and on the bare ring-shaped sensing element is comparatively studied by repeatability and hysteresis tests. The effectiveness of sensitivity enhancement of the bio-inspired tactile sensor with solder ball attached is verified from the experimental results, in which the fractional change in resistance in normal direction is improved from 1.34 percent to 2.8 percent per micro meter.

Following, a miniaturized resonant tactile sensor is developed for material elasticity measurement. The sensor is based on a MEMS silicon bulk acoustic wave (BAW) resonator, incorporated with a rounded sensing tip and mechanical stoppers to limit the maximum loading force and contact area. Upon contact with the material, the resonant frequency shifts corresponding to the amount of mass and stiffness loading from the contact object. The BAW resonator is excited in the square-extensional (SE) mode and is fabricated using a silicon-on-insulator (SOI) multi-user MEMS process through MEMSCAP. The resonant frequency of the resonator is 2.2 MHz and has a Q-factor of 10430 in air. The ability of the sensor to measure the relative hardness of the contact object is tested with experiments by contacting the sensor with materials of different degrees of Young's modulus. Results show that the BAW resonant tactile sensor can provide information not only about the stiffness of the materials in contact with the sensor, but also the extent of force/stress applied on the sensing tip. The sensor is demonstrated to be able to differentiate materials of Young's modulus in the GPa range.

Chapter 1 Introduction

This chapter gives a brief introduction to the background of the minimally invasive surgery (MIS) along with the motivation of developing MEMS force sensors for guidewires and tactile sensors for tissue elasticity measurement in biomedical applications. The research objectives and technical requirements regarding the development of force and tactile sensors are addressed, followed by the overall structure of this dissertation.

1.1 Motivation

MIS has been widely practiced in hospitals since the 1980s. Instead of making large incisions on patient to gain access to internal anatomy, MIS operates through small incisions with long surgical instruments and viewing equipment. Nowadays this technique has been widely adopted in various surgical procedures including laparoscopy for abdominal cavity, thoracoscopy for chest cavity, arthroscopy for joints, pelviscopy for pelvis, and angioscopy for blood vessels, because it has numerous advantages over the traditional open surgery, such as less access trauma, reduced postoperative pain, shorter hospital stays and faster recovery time, which are cost-effective for both hospitals and patients [1].

Despite these advantages, MIS introduces drawbacks for surgeons in performing operations through small incisions such as constrains on the manoeuvrability of the surgical tool, restriction of the visual feedback, and the impossibility of directly manipulating tissue. This latter shortcoming is one of the main limitations of MIS [2]. In traditional open surgery, sense of touch plays an important role. It enables the surgeons to interact with the organs or tissues and manipulate them freely, e.g. feel the tissue hardness, measure the tissue properties, and apply appropriate force control actions for safe tissue manipulation [3]. However, in the MIS, the surgeon's ability of

perceiving force (kinesthetic) and tactile information, referred to as haptic perception when the tissue interacting with the medical instrument, is severely impaired. Performing the surgery without such sensory information could lead to the exertion of excessive forces, which can cause unintentional damage and stress to healthy tissues, or an accidental cut of blood vessels or nerves [4]. Furthermore, MIS restricts the surgeon's ability to sense the tissue hardness which is essential to identify the tissue abnormality. Therefore, in order to restore the surgeon's perceptual capability, knowledge of contact force and tissue elasticity is vital.

Consequently, restoring the sense of touch in MIS has been an active research topic and various sensing techniques have been developed [5-10] utilizing transduction principles such as piezoresistive, capacitive, optical, piezoelectric, acoustic, and magnetic. The choice and placement of the sensing elements are important matters that can greatly influence the quality of force measurement. Because forces are transferred through mechanical linkages, the effects of friction, backlash, gravity, and inertia are factors that must be taken into account when high measurement accuracy is needed. Therefore, sensors should be placed close to the locations that generate forces. Initially, force and tactile sensors have been integrated in the instrument shaft outside the patient's body, as the sensor size is less restricted outside the body. However, the friction forces with the trocar (point where the instrument enters the body) and inertial forces on the instrument shaft disturb the measurement. The best solution, but technically the most challenging, is to integrate sensors at the tip of instruments. To integrate sensors on the tip of surgical tools, miniaturization is one of the most challenging factors for medical instruments such as guidewires and catheters, where ultra-miniaturization and high sensitivity are essential.

1.1.1 Force sensor for contact force measurement

Catheterization is a conventional process for diagnostic and interventional treatment of blood vessels. In this procedure, medical instruments such as catheters and guidewires are inserted into patient body passing through blood vessels to treat tissues and organs, and the instruments are controlled outside of the patient body at their distant end as

Figure 1-1: Passage of the guidewire © 2004 IEEE [5]. Inset shows the passage of guidewire under radiation.

Figure 1-2: Screen shot captured from angiogram during surgery. The guidewire passes through the lesion and the balloon catheter opens up the blocked artery to allow the artery to re-open (pictures from the National University Hospital of Singapore).

illustrated in Figure 1-1. Limitations such as the totally lost of force sensation, insufficient vision information from 2-D images [3], and exposure of patients and surgeons to radiations during X-ray fluoroscopy, make this procedure challenging which requires certain level of operational skills from surgeons, especially in the cases of stenosed lesion and blood vessels of complex configurations as illustrated in Figure 1-2.

In order to increase the navigation safety and accuracy of the catheter and guidewire inside blood vessels, many techniques have been developed including magnetic resonance imaging (MRI) and intravascular ultrasound (IVUS). MRI technique has

being greatly used to guide the catheter and guidewire insertion [11] with the advantage of providing high quality 3-D images without harmful radiations.

IVUS devices are also becoming popular nowadays as it can offer high-resolution anatomical images of the blood vessel from inside body [12, 13]. Despite of the aforementioned advantages, both MRI and IVUS techniques may not be able to identify tumours, nerves, vessels, and hard organic structures buried under the tissue surfaces [2]. In the case that imaging techniques are not capable of providing a clear view of tissue lesions, the touch sensation can then be used as another perceptual modality to gain necessary information [14-18]. Meanwhile, studies have shown that the contact force information is a determinant factor for determining lesion size in the catheterization procedures [19-21]. Thus, development of catheters and guidewires with tactile sensing capability by integrating force sensors at tip has become of great interest.

The ultra-miniaturization and sensitivity requirements for sensors to be integrated on guidewires are challenging as the tip diameter of a typical guidewire is only 0.36 mm, which is among the smallest instruments inserted into human body. A number of studies have reported good results on sensor miniaturization and have demonstrated the successful integration of sensors on guidewire. However, quite a number of the developed sensors turn out to be complex and difficult to be integrated on guidewires. Moreover, most of them have changed the standard manufacturing flow of guidewires to fit in with specific sensor designs.

Therefore, in this thesis, it is necessary to develop a miniaturized tri-axial force sensor which can be appropriately mounted at the distal tip of a guidewire. The sensor is required to be highly miniaturized (dimension $< 400\text{ }\mu\text{m}$) to be fit into the distal tip of a standard guidewire, highly sensitive to detect forces, and biocompatible. Together with the movable core wire, the sensor should be able to detect three axis (pressure and shear) contact forces while maintaining the original welded tip which is an essential part of the standard guidewire assembly.

1.1.2 Resonant tactile sensor for tissue elasticity measurement

In order to perform the MIS surgery effectively, the knowledge of tissue elasticity is important, based on the fact that the elasticity or stiffness of tissues is an index of

normal/abnormal of the soft tissues, which can be used as an indicator of human diseases. To measure the tissue elasticity, many methods have been previously developed, such as MRI methods, deformation-based methods and vibration or resonance-based methods. Most of the deformation and vibration based methods involve stiffness probes to provide a controlled deformation or force, which have limited the miniaturization of the measurement devices and also have the problem of fabrication complexity. It becomes even more challenging for applications within a confined and limited space, such as in MIS. Regarding to these drawbacks, it is necessary to develop miniaturized elasticity sensors of high sensitivity and able to be scaled up into arrays with good spatial resolution.

This dissertation presents a novel silicon bulk-mode resonant tactile sensor which has been realized based on the principle of changes in the resonance frequency when its vibrating tip is brought into contact with an object. Upon contact with the material/object, the resonant frequency shifts corresponding to the amount of mass and stiffness loading from the contact object. Unlike the conventional piezoelectric vibrating tactile sensors that generally consist of two piezoelectric elements as transducer and vibration pick up, the actuation and sensation of the bulk-mode resonator can be integrated together, which allows further miniaturization and possible MIS tool integration. In the previous work, silicon BAW resonators of different geometric structures have been intensively studied, such as disk [22-24], ring [25, 26] and square [27, 28] shape resonators. They show to have high resonance frequencies, have quality factors larger than 10000 in air and have the possibility to be scaled up into arrays. These features make the bulk-mode resonant tactile sensor a promising solution for material elasticity measurement.

1.2 Research objectives

As discussed above, the objective of the work reported in this dissertation is to develop MEMS-based miniaturized i) force and ii) tactile sensors that can be used for tissue palpation in minimally invasive surgery, measuring tool-tissue interaction forces at the tip of guidewire and measuring tissue elasticity. Both types of the sensors are developed based on MEMS technology.

The developed sensors should not be limited to MIS application but also have the potential for use in other biomedical applications such as in robotics assisted procedures and teleoperations wherein haptic feedback is required, to measure the contact forces between tissue and the surgical instrument, to enable tissue stiffness measurement and for other possible applications.

1.3 Dissertation organization

This dissertation is divided into six main chapters detailing the development of MEMS based force and tactile sensors. This introduction is the first chapter.

Chapter 2 provides a literature review of previous research work related to the topic of this dissertation, including the review of MEMS sensors for guidewires, the review of MEMS tri-axial force sensing technology and the existing technology for elasticity measurement. Meanwhile, this chapter also addresses how the sensors developed in this dissertation are differentiated from the previous research work.

Chapter 3 presents the design of a silicon nanowire based ring-shaped tactile sensor. The device sensing principle and fabrication process are described. In addition, the experimental characterization results and discussion are elaborated.

Chapter 4 introduces a biomimetic tactile sensor that inspired from the Meissner's corpuscles mechanoreceptor in human skin. A ball is attached on top of the ring sensor to increase the shear force sensibility. The design and packing process of the sensor is elaborated following by the discussion of the experimental characterization results.

Chapter 5 describes the design concept of a MEMS resonant tactile sensor that is based on the principle of bulk acoustic wave resonance. Numerical (Finite Element Analysis) and analytical methods are used to study the motion of the resonator and the output characteristics. The sensor is electrically characterized by electrostatic driving, capacitive and piezoresistive sensing methods and the important resonator parameters are extracted. The material contact behaviour of the developed sensor is presented.

Finally, a general conclusion is given in Chapter 6, along with an outlook for future work.

Chapter 2 Literature review

The objective of this dissertation is introduced in Chapter 1, i.e. to develop MEMS force sensors for guidewire tip contact force measurement and tactile sensors for elasticity measurement. In view of the objective, this chapter presents a thorough literature review of previous research work along with the benchmark study through the review of MEMS sensors for guidewires, the review of MEMS tri-axial force sensing technology and the existing technology for elasticity measurement. Meanwhile, this chapter also addresses how the sensors developed in this dissertation are differentiated from the previous research work. While many types of MEMS force and tactile sensors have been developed for applications related to minimally invasive surgery during the past two decades, only those designs that are directly relevant to this project are reviewed, focusing on the transduction principles of piezoresistance and bulk acoustic resonance (BAW). For a complete review of the tactile sensing mechanism and the applications of tactile sensors in MIS, review articles are available by Puangmali et al. [8], Tiwana et al. [10], Fearing [29], Eltaib and Hewit [30], Lee and Nicholls [31].

2.1 Review of force sensing technology

2.1.1 Force sensors for catheters and guidewires

As discussed in Chapter 1, development of catheters and guidewires with tactile sensing capability by integrating force sensors at tip has become of great interest. The forces experienced by surgeons originate from two sources, i.e. contact force between the guidewire tip and the blood vessel, and the friction force between the guidewire and the blood vessel walls [32]. To measure the force interaction between a catheter or guidewire and blood vessels, a sensor should be integrated at the tip and / or sides of the catheter or guidewire. Up to date, various kinds of force and pressure sensors have been

reported with the attempts to integrate those sensors on tip of catheters and guidewires [5, 10, 20, 33-41]. Meanwhile, a number of difficulties arise from both the aspects of the integration requirements and the sensor specifications.

One of the most important concerns is associated with the miniaturization process. In order to integrate a sensor on a catheter or guidewire, its dimensions are constrained by the diameter of the catheter or guidewire (usually less than 2.3 mm for catheter). Due to this limitation, several force sensors that are developed for other MIS applications, such as [42-45], are not suitable for catheter-based application. Commercial systems with force measurement abilities at catheters are available recently [46] [47]. Fibre-optic-based sensing has been proved to be a very promising technology for force measurement in MIS because of high sensitivity and compatibility with MRI. A series of fibre optic sensors have been developed for integration within cardiac catheters to provide force or pressure feedback from the catheter tip [20, 48, 49]. However, to integrate sensors on the tip of guidewires, the ultra-miniaturization requirement is more challengeable as the tip diameter of a typical guidewire is only 0.36 mm, which is among the smallest instruments inserted into human body.

The sensor specifications include the force measurement range, resolution, and sensitivity. Depending on the tip flexibility, guidewires are generally classified into three types, i.e. floppy (tip load < 0.5 g), balanced (tip load 0.5 g ~ 0.9 g) and extra support (tip load > 0.9 g). The tip flexibility is measured as tip load, that the amount of force (or load) it requires to bend the guidewire a certain amount to avoid excessive forces at the tip which may result in bleeding and stroke. The force measurement range of a sensor can be then determined according to the tip load requirement of a given type of guidewire. Higher resolution sensors are helpful to distinguish small changes in force. In addition, high sensitivity is required for a sensor to detect the initial contact forces. Furthermore, the sensor should also have high linearity and low level of hysteresis. Piezoresistive sensors are promising candidates as they can satisfy the criteria requirements on the miniaturization and the sensor specifications.

A number of studies have reported good results on sensor miniaturization with piezoresistive sensing principle and have demonstrated successfully the integration of such sensors on guidewires. Kalvesten et al [35] reported firstly the integration of a

miniaturized sensor on a guidewire for balloon angioplasty application. In this work, a surface micromachined piezoresistive pressure sensor was mounted in a recess near the tip of a guidewire with an outer diameter of 0.36 mm, from the work of which a commercialized guidewire named Pressure-Wire was produced by St Jude Medical Systems, as shown in Figure 2-1[50]. The reported sensor has dimensions of $100\text{ }\mu\text{m} \times 150\text{ }\mu\text{m} \times 1300\text{ }\mu\text{m}$ and a polysilicon diaphragm area of $103 \times 103\text{ }\mu\text{m}$. The sensor integrated guidewire was demonstrated to be able to detect the location and the severity of the stenosis inside blood vessel. The sensor has a sensitivity of $2.0\text{ }\mu\text{V}\text{V}^{-1}\text{mmHg}^{-1}$, the operation range from -30 to +300 mmHg and a resolution of $\pm 1\text{mmHg}$. Comparing to optical type of sensors, the advantage of this piezoresistive sensor is that the proximal (out-of-body) contact of the guide-wire pressure sensor is disconnectable, which means the pressure can be measured both before and after the balloon has been guided into a coronary artery, keeping the guidewire at the same place during the operation. As another example, T. Meiss et al [51] reported a MEMS force sensor integrated on a $360\text{ }\mu\text{m}$ diameter guidewire, named *HapCath* system, to provide haptic feedback of the forces acting on a guidewire's tip during vascular catheterization. This piezoresistive force sensor has a diameter of $200\text{ }\mu\text{m} \times 200\text{ }\mu\text{m} \times 640\text{ }\mu\text{m}$ and a reported force resolution of 1 mN. The sensor was successfully integrated and encapsulated at the tip of guidewire with an elastomer, which resulted in a different tip structure with commercial guidewire.

Those early research attempts on developing MEMS sensors for guidewire-related applications were mainly focused on mounting pressure sensors near the tip of the guidewires and designed force sensors on the tip of the guidewires in which only one dimensional contact force or pressure were possibly acquired whereas tri-axial force information is more desirable in catheterization procedures. Moreover, most of the reported sensors have changed the standard manufacturing flow of guidewires to fit in with their specific sensor designs. In this work, we report a novel ring-shaped tri-axial force sensor for easy integration with guidewire. The specific sensor requirements includes: (1) the total size of the sensor is smaller than the inner diameter of the guidewire, which is $360\text{ }\mu\text{m}$; (2) the sensor is able to detect forces as large as 20 mN, with a minimal detectable force of 0.1 mN; (3) the sensor is robust to stand a maximum

force of at least 2 N; (4) meanwhile, the sensor is required to have linear response and minimal hysteresis.

Figure 2-1: Illustration of pressure wire with a transducer near the distal tip of it. Image courtesy St. Jude Medical [50].

Figure 2-2: (a) & (b) Illustration of the working principles of the sensor element in Hapcath and (c) the sensor encapsulated into the guidewire tip by elastomer [51].

With the ring-shaped structure design, the sensor can be easily mounted on the circular core wire at the tip of a guidewire. Together with the movable core wire, the sensor should be able to detect three axis (pressure and shear) contact forces while maintaining the original welded tip which is an essential part of the standard guidewire assembly. Comparing with previous reported tri-axial force sensor designs in [37, 39, 52] the ring shape design is easy for sensor-guidewire assembly and adaptable to the commercial guidewire structures. In addition, the assembly configuration of a ring-shaped sensor

minimizes the bending artifacts as it is directly integrated on the distal tip of the guidewire, whereas most of the previous reported sensors for catheterization applications were mounted in a recess near the tip of the core wire or on the wall of catheter [35, 36, 53].

In the following section, the up-to-date development of the piezoresistive tri-axial force sensor will be explored and benchmarked.

2.1.2 Tri-axial force sensors worldwide

Regarding to the tri-axial force sensors, a considerable amount of research has been conducted on developing tri-axial force sensors to detect both normal and shear forces in robotics and biomedical applications [52, 54-61]. Table 2-1 summarizes the recently literature reported tri-axial force sensor designs. Kane et al [55] proposed a poly-silicon based piezoresistive complementary metal-oxide semiconductor CMOS - compatible tactile sensor for grasping applications in robotic manipulators, where poly piezoresistors are embedded in the four bridges as sensing elements. A sensitivity of 1.59 mV kPa^{-1} in normal direction and 0.32 mV kPa^{-1} in shear direction was reported. However, due to the flat shuttle plate design without additional shear force enhancing mechanism, the shear sensitivity was reported five times lower than the normal sensitivity. To increase the shear force sensitivity, force sensors of various types of structure designs have been developed, including diaphragm-type sensors designs with additional mesa acting as shear force transferring element [52, 56-58, 60] and biomimetic tactile sensors designs with epidermal ridges [61]. Vasarhelyi et al [59] reported a tri-axial force sensor that has an assembled epoxy mesa across a hole of a suspended membrane. Boron implanted p+ single crystalline silicon piezoresistor was formed at the edge of cantilever as sensing material. Beccai et al [52] developed a sensor with a fabricated mesa as part of the sensor. P-type piezoresistors with a size of $6\text{-}10 \text{ }\mu\text{m} \times 30\text{-}50 \text{ }\mu\text{m}$ are fabricated on SOI wafer. Hu et al [60] reported a hair cell inspired sensor for minimally invasive surgery with a fabricated mesa at middle. Polysilicon piezoresistors are located at the edge of membrane for force detection. Nevertheless, most of the previous reported tri-axial force sensors are designed for robotics and biomedical applications and cannot be directly adapted for catheterization applications due to their incompatible mechanical structure designs, as well as their bulk

dimensions. As indicated in the third column of Table 2-1, the sensing elements of the reported piezoresistive tri-axial force sensors are mostly made from polysilicon or boron implanted piezoresistors whereas the size of the sensing elements themselves have limited the sensor structure to be further scaled down.

The recent research on nano-scale transduction mechanisms such as silicon nanowires (SiNWs) and carbon nanotubes have provided great opportunity to significantly reduce the size of MEMS devices for biomedical applications where ultra-miniaturization is essential, such as the guidewire applications. Compared to the conventional metal strain gauge and bulk implanted piezoresistors, the outstanding piezoresistive performance of single crystal silicon nanowires has been reported by many research groups [62-67] and the scalability of the SiNW-based device has been successfully demonstrated in many MEMS device designs [68-71]. Therefore in this work, to achieve high sensitivity and ultra-miniaturization requirement, silicon nanowires (SiNWs) as nanometer scale piezoresistors are adopted as sensing elements. In the following section, principle of piezoresistance and a review of SiNW technique will be presented.

2.1.3 Principle of piezoresistance

2.1.3.1 Piezoresistive effect

Piezoresistance is a material property defined as the change in bulk resistivity due to an applied stress or strain. Piezoresistive behaviour has been observed in many materials, but much of today's research focuses on the piezoresistive behaviour of silicon, as silicon is the prime material used in MEMS fabrication processes. The first report of piezoresistance was by William Thomson in 1856 with regard to resistance change in iron and copper during their elongation [72]. The problem of telegraph wire signal propagation changes that occurred to telegraph companies have further motivated the

Table 2-1: Summary of recently reported tri-axial force sensor designs

Shear force transfer method	Application	Sensing element	Size (μm)	Sensitivity	Linearity	Force range	References
Fabricated silicon rod	Minimally Invasive Surgery	Polysilicon Piezoresistor	Diaphragm: $420 \times 420 \times 7$ Post: H 470	$S_{Dz} = 6.0 \times 10^{-3} \text{ um}^{-1}$ $S_{Dx} = 1.2 \times 10^{-3} \text{ um}^{-1}$ $S_{Fz} = 3.5 \text{ N}^{-1}$ $S_{Fx} = 10.8 \text{ N}^{-1}$	n.a.	0-0.05N	Hu, et al. 2010 [60]
Added epoxy mesa	Robotic control	Boron implanted p+ piezoresistor	$300 \times 300 \times 45$	4-6 mV/mN/V	n.a.	0-3mN	Vasarhelyi, et al. 2006 [59]
Silicon mesa fabricated on the sensor	amputees prosthesis	Boron implanted p+ piezoresistor	$2.3 \text{ mm} \times 2.3 \text{ mm} \times 1.33 \text{ mm}$	$S_z = 0.032 \pm 0.001 \text{ N}^{-1}$ $S_x = S_y = 0.0323 \text{ N}^{-1}$	0.997	0-2N	Beccai, et al. 2005 [52]
planar structure	Grasping in robotic manipulator	Polysilicon Piezoresistor	Array: $19.2 \text{ mm} \times 19.2 \text{ mm}$	$S_z = 1.59 \text{ mV kPa}^{-1}$ $S_x = S_y = 0.32 \text{ mV kPa}^{-1}$	0.999	Normal: 0.35 mN^a Shear: 0.6 mN^a	Kane et al. 2000 [55]
Attached SU8 square mesa	Hand grasping activity	Boron implanted p+ piezoresistor	Diaphragm: $1.9 \text{ mm} \times 1.9 \text{ mm} \times 50$ Mesa: H 60	$S_z = 1.57 \text{ N}^{-1}$ S_x, S_y varies with loading conditions	0.98	0-3N	Wang and Beebe 2000 [56]
Stylus attached by epoxy resin	Material characterization	Piezoresistor	$5 \text{ mm} \times 5 \text{ mm} \times 0.36 \text{ mm}$	$S_z = 0.36 \text{ N}^{-1}$ $S_x = S_y = 13.2 \text{ N}^{-1}$	n.a.	n.a.	Bütefisch et al. 2001 [57]
Guidewire or micro pillar	Guidewire navigation	Silicon nanowire	$380 \times 380 \times 10$	$S_{Dz} = 13.4 \times 10^{-3} \text{ um}^{-1}$ $S_{Fz} = 0.734 \text{ N}^{-1}$	0.999	20 mN	This work

^a Unit converted from normal force range of 35 kPa and shear force range of 60 kPa to milli Newton in order to be identical with other design

Figure 2-3: Illustration of a simple resistor subjected to a longitudinal stress, σ_x

research on conductivity under strain [73-75]. In 1954, Smith firstly reported the piezoresistive effects in silicon and germanium [76]. After the piezoresistance in semiconductor materials being discovered, in 1957, Mason and Thurston first reported the application of using silicon strain gauges for measuring displacement, force and torque. Their work was considered a leap forward in sensing technology with the discovery of the fact that the sensitivity of semiconductor strain gauges is fifty times higher than of conventional metal strain gauges [77].

For a rectangular resistor of isotropic material as shown in Figure 2-3, the material resistance is given by

$$R = \rho \frac{l}{wh} \quad (2.1)$$

where R is the resistance, ρ is the resistivity of the material, l , w , and h are the length, width and thickness of the resistor respectively. As observed from the equation (2.1), the resistance is a function of the resistor dimensions and the resistivity ρ .

When a voltage, U , is applied at the two ends of the resistor, according to Ohms law, the resulting current, I , is related to the electrical resistance by the following equation

$$U = RI \quad (2.2)$$

When the resistor is subjected to an applied stress σ_l in x -direction as shown in Figure 2-3, by differentiating the equation (2.2) in both sides, the change in resistance is given by

$$\begin{aligned}
dR &= \frac{\partial R}{\partial l} dl + \frac{\partial R}{\partial w} dw + \frac{\partial R}{\partial h} dh + \frac{\partial R}{\partial \rho} d\rho \\
dR &= \frac{\rho}{wh} dl - \frac{l\rho}{w^2h} dw - \frac{l\rho}{wh^2} dh + \frac{l}{wh} d\rho \\
\frac{dR}{R} &= \frac{1}{l} dl - \frac{1}{w} dw - \frac{1}{h} dh + \frac{1}{\rho} d\rho
\end{aligned} \tag{2.3}$$

For a finite change in the resistance, equation (2.3) turns into

$$\frac{\Delta R}{R} = \frac{\Delta l}{l} - \frac{\Delta w}{w} - \frac{\Delta h}{h} + \frac{\Delta \rho}{\rho} \tag{2.4}$$

With the Poisson's ratio equations,

$$\begin{aligned}
\Delta l &= \varepsilon_l \cdot l \\
\Delta w &= -\nu \varepsilon_l \cdot w \\
\Delta h &= -\nu \varepsilon_l \cdot h
\end{aligned} \tag{2.5}$$

The resistance change can be written in the following form:

$$\frac{\Delta R}{R} = (1 + 2\nu) \varepsilon_l + \frac{\Delta \rho}{\rho} \tag{2.6}$$

where ν is Poisson's ratio, ε_l is the stress induced strain in x-direction.

The first term in equation (2.6) is the contribution from the change in geometry $(1+2\nu)$ and the second is that from the resistivity change $(\Delta\rho/\rho)$ [78]. For metal foil strain gauges, resistance change is mainly due to the geometric effect. The Poisson's ratio of most metals ranges from 0.20 to 0.35. For semiconductors, the change of resistance is mainly due to the change in resistivity.

The relative resistance change in an isotropic material can be expressed as a function of the applied stress σ_x

$$\frac{\Delta R}{R} = \frac{\Delta \rho}{\rho} = \sigma_x \pi \tag{2.7}$$

where π is the piezocoefficient.

The piezoresistive effect, i.e. the gauge factor of a material is defined mathematically as:

$$GF = \frac{\Delta R / R}{\varepsilon} = (1 + 2\nu) + \frac{\Delta \rho / \rho}{\varepsilon} \quad (2.8)$$

For anisotropic semiconductor materials such as silicon, the piezoresistivity and elasticity varies for different spatial directions in the crystallographic lattice [79]. The piezoresistive effect can be described using a piezocoefficient tensor π . The piezoresistive coefficient tensor of silicon has a simple form in a crystallographic coordinate system. Smith has fully characterized the piezoresistive tensor of silicon at a doping resistivity of $7.8 \, \Omega \cdot \text{cm}$ by using a designed test configuration.

$$\begin{aligned} \pi &= \begin{pmatrix} \pi_{11} & \pi_{12} & \pi_{12} & 0 & 0 & 0 \\ \pi_{12} & \pi_{11} & \pi_{12} & 0 & 0 & 0 \\ \pi_{12} & \pi_{12} & \pi_{11} & 0 & 0 & 0 \\ 0 & 0 & 0 & \pi_{44} & 0 & 0 \\ 0 & 0 & 0 & 0 & \pi_{44} & 0 \\ 0 & 0 & 0 & 0 & 0 & \pi_{44} \end{pmatrix} \\ &= \begin{pmatrix} 6.6 & -1.1 & -1.1 & 0 & 0 & 0 \\ -1.1 & 6.6 & -1.1 & 0 & 0 & 0 \\ -1.1 & -1.1 & 6.6 & 0 & 0 & 0 \\ 0 & 0 & 0 & 138.1 & 0 & 0 \\ 0 & 0 & 0 & 0 & 138.1 & 0 \\ 0 & 0 & 0 & 0 & 0 & 138.1 \end{pmatrix} \\ &\quad \times 10^{-11} \text{Pa}^{-1} \end{aligned} \quad (2.9)$$

After the first experimental measurement of values of π_{11} , π_{12} and π_{44} by Smith in 1954, the piezocoefficients have been extensively studied. It has been found that the piezoresistive effect depends on many factors including the temperature, the doping type, the doping level and the pre-strain in the crystal lattice [65]. Usually the piezoresistors are designed along a specific direction for the purpose of obtaining maximum piezocoefficients in real applications. The common configuration of a piezoresistive element is to form a relatively long narrow resistor defined in a planar structure [65, 76]. The primary electric field and current direction are usually confined

along the longitudinal axis of the resistor. Usually the piezoresistors are defined along $\langle 100 \rangle$ or $\langle 110 \rangle$ direction on a (100) wafer. This is because firstly (100) wafers are most commonly used due to their relatively low cost, and secondly the two orientations, i.e. $\langle 100 \rangle$ on n-type and $\langle 110 \rangle$ direction on p-type silicon wafer shows the highest piezoresistive effect in comparison with other configured piezoresistors on the (100) wafer plane [65]. In this work the piezoresistive silicon nanowires are fabricated on $\langle 110 \rangle$ direction of a p-type wafer at (100) plane to maximize the sensitivity.

2.1.3.2 SiNW piezoresistance

Silicon nanowire (SiNW) is one of the nanoscale structures that made from silicon and usually have a lateral dimensions less than 400 nm (width and height) with unconstrained length. With the potential in downsizing devices, the SiNW has significant impact in the migration of Micro-electro-mechanical-system (MEMS) to Nano-electro-mechanical-system (NEMS). During the past ten years, many interesting and remarkable properties of SiNW have been reported [62, 66, 80-83], and because of the merits such as relative ease of scaling down, high sensitivity, and CMOS compatibility, etc., various potential sensor applications have been demonstrated. He and Yang reported an dramatically increased piezoresistive effect, known as giant piezoresistance effect in p-type silicon nanowires in the $\langle 111 \rangle$ direction [62]. They found that the piezoresistance depends on the SiNW diameters and resistivity and the largest piezoresistive coefficient value of $3550 \times 10^{-11} \text{Pa}^{-1}$ were found in the longitudinal direction. Inspired by the experimental results of He and Yang, Reck et al [82] designed and fabricated silicon chips to study the piezoresistive properties of crystalline and polycrystalline nanowires using a top-down approach. Compared to the bulk silicon's piezoresistive effect, they found a 633% increase in crystalline nanowires and 34% increase in polysilicon nanowires with decreasing dimensions.

The effect of giant piezoresistance is still debatable among research groups, but migrating from bulk piezoresistors to SiNWs will increase the initial piezoresistance, thus, a lower power consumption at a given supply voltage [71]. Furthermore, the scalability of the device will be significantly improved and such improvement has been demonstrated in many MEMS pressure and force sensor designs recently [69, 71, 84, 85]. Kim et al [86] reported a pressure sensor using suspended SiNWs. They claimed

that the SiNW of 140 nm x 200 nm shows seven times higher piezoresistive effect than that of bulk silicon. However, the fabricated SiNW has a rough surface profile and the suspended structure in the sensor design need to be protected from fatigue. Meanwhile, various works have also been reported toward increasing the sensor robustness without decreasing the sensitivity. W.T. Park et al [84] reported a tri-axial force sensor design utilizing SiNW as sensing element for guidewire applications as shown in Figure 2-4. The sensor is protected by adding a mechanical stopper through post-package. Unfortunately the sensor tip was too fragile to be integrated into real assembly process of commercial guidewires.

Figure 2-4: (a) Schematic of force sensor design and position of SiNW. (b) Cross-sectional schematic demonstrating the vertical Si rod connecting the cross beams and the mechanical stopper [84].

There are many reported methods for the integration of nanomaterials onto MEMS devices to utilize sensing functionality. The silicon nanowires can be either fabricated using the bottom-up approach and then transferred onto the pre-fabricated MEMS device [87], or can be fabricated by top-down technique directionally along with the MEMS device fabrication [88]. In this work, SiNWs with a cross section of 90 nm x 90 nm will be fabricated by the top-down approach using lithographic pattern transfer and thermal dry oxidation processes.

2.2 Resonant tactile sensor

2.2.1 Technologies for tissue elasticity measurement

The change of tissue stiffness is an important index for many diseases such as breast, liver and prostate cancer [18, 89-93]. Since the stiffness of malignant tissue such as a tumour is typically higher than the surrounding healthy tissue, a surgeon can use the measurement of stiffness to identify abnormal tissue regions. Tissue stiffness can be defined by the resistance of a rigid body to deformation by an external force [94]. Young's modulus, defined as the ratio of stress over strain in a region, is generally used to describe the stiffness of soft tissue. The force sensors described in previous section are mainly designed for contact force measurement and cannot be directly used to measure the tissue elasticity. In order to measure the tissue elasticity, many methods have been developed and can be generally classified into three categories, i.e. magnetic resonance imaging (MRI) methods [95-97], deformation-based methods [98-104] and vibration or resonance-based methods [105-111].

MRI is a clinically valuable imaging technique that can provide exquisite in vivo anatomical images of soft tissues. In 1971, Damadian [95] was firstly demonstrated the ability of using MRI to differentiate cancer tissue from noncancerous tissue. Since then, many investigators have duplicated this type of study to obtain the quantitative information on biological and physical properties of tissues [97]. MRI enables noninvasive in vivo measurement of the elasticity of tissues. However, the high cost has prevented the frequent use of this technique.

Another technology for elasticity or stiffness measurement of tissues is based on measurement of force-deformation response. Researcher have performed experiments on human tissue using this technique, such as breast [101], brain [112], and liver [113] tissues, by applying loading on tissue surface and recoding the corresponding reaction force, and then reconstruct the tissue Young's modulus from the measurement. Wellman et al [101] performed the deformation type tissue stiffness tests on cancerous breast tissue by estimating elastic moduli from measured force displacement curves. The early developed stiffness measurement instruments based on indentation are generally hand-held devices and presents difficulties to be used in MIS applications due

to its poor spatial resolution inherent to its large form factor which cannot fulfil the miniaturization requirement for MIS applications.

Another approach of measuring stiffness or elasticity of materials is based on the resonance vibration technique. This technology is based on the fact that the resonance frequency of the sensor shifts relative to the stiffness of the contacted material. Various resonant tactile sensors are developed based on this principle by measuring the frequency shift between a freely vibrating resonator's resonance frequency and the resonance frequency measured when the sensor makes contact to an object. This type of tactile sensors have also been applied in multiple applications, such as for detection of breast cancer [114], for measuring muscular elasticity [105], characterization stiffness of human ovum [115, 116], and also for prostate tissue stiffness measurement [117].

2.2.2 Resonant tactile sensors worldwide

The resonant vibrating sensors consist of an actuating element which is an essential component in the measurement system. In 1992, Omata and Terunuma [105] firstly developed a resonant tactile sensor that consists of two piezoelectric elements, one as transducer and the other as vibration pick up. As shown in Figure 2-5 (a), when an alternating voltage is applied across its electrode, the PZT element will vibrate in the direction of its length. The pick-up detects this vibration and feeds the alternating signal to driving amplifier in a feedback circuit. The circuit drives the PZT -element and the system oscillates at its resonance frequency. If the free end of the PZT element, or a contact piece attached to that end, touches an object the resonance frequency characteristics and frequency of the system will change. The amount of change depends on the acoustic impedance of the object. The sensor system output signal is the shift of the oscillation frequency from unloaded to loaded condition as illustrated Figure 2-5 (b). This sensor shows that it could sense hardness or softness of an object like human hands dose. Since then, a number of piezoelectric tactile sensors based on the resonant sensing technique have been developed and evaluated [108, 114-116, 118, 119]. For example, a piezoelectric bimorph tactile sensor was reported by Hemsell et al [120] as shown in Figure 2-6. The device consists of an ultrasound scaler with a modified contact ball to increase contact surface area and a piezoelectric sensor used as a sensing element. The resonance frequency of the vibrating element change with varying loads, which allows

the calculation of mechanical load parameters. As illustrated in Figure 2-5 (c) and Figure 2-6, most of the early developed stiffness probes which include a vibrating piezoelectric element to provide a controlled deformation or force, have limited the miniaturization of such probes and also increases the fabrication complexity. It becomes even more challenging in applications within a confined and limited space, such as in MIS. Regarding to these drawbacks, researcher have been trying to develop miniaturized elasticity sensors of high sensitivity and able to scaled up into arrays with good spatial resolution.

Figure 2-5: (a) Schematic of the PZT resonant vibrating sensor [105, 114]. (b) Resonance frequency curve shifts when the sensor is in contact with object. (c) A prototype of the resonant sensor probe [121].

Figure 2-6: Piezoelectric bimorph sensor, arrows indicating direction of vibration [120].

Figure 2-7: (a) Schematic view of magnetically driven active tactile sensor [106]. (b) Principle of detecting elastic and damping coefficient of the sensor. (c) Resonant characteristics of diaphragm vibration.

Figure 2-8: (a) Schematic of the IPMC resonant vibrating tactile probe [109]. (b) Prototype of the IPMC sensor. Two strips of IPMC used, one working as actuator and one as sensor along with the added mass for the tuning of the resonant frequency.

Lately, Hasegawa et al. [106] reported an magnetic actuated tactile sensor that able to detect the hardness of an object. As shown in Figure 2-7 (a), the sensor consists of a diaphragm with a mesa at the centre, piezoresistors at the periphery of diaphragm, and a chamber for pneumatic actuation. To detect the hardness distribution, the contacted mesa element is pneumatically driven towards the object as illustrated in Figure 2-7 (b). The contacted region of the object is deformed according to the driving force of the mesa element and the hardness of the object. Then, from the relationship between the resulting deformation and frequency shift, the hardness of the contact object can be evaluated Figure 2-7 (c). However, precise pneumatic drive and control of a matrix of these sensors could be a difficult task. Brunetto et al. [109] worked on developing polymer-based resonant tactile sensor based on ionic polymer metal composites (IPMCs) as an alternative solution (Figure 2-8). The sensor consists of two strips of IPMC that one used as actuator and the other one as sensor along with the added mass for the tuning of the resonance frequency as shown in Figure 2-8 (b). If mechanical deformation occurred, a low voltage will be generated across their thickness and causes the IPMCs bend. Their reported results show that the IMPC tactile probe can be used to distinguish the different materials. Table 2-2 summarizes the recently literature reported tactile sensors for material elasticity measurement.

In this dissertation, a new type of miniaturized resonant tactile sensor based on MEMS silicon bulk-mode resonator is proposed for material elasticity measurement which will be reported in Chapter 5. The sensor is developed based on simple MEMS fabricated bulk acoustic wave (BAW) resonator, incorporated with rounded sensing tip and mechanical stopper to prevent over loading. Previously, silicon BAW resonators, like disk [22-24], ring [25, 26] or square [27, 28] shape resonators, have been intensively studied for more than a decade. They feature high resonance frequencies, shown to have quality factor of >10000 in air [122] and can be easily scaled up into arrays. These features make the BAW resonant tactile sensor a promising solution for precise and high-resolution 2-D material elasticity mapping.

Table 2-2: Summary of recently reported tactile sensors for material elasticity measurement

Reference	Peng et al. 2010 [98]	Hasegawa et al. 2006 [106]	Kudo et al 2007&2010 [118, 123]	Kudo et al. 2006 [124]	This work (Chapter 5)
Device characteristics
	Based on the measurement of force-capacitance response	Based on force-deformation response	Based on measurement of contact impedance	Based on measurement of resonance frequency change	Based on resonance frequency change measurement, electrostatic driven
Feature	Compose two capacitive sensing bumps with different stiffness values	Piezoresistor sensing and magnetic actuating tactile sensor	Longitudinal bar-type piezoelectric resonator	Combined a longitudinal bar resonator and a double-ended tuning fork resonator as tactile and force sensor	Consist of a silicon bulk acoustic wave resonator designed with a mechanical stopper
Fabrication	Surface micromachining; polymer MEMS process	MEMS micromachining and micro-assembly process	fabricated from metal alloy and piezoelectric ceramics	Manufactured from SUS304 stainless steel by electric discharge machining; piezoelectric ceramic plates attached	Silicon micromachining
Dimensions	1 mm x 1 mm	6 mm x 6 mm	16 mm x 4 mm	55mm x 6 mm	2 mm x 2 mm

Chapter 3 SiNW-based tri-axial force sensor for sensorized guidewire

The trend towards minimally invasive medical procedures drives the development of ultra-miniaturized sensors that are highly sensitive and physically robust. Minimally invasive-based catheterization procedures, such as cardiovascular and thoracic interventional procedure, are becoming increasingly popular in the medical practice for both diagnosis and surgery. In these procedures, medical instruments such as catheters and guidewires are inserted into patient body passing through blood vessels to treat tissues and organs. Comparing with the traditional open surgery, there are numerous advantages of the minimally invasive procedure. However, because the medical instruments such as catheters and guidewires are controlled outside of the patient body at the distant end, limitations such as the totally lost of force sensation, insufficient vision information from 2-D images, and exposure of patients and surgeons to radiations during X-ray fluoroscopy, make the procedure challengeable and requires certain level of operational skills from surgeons, especially in the cases of stenosed lesion and blood vessels of complex configurations. Therefore, there is a need to develop a miniaturized tri-axial force sensor to be integrated at the tip of guidewires.

The sensor is required to be highly miniaturized (dimension less than $360\text{ }\mu\text{m}$) to be fit into the distal tip of a standard guidewire, the sensor should be able to detect three axis (pressure and shear) contact forces while maintaining the original welded tip which is an essential part of the standard guidewire assembly. From the technical requirement, the designed sensor should satisfy the following requirements: (a) The total size of the sensor should be smaller than the inner diameter of the guidewire, which is $360\text{ }\mu\text{m}$; (b) The sensor should be able to detect forces as large as 20 mN , with a minimal detectable

force of 1 mN; (c) Meanwhile, the sensor is required to have linear response and minimal hysteresis.

This chapter presents the design, modelling, fabrication and characterization of the novel ring-shaped tri-axial force sensor for guidewire application. By using MEMS technology, a force sensor that utilizing silicon nanowires (SiNWs) as piezoresistive sensing elements is developed. With a ring-shaped structure design, this sensor can be easily mounted on the circular core wire at the tip of a guidewire. Together with the movable core wire of the guidewire, the sensor can detect contact forces while maintaining the original welded tip which is an essential part of the standard guidewire assembly. The overall sensing area including cantilever beams of this sensor is designed to be $200\text{ }\mu\text{m} \times 200\text{ }\mu\text{m}$ whereas the total area of the sensor unit is less than $360\text{ }\mu\text{m} \times 360\text{ }\mu\text{m}$. This dimensional miniaturization is an essential design requirement to assure it is capable to be integrated on the distal tip of commercial available guidewires.

Finite element analysis is used to determine the location of maximum stress and to estimate the performance of the sensor when it is integrated with a guidewire. The simulation results are also used to be verified with the experimental measurements from sensor characterization. SiNWs of a cross section of $90\text{ nm} \times 90\text{ nm}$ are fabricated by the top-down technique using lithographic pattern transfer and thermal dry oxidation processes on a SOI wafer directionally along with the MEMS device fabrication. By taking advantages of the high sensitivity of SiNWs, the fabricated sensor has been demonstrated to be capable of detecting small displacement in nanometer ranges with a sensitivity of $13.4 \times 10^{-3}\text{ }\mu\text{m}^{-1}$ in the z-direction which yields a normal force sensitivity of 0.734 N^{-1} with a resolution of 0.383 mN. The sensor has shown to meet the designed requirement of minimal detectable force of 1 mN. The sensor also has shown to exhibit a linear response to applied force. A high linearity ($> 99.9\%$) between the piezoresistance changes with respect to the applied loading on the sensor was obtained. No obvious hysteresis is observed from the characterization results. This developed SiNW based force sensor provides new opportunities to implement force sensing capability on minimally invasive surgical instruments, including but not limited to guidewires and robotic surgical grippers, where ultra-miniaturization and high sensitivity are essential.

3.1 Ring-shaped force sensor design

3.1.1 Sensor design

The ring-shaped sensor design is inspired from the requirement of mounting a force sensor on the tip of a movable core guidewire. Meanwhile, the sensor should be adaptable to standard guidewire manufacturing process flow without changing the commercial guidewire structures. Generally the structure of a guidewire comprises a distal tip, a central core wire, a coil, and elastic cover. According to the structural design of the central core wire, commercial guidewires are mainly divided into two types, fixed core guidewire and movable core guidewire [67, 125] as shown in Figure 3-1. The fixed core guidewire or 1- piece core guidewire as shown in Figure 3-1 (a) has a tapered core wire that connects the weld distal tip and the proximal end. The movable core guidewire, which was introduced by Simpson et al in 1982 [126], has an independently movable, flexible-tipped core wire as illustrated in Figure 3-1 (b). The moveable core guidewire is more flexible and steerable than the fixed core guidewire, and enables to reduce the trauma while traversing along blood vessels. In general the tip of the standard movable guidewire is straight at the natural state and moveable in all directions. During insertion, the tip of the guidewire can be bent into different shape configurations under outside control of surgeons. Comparing to the fixed core configuration, the force sensation from the flexible tip of the moveable guidewire is hardly perceived. As shown in Figure 3-1 (c), a force sensor is designed to be integrated on the movable part of core wire to measure the tip contact force between guidewire and vessel walls. When external forces are applied at the distal tip of the guidewire, displacement differences between the distal tip and the coil will induce a deformation on the sensor structure. The sensor measures forces in both normal and shear directions as shown in Figure 3-1 (d). An interface ASIC is integrated with the force sensor to read the force information applied on the sensor in the form of resistance and converts it to current pulse [127]. This ASIC interface also functions to reduce the number of the input and output signal lines.

Figure 3-1: Illustration of the distal tip of the sensorized guidewire design: (a) standard fixed core guidewire (b) standard movable core guidewire (c) standard movable core guidewire with integrated MEMS force sensor and ASIC (d) sensorized guidewire under forces applied at distal tip (e) Illustration of double-coil design with integrated MEMS force sensor.

Since the force sensors are fabricated from silicon, it is mechanically fragile. To prevent the sensor from breaking off, an integration scheme can be designed to avoid the force overloading. The integration of the designed sensor and the movable core guidewire is illustrated in Figure 3-1 (e). A two coil structure is designed with the purpose of inducing a differential displacement between the movable core wire and the first coil to lead the ring deformation. The sensor is to be integrated with the movable core wire which is connected with the distal tip and located in the first coil. When external force

applied at the end of the guidewire, displacement differences between the distal tip and the first coil will induce a deformation of the sensor structure. The spring stiffness of the two coils can be appropriately designed to limit the displacement of the ring in a safety range such that the robustness of the sensor can be guaranteed.

The designed force sensor consists of a suspended ring-shaped structure located at the centre of four suspended cantilever beams with the axes perpendicular to each other, as shown in Figure 3-2 (a). Single crystal silicon nanowires (SiNWs) are embedded at the end of each beam as the piezoresistive sensing element to leverage its high gauge factor due to their high sensitivity and miniaturization achievable. Comparing with previous reported tri-axial force sensor designs in [37, 39, 52], the ring shape force design is easy for sensor-guidewire assembly and adaptable to the commercial guidewire structures. In addition, the assembly configuration of the ring-shaped sensor minimizes the bending artifacts as it is directly integrated on the distal tip of the guidewire, whereas most of the previous reported sensors for catheterization applications were mounted in a recess near the tip of the core wire or on the wall of catheter [35, 36, 53].

Depending on the tip flexibility, guidewires are generally classified into three types, Floppy (tip load < 0.5g), Balanced (tip load 0.5g ~ 0.9g) and Extra support (tip load > 0.9g). For each type there are many available products on market. The tip flexibility is measured as tip load, that the amount of force (or load) it requires to bend the guidewire a certain amount to avoid damaging the vessel wall, for example in the cerebral vasculature surgery usually the more flexible of the tip the better. The force measurement range of a sensor is generally determined according to the tip load requirement of a given type of guidewire. For the force sensor designed in this work, a force measurement range of 20 mN force is required from a given guidewire of 2g tip load. In order to reduce the effect quantization error, the sensor resolution should be one order of magnitude less than the force difference that needs to be resolved. Higher resolution sensors are helpful to distinguish small changes in force. In this work, the minimal detectable force of 1 mN is required. .

3.1.2 Force sensing principle

The relationship between the applied external force and the displacement on the SiNW ring shaped sensor is

$$F = K_e \Delta x \quad (3.1)$$

where F is the applied force, K_e is the equivalent spring stiffness of the sensor structure, and Δx is the ring displacement.

The piezoresistors change their resistance due to the beam deformation and the induced strain. The resistance change is given by

$$\frac{\Delta R}{R} = \pi_l \sigma_l + \pi_t \sigma_t \quad (3.2)$$

where R is the resistance without applied stress, ΔR is the resistance change with applied stress, π and σ represent the piezoresistive coefficient and stress respectively, and the subscripts l and t refer to longitudinal and transverse components [128].

The resistance change corresponding to the X-Y-Z direction are given by

$$\begin{aligned} S_x &= \left(\frac{\Delta R}{R} \right)_x = \frac{\Delta R_3}{R_3} - \frac{\Delta R_4}{R_4} \\ S_y &= \left(\frac{\Delta R}{R} \right)_y = \frac{\Delta R_1}{R_1} - \frac{\Delta R_2}{R_2} \\ S_z &= \left(\frac{\Delta R}{R} \right)_z = \frac{\Delta R_1}{R_1} + \frac{\Delta R_2}{R_2} + \frac{\Delta R_3}{R_3} + \frac{\Delta R_4}{R_4} \end{aligned} \quad (3.3)$$

where R_1 , R_2 , R_3 , and R_4 are the resistance of each of the resistors as labelled in Figure 3-2 (a). When shear force is applied, the opposite piezoresistors such as R_1 and R_2 experience opposite strain (i.e. R_1 will experience tensile strain whereas R_2 will experience compressive strain, or vice versa). When normal loading is applied, the resistance of each four piezoresistors has the same sign of strain (either tensile or compressive). The Z axis sensitivity is defined as the total resistance change of the four piezoresistors.

The total dimension of the sensor was firstly determined according to the structural dimension of the guidewire which is typically 360 μm at the tip. Considering restrictions such as the overall dimension of the designed sensor should not exceed the inner diameter of the coil and, the diameter of the inner ring should be larger than the diameter of the movable core wire, the inner ring diameter was designed to be 100 μm

and the diameter of the outer ring was designed to be $116\text{ }\mu\text{m}$. The cantilever beam has a size of $42\text{ }\mu\text{m} \times 8\text{ }\mu\text{m} \times 10\text{ }\mu\text{m}$ in length, width, and thickness respectively as shown in Figure 3-2 (b). The overall sensing area including cantilever beams is $200\text{ }\mu\text{m} \times 200\text{ }\mu\text{m}$. In the future design of the device, the critical structural parameters of the sensor can be optimized to maximize the sensor performance, such as the inner and outer diameters of the ring, and the cantilever length, width and thickness.

Figure 3-2: (a) Schematic of the ring-shaped force sensor design. (b) Dimensions of the sensor structure: beam $42\text{ }\mu\text{m} \times 8\text{ }\mu\text{m} \times 10\text{ }\mu\text{m}$, inner ring diameter $100\text{ }\mu\text{m}$, and outer ring diameter $116\text{ }\mu\text{m}$.

3.1.3 FEM simulation

A finite element analysis simulation tool, COMSOL[®] V4.2, was used to study the mechanical behaviour of the structure. The material properties used in the numerical analysis are shown in Table 3-1 [129-131]. The ring was modelled as a double layer structure with a layer of $9\text{ }\mu\text{m}$ silicon and $1\text{ }\mu\text{m}$ silicon oxide (SiO_2) on top. In order to find the location with the maximum sensitivity to implement the piezoresistors, regions of highest stress on the beams are identified by finite element modelling (FEM) as shown in Figure 3-3. When normal force is applied on the top surface of the ring, the sensor beams bend downward and the maximum stress is found at the bases of the beams. SiNWs are embedded between the layers of silicon and SiO_2 at these high stress

regions in order to obtain the maximum piezoresistance output. To avoid the fracture of the structure, a value of 2.8 GPa [132] was used as the silicon yield stress in the FEM analysis, and the maximum force that applied on the sensor was 20 mN which was deduced according to the tip load requirement of the given type of guidewire , i.e. 2g tip load in this case.

A FEM simulation of the designed ring-shaped sensor integrated with a steel wire is presented in Figure 3-4, where the steel wire represents the distal tip of a movable core guidewire. For simplification, the function of coil spring and the assembled distal tip of the guidewire were not considered in the simulation. As shown in Figure 3-4 (a), the steel wire passes through the ring and they are firmly attached by epoxy. In this case, the steel wire acts as force transferring element when shear load is applied. The stress field for normal and shear load is presented in Figure 3-4 (b). When a normal load is applied, the lateral stress components in the four SiNWs are equal in magnitude with same direction (Figure 3-4 (b) left). When a transverse loading is applied in the x- or y-direction, the two SiNWs along with the loading direction experience stress of the same magnitude with opposite signs while the other two SiNWs perpendicular to the loading undergo torsion (Figure 3-4 (b) right).

Table 3-1: Material properties for guidewire & sensor integration modeling

Material	Young's Modulus	Poisson Ratio	Density
Silicon	169 GPa	0.278	2330 Kg/m ³
SiO ₂	70 GPa	0.17	2200 Kg/m ³
Steel	200 GPa	0.33	7850 Kg/m ³
Epoxy	3.5 GPa	0.069	1250 Kg/m ³

Figure 3-3: COMSOL modelling results of the ring structure under vertical load. The right colour table shows the von Mises stress field of the sensor in response to 5 mN force applied in the z-direction. The maximum stress is found at the bases of the beams wherein the SiNWs are embedded.

Figure 3-4: (a) Illustration of the ring sensor and steel wire integration by epoxy (b) Sensor deformation under applied forces F_z (left) and F_x (right). The colour table shows the von Mises stress field in response to 20 mN force applied in the z- and x- direction respectively.

3.2 Device fabrication

The force sensor fabrication involves two main steps which are the SiNW fabrication and cantilever release. The critical fabrication steps are summarized in Table 3-2.

The starting material was a (100) - oriented p-type silicon-on-insulator (SOI) wafer of a 117 nm thick top silicon layer on a 145 nm thick buried oxide (BOX) layer (Table 3-2 (a)). The top silicon layer was pre-doped with the resistivity of $13.5 \Omega \cdot \text{cm}$. The SiNWs were created by standard photolithography using Nikon KrF DUV lithography scanner (model NSR-203B) along the $\langle 110 \rangle$ crystal orientation (Table 3-2 (b)). Figure 3-5 demonstrates the SiNW piezoresistor orientation with respect to the [110] wafer crystallographic axes orientation on the (100) SOI wafer plane. The possible alignment error arising from the crystal orientation variation between the SOI device layer and the wafer notch position and its effect on the SiNW piezoresistance can be found in Appendix A.

The mask critical dimension (CD) for NW was 160 nm. Once the silicon layer was patterned, the critical dimension of the NW line width was trimmed down to approximately 110 nm by etching the photoresist with a plasma-induced feeding gas ($\text{He}/\text{O}_2 + \text{N}_2$). After the silicon etching and resist removal process, a two hours thermal oxidation process was carried out at 875°C to grow SiO_2 on the patterned SiNW. At the end of this step, SiNWs with an approximate length of 6 μm and a cross section of 90 nm x 90 nm were achieved along the $\langle 110 \rangle$ crystal orientation. After that, p-type ion implantation was performed at the SiNW regions by BF_2^+ with a dosage of 1×10^{14} ions cm^{-2} and energy of 100 keV (tilt 7° and twist 22°). Then a photolithography was performed to protect the SiNWs from the later high doped implantation at the resistor connecting regions (paddle regions). A higher doped implantation with a dosage of 1×10^{15} ions cm^{-2} and an energy of 35 keV (tilt 7° and twist 0°) was performed at the three paddle regions for low resistivity silicon trace and silicon to metal contact. The three paddles as indicated in Table 3-2 (c)-(e)) were used to connect two nanowires and form an electrical signal path. Then the dopants were activated by rapid thermal annealing (RTA) at 1000°C for 30 s, and a 0.4 μm silicon oxide premetal dielectric was deposited (Table 3-2 (c)) followed by photolithography and silicon oxide etch for contact opening. After metal deposition, the metal was patterned by photolithography and metal etch.

The process was continued by silicon oxide passivation and pads opening by photolithography and oxide etch (Table 3-2 (d)).

After the SiNWs fabrication, the cantilevers and the ring of the sensor were released by the Deep Reactive Ion Etching (DRIE) process. This fabrication process was started with patterning a 2.5 μm photoresist sacrificial layer on the silicon oxide layer by standard photolithography as shown in Table 3-2 (e). Next, silicon oxide was etched followed by a 10 μm silicon DRIE process. The thickness of the cantilever and the ring was defined as 10 μm at this step. After resist removal, a thin layer of silicon oxide was deposited by plasma-enhanced chemical vapor deposition (PECVD) process. The purpose of the PECVD process was to protect the sidewall from the silicon isotropic DRIE process during final cantilever release. After the PECVD process, the oxide was etched back by oxide dry etching process to open the silicon area for isotropic plasma etch release. The oxide on the floor area was etched out while the sidewall oxide was remained. Then, a 5 μm anisotropic DRIE process followed by 5 μm isotropic plasma etching (using DRIE etcher) was processed to release the cantilever ring structure at the front. Then grinding and polishing were carried out at the wafer backside before the final back DRIE releasing process. The force sensor structure was finally released by backside DRIE process up to the BOX layer as shown in Table 3-2 (f). SEM images of the fabricated silicon nanowire and the ring shaped force sensor with four embedded silicon nanowire piezoresistors are shown in Figure 3-6.

Figure 3-5: Diagram illustrating the orientation of the fabricated force sensor on the (100) silicon wafer the SiNW piezoresistor orientation with respect to the [110] wafer crystallographic axes orientation on the (100) SOI wafer plane.

Figure 3-6: SEM of the fabricated silicon nanowire and the ring shaped tactile sensor with four embedded silicon nanowire piezoresistors.

Table 3-2: Fabrication process flow for the SiNW-based ring shape force sensor.

Main fabrication flow	Process	Description
(a) Wafer inspection	SOI wafer, 117 nm buried oxide (BOX) layer, 145 nm top silicon (device) layer lowly doped of boron (p-type) with resistivity of 13.5
(b) SiNW pattern and thermal dry oxidation	Photolithography for SiNW patterning (160nm) and resist trimming to 90nm. 2 hours thermal oxidation, followed by low doped implantation.
(c) SiO ₂ PMD deposition	High doped implantation at paddle regions and 0.4um Silicon oxide PMD (Premetal Dielectric) deposition
(d) Contact opening and metallization	Metal deposition, photolithography pattern and metal etch., then silicon oxide passivation and pad opening by photolithography and oxide etch
(e) ring structure pattern	Photolithography, silicon oxide etch, and 10um DRIE to form the ring structure at front
(f) front and back side release	A 5 μm anisotropic and 5 μm isotropic DRIE process for front side release and back side DRIE up to BOX layer is carried out for final release						
<table border="0"> <tr> <td> Silicon (Substrate)</td> <td> Silicon (Device layer)</td> <td> Photoresist</td> </tr> <tr> <td> SiO₂ (BOX layer)</td> <td> SiO₂ (PECVD)</td> <td> Aluminum</td> </tr> </table>			Silicon (Substrate)	Silicon (Device layer)	Photoresist	SiO ₂ (BOX layer)	SiO ₂ (PECVD)	Aluminum
Silicon (Substrate)	Silicon (Device layer)	Photoresist
SiO ₂ (BOX layer)	SiO ₂ (PECVD)	Aluminum						

3.3 Sensor characterization

3.3.1 Measurement setup

In order to calibrate and evaluate the functionality of the fabricated SiNW force sensor, characterization experiments have been conducted and the simulation results of characterization were used to verify experimental measurements. A schematic diagram of the experimental setup is shown in Figure 3-7. The characterization system consists of a probe station with a microscope for visual control, a piezoactuator (PI-245) connected to a piezo motion controller (E517) for displacement loading control, and a Keithley 4200-SCS semiconductor analyzer to obtain the experimental data and, meanwhile to control the piezo-controller. As demonstrated in Figure 3-7, the sensor wire that bonded on a PCB board was fixed on the probe station through vacuum suction. The resistance of the SiNW was measured through the pads on PCB and recorded by the semiconductor analyzer. Displacement loadings were applied through a piezoactuator (PI-245) which was controlled by the piezo-controller (E517). The resolution of the piezo-controller is 1.6 nm with a displacement range of 0-80 μm which allows precise control of the step displacement. As shown in Figure 3-8 (a), the displacement loading with various step sizes was applied through a probe needle at the end of one cantilever beam. The needle was firmly fixed in the probe holder which is connected with the motion controller. The displacement of the structure was obtained by recording the displacement of the probe needle. To reduce the measurement uncertainty, we have considered the following conditions during the experiment,

- Use a probe needle with large stiffness, and can firmly attach on the probe holder,
- Use a probe with a relatively large tip diameter,
- Decrease the approaching speed of the needle to increase the contact stability.

A tungsten needle with a tip diameter of 3.5 μm was chosen as the loading probe for the measurement. The loading speed was considerably controlled to avoid the needle slipping during contact.

Figure 3-7: Schematic diagram of the experimental setup for the ring-shaped force sensor characterization. The inset is the enlarged view of the loading probe on top of the sensor.

Figure 3-8: Demonstration of the probe loading at the end of one cantilever under the view of microscope (a) and simulation (b), where R1, R2, R3, and R4 represent the resistance of SiNW resistors embedded at the anchor of each cantilever. The colour bar in (b) represents the stress of the sensor in response to 5 μm prescribed displacement applied in the z direction.

3.3.2 Results and discussion

The repeatability and hysteresis test results for one SiNW piezoresistor were plotted in Figure 3-9 with different displacement step size of (a) 50 nm, (b) 0.5 μm , and (c) 1 μm respectively. In each case, the normal displacement was applied with an increasing step of (a) 50 nm, (b) 0.5 μm and (c) 1 μm and then decreased back to zero with the same step size along the z-direction. The loading cycle was repeated two times for each case. Figure 3-9 (left) plots the resistance changes with respect to time. Linear fitting equations and adjacent R^2 values were calculated for the loading and unloading phases to obtain the hysteresis cycle as shown in Figure 3-9 (right). In Figure 3-9 (a), the SiNW ring shaped force sensor showed nano-scale displacement detection in a 50 nm step size with high repeatability. From Figure 3-9 (a) right, the end point of unloading cycle was not exactly zero. The error may be caused by resistance drift due to temperature as there was no temperature control implemented for the setup. From the repeatability and hysteresis test results, the sensor demonstrated high repeatability, high linearity (> 99.9%) and no obvious hysteresis were observed as the two slopes of loading and unloading were almost coincident with each other as shown in Figure 3-9 (right). In addition, A resulting sensitivity of $S_{Dz} = 13.4 \times 10^{-3} \mu\text{m}^{-1}$ was obtained in z-direction. From the noise level of the output resistance value as shown in Figure 3-10, the displacement detection resolution of the force sensor was determined to be 20.7 nm at a sampling rate of 8Hz. To compare the work with recently reported force sensor designs, the sensitivity and linearity of the ring shaped sensor were summarized in Table 2-1 with other recently reported piezoresistive tri-axial force sensors. As shown in the column five and six of Table 2-1, comparing with the other recently reported force sensor designs using bulk implanted piezoresistors as sensing elements, the linearity and sensitivity of the SiNW based force sensor have been improved.

Figure 3-9: Repeatability and hysteresis tests for one SiNW piezoresistor on the sensor with displacement step size of (a) 50 nm, (b) 0.5 μm , and (c) 1 μm respectively. The linear fitting equations and adjacent R^2 values are shown in the right figures for the loading and unloading phases of the hysteresis cycle. Nano scale displacement detection in a 50 nm step size with high repeatability is shown in (a). High linearity (99.9%) and no obvious hysteresis are demonstrated.

Figure 3-10: Noise level of the output resistance value from the force sensor.

The resistance change ratio of four SiNW piezoresistors with respect to displacement is plotted in Figure 3-11 (a). A normal loading was applied on piezoresistor R_1 as demonstrated in Figure 3-8 (a). Obtained experimental results on $\Delta R/R$ versus normal displacement in Figure 3-11 (a) showed qualitative agreement with the stress-strain simulation of the ring shaped force sensor as shown in Figure 3-11 (b). Ideally when normal loading is applied, four piezoresistors under tensile stress should have the same signs of strain as shown in Figure 3-11 (b). However, a negative value of resistance change was observed for $\Delta R_2/R_2$ in Figure 3-11 (a) which means R_2 was under compressive stress during the loading cycle. This was caused mainly due to the error in vertical loading on the sensor that the angle between the loading probe and the sensor was not ideally 90° .

The force sensitivity of the device can be obtained from the displacement sensitivity characterization based on the stiffness of the beams. Each four of the identical cantilever can be modelled as a fix-guided suspended beam that one end is fixed at the support silicon and the other end is guide-fixed on the silicon ring. A vertical loading force F_z is applied at the guided end of the suspended beam. The relationship between the applied force and the resultant bending of the cantilever follows Hook's law

$$F = K_e \Delta x \quad (3.4)$$

where F_z is the loading force, K_e is the spring stiffness of the suspended beam, and Δx is the displacement of silicon ring.

The stiffness of each guided-beam with respect to normal loading force is given by

$$k_e = \frac{Ewt^3}{L^3} = 18.25 \times 10^3 \text{ N / m} \quad (3.5)$$

where E is the Young's modulus, L , w , and t are the length, width, and thickness of the rectangular beam, respectively. Thus, with the known normal displacement sensitivity S_{D_z} of loading displacement D_z , the change in resistance per input force in the normal direction can be expressed with respect to the obtained resistance change by following

$$S_{F_z} = \frac{S_{D_z}}{k_e} = \frac{\Delta R / R}{F_z} = 0.734 \text{ N}^{-1} \quad (3.6)$$

By applying (3.6) to the characterization results in Figure 3-10 (a), the normal displacement sensitivity of $13.4 \times 10^{-3} \mu\text{m}^{-1}$ yields a normal force sensitivity of 0.734 N^{-1} with a resolution of 0.383 mN, which has shown to meet the designed requirement of minimal detectable force of 1 mN. Generally SiNW piezoresistors suffer from non-linear behaviour. From the previous normal loading tests, the piezoresistance change ratio of the SiNW was found to be linear with a normal displacement loading applied up to 5 μm in the z-direction. The failure test was carried out on the sensor by increasing displacement loading at the end point of the beam until the failure point. The testing result is shown in Figure 3-12. The maximum standing force of the sensor should be obtained from the failure test. Unfortunately, due to the experimental limitation that no force gauge was available during the testing, the maximum force range was not obtained. The completed force loading characterization of the sensor would be done in the following work, and the linearity behavior of the SiNW under large force loading conditions should be discussed accordingly.

Future characterization will be performed with a cylindrical mesa passing through the ring of the sensor acting as the guidewire. The mesa acts as the shear force transferring

element in order to acquire the three components of the loading force, thus completing characterization of the sensor.

Figure 3-11: Experimental and simulation results: (a) Experimental results of four SiNW piezoresistance change when load applied near R_1 , (b) simulation results of von Mises stress at four SiNW piezoresistors.

Figure 3-12: Failure test of the force sensor under displacement loading

3.4 Influences of blood flow

In practical application, due to the complexity of the vessel shape and viscosity of the blood, the sensor that integrated on the guidewire distal tip will measure not only the contact force between the tip of guidewire and the vesicular wall, but also the resistance force that the blood flow imposed on the tip of guidewire. As the guidewire advancing direction is generally opposite to the direction of blood flow, the resistance raised from the blood flow environment will inevitably affect the performance of the force sensor. Therefore, by analyzing the resistant force that acting on the guidewire when it moving into the body blood vessel during the minimally invasive cardiovascular interventional surgery, the influences of the blood flow on the performance of the force sensor will be known.

The resistant force, impacting on the tip of guidewire, can be modeled the Stokes' drag equation, known as the force acting opposite to the relative motion of object moving with respect to its surround fluid

$$F_d = -6\pi\mu rV$$

where μ is the dynamic viscosity, r is the radius of the spherical distal tip of the guidewire, and V is the blood flow velocity relative to the guidewire travel speed. The

blood viscosity is about $3.5 \times 10^{-3} \text{Pa} \cdot \text{s}$, and for a commercial 360 μm guidewire, the spherical tip radius is 0.114mm (tip diameter 0.009in). The mean blood flow rate is 15 cm/s in a blood vessel that located 2 cm proximal to the carotid bifurcation. And the travel speed of a guidewire is generally operated at 1 cm/s. Therefore, the relative velocity of the flow is approximately 16 cm/s.

From the calculation, a resistant force of $1.2 \times 10^{-6} \text{N}$ is obtained. According to the characterization results, the resolution of the sensor was 0.383 mN, which means that the resistant force from the blood flow is too small to be detected by the force sensor. Therefore, the blood flow will not largely affect the measurement results of the force sensor.

3.5 Summary

In this chapter, a ring shaped tri-axial force sensor has been developed to be used on a guidewire in catheterization applications. The sensor has been designed with a ring structure for simple and smart integration with the distal tip of guidewire. The mechanical behaviour of the ring sensor with an integrated steel guidewire was simulated with FEM analysis. SiNWs with a length of 6 μm and a cross section of 90 nm x 90 nm have been successfully fabricated by top down approach on a SOI wafer. The device functionality has been verified by repeatability and hysteresis tests with different displacement step sizes including 50 nm, 0.5 μm and 1 μm . The sensor has been demonstrated to be capable of fulfilling the performance requirement of high repeatability, high linearity (> 99.9%) and negligible hysteresis with a resistance change of 1.34 percent per micron displacement in z- direction. In addition, the sensor shows nano-scale displacement detection in a 50 nm step size with high repeatability.

Chapter 4 Biomimetic micro-tactile sensor

Tactile sensors of flat membrane alone as sensing structure have their limitations on detecting forces in shear directions. As reported by Kane et al [55], the force sensitivity in shear direction is approximately five times lower than in normal direction for a sensor composed of a free-standing central shuttle planet as sensing structure. To increase the shear force sensitivity, tactile sensors of various types of structure designs have been developed, including diaphragm-type sensors designed with additional mesa [52, 59, 60] and biomimetic tactile sensors designed with epidermal ridges [61].

This Chapter presents the development of a microscale biomimetic tactile sensor that inspired from the Meissner corpuscle mechanoreceptor in the human glabrous skin, to enhance the force sensitivity of the ring-shaped sensor of flat surface that developed in Chapter 3. The Meissner corpuscle inspired tactile sensor is designed with a rigid spherical ball attached on top of the suspended ring structure located at the centre of four suspended beams which mimics the biological structure of the Meissner corpuscles. Silicon nanowires (SiNWs) embedded at the anchor of four beams can be used as sensing elements to represent the neural signal transmission between the mechanoreceptor and the somatosensory area in brain. The sensor was successfully developed with a simplified fabrication and packaging process by dropping spherical balls on the SiNW-based ring structure. Finite element analysis (FEA) has been used to predict the mechanical behaviours of tactile sensor with rigid solder ball and with elastic bump covering of rigid solder ball underneath. The performance of the SiNWs as piezoresistive sensing elements on the bio-inspired sensor and on the bare ring-shaped sensing element has been comparatively studied by repeatability and hysteresis tests. The effectiveness of sensitivity enhancement of the bio-inspired tactile sensor with solder ball attached is verified from the experimental results, in that the fractional change in resistance in normal direction is improved from 1.34 percent to 2.8 percent pe

micro meter with a linear response (>99.9 %). The designed artificial tactile sensor offers further potential for a variety of applications, such as integrating on robotic hands, biomedical devices or embedding into artificial skins as sensing elements

4.1 Introduction

The highly optimized performance of nature has always provided humans with the source of inspiration and models for creation. Accordingly, humans have been increasingly making efforts to understand the nature and to mimic or being inspired to make artificial counterparts from the nature [133]. In particular, the unique functionalities and properties of the human skin have led to extensive research on developing skin inspired designs [134, 135]. Along with the understanding of the functional morphology and the mechanotransduction mechanism of the human skin, various types of biomimetic tactile sensors have been developed to acquire the touch information, such as material properties, contact events, and normal and shear forces. These tactile sensors have been widely used in various applications such as minimally invasive medical devices [8, 10], robotics [9, 136, 137] and biomedical prosthetics [138, 139].

There have been various studies conducted on the importance of the mechanical properties and neural mechanisms of the cutaneous surface and the subcutaneous mechanoreceptors in human skin [140-144]. Maeno et al. [145] analyzed the relationship between the structure of human finger skin and the location of the cutaneous mechanoreceptors using finite element analysis and found that both the geometry of the papillae and the epidermal ridges are important to increase the sensitivity of tactile sensation. T.Iwanaga et al. [146] experimentally studied the fine structure of Meissner corpuscles (MCs) in the glabrous skin of a monkey and found that each MC was completely covered by a connective tissue capsule and linked with the epidermis by dermal collagen fibers. A possible stimulus propagation mechanism was indicated from their findings: external forces imposed on the epidermis caused the deformation of the collagen fibers, through which the deformation was transferred into

the corpuscles. Based on Iwanaga's finding, Kuroki et al. built a structural model of MCs and proposed that the MCs can encode the normal strain by detecting the transverse normal strain [147], which indicated a directional sensibility of the MCs to the skin deformation. Meanwhile, various methods have been built to analyze the geometric properties of the cutaneous mechanoreceptors.

The development of biomimetic tactile sensors and systems have been guided by incorporating these mechanical features and functional characteristics of the human skin in the designs, such as including multiple sensing layers of variable stiffness [148, 149], emulating the epidermal ridges and fingerprint like structures as possible tactile amplification mechanism [61, 150], and mimicking the mechanical properties and distributed mechanoreceptors like human skin [138]. For example, as one of the pioneer works on the development of anthropomorphic tactile sensors, Dario et al. [149] reported a tactile transducer for use in prosthetics and robotics. The sensor has a composite, multi-layered structure of different layer stiffness, which mimics the dermal and epidermal skin layers. Zhang [61] reported a biomimetic tactile sensor with artificial epidermal ridges inspired from the glabrous skin to increase the force detection sensitivity. From experimental results, they demonstrated a 1.8 time increase on the sensitivity of the tactile sensor with the artificial epidermal ridges. Besides the improvement in the appearance and mechanics, tactile sensors were also designed to mimic functional attributes of human tactile mechanoreceptors. A tactile sensor array developed by Wettels et al. was reported in [138]. It was designed to mimic the mechanical properties and the positional distributions of tactile receptors in human skin. The sensor array was capable to detect a force range of 0.1 to 30 N with a spatial resolution below 2 mm.

As one of the four main mechanoreceptors, Meissner corpuscles (MCs) are rapidly adaptive type receptors located in the dermal papillae of glabrous skin and are shaped as ellipsoidal bodies. The MC is hypothesized to play a key role in response to dynamic skin deformation and transmit information of skin motion and slip detection [141]. Recently, X. Chen et al reported a biomimetic tactile sensor utilizing carbon micro-coils as sensing material [151]. The tactile sensor was made from carbon micro-coils (CMC)

that embedded into an elastic resin to mimic the morphology and dimensions of Meissner corpuscles mechanoreceptor. Although the CMC sensor has its advantage on material properties, it was also noticed that the hysteresis issue [152, 153] makes it challengeable to utilize the sensor in practical applications.

In this work, the design of a MEMS-based tri-axial tactile sensor that inspired from the structure and functionality of the Meissner corpuscles in human skin is presented. The MC-inspired tactile sensor is designed with a rigid spherical ball located at the centre of a suspended ring structure which emulates the ellipsoidal body of the Meissner corpuscles. Meanwhile, the spherical ball structure acts as a force transmitting element. Silicon nanowire (SiNW) fabricated by top-down approach is embedded at the four anchors of the ring structure and as the sensing element it represents the neural signal transmission between the mechanoreceptor and the somatosensory area in brain. By taking advantages of the MEMS technology, it allows to fabricate tactile sensors with similar attributes to the biological sensors, such as miniaturized sensor dimension and high spatial resolution in sensory array. In the following section, the design concept of the sensor is reported.

4.2 Sensor design

4.2.1 Tactile sensing in human skin

Figure 4-1 illustrates the subcutaneous structure of human glabrous skin and the location of the tactile mechanoreceptors [141, 142]. The human skin is identified as a viscoelastic material consisting of three layers - epidermis, dermis and subcutaneous layers. When a mechanical stress is applied on the skin, the stress is transmitted through the skin to the embedded mechanoreceptors, wherein the mechanoreceptors act as biological transducers which convert the mechanical stimulus into electrical signals in the sensory neuron.

There are four primary tactile mechanoreceptors distributed throughout the skin that specialized to provide tactile information of touch, force, vibration and cutaneous

tension, which are Meissner corpuscles, Pacinian corpuscles, Merkel discs, and Ruffini endings. These four types of mechanoreceptors have very different morphology at their nerve endings as illustrated in Figure 4-1, and meanwhile each of these mechanoreceptors differs with regard to their functional properties, such as the sensitivity to static and dynamic events, size and structure of their receptive fields, numbers and densities within the separate sub-regions of the glabrous skin area, and their perceptive effects [140]. Generally, the mechanoreceptors can be classified into two types - rapidly-adapting (RA) receptors for the Meissner corpuscles and Pacinian corpuscles, and slow-adapting (SA) receptors for the Merkel discs and Ruffini endings. The RA receptors are highly sensitive to the dynamic skin deformation and play an important role in contact and forces detection. In contrast, the SA receptors are mainly responsible for light pressure sensing and static discrimination such as detecting shapes, edges and rough textures [142].

Meissner corpuscles are the most common mechanoreceptors in glabrous skin which account for about 40% of the sensory innervation of the human hand. The dimension of the Meissner corpuscles is around 150 μm in height and 40-70 μm in diameter, and they are located in the papillary dermis of glabrous skin shaped as ellipsoidal bodies. As a RA type receptor, the Meissner corpuscles are known to play a key role in response to dynamic skin deformation and transmit information of skin motion and slip detection [154]. A observation from the study of Birznieks et al. reported that different types of mechanoreceptor have different direction-dependent sensitivity by applying directional forces to the finger pad [155]. Recently, T.Iwanaga et al. examined the fine structure of Meissner corpuscles in monkey [146] and suggested that the disk shape of MCs are susceptible to mechanical deformations of surrounding tissues and are possibly directional sensitive to the skin deformation.

Figure 4-1: Schematic view of the human glabrous skin illustrating the four types of mechanoreceptors and the neuron connection to the spinal cord [141, 142]. The mechanoreceptors are innervated by somatosensory neurons that project to the spinal cord.

4.2.2 Biomimetic tactile sensor

The miniaturized tri-axial tactile sensor is designed from the inspiration of the morphology structure and functional properties of the Meissner corpuscles mechanoreceptor as illustrated in Figure 4-2 (c). Figure 4-2 (a) shows a micrograph image of Meissner corpuscles in human glabrous skin [156]. A schematic drawing of the Meissner's corpuscles is illustrated in Figure 4-2 (b) [157].

The core part of the sensor is composed of a ring-shaped piezoresistive sensing element and a rigid spherical element that sits on the center of the ring structure which emulates the ellipsoidal body of the Meissner corpuscles. SiNW is embedded at the four anchors of the ring structure and as the sensing element it represents the neural signal transmission between the mechanoreceptor and the brain. When force is applied on the sensor, the resistance value will be changed accordingly due to the displacement induced strain on the sensing element. The ring-shaped tactile sensing element consists of a suspended ring structure located at the center of four suspended beams as illustrated in Figure 3-2. The overall dimension of the sensor with the spherical ball attached is $380\ \mu\text{m} \times 380\ \mu\text{m} \times 500\ \mu\text{m}$. The force sensing principle of the ring-shaped tactile element can be referred to section 3.1.2.

Comparing with tactile sensors of flat bulk surface design [55], a primary advantage of the Meissner corpuscles' inspired sensor designed with a rigid spherical element is its better sensitivity to normal and lateral forces. The placement of rigid ball on the ring can largely increase the force sensitivity on shear direction comparing to the flat ring-shaped structure, as the stress is accumulated in one cantilever, instead of equally distributed in two opposite cantilevers. Another unique property from the spherical geometry design is that the contact between the sensor and an arbitrary object is at a point, which concentrates the contact stress at the top of the sphere and makes the sensor appropriate to contact with arbitrary shapes. In the following section, a FEM model is created to analyze the mechanical behaviour of the sensor design.

Figure 4-2: Micrograph image of Meissner corpuscles in human glabrous skin [156]. (b) Schematic drawing of Meissner corpuscle [157], and (c) Tactile sensor design inspired from the Meissner corpuscle mechanoreceptor. SiNWs function as nerve plexus to transmit the touch information to a central unit.

Table 4-1: Material property for the biomimetic tactile sensor modelling.

Material	Young's Modulus	Poisson Ratio	Density
Silicon	169 GPa [131]	0.28	2330 Kg/m ³
Solder, (60Sn-40Pb)	10 GPa	0.4	9000 Kg/m ³
Stand steel	200 GPa	0.33	7850 Kg/m ³
Elastic cover	18 MPa [158]	0.49 [159]	230 Kg/m ³

Figure 4-3: Finite element model mesh of the MC-inspired tactile sensor with a rigid solder ball on top of the ring structure. The flat and rigid indentation touches the model when external loads apply on (a) Z-direction and (b) XY-direction.

Figure 4-4: Finite element modelling results - stress distribution at the region of silicon nanowires, caused by forces of different directions. The von misses stress value is shown in the left color bar of each image, while the force directions is depicted by the red arrows on the indentation. (a) and (b) show the tactile sensor with solder ball on top of the ring-shaped sensing element under normal and shear loading conditions, (c) and (d) show the tactile sensor with elastic bump covering under normal and shear loading conditions, respectively.

4.2.3 FEM modelling

A finite element model was created in COMSOL® V4.2, commercial FEM software, to conduct a contact analyze of the described sensor. The FEM analysis of the ring-shaped sensing element is studied in section 3.1.3, whereas the regions of highest stress on the beams were identified through the FEM modelling. SiNWs were implemented at the regions of highest stress to obtain maximum sensitivity.

Finite element models of the ring sensing element with rigid ball and elastic bump covering, subjected to the normal and shear forces applied through a contact object, are comparatively studied to analyze the mechanical behaviours of the sensor structures. Figure 4-3 illustrates the finite element model of the MC-inspired tactile sensor with a rigid ball on top of the ring structure. A flat object, defined as a rigid indentation to mimic the contact object, was placed to be in touch with the model from the top (Figure 4-3 (a)) and the side (Figure 4-3 (b)) when external loads apply on Z - and XY - directions. P1 and P2 indicate the location of SiNW piezoresistors on the cantilever beam. Table 4-1 lists the material properties that are used in the numerical analysis [131, 158, 159]. The elastic bump material is assumed to be homogeneous and isotropic, with a Young's modulus of 18 MPa and Poisson's ratio of 0.49 which mimics the dermis properties of human skin [158].

Figure 4-4 (a) and (b) shows the stress distribution on the tactile sensor with a solder ball on top of the ring-shaped sensing element, and Figure 4-4 (c) and (d) shows the tactile sensor with elastic bump covering. It can be seen that in both cases, when a normal force is applied through the contact object, SiNWs are under lateral stresses of equal magnitude and same direction (Figure 4-4 (a) and (c)), and when a transverse loading is applied in the xy-direction, the two SiNWs along with the loading direction experience stress of different magnitude as illustrated in Figure 4-4 (b) and (d). The FEM results of the von Mises stress at the locations of two SiNW piezoresistors (i.e. P1 and P2) are plotted in Figure 4-5. Figure 4-5 (a) shows the von Mises stress values when external force is applied at the Z - direction on the rigid ball and on the elastic bump, whereas Figure 4-5 (a) shows the results when external force is applied at the XY –

direction. It can be seen from both the normal and shear directional cases that for the same applied force, the stresses at the location of SiNWs are approximately two times higher for the sensor with rigid solder ball than with the elastic bump.

Figure 4-5: FEM results comparison between tactile sensors of solder ball and elastic bump covering. The von Mises stress at the locations of two SiNW piezoresistors (i.e. P1 and P2) are plotted when external force is applied at (a) the Z - direction and (b) XY – direction. For the same applied force, the stresses for sensor with rigid solder ball are approximately two times higher than the structure with elastic bump on top

4.3 Sensor fabrication

There were two main steps involved in the tactile sensor fabrication which are the microfabrication of the ring-shaped sensing element and the packaging with the rigid solder ball. The microfabrication process of the ring-shaped sensing element has been summarized in Table 3-2. It started on a p-type silicon-on-insulator (SOI) wafer at (100) plane with a 117 nm thick top silicon layer and a 145 nm thick buried oxide (BOX) layer. Two main fabrication steps were involved – SiNW piezoresistor formation and the ring shaped structure release. The SiNWs were created by standard photolithography and the resulting SiNWs have a length of 6 μm and a cross section of 90 nm x 90 nm. After the SiNWs fabrication, the cantilevers and the ring structure of the sensor were

released by Deep Reactive Ion Etching (DRIE) process followed by an isotropic plasma etching process.

Figure 4-6 illustrates the packaging process flow to assemble the bio-inspired sensor from the fabricated sensing elements. The assembly process started from attaching the tactile sensing elements on a PCB board using non-conductive epoxy as shown in Figure 4-6 (a). Then the ring area on the sensor was coated with flux before dropping the ball to protect the structure from fracture. Several types of flux from Indium Corporation were tested including NC 506, WS 366 and NC-026. One critical step in the packaging process was to choose the proper flux with minimal residual stress and appreciate viscosity. Otherwise, devices might be broken if using flux with large residual stress as shown in Figure 4-7 (b). Finally the flux of NC-026, with a residue of 5% was found to be the optimal solution with minimal residual stress and appreciate viscosity. After coating with flux, solder balls were dropped on sensing elements by using the solder jetting machine -SB² solder jetting tool at the room temperature. The SAC solder balls with the diameter of 120 μm were used with 140 μm solder jetting capillary as shown in Figure 4-6 (b). Unlike the case of placing balls manually, accurate alignment can be achieved at this step and the breakage of the cantilever due to misalignment can be also prevented. After the end of this step, the sensor chip with ball attached were put into the baking reflow machine for 10 minutes with a temperature of 260 degree to remove the flux residue. Then the chip was baked inside an oven to firmly attach the ball and the sensing element together. The tactile sensing element bonding pads were then wire bonded on a PCB board for electrical connection as shown in Figure 4-6 (c). Figure 4-8 shows the SEM picture of the packaged tactile sensors with rigid solder ball on the centre of the ring-shaped sensing elements.

Figure 4-6: Packaging process of the bio-inspired tactile sensor (a) The tactile sensing element is attached on a PCB board; (b) SAC solder ball is dropped and attached on the sensing element using solder jetting equipment; (c) Wire bonding the MEMS pad on PCB board.

Figure 4-7: (a) Flux residue on the device surface (WS 366). (b) Structure fracture due to large residual stress of flux (NC 506). The white material around the solder ball is the residue of flux. Charging effect happens during SEM

Figure 4-8: Fabricated tactile sensor with rigid solder ball on the centre of ring structure.

Figure 4-9: Schematic diagram of the experimental setup for tactile sensor characterization; loading probe is vertical to the sensor surface when a normal loading applies (b) on the ring-shaped sensing element and (c) on the biomimetic tactile sensor with rigid spherical ball attached under the view of a microscope

Figure 4-10: Plot of the repeatability / hysteresis tests for one SiNW piezoresistor on the bare ring shaped sensing element and on the bio-inspired sensor of solder ball, with a displacement step size of 0.5 μm.

Figure 4-11: (a) Needle applies force on top of the sensor, (b) Illustration of the sliding effect during loading

4.4 Characterization and experimental results

Figure 4-9 (a) shows the schematic of the characterization system setup. The sensor was firstly wire bonded on a PCB board and then placed on a probe station. External loading was applied through a loading probe that connected with a piezoactuator – PI245. The displacement of the piezoactuator was controlled by a piezo-controller, the resolution of which is 1.6 nm to apply precise step displacement. A semiconductor analyzer was used to record the resistance change of SiNW, and meanwhile apply command on the piezocontroller. Figure 4-9 (b) and (c) demonstrate that the loading probe is vertical to the sensor surface when a normal loading applies to the sensing element and the ball-attached sensor respectively.

The effectiveness of the SiNWs as piezoresistive sensing elements on the bionic Meissner corpuscles mechanoreceptor and on the bare ring-shaped sensing element has been comparatively studied by repeatability and hysteresis tests. The experimental results are plotted in Figure 4-10. A normal displacement was applied on the bare ring-shaped sensing element and on the bionic tactile sensor with solder ball attached on top respectively, with an increasing step of 0.5 μm and then decreased back to zero with the same step size after several steps. The loading cycle was repeated two times in each case. Figure 4-10 (a) plots the SiNW resistance changes with respect to time for the bare ring shaped sensing element and the bionic sensor. Linear fitting equations and adjacent R^2 values were calculated for the loading and unloading phases to obtain the hysteresis cycle as shown in Figure 4-10 (b). From the test results of the bare sensing element, the SiNW demonstrated good repeatability; linearity ($> 99.9\%$) and no obvious hysteresis were observed as shown in the Figure 4-10 (b) top. A resulting sensitivity of $S_{Dz} = 13.4 \times 10^{-3} \mu\text{m}^{-1}$ was obtained in z-direction for the bare sensing element.

From the repeatability and hysteresis test results, the sensor with solder ball shows hysteresis while in the bare ring sensor test there is no obvious hysteresis observed. This is mainly due to the misalignment between the tip position of the loading probe and the loading point on the ball. Sliding happens during the loading since the probe position is not precisely touching the top point on the ball. As a result, with the solder ball attached

on the ring-shaped sensing element, a resulting sensitivity of $28 \times 10^{-3} \mu\text{m}^{-1}$ was obtained in normal direction, which is two times higher than the bare ring-shaped sensing element of flat surface. This sensitivity incensement mainly due to the sliding effected during the needle loading. As illustrated in Figure 4-11, the needle may not precisely located at the top point of the ball which caused the ball rotate during loading. As a result, the piezoresistor at one side suffered an increased tensional stress and the piezoresistor at the opposite side suffered a higher compressional stress, which yields an increase piezoresistance change ratio.

4.5 Summary

The main focus of this chapter is to engineer a Meissner corpuscles inspired tactile sensor with the combination of MEMS and SiNWs. The performance of the SiNWs as piezoresistive sensing elements on the bio-inspired tactile sensor and the bare ring-shaped sensing element has been comparatively studied by repeatability and hysteresis tests. The effectiveness of sensitivity enhancement of the MC-inspired tactile sensor with solder ball attached is verified from the preliminary characterization results, in that the fractional change in resistance in normal direction is improved from 1.34 percent to 2.8 percent per micro meter. As a result, the bionic MC tactile sensor design showed two times enhancement in the force sensitivity with flat surface structure by attaching a rigid solder ball on a ring sensing structure.

The sensor developed in this chapter tries to capture the characteristics of biological tactile receptors that considered necessary for force sensation. For a precisely emulation of the human skin structure, the sensor can be further designed with a coverage of an elastic material layer (e.g. PDMS), to study the compliant behaviour of the skin by mimicking the viscoelasticity of skin dermis and epidermis layer. The characteristic study of the MC-inspired tactile sensor with elastic coverage will remain as a future work.

Chapter 5 Resonant tactile sensor

In this chapter, a new type of miniaturized resonant tactile sensor based on MEMS silicon bulk-mode resonator is reported for material elasticity measurement. The sensor is developed based on a MEMS fabricated bulk acoustic wave (BAW) resonator, incorporated with a rounded sensing tip and mechanical stoppers to limit the maximum loading force and contact area. Upon contact with the material / object, the resonant frequency shifts corresponding to the amount of mass and stiffness loading from the contact object. The BAW resonator is designed as a square plate that can be excited in the square-extensional (SE) mode and is fabricated using a silicon-on-insulator (SOI) multi-user MEMS process through MEMSCAP. Unlike the conventional piezoelectric tactile sensors discussed in Chapter 2, which consists of two piezoelectric elements as transducer and vibration pick up, the actuation and sensation of the BAW resonator are integrated together, which allows further miniaturization and possible MIS tool integration. The resonator is driven using capacitive to piezoresistive transduction to enhance the motional signal. In the previous work, silicon BAW resonators of different geometric structures have been intensively studied, such as disk [22-24], ring [25, 26] and square [27, 28] shape resonators. They show to have high resonance frequencies, have quality factors larger than 10000 in air [122] and have the possibility to be scaled up into arrays. These features make the BAW resonant tactile sensor a promising solution for precise and high-resolution 2-D material elasticity mapping. The performance of the tactile sensor has been tested for different types of objects placed in contact with the sensor under a laboratory setup. The testing results show that the current prototype of the resonant sensor is able to recognize materials of different young's modulus in the GPa range.

In this chapter, the sensor design concept and the working principle are firstly described. The microfabrication and packaging of the sensor are presented after a

discussion of the BAW resonator and the resonant tactile system modelling. Following the characterization of the resonator is reported along with the parameter extraction. Finally the material elasticity measurement capability of the resonant tactile sensor is demonstrated using a 2.2 MHz BAW resonator.

5.1 Tactile sensor design and working principle

The conceptual design of the MEMS resonant tactile sensor is presented in Figure 5-1. The tactile sensor consists of a square silicon resonator, a rounded sensing tip and two mechanical stoppers. The square BAW resonator is anchored at the center of each edge, which allows the resonator to be operated in the square extensional (SE) mode at a given driving frequency. Electrodes are placed at each corner of the resonator for electrostatic transduction of the resonator. The rounded sensing tip is placed at a maximum of displacement on the resonator in order to obtain the vertical motion of the tip when the resonator is in contact with an object. Mechanical stoppers are placed near the sensing tip to limit the maximum force and contact area that is loaded onto the resonator. The motion of the resonator is driven through the three lateral capacitive-gap electrodes at the corner of the structure as shown in Figure 5-1. And the motion of the resonator is detected through a motional current that results from the capacitance changes across the transduction gaps or the piezoresistance changes of the resonator structure.

The working principle of the bulk acoustic wave resonant tactile sensor is illustrated in Figure 5-2. Upon contact with a test material/object, the resonant frequency shifts corresponding to the amount of mass and stiffness loading from the contacting object. When the tip of the sensor comes into contact with an object, the resonant frequency of the resonator will be changed, $\Delta f = f - f_0$, where f_0 is the resonance frequency of the resonator at natural state, and f is the resonance frequency of the resonator when it is in contact with an object. The sign and magnitude of Δf depends on the stiffness of the material that the sensor comes into contact with. A soft material that can be deformed

by the sensor will result in a negative shift, while a hard object will give a frequency shift with a positive sign, as reported by Omata's et al. in [105].

Compared with previous dynamic sensing techniques which require the measurement of contact force for the calculation of material elasticity [120, 160, 161], this design utilizes the mechanical stopper to fix the material deformation displacement and contact force for determining the elastic properties of the contacting object. This design limits the maximum contact area to $500 \mu\text{m}^2$ and contact force to 150 mN, which protects the sensor from overloading. The sensors can be packaged into a 1-D array and further stacked up to form a 2-D array to fit inside a probe. Additionally, the bulk mode resonator frequency scales linearly with dimension, thus various sizes and frequencies resonators can be developed. It is also possible to design specific resonators to have optimum response to a specific range of young's modulus.

Figure 5-1: Illustration of the design of the resonant tactile sensor. The sensor consists of a square shape BAW resonator, a rounded sensing tip and mechanical stoppers. Mechanical stoppers are placed near the sensing tip to limit the maximum force and contact area that is loaded onto the resonator.

Figure 5-2: Working principle of the bulk acoustic wave resonant tactile sensor, upon contact with the test object, its resonant frequency shifts corresponding to the amount of mass and stiffness loading.

5.2 BAW resonator modelling

Figure 5-3 shows the square extensional (SE) mode shape of the resonator obtained with FEM simulation in COMSOL[®] V4.2. The resonator was simulated using a 3D model and the anisotropic elasticity of single crystal silicon was considered to determine the resonant frequencies accurately. Table 5-1 summarizes the elastic values of silicon that used in the resonator simulation [131, 162], which are given with respect to the xyz-axis of a <100> crystal. The orientation of the 3-D model that aligns the same as the fabricated square plate resonator is illustrated in Figure 5-4. The resonance frequency of a 2 mm x 2 mm SE mode resonator is found at 2.2 MHz from the FE modelling.

Table 5-1: Compliance and stiffness constants of single-crystal silicon

$C_{ij} : 10^9 Pa$	C_{11}	C_{12}	C_{44}	S_{11}	S_{12}	S_{44}
$S_{ij} : 10^{-12} Pa$						
Silicon	165.7	63.9	79.6	7.68	-2.14	12.6

Figure 5-3: Square extensional (SE) mode shape of the plate resonator obtained with 3D anisotropic FEM simulation in COMSOL. The resonance frequency is found at 2.2 MHz.

Figure 5-4: Illustration of the orientation of the simulated and fabricated square plate resonator on the (100) silicon wafer plane.

5.2.1 Motion of resonator

In general, the resonator can exhibit a three dimensional motion each of which has its own mode. For the SE mode of a plate resonator, the mode shape can be described as a plate that is extending and contracting symmetrically in four directions within a plane, and it can be approximated using two orthogonal sinusoid wave functions [163] as following:

$$u_x = A \sin\left(\frac{\pi x}{L}\right) \quad u_y = A \sin\left(\frac{\pi y}{L}\right) \quad (5.1)$$

where A is the vibration amplitude, L is the side length of the square plate, and x , y denotes the direction of the plate extension.

The dynamic response of the resonator is generally described by a mass-spring-damper model. The motion of the system is expressed by a second order ordinary differential equation:

$$m \frac{d^2 x(t)}{dt^2} + \gamma \frac{dx(t)}{dt} + kx(t) = f(t) \quad (5.2)$$

where m is the effective mass of the system, k is the effective stiffness, γ is the damping coefficient that related to the resonator quality factor $\gamma = \sqrt{km} / Q$, $x(t)$ is the resonator displacement function, and $f(t)$ is the time dependent drive force. The effective mass and effective stiffness which related to the device geometry can be given by [164]

$$\begin{aligned} m &= \rho h L^2 \\ k &= \pi^2 E_{se} h \end{aligned} \quad (5.3)$$

where ρ is the silicon mass density, h is the device thickness, and E_{se} denotes effective Young's modulus for the SE mode expansion. For the silicon structural layer of SOIMUMPs, the value of E_{se} on the (100) plane is 180GPa, $\rho = 2330 \text{ kg/m}^3$. The resonance frequency of the SE mode is then given by

$$f_0 = \frac{1}{2L} \sqrt{\frac{E_{SE}}{\rho}} \quad (5.4)$$

For a square plate with the side length of 2 mm, the predicted resonant frequency from (5.4) is approximately 2.2 MHz. This value agrees well with the solution obtained from finite element simulation, and also matches with experimental results that will show later in the resonator characterization section.

5.2.2 Electrostatic transduction

Electrostatic actuation of the resonator is achieved by applying a voltage across a capacitive gap which results in an electrostatic force to drive the resonator into motion. The capacitive actuation force can be given by:

$$F = \frac{1}{2} \left(\frac{\partial C}{\partial x} \right) V^2 \quad (5.5)$$

where V is the sum of the applied voltage and x is the resonator displacement. The capacitance of one parallel plate can be written as:

$$C = \left(\frac{\epsilon_0 A}{g} \right) \quad (5.6)$$

where ϵ_0, g, A are the permittivity, transduction gap and transduction area, respectively.

For fundamental frequency driving, the applied voltage can be written as the sum of the applied dc voltage V_{DC} and the ac voltage $V_{AC} = |V_{AC}| \cos(\omega_0 t)$ at resonant frequency ω_0 , yielding the expression for actuator force as following:

$$\begin{aligned} F &= \frac{1}{2} \left(\frac{\partial C}{\partial x} \right) [V_{DC} + |V_{AC}| \cos(\omega_0 t)]^2 \\ &= \frac{1}{2} \left(\frac{\partial C}{\partial x} \right) \times \left[V_{DC}^2 + 2V_{DC} |V_{AC}| \cos(\omega_0 t) + \frac{|V_{AC}|^2}{2} + \frac{|V_{AC}|^2}{2} \cos(2\omega_0 t) \right] \end{aligned} \quad (5.7)$$

The force component occurs at the resonant frequency ω_0 is:

$$\begin{aligned}
F &= \frac{1}{2} \left(\frac{\partial C}{\partial x} \right) \times (2V_{DC} |V_{AC}| \cos(\omega_0 t)) \\
&= V_{DC} |V_{AC}| \cos(\omega_0 t) \left(\frac{\partial C}{\partial x} \right) \\
&= F_0 \cos(\omega_0 t)
\end{aligned} \tag{5.8}$$

where

$$F_0 = V_{DC} |V_{AC}| \left(\frac{\partial C}{\partial x} \right) = V_{DC} |V_{AC}| \frac{\epsilon_0 A}{g^2} \tag{5.9}$$

In the capacitive driving and capacitive sensing method, the measured motional current is the result of the modulation of the capacitance due to changes in the gap size from the vibration motion, given by

$$i_m = V_{dc} \left(\frac{\partial C}{\partial x} \right) \dot{x} = \eta \dot{x} \tag{5.10}$$

where $\eta = V_{DC} \left(\frac{\partial C}{\partial x} \right)$ is defined as the electromechanical coupling coefficient, which results in the relation between electrical current and mechanical transduction velocity as shown in equation (5.10).

In the capacitive driving and piezoresistive detection method, the motional current that measured can be calculated as [165]:

$$I_m \approx \frac{V_d}{R} \left(\frac{\Delta R_r}{R_r} \right) \tag{5.11}$$

where V_d is a applied voltage differences, and $R = R_r + R_s$ where R_r , R_s are the resistance of the resonator and the resistance of an external resistor respectively.

5.3 Resonant tactile system modelling

Figure 5-5: Mechanical model for the resonant tactile sensor system when interacting with an external material.

The resonant frequency of the resonant sensor at unloaded status is given by:

$$f_0 = \frac{1}{2\pi} \sqrt{\frac{K_R}{M_R}} \quad (5.12)$$

where K_R, M_R is the effective spring constant and mass of the unloaded resonator. When the resonator is in contact with an object, both the effective spring and mass are changed. The contact of the target object can be modelled as an additional spring, mass and damper added to the resonating system as shown in Figure 5-5. The motional equation to describe the interaction between the resonator and the external material can be written as:

$$(M_R + M_L)\ddot{x} + (C_R + C_L)\dot{x} + (K_R + K_L)x = F \quad (5.13)$$

where the parameters M_R, C_R and K_R are the characteristics of the resonator, and M_L, C_L and K_L represents the characteristics of the material in contact with the resonant sensor. Taking the Laplace transform at both sides of the equation (5.13):

$$s^2(M_R + M_L)x(s) + s(C_R + C_L)x(s) + (K_R + K_L)x(s) = F(s) \quad (5.14)$$

with initial conditions that $x(s)=0$, and $\dot{x}(s)=0$, i. e. initial position and velocity of the system are zero. Solving for the transfer function:

$$\frac{x(s)}{F(s)} = \frac{1/(M_R + M_L)}{s^2 + s \frac{C_R + C_L}{M_R + M_L} + \frac{K_R + K_L}{M_R + M_L}} \quad (5.15)$$

The above equation can be simplified by applying the following definitions:

- 1) $\omega_r^2 = \frac{K_R + K_L}{M_R + M_L}$, where ω_r is the resonance frequency of the vibrating system and can be written as:

$$f_r = \frac{1}{2\pi} \sqrt{\frac{K_R + K_L}{M_R + M_L}} \quad (5.16)$$

- 2) $c_{cr} = 2\sqrt{(K_R + K_L)(M_R + M_L)}$, where c_{cr} is the critical damping

ζ is the amount of proportional damping, typically stated as a percentage of critical damping

- 3) $2\zeta\omega_r$ is the multiplier of the velocity term $\dot{x}$, developed below

$$2\zeta\omega_r = 2 \frac{C_R + C_L}{c_{cr}} \sqrt{\frac{K_R + K_L}{M_R + M_L}} = \frac{C_R + C_L}{M_R + M_L} \quad (5.17)$$

Using the above substitutions, the transfer function (5.15) can be written as:

$$\frac{x(s)}{F(s)} = \frac{1/(M_R + M_L)}{s^2 + 2\zeta\omega_r s + \omega_r^2} \quad (5.18)$$

If the stiffness of the contacting object is much higher than that of the resonator (hard object), the object does not move with the resonator and the effective load will be a pre-stress condition to the sensor, resulting in a positive shift of the resonant frequency, given by:

Figure 5-6: Simulated electrical admittance of the designed BAW sensor for different sets of load parameters. (a) $K_L > 0$, $M_L = 0 \text{ Kg}$, $C_L = 0 \text{ Ns/m}$, (b) $C_L > 0$, $M_L = 0 \text{ Kg}$, $K_L = 0 \text{ N/m}$, (c) $M_L > 0$, $C_L = 0 \text{ Ns/m}$, $K_L = 0 \text{ N/m}$

$$\Delta f = f_0 \frac{\Delta k}{2k} \quad (5.19)$$

On the other hand, if the stiffness of the contacting object is much lower than the spring of the resonator (soft object), the material will likely move with the resonator and hence the effect of added mass will be dominating, decreasing the resonant frequency, given by:

$$\Delta f = -f_0 \frac{\Delta m}{2m} \quad (5.20)$$

In the case of contacting with a material with intermediate hardness, the effect of damping will be dominating which decreases the quality factor of the system. From both the frequency shift and change in quality factor, the information on the stiffness of the material under test can be obtained. Simulation results of the models introduced by using different loads are illustrated in Figure 5-6 to demonstrate the capabilities of the system to recognize different materials. The Matlab codes used to generate Figure 5-6 plots can be found in Appendix B.1.

5.4 Microfabrication

The resonators were fabricated using commercially available Silicon on Insulator Multi-User MEMS process (SOIMUMPs) provided by MEMSCAP Inc. The cross sectional view of the fabrication process is illustrated in Figure 5-7. The fabrication is a simple four-mask level patterning and etching process based on SOI wafer which consisted of three layers, i.e. 10 μm or 25 μm thick top device layer, a 1 μm thick buried oxide (BOX) layer, and a 400 μm thick silicon substrate. The fabrication process begins with 150 mm n-type double-side polished SOI wafers, the top surface of which is doped by depositing a phosphosilicate glass (PSG) layer and annealing at 1050 $^{\circ}\text{C}$ for 1 hour in Argon. This PSG layer is then removed via wet chemical etching. Following is the metallization process to define the bond pads. A metal stack of 20nm of chrome and 500nm of gold is patterned through a liftoff process. Then the resonator structure including the square

plate, electrodes and anchors are lithographically patterned and etched by Deep Reactive Ion Etching (DRIE) from the front side.

Figure 5-7: Cross sectional view showing different layers of the SOI-MUMPs fabrication process by MEMSCAP Inc. Courtesy of MEMSCAP

The structures on the front side are subsequently protected by a polyimide layer coating, and then the wafer is reversed for backside release. The substrate layer is lithographically patterned from the bottom side and etched by DRIE until the bottom BOX layer. The BOX layer is then removed using a wet oxide etching process. The suspended device structure is fully released after the front side polyimide protective layer is stripped in a dry etch process. The remaining BOX layer that is exposed from the front side is removed using a vapor HF process for patterning a metal pad on the substrate in the following step. The final process is to pattern a metal pad on the substrate from the topside using a shadow mask to provide an electrical contact to the substrate. The removal of the underlying substrate can largely reduce parasitic capacitance between the device layer and the substrate, and it is beneficial for sensing small motional currents.. The wafer is then laser diced into smaller size dies for later testing. An optical micrograph of the fabricated square plate resonator is shown in Figure 5-8.

Figure 5-8: Optical micrograph of the fabricated 2 mm BAW tactile sensor showing the rounded sensing tip and the mechanical stopper.

Figure 5-9: Circuit board with the packaged chip and the connection diagram of the board with testing instruments.

Table 5-2: Resonator dimensions and characteristic parameters measured at $V_{DC} = 80$ V.

Parameter	Symbol	Value
Resonator side length	L	2000 μm
Resonator thickness	h	25 μm
Transduction gap	g	2 μm
Natural frequency	f_0	2.2 MHz
Effective stiffness	k	4.466×10^7 N/m
Effective mass	m	2.33×10^{-7} Kg
Q-factor	Q	10430

5.5 Resonator characterization

The fabricated devices have been packaged through wire bonding in a 44-pin ceramic leadless chip carrier (LCC44) supplied by Spectrum Semiconductor Materials. The packaged devices are then fitted into a testing socket from Plantronics. The testing socket is mounted on a custom designed printed circuit board (PCB). Then the electrical transmission characteristics of the devices are tested based on the PCB connection diagram as illustrated in Figure 5-9. The resonator parameters are summarized in Table 5-2. The length of the plate structure is 2 mm. The gap between electrode and square plate is 2 μm , which is the minimum allowable gap size in this SOIMUMPs process. The effective stiffness and effective mass were calculated based on equation (5.3).

5.5.1 Capacitive driving and capacitive sensing

Figure 5-10 shows a circuit schematic to illustrate the measurement setup used for obtaining the transmission characteristics of the resonator. In this one-port capacitive transduced measurement setup, three electrodes are connected together with the same DC and AC drive voltages applied to generate a time varying electrostatic force that drives the resonator into motion. The resonator body and substrate are grounded to minimize the parasitic capacitance. Bias voltages ranging from 40 V to 100 V were applied to the resonator in incremental steps of 20 V. Figure 5-12 shows the frequency

response curves in terms of magnitude and phase obtained when measuring in air, each corresponding to the bias voltage applied.

Figure 5-10: Schematic diagram showing the resonator measurement setup for one-port capacitive driving and capacitive sensing of the sensor.

Figure 5-11: Equivalent circuit model of a resonator including the feedthrough capacitor C_f located parallel to the resonator.

Figure 5-11 shows a schematic of the equivalent circuit model of a typical resonator. The associated electromechanical parameters can be calculated as following:

$$\begin{aligned}
R_m &= \sqrt{\frac{km}{(\eta V_{DC})^2 Q}} \\
C_m &= \frac{(\eta V_{DC})^2}{k} \\
I_m &= \frac{m}{(\eta V_{DC})^2}
\end{aligned} \tag{5.21}$$

The measured admittance of the system can be expressed as:

$$Y_m = \frac{I_m}{V} = \frac{j\omega C_m}{[1 - (\omega/\omega_0)^2] + j(\omega/\omega_0)/Q} + j\omega C_f \tag{5.22}$$

In the model, a feedthrough capacitor C_f is located in parallel with the resonator, which has been modelled as an LRC series resonant circuit. The physical sources of this feedthrough capacitor have been identified in the previous studies [166, 167], including the direct overlap capacitance of the transducer, capacitive coupling through the substrate, interconnects, and the electrical package [167]. As previously studied, in macro-scale applications, the feedthrough capacitance is typically negligible, but at the microscale level, this capacitance becomes significant. There are two important consequences of the feedthrough capacitance. Firstly it determines the noise floor of the resonator sweep measurement and also affects the minimum signal level. Secondly, it differs the electrical measurement of the resonator significantly from the ideal resonator response. As observed from the frequency response shown in Figure 5-12, the peak magnitude of the measured electrical transmission of the resonator is approximately 0.15 dB which is far less than -3 dB, which makes it impossible to deduce important resonator parameters such as quality factor or motional resistance directly from the capacitive transduction measurement results.

As reported previously by Joshua et al. [167], in order to extract the resonator parameters, the feedthrough signal has to be de-embedded from the measured response first. This can be done by extracting the resonator admittance response from the total measurement by measuring the value of the signal contribution from C_f (without applied bias) and then subtracting it from the measured total response (with DC bias

applied). This procedure was applied to the raw experimental data as plotted in Figure 5-12 to plot the restored frequency responses corresponding to the different applied DC voltages. From the measurement, the resonant frequency of the resonator is at 2.19 MHz, and the obtained Q-factor is 10430 in air. The parameters are summarized in Table 5 2. The MATLAB code for performing feedthrough de-embedding is included in Appendix B.2.

As shown in Figure 5-13, even though the magnitude of the electrical transmission is largely improved which makes it possible to extract the resonator parameters from the measurement; however, the magnitude and phase responses are noisy across a large bandwidth range. For the application of using the resonator as tactile sensor for material property measurement, as discussed in Chapter 5.1, the frequency shifts when the resonator is in contact with objects need to be abstained, which requires to further reduce the capacitive feedthrough noise and increase the motional current signal. Previously piezoresistive sensing method has been studied to enhance the motional to feedthrough current ratio and showed enhanced transduction over capacitive sensing with efficiencies that scale better scaling with frequency [165]. In the following section, capacitive to piezoresistive transduction method is applied to obtain the frequency response of the square resonator.

(a) Magnitude

(b) Phase

Figure 5-12: Measured frequency response curves using capacitive drive and capacitive sensing in terms of (a) magnitude and (b) phase for a range of applied DC bias.

(a) Magnitude

(b) Phase

Figure 5-13: Restored frequency response curves using capacitive drive and capacitive sensing in terms of (a) magnitude and (b) phase for a range of applied DC bias.

5.5.2 Capacitive driving and piezoresistive sensing

In the piezoresistive transduction, the resonator motion is detected through the variation in the electrical resistance of the vibrating structure due to the piezoresistive effect of single crystal silicon. Figure 5-10 shows the schematic diagram of the resonator measurement setup for capacitive driving and piezoresistive sensing method. The electrical resistance between two anchors of the resonator structure is monitored by applying a voltage differences V_d .

The measured admittance of the system can be expressed as:

$$Y_m = \frac{I_m}{V} = \frac{j\omega C_m}{[1 - (\omega/\omega_0)^2] + j(\omega/\omega_0)/Q} + j\omega C_f + g_m \quad (5.23)$$

where g_m is the transconductance given by:

$$g_m = \frac{I_m}{V_{AC}} = \frac{V_d}{V_{AC} R} \left(\frac{\Delta R_r}{R_r} \right) \quad (5.24)$$

The change in the resonator resistance follows equation (3.2), which is

$$\frac{\Delta R_r}{R_r} = \pi_l \sigma_l + \pi_t \sigma_t \quad (5.25)$$

Figure 5-14 shows the frequency response curves for the one-port capacitive driving and piezoresistive sensing measurements, each curve corresponding to the bias voltage applied. A peak magnitude of 0.85 dB is obtained with a 3 V differential voltage applied (V_d) and the bias voltage at 80 V. A clear motional signal improvement can be seen comparing to the capacitive driving and capacitive sensing measurement results as shown in Figure 5-12. In the following section, the resonator is characterized to be used as resonance tactile sensor, whereas the transmission curves are obtained based on the capacitive driving and piezoresistive sensing method.

(a) Magnitude

(b) Phase

Figure 5-14: Measured frequency response curves using capacitive drive and piezoresistive sensing in terms of (a) magnitude and (b) phase for a range of applied DC bias. The results was obtained with the setting of $V_{AC} = 0dBm$, and $V_d = 3V$.

Figure 5-15: Schematic diagram showing the resonator measurement setup for one-port capacitive driving and piezoresistive sensing of the sensor.

5.6 Resonant tactile sensor characterization

5.6.1 Testing setup

To evaluate the performance of the resonant tactile sensor, the fabricated resonator was characterized to obtain its resonance frequency shift when contacting with materials of different Young's modulus. The SE mode resonator was driven using capacitive to piezoresistive transduction method as discussed in previous section. The setup is illustrated in Figure 5-16. The characterization system consists of a microscope for visual control, a piezoactuator (PI-245) connected to a piezo motion controller (E517) for position control of the testing materials, a Agilent 5071B network analyzer to obtain the frequency response of the sensor, and a computer to control the piezo-controller. As demonstrated in Figure 5-17, the resonant sensor that packaged on a PCB board was fixed on the experimental table. The test material was placed at the tip of a tungsten probe needle, whose displacement was controlled using the piezo-motion controller

(E517). The resolution of the piezo-controller is 1.6 nm with a displacement range of 0-80 μm which allows precise control of the sensor-material contact. As shown in Figure 5-18, the testing material approaches the sensing tip with different contact stages under the control of the motion controller. The frequency response of the resonator was recorded by the network analyzer when the resonator is in contact with the testing material. The DC voltage used to drive the resonator is $V_{\text{DC}} = 80 \text{ V}$; and a $V_{\text{d}} = 2 \text{ V}$ is applied for piezoresistive measurement. The applied AC voltage is $V_{\text{AC}} = 0 \text{ dBm}$.

The center frequency was obtained at 2.19 MHz. Then the frequency response of the tactile sensor was tested for five different type of materials placed in contact with the sensor. The sensor was tested under different extents of stress applied through the object by pushing it toward the contact tip with a step movement from 1 μm to 5 μm . Top view of the sensor-object contact is illustrated in Figure 5-18.

Figure 5-16: Laboratory setup showing the equipment setup for the testing.

Figure 5-17: Experimental setup developed for sensor characterization. The sensor is connected on PCB through chip carrier and the test object is carried on a probe needle to be in touch with the resonator sensing tip.

Figure 5-18: Optical image showing object contacting with the sensing tip.

5.6.2 Results and discussion

Figure 5-19 shows the measured frequency response of the sensor when a polystyrene bead was in-contact with the sensing tip. From the graph it can be seen that when the tactile sensor is brought into contact with the object, the resonance frequency of the resonator increases as a result of an additional mass effect. The value of the frequency shift also increases along with the contact depth increasing due to the increased stress applied on the sensing tip, as illustrated in Figure 5-21. The frequency shift was expressed as $\Delta f (= f - f_0)$, where f is the resonance frequency when a load was applied and f_0 is the resonance frequency under no loading condition. Figure 5-18 shows the optical image of the resonator when in contact with the polystyrene bead. It can be seen that the mechanical stopper limits the maximum contacting force and displacement of the polystyrene bead. The measurement result for polypropylene test object is shown in Figure 5-20. From the graph it can be seen that with higher contact force, the frequency increases and meanwhile the damping increases. The polypropylene test shows the resonant characteristics when the contacting object has comparable spring constant to that of the resonator, resulting in significant acoustic energy loss propagating out of the sensor, thereby decreasing the quality factor of the sensor. The different frequency responses of the two test objects clearly demonstrate the functionality of the sensor for material elasticity measurement. The measured frequency responses of the other three materials can be found in the Appendix C Figure A.2.

The frequency shift ratio $|\frac{\Delta f}{f_0}|$, which corresponds to the sensitivity of the tactile sensor, is plotted in Figure 5-22 with respect to different contact depths. It shows the frequency shift ratio of the BAW resonant tactile sensor when in contact with all the five materials of different Young's modulus: (a) polypropylene ($E = 1.5 \sim 2$ GPa) bead, (b) polystyrene ($E = 3 \sim 3.5$ GPa) bead, (c) poly (ethyl methacrylate) (PEMA) bead ($E = 1.1 \sim 1.3$ GPa), (d) Sn63/Pb37 bead ($E = 33.6$ GPa), and (e) Sn96.5/Ag3.0/Cu0.5 bead ($E = 51.2$ GPa). The plot of the frequency change ratio in respect to the Young's modulus of the testing materials is shown in Figure 5-23. Figure 5-24 shows the optical images of the five different types of materials that in contact with the tactile sensor.

From the testing results we can see that the sensor is able to differentiate materials of different young's modulus in the GPa range. The frequency shift ratio is in correspondence with the increment of the material stiffness except for the material beads of Sn/Pn and PEMA. Even though the stiffness of the Sn/Pn is lower than that of the Sn/Ag/Cu and the stiffness of the PEMS is lower than that of the Polypropylene and Polystyrene beads, the frequency shift of Sn/Pn is larger than the frequency shift of Sn/Ag/cu and the same trend appears for PEMA, which shows an opposite frequency shift ratio with the increment of the material stiffness. The reason that caused the discrepancy of the frequency response differences between Sn/Ag/Cu and Sn/Pn is because of the size inconsistent of the material beads used in the testing. As illustrated of Figure 5-24, the sizes of the testing beads Sn/Pn and PEMA are comparatively smaller than the other three testing beads. It is believed that the larger frequency shift ratio of Sn63 and PEMA materials are due to the combination of stiffness between the testing material and the tungsten needle, which means that the material stiffness measured by the sensor may not be the purely material stiffness but the stiffness contributed from both the material and the tungsten needle. As illustrated in Figure 5-18 the testing material was manually fixed on the tip of a tungsten probe needle. Comparing to the young's modulus of the testing material in few GPa ranges, the tungsten has a much higher Young's modulus of 410 GPa, such that the tungsten needle is much stiffer than the testing materials. So when a smaller material ball contacts with the resonator, the distance between the needle and the sensor contact point is smaller than large size object, such that the needle impose larger effect on the stiffness measurement which results in a larger frequency shift. In future work, new testing schematics need to be properly designed to minimize the extrinsic factors that might affect the material properties.

Figure 5-19: Magnitude and phase response of the sensor when in contact with polystyrene of Young's modulus 3 ~ 3.5 GPa. Contact steps applied from 1um to 5um.

Figure 5-20: Magnitude and phase response of the sensor when in contact with polypropylene of Young's modulus 1.5 ~ 2 GPa. Contact steps applied from 1um to 5um.

Figure 5-21: Frequency shift increases with increasing contact depth thus an increased stress applied on the sensing tip. Linear response was obtained ($R^2 > 99.6\%$).

Figure 5-22: Summary plot showing the frequency shift ratio of the BAW resonant tactile sensor when in contact with materials of different Young's modulus.

Figure 5-23: Plot of the frequency change ratio in respect to the Young's modulus of the testing materials.

Figure 5-24: Illustration of the five different types of materials with different Young's modulus used in the BAW resonant tactile sensor characterization.

5.7 Summary

In this chapter, the design, modeling, fabrication and characterization of a BAW resonant tactile sensor based on MEMS bulk-mode resonator is presented for material elasticity measurement. The resonator was fabricated using a silicon-on-insulator (SOI) multi-user MEMS process through MEMSCAP. The important resonator parameters such as the resonance frequency and Q factor of the resonator has extracted through a de-embedding approach of the feedthrough capacitance from the measured total transmission response. The Q factor 10430 was obtained in air.

The performance of the resonator used as tactile sensor has been tested with five different types of materials placed in contact with the sensor under a laboratory setup. The output response study carried out on the sensors indicated that the BAW resonant tactile sensor can provide information not only about the extent of force/stress applied on the sensing tip, but also the stiffness of the material which is in contact with the sensor. The testing results show that the sensor is able to differentiate materials of young's modulus in the GPa range.

The proposed BAW resonant tactile sensor demonstrated a first step toward material elasticity measurement that in a further time could be miniaturized and opens up applications such as robotics, minimally invasive surgery monitoring and 2D mapping of tissue elasticity for cancer tumor detection.

Chapter 6 Conclusions and future work

6.1 Summary and conclusions

The aim of the work carried out in this thesis was to develop miniaturized force and tactile sensors that are able to measure the contact force and relative hardness of contact objects, targeted on MIS applications.

To achieve this goal, literature reviews of previous research work on the development of MEMS sensors for guidewires and the existing technologies for elasticity measurement are initially studied. The device miniaturization was identified as one of the most important concerns for integrating a force sensor on the tip of a guidewire as the tip diameter of a typical guidewire is only 0.36 mm. Besides, it is also important to satisfy the sensitivity, resolution, and force range requirements.

In order to develop force sensors that are adaptable to standard manufacturing process flow, a ring-shaped force sensor was designed to be integrated on the distal tip of a standard movable core guidewire. The sensor has a sensory area of $200\mu\text{m} \times 200\mu\text{m}$, and silicon nanowires (SiNWs) are utilized as piezoresistive sensing element by taking advantage of its ultra-miniaturization and high sensitivity. With this ring-shaped structure design, the tri-axial force sensor can be easily mounted on the circular core wire at the tip of a guidewire. Together with the movable core wire, the sensor can detect contact forces while maintaining the original welded tip which is an essential part of the standard guidewire assembly. Meanwhile, this assembly configuration of the ring-shaped sensor can minimize the bending artifacts as it is directly integrated on the distal tip of the guidewire, whereas most of the sensors reported in the earlier work for catheterization applications were mounted in a recess near the tip of the core wire or on the wall of catheter.

Then a finite element (FE) model that built in COMSOL Multiphysics environment was used to find the maximum stress location in order to put the SiNWs at the high stress regions and the mechanical behaviour of the sensor-guidewire interaction was studied by using this model. When a normal force was applied on the top surface of the ring, the sensor beams bent downward and the maximum stress was found at the bases of the beams. SiNWs were embedded between the layers of silicon and SiO₂ at these high stress regions in order to obtain the maximum piezoresistance output.

The sensor structure and SiNWs with a length of 6 μm and a cross section of 90 nm x 90 nm have been successfully fabricated by a complementary metal-oxide semiconductor (CMOS) - compatible process in IME (Institute of Microelectronics, A*STAR). In order to calibrate and evaluate the functionality of the fabricated SiNW force sensor, characterization experiments have been conducted and the simulation results of characterization were used to verify experimental measurements. Repeatability and hysteresis tests have been carried out on the ring tactile sensor with different displacement step sizes of 50 nm, 0.5 μm and 1 μm . The sensor has been showed capable of fulfilling the performance requirement of high repeatability, high linearity ($\approx 99.99\%$) and negligible hysteresis with a resulting sensitivity of $13.4 \times 10^{-3} \mu\text{m}^{-1}$ in z- direction. In addition, the sensor was able to detect nanoscale displacement in a 50 nm step size with high repeatability.

Tactile sensors of flat membrane alone as sensing structure have their limitations on detecting forces in shear directions. To enhance the force sensitivity of the ring-shaped sensor that of flat surface, a Meissner corpuscles inspired tactile sensor was then developed. The Meissner corpuscle inspired sensor was designed with a rigid spherical ball attached on top of the suspended ring structure located at the centre of four suspended beams which mimics the biological structure of the Meissner corpuscles. The sensor was developed with a simplified fabrication and packaging process by dropping spherical balls on the SiNW-based ring structure. Finite element analysis (FEA) was used to predict the mechanical behaviours of the tactile sensor with rigid solder ball and with elastic bump covering of rigid solder ball underneath. The performance of the SiNWs as piezoresistive sensing elements on the bio-inspired sensor and on the bare

ring-shaped sensing element was comparatively studied by repeatability and hysteresis tests in Chapter 4. The effectiveness of sensitivity enhancement of the bio-inspired tactile sensor with solder ball attached was verified from the experimental results, in that the fractional change in resistance in normal direction was improved from 1.34 percent to 2.8 percent per micro meter.

This biomimetic sensor tried to capture the characteristics of biological tactile receptors that considered necessary for force sensation. For a precisely emulation of the human skin structure, the sensor need to be further designed with a coverage of an elastic material layer (e.g. PDMS), to study the compliant behaviour of the skin by mimicking the viscoelasticity of skin dermis and epidermis layer. The characteristic of the MC-inspired tactile sensor with elastic coverage can be further studied in the future.

To measure the material elasticity, a miniaturized resonant tactile sensor based on MEMS silicon bulk-mode resonator was developed in Chapter 5. The sensor was based on a MEMS silicon bulk acoustic wave (BAW) resonator, incorporated with a rounded sensing tip and mechanical stoppers to limit the maximum loading force and contact area. Upon contact with the material / object, the resonant frequency shifts corresponding to the amount of mass and stiffness loading from the contact object. The advantages of BAW resonator used as sensor have been widely studied, such as high resonance frequencies, high quality factors (> 10000) in air, and can be easily scaled up into arrays.

The square shape BAW resonator was excited in the square-extensional (SE) mode and was successfully fabricated through the silicon-on-insulator (SOI) multi-user MEMS process through MEMSCAP. To extract the performance parameters of the resonator, capacitive to piezoresistance transduction was used in the resonator characterization. The resonant frequency of the resonator was obtained at 2.2MHz and had a Q-factor of 10430 in air. Following, the ability of the sensor to measure the relative hardness of the contact object was tested with experiments by contacting the sensor with materials of different degrees of Young's modulus. The output response study carried out on the sensors indicated that the BAW resonant tactile sensor can provide information not only

about the stiffness of the material which is in contact with the sensor, but also the extent of force/stress applied on the sensing tip. The testing results showed that the sensor is capable to differentiate materials of Young's modulus in the GPa range.

This developed BAW resonant tactile sensor demonstrated a first step toward material elasticity measurement that in a further time could be miniaturized and opens up applications such as robotics, minimally invasive surgery monitoring and 2D mapping of tissue elasticity for cancer tumour detection.

The MEMS force and tactile sensors developed in this thesis have end applications in surgical procedures, particularly in the minimally invasive surgery including robotic surgery and teleoperation for tissue palpation, and measuring tool-tissue interaction forces and tissue elasticity at the guidewire tip. It also has the potential for using in gaming and mechatronic applications. The developed technique can also be extended to other types of guidewires and guiding catheters by modifying the geometric parameters of the sensor structure.

6.2 Limitations and Future work

6.2.1 Force sensor further characterization and practical mounting on guidewire

The force sensor developed in this dissertation has been preliminary characterized to demonstrate a proof of concept. Future characterization can be performed with a cylindrical mesa passing through the ring of the sensor acting as a force transmitting element. The mesa that acts as the shear force transferring element can be used to acquire the three components of the loading force, thus further completing the characterization of the sensor.

As a force characterization setup will be different with the displacement loading case discussed in Chapter 3, the force sensor need to be firstly wire bonded on a PCB and

then sold wire out to connect with the probes to get the SiNW piezoresistance value. Practically, depending on the metal pad quality of the sensor and PCB board surface finishing performance, the difficulties of the wire bonding process various. In the current sensor packaging, the sensor was wire bonded to a customized chip carrier. For this PCB of hard gold surface finishing, high temperature and high intensity were required during the wire bonding process which decreased the metal pads quality of the sensor die. In future, a PBC of better surface finishing ($0.3\ \mu\text{m} - 0.5\ \mu\text{m}$ soft gold surface finishing) can be designed and used for the sensor packaging. A Wheatstone bridge circuit can also be designed along with the PCB fabrication to minimize the temperature effect on SiNW piezoresistors.

Feasibility of mounting the tri-axial force sensor on the tip of a standard move core guidewire can be demonstrated in the future. Even if dimensions of the reported prototype are still larger to allow practical integration, scalability of the proposed device is possible through the developed fabrication process and it can be pursued in future work. Safety and sterilization issue must also be addressed in the practical use.

6.2.2 Biomimetic tactile sensor with elastic coverage

The biomimetic sensor reported in Chapter 4 tried to capture the characteristics of biological tactile receptors that considered necessary for force sensation. For a precisely emulation of the human skin structure, the sensor can be further designed with a coverage of an elastic material layer (e.g. PDMS), to study the compliant behaviour of the skin by mimicking the viscoelasticity of skin dermis and epidermis layer. The characteristic of the MC-inspired tactile sensor with elastic coverage can be further studied in the future.

6.2.3 Further development on resonant tactile sensor

The resonant tactile sensor developed in Chapter 5 has demonstrated can differentiate materials of Young's modulus in GPa ranges. To practically be able to measure the elasticity of tissues, the sensor need to be further tested with softer materials (Young's

modulus in MPa ranges) and new testing schematics need to be properly designed to minimize the extrinsic factors that might affect the material properties.

The sensors can also be designed to be packaged into a 1-D array and further stacked up to form a 2-D array to fit inside a probe. Additionally, the bulk mode resonator frequency scales linearly with dimension, thus various sizes and frequencies resonators can be developed in the future. It is also possible to design specific resonators to have optimum response to a specific range of young's modulus.

Appendix A SiNW alignment on SOI wafer

The SiNW-based ring-shaped force sensor was fabricated by using top-down technique directionally along with the MEMS device fabrication. Standard scanner lithography (Nikon KrF DUV lithography – model NSR-203B) was used to fabricate the silicon nanowire piezoresistors along the $\langle 110 \rangle$ direction on the SOI wafer. The SiNW mask was the first mask in the process flow and the mask alignment was done with respect to the SOI wafer notch position.

In the following section, the effect of the possible alignment error on the SiNW piezoresistance measurement in the fabricated force sensor is analysed, as there are concerns arising from the crystal orientation variation between the SOI device layer and the wafer notch position. As reported by [168], alignment errors may arise from aligning the wafer notch to the crystallographic axes and from aligning the lithographic mask to the notch during the fabrication, and the crystallographic axis orientation relative to the notch can vary up to $\pm 1^\circ$, which dominates the overall alignment errors.

For a piezoresistor on (100) wafer plane and oriented at an angle of ϕ with respect to the [110] crystallographic axes orientation, when subjected to an arbitrary state of stress, the fractional changes in resistance of the piezoresistor can be expressed as [169, 170]

$$\begin{aligned} \frac{\Delta R}{R} = & \left[\left(\frac{\pi_{11} + \pi_{12} + \pi_{44}}{2} \right) \sigma_{11} + \left(\frac{\pi_{11} + \pi_{12} - \pi_{44}}{2} \right) \sigma_{22} \right] \cos^2 \phi \\ & + \left[\left(\frac{\pi_{11} + \pi_{12} - \pi_{44}}{2} \right) \sigma_{11} + \left(\frac{\pi_{11} + \pi_{12} + \pi_{44}}{2} \right) \sigma_{22} \right] \sin^2 \phi \\ & + (\pi_{11} - \pi_{12}) \sigma_{12} \sin 2\phi + \alpha_1 \Delta T + \alpha_2 \Delta T^2 \end{aligned} \quad (\text{A-1})$$

where R is the initial resistance and ΔR is the resistance changes, σ_{11} and σ_{22} are the normal stresses along the $[110]$ and $[\bar{1}10]$ directions, respectively, and σ_{12} is the shear stress. π_{11} , π_{12} and π_{44} are the three piezoresistive coefficients. α_1 and α_2 are the first and second order temperature coefficients respectively, and ΔT is the temperature difference.

We assume that all of the SiNWs are simultaneously misaligned by a same angle of $\Delta\phi$ respect to the crystallographic axis on the device layer (as shown in Figure A.1). The dependence of the sensor behavior on SiNW piezoresistor misalignment can be expressed by replacing ϕ with an angle of $(\phi + \Delta\phi)$ in equation (A-1) and by ignoring the temperature influence:

$$\begin{aligned} \frac{\Delta R}{R} = & \left[\left(\frac{\pi_{11} + \pi_{12} + \pi_{44}}{2} \right) \sigma_{11} + \left(\frac{\pi_{11} + \pi_{12} - \pi_{44}}{2} \right) \sigma_{22} \right] \cos^2(\phi + \Delta\phi) \\ & + \left[\left(\frac{\pi_{11} + \pi_{12} - \pi_{44}}{2} \right) \sigma_{11} + \left(\frac{\pi_{11} + \pi_{12} + \pi_{44}}{2} \right) \sigma_{22} \right] \sin^2(\phi + \Delta\phi) \\ & + (\pi_{11} - \pi_{12}) \sigma_{12} \sin 2(\phi + \Delta\phi) \end{aligned} \quad (A-2)$$

In the ring sensor design, the four SiNW piezoresistors are in 0° - 90° configurations with respect to the $[110]$ wafer crystallographic axes orientation. Then for 0° - 90° p-type piezoresistors with a small rotational alignment error $\Delta\phi$, we can obtain from equation (A-2) that [168, 171]

$$\frac{\Delta R_4}{R_4} - \frac{\Delta R_2}{R_2} = \pi_{44} \left[(\sigma_{11} - \sigma_{22}) + 4 \frac{\pi_{11} - \pi_{12}}{\pi_{44}} \sigma_{12} \Delta\phi \right] \quad (A-3)$$

As the value of π_{44} is much larger than that of $(\pi_{11} - \pi_{12})$ ($\pi_{44} = 138.1 \times 10^{-11} \text{ Pa}^{-1}$ and $\pi_{11} - \pi_{12} = 7.7 \times 10^{-11} \text{ Pa}^{-1}$ as in [76]), the misalignment error causes very small measurement errors for the SiNW piezoresistance differences. Therefore, the influence of the misalignment errors on the p-type SiNW piezoresistance measurement can be

neglected reasonably. Since our nanowire sensors do not rely on ‘giant piezoresistance’, the difference due to the orientation error should be similar to Smith and Jaeger.

Figure A.1: SiNW piezoresistor orientation with respect to the $[110]$ wafer crystallographic axes orientation on (100) SOI wafer plane

Appendix B Matlab code

B.1 Matlab codes used to simulate the electrical admittance of the BAW sensor for different sets of load parameters

```
% resonator contact.m plotting frequency responses of a mass-damper-spring model for different
% parameter values
clear all;

% calculate the effective mass, effective stiffness, percentage of critical damping,
rho = 2330; % density of resonator (silicon) in kg/m^3
h = 25e-6; % resonator height 25um
L = 2e-3; % resonator side length 2mm
Y2d = 181e9; % effective elastic modulus for the 2-D expansion -  $Y_{2d} = c_{11} + c_{12} - 2c_{12}^2/c_{11}$ 
M_R = rho*h*L^2; % effective mass of the resonator
K_R = pi^2*Y2d*h; % effective spring constant of the resonator
zeta = 0.01:0.01:0.01; % 0.01 = 1% of critical damping
C_cr = 2*sqrt(K_R*M_R); % critical damping
C_R = zeta*C_cr;
w0 = sqrt(K_R/M_R);
f0 = w0/(2*pi);

% define contact material properties
M_L = 0:0.05*M_R:0.5*M_R;
K_L = 0;
C_L = 0;
%K_L = 0:0.2*K_R:2*K_R;

% defines a vector of frequencies to use, radians/sec. The logspace command
% uses the log10 values as limits, i.e. -1 is  $10^{-1} = 0.1$  rad/sec, and 1 is
%  $10^1 = 10$  rad/sec. the 400 defines 400 frequency points.
```

```

%w = logspace (6, 8, 400);
f = 1.5e6:0.2e3:4e6;
w = 2*pi*f;
% pre-calculate the radians to degree conversion
rad2deg = 180/pi;

% define s as the imaginary operator times the radian frequency vector
s = j*w;

% define a for loop to cycle through all the damping values for calculating magnitude and phase

for cnt = 1:length(M_L)

    % calculate system resonance frequency
    wr = sqrt((K_R+K_L)/(M_R+M_L(cnt)));
    fr = wr/(2*pi);

    % define the frequency response to be evaluated
    xfer(cnt, :) = (1/(M_R+M_L(cnt)))/(s.^2 + s.*((C_R+C_L)/(M_R+M_L(cnt))) +
(K_R+K_L)/(M_R+M_L(cnt)));

    % calculate the magnitude and phase of each frequency response
    mag(cnt, :) = abs(xfer(cnt, :));
    phs(cnt, :) = angle(xfer(cnt, :))*rad2deg;
    mag_db(cnt, :) = mag2db(mag(cnt, :));
end

% define a for loop to cycle through all the damping values for plotting
% magnitude
%cc=hsv(12);
ColorSet = varycolor(cnt);
set(gca, 'ColorOrder', ColorSet);
% hold all;

% Change default axes fonts.
set(0,'DefaultAxesFontName', 'Times New Roman')
set(0,'DefaultAxesFontSize', 14)

% Change default text fonts.

```

```

set(0,'DefaultTextFontname', 'Times New Roman')
set(0,'DefaultTextFontSize', 30)

for cnt = 1:length(M_L)
    subplot (2,1,1)

    % set(gca, 'ColorOrder', ColorSet);

    % plot(f, mag_db(cnt, :),'color',cc(cnt,:))
    plot(f, mag_db(cnt, :),'Color', ColorSet(cnt,:), 'LineWidth', 1.3)
    title('Simulation of the resonance tactile sensor system for different sets of loading parameters: M_L >
0, C_L = 0, K_L = 0')
    xlabel('Frequency (Hz)')
    ylabel('Magnitude (dB)')
    legend('M_L=0', 'M_L=0.1M_R', 'M_L=0.2M_R', ..., 'M_L=M_R')
    set(gca, 'XTick', [1.6e6:0.2e6:4e6])
    set(gca, 'YTick', [-165:10:-115])
    grid on
    hold on

    subplot(2,1,2)
    % plot phase
    % plot(f, phs(cnt, :), [coVal liVal], 'LineWidth', 1.2)
    plot(f, phs(cnt, :), 'Color', ColorSet(cnt,:), 'LineWidth', 1.3)
    %title('2mm resonator frequency response phase for 1% of critical damping')
    xlabel('Frequency (Hz)')
    ylabel('Phase (degree)')

    ylim([-200, 20]) %Éè¶“yÖáµÄ»æÖÆ·¶Î§,
axis([xmin,xmax,ymin,ymax])£¬ÖÄÖâ,öÖř¼ä¿ÉÖÔ¶¶Öx£¬yÖáµÄÉ¶¶ÖéĬÄĬP»æÖÆ·¶Î§Ö»Æð×ö¿ÖÖ
Æ
    legend('M_L=0', 'M_L=0.1M_R', 'M_L=0.2M_R', ..., 'M_L=M_R') %ĬßĬö×çÊĬ¶ĬăĬöµÄ»°:
legend('xxx', 'xxx2', 'xxx3'£©
    set(gca, 'XTick', [1.6e6:0.2e6:4e6]) %,Ä±äxÖá×ø±ê¼ä,ôĬÖÊ¾ ÖâÀĬ¼ä,ôĬ²2
    grid on
    hold on
end
hold off
grid on

```

B.2 Matlab function to de-embed the capacitive feedthrough from the measurement signals

```
function res= feedthru( tot,cap )
% This function calculate the "true" resonator response assuring that the noise floor is due to a large
% capacitor in parallel with the resonator (the "feedthrough capacitance", Cf).
%  $Z = R \cdot \exp(i \cdot \theta)$ 
% rad = deg*2*pi/360
sym i;

% identify frequency axis - assume it is the same for both tot & cap
xfreq=tot.frequency;

% converts the magnitude from decibel (dB) to linear units. The relationship between the decibel unit and
% the linear unit is:  $Y_{db}=20 \cdot \log_{10}(Y)$ .
tmag=10.^(tot.magnitude./20); % or tmag=db2mag(tot.trace1.y);
cmag=10.^(cap.magnitude./20); % or cmag=db2mag(cap.trace1.y);

% convert phase to radians
tphase=tot.phase*2*pi/360; % or tphase=deg2rad(tot.trace.y);
cphase=cap.phase*2*pi/360;

% convert to complex quantities
tcmx=tmag.*exp(i*tphase);
ccmx=cmag.*exp(i*cphase);

% subtract to get resonator response
rcmx=tcmx-ccmx;

% convert back to phasor
rmag=abs(rcmx);
rphase=angle(rcmx);

% convert to dB
rdb=20*log10(rmag);

% convert to degrees
rdeg=rphase*360/2/pi;

res.frequency=xfreq;
res.magnitude=rdb;
res.phase=rdeg
end
```

Appendix C Resonant tactile sensor material testing results

(a) poly (ethyl methacrylate) (PEMA) ($E = 1.1 \sim 1.3$ GPa).

(b) Sn63/Pb37 ($E = 33.6$ GPa)

(c) Sn96.5/Ag3.0/Cu0.5 bead ($E = 51.2$ GPa)

Figure C.1: Magnitude and phase response of the sensor when in contact with materials with different Young's modulus

Appendix D List of Publications

A large part of the work presented in this thesis have been published in peer-reviewed journals and conference proceedings, below is a list of the publications that are related to the work presented in this thesis.

D.1 Journal papers

- [1] **Han, B.**, Yoon, Y.-J., Hamidullah, M., Lin, A. T.-H., and Park, W.-T., “Silicon nanowire-based ring-shaped tri-axial force sensor for smart integration on guidewire,” *Journal of Micromechanics and Microengineering*, vol. 24, no. 6, 2014 (cover page).
- [2] **Han, B.**, Yoon, Y.-J., Hamidullah, M., Lin, A. T.-H., and Park, W.-T., “Bionic Meissner Corpuscles Mechanoreceptor: Silicon Nanowire (SiNW) Based Biomimetic Micro-tactile Sensor”, *Bioinspiration & Biomimetics* (under review)
- [3] **Han, B.**, Yoon, Y.-J., Park, W.-T., Wang N., A. Gu Y., A. Lin T.-H., “Resonance tactile sensors using the resonant frequency shift to detect material hardness,” *Journal of Micromechanics and Microengineering* (in submission)
- [4] **Han, B.**, Yoon, Y.-J., Park, W.-T., Lim R., Wang N., A. Gu Y., A. Lin T.-H., “A MEMS resonant tactile sensor for material elasticity measurement,” *Applied Physics Letter* (in submission)

D.2 Conference proceedings

- [1] **Han, B.**, Yoon, Y.-J., Park, W.-T., Lim R., Wang N., A. Gu Y., A. Lin T.-H., “A silicon ring resonant tactile sensor for material elasticity measurement,” *2015 IEEE Conference on Electron Devices and Solid-State Circuits (EDSSC2015)*

- [2] **B. Han**, Y.-J. Yoon, M. Hamidullah, Angel T.-H. Lin, and W.-T. Park, 2013, “Silicon nanowire based ring shape force sensor for sensorized guidewires”, *The 17th International Conference on Solid-State Sensors, Actuators and Microsystems (IEEE Transducers 2013)*, 16 - 20 Jun 2013, CCIB Barcelona, Spain
- [3] **B. Han**, Y.-J. Yoon, M. Hamidullah, Angel T.-H. Lin, and W.-T. Park, 2013, “Characterization of a bio-inspired tri-axial piezoresistive silicon nanowire tactile sensor”, *7th International Conference on Materials for Advanced Technologies (ICMAT 2013)*, 30 June to 5 July 2013, Suntec Singapore
- [4] **B. Han**, Y.-J. Yoon, M. Hamidullah, Angel T.-H. Lin, and W.-T. Park, 2012, “A ring-shaped tri-axial force sensor for minimally invasive surgery”, *International Conference on Medical and Biological Engineering (ICMBE 2012)*, September 12-13, 2012, River View Hotel Singapore
- [5] **B. Han**, Y.-J. Yoon, M. Hamidullah, Angel T.-H. Lin, and W.-T. Park, 2013, “Development and characterization of a silicon nanowire based ring shape force sensor for guidewire application”, *3rd Asia-Korea Conference on Science and Technology 2013 (AKC 2013)*, 12-13 November 2013, Resort World Sentosa, Singapore
- [6] **B. Han**, Y.-J. Yoon, M. Hamidullah, Angel T.-H. Lin, and W.-T. Park, 2012, “Development of a ring-shaped tri-axial tactile sensor for minimally invasive surgery applications”, *Bio4Apps 2012*, 19-20 Nov 2012, NUS, Singapore
- [7] M. Hamidullah, **B. Han**, Angel T.-H. Lin, Y.-J. Yoon, and W.-T. Park, 2012, “Bioinspired 3-d piezoresistive micro-tactile sensor array”, *The 6th Asia-Pacific Conference on Transducers and Micro/Nano Technologies (APCOT 2012)*, July 8-11, 2012, Nanjing China
- [8] M. Hamidullah, Angel Lin, **B. Han**, and Y.-J. Yoon, “A sensorized surgical needle with miniaturized mems tri-axial force sensor for robotic assisted minimally invasive surgery”, *14th Electronics Packaging Technology conference (EPTC 2012)*, 5-7 December 2012, Resorts World Sentosa, Singapore

D.3 Funding application

MoE Tier 1 research fund, ‘MEMS tactile sensors for minimally invasive surgery application’ (2013 Awarded)

References

- [1] G. S. Guthart and J. K. Salisbury, Jr., "The IntuitiveTM telesurgery system: overview and application," in *IEEE International Conference on Robotics and Automation (ICRA), 2000. Proceedings.*, 2000, pp. 618 -621
- [2] J. C. Gwilliam, Z. Pezzementi, E. Jantho, A. M. Okamura, and S. Hsiao, "Human vs. robotic tactile sensing: Detecting lumps in soft tissue," in *Haptics Symposium, 2010 IEEE*, 2010, pp. 21-28.
- [3] B. Demi, T. Ortmaier, and U. Seibold, "The touch and feel in minimally invasive surgery," in *Haptic Audio Visual Environments and their Applications, 2005. IEEE International Workshop on*, 2005, p. 6 pp.
- [4] M. Hashizume, M. Shimada, M. Tomikawa, Y. Ikeda, I. Takahashi, R. Abe, *et al.*, "Early experiences of endoscopic procedures in general surgery assisted by a computer-enhanced surgical system," *Surgical Endoscopy and Other Interventional Techniques*, vol. 16, pp. 1187-1191, Aug 2002.
- [5] Y. Haga and M. Esashi, "Biomedical microsystems for minimally invasive diagnosis and treatment," *Proceedings of the IEEE*, vol. 92, pp. 98-114, 2004.
- [6] K. J. Rebello, "Applications of MEMS in surgery," *Proceedings of the IEEE*, vol. 92, pp. 43 - 55 2004.
- [7] R. A. M. Receveur, F. W. Lindemans, and N. F. d. Rooij, "Microsystem technologies for implantable applications," *Journal of Micromechanics and Microengineering*, vol. 17, pp. R50-R80, 2007.
- [8] P. Puangmali, K. Althoefer, L. D. Seneviratne, D. Murphy, and P. Dasgupta, "State-of-the-Art in Force and Tactile Sensing for Minimally Invasive Surgery," *IEEE Sensors Journal*, vol. 8, pp. 371 -381, 2008.
- [9] R. S. Dahiya, G. Metta, M. Valle, and G. Sandini, "Tactile sensing—from humans to humanoids," *IEEE Transactions on Robotics*, vol. 26, pp. 1 -20 2010.
- [10] M. I. Tiwana, S. J. Redmond, and N. H. Lovell, "A review of tactile sensing technologies with applications in biomedical engineering," *Sensors and Actuators A: Physical*, vol. 179, pp. 17-31 2012.
- [11] R. Razavi, D. L. G. Hill, S. F. Keevil, M. E. Miquel, V. Muthurangu, S. Hegde, *et al.*, "Cardiac catheterisation guided by MRI in children and adults with congenital heart disease," *Lancet*, vol. 362, pp. 1877-1882, Dec 2003.

-
- [12] S. E. Nissen and P. Yock, "Intravascular ultrasound - Novel pathophysiological insights and current clinical applications," *Circulation*, vol. 103, pp. 604-616, Jan 2001.
- [13] Y. Haga, M. Fujita, K. Nakamura, C. J. Kim, and M. Esashi, "Batch fabrication of intravascular forward-looking ultrasonic probe," *Sensors and Actuators a-Physical*, vol. 104, pp. 40-43, Mar 2003.
- [14] T. Ohtsuka, A. Furuse, T. Kohno, J. Nakajima, K. Yagyu, and S. Omata, "Application of a new tactile sensor to thoracoscopic surgery - experimental and clinical-study," *Annals of Thoracic Surgery*, vol. 60, pp. 610-614, Sep 1995.
- [15] D. P. Noonan, H. Liu, Y. H. Zweiri, K. A. Althoefer, and L. D. Seneviratne, "A Dual-Function Wheeled Probe for Tissue Viscoelastic Property Identification during Minimally Invasive Surgery," in *Robotics and Automation, 2007 IEEE International Conference on*, 2007, pp. 2629-2634.
- [16] A. L. Trejos, J. Jayender, M. T. Perri, M. D. Naish, R. V. Patel, and R. A. Malthaner, "Robot-assisted Tactile Sensing for Minimally Invasive Tumor Localization," *International Journal of Robotics Research*, vol. 28, pp. 1118-1133, Sep 2009.
- [17] T. Yamamoto, B. Vagvolgyi, K. Balaji, L. L. Whitcomb, and A. M. Okamura, "Tissue Property Estimation and Graphical Display for Teleoperated Robot-Assisted Surgery," in *ICRA: 2009 IEEE International Conference on Robotics and Automation, Vols 1-7*, ed, 2009, pp. 3117-3123.
- [18] H. Liu, D. P. Noonan, B. J. Challacombe, P. Dasgupta, L. D. Seneviratne, and K. Althoefer, "Rolling Mechanical Imaging for Tissue Abnormality Localization During Minimally Invasive Surgery," *IEEE Transactions on Biomedical Engineering*, vol. 57, pp. 404-414, Feb 2010.
- [19] Y. Okumura, S. B. Johnson, T. J. Bunch, B. D. Henz, C. J. O'Brien, and D. L. Packer, "A systematical analysis of in vivo contact forces on virtual catheter tip/tissue surface contact during cardiac mapping and intervention," *Journal of Cardiovascular Electrophysiology*, vol. 19, pp. 632-640, Jun 2008.
- [20] K. Yokoyama, H. Nakagawa, D. C. Shah, H. Lambert, G. Leo, N. Aeby, *et al.*, "Novel Contact Force Sensor Incorporated in Irrigated Radiofrequency Ablation Catheter Predicts Lesion Size and Incidence of Steam Pop and Thrombus," *Circulation-Arrhythmia and Electrophysiology*, vol. 1, pp. 354-362, Dec 2008.
- [21] A. Thiagalingam, A. D'Avila, L. Foley, J. L. Guerrero, H. Lambert, G. Leo, *et al.*, "Importance of Catheter Contact Force During Irrigated Radiofrequency Ablation: Evaluation in a Porcine Ex Vivo Model Using a Force-Sensing Catheter," *Journal of Cardiovascular Electrophysiology*, vol. 21, pp. 806-811, Jul 2010.
- [22] J. R. Clark, W. T. Hsu, M. A. Abdelmoneum, and C. T. C. Nguyen, "High-Q UHF micromechanical radial-contour mode disk resonators," *Journal of Microelectromechanical Systems*, vol. 14, pp. 1298-1310, Dec 2005.
- [23] S. Pourkamali, Z. Hao, and F. Ayazi, "VHF single crystal silicon capacitive elliptic bulk-mode disk resonators - Part II: Implementation and characterization," *Journal of Microelectromechanical Systems*, vol. 13, pp. 1054-1062, Dec 2004.

References

- [24] J. E. Y. Lee and A. A. Seshia, "5.4-MHz single-crystal silicon wine glass mode disk resonator with quality factor of 2 million," *Sensors and Actuators a-Physical*, vol. 156, pp. 28-35, Nov 2009.
- [25] E. Algre, X. Zhuang, M. Faucher, B. Walter, L. Buchaillot, and B. Legrand, "MEMS Ring Resonators for Laserless AFM With Sub-nanoNewton Force Resolution," *Microelectromechanical Systems, Journal of*, vol. 21, pp. 385-397, 2012.
- [26] Z. Xiong, B. Walter, E. Mairiaux, M. Faucher, L. Buchaillot, and B. Legrand, "MEMS piezoresistive ring resonator for AFM imaging with pico-Newton force resolution," *Journal of Micromechanics and Microengineering*, vol. 23, p. 035016, 2013.
- [27] V. Kaajakari, T. Mattila, A. Oja, J. Kiihamaki, and H. Seppa, "Square-extensional mode single-crystal silicon micromechanical resonator for low-phase-noise oscillator applications," *Electron Device Letters, IEEE*, vol. 25, pp. 173-175, 2004.
- [28] J. E. Y. Lee, Y. Zhu, and A. A. Seshia, "A bulk acoustic mode single-crystal silicon microresonator with a high-quality factor," *Journal of Micromechanics and Microengineering*, vol. 18, p. 064001, 2008.
- [29] R. S. Fearing, "Tactile sensing mechanisms," *International Journal of Robotics Research*, vol. 9, pp. 3-23, Jun 1990.
- [30] M. E. H. Eltaib and J. R. Hewit, "Tactile sensing technology for minimal access surgery - a review," *Mechatronics*, vol. 13, pp. 1163-1177, Dec 2003.
- [31] M. H. Lee and H. R. Nicholls, "Tactile sensing for mechatronics - a state of the art survey," *Mechatronics*, vol. 9, pp. 1-31, Feb 1999.
- [32] L. Da, D. Zhang, and T. Wang, "Overview of the vascular interventional robot," *International Journal of Medical Robotics and Computer Assisted Surgery*, vol. 4, pp. 289-294, Dec 2008.
- [33] P. Polygerinos, D. Zbyszewski, T. Schaeffter, R. Razavi, L. D. Seneviratne, and K. Althoefer, "MRI-Compatible Fiber-Optic Force Sensors for Catheterization Procedures," *IEEE Sensors Journal*, vol. 10, pp. 1598 -1608 2010.
- [34] M. Esashi, H. Komatsu, T. Matsuo, M. Takahashi, T. Takishima, K. Imabayashi, *et al.*, "Fabrication of catheter-tip and sidewall miniature pressure sensors," *IEEE Transactions on Electron Devices*, vol. 29, pp. 57-63, 1982.
- [35] E. Kalvesten, L. Smith, L. Tenerz, and G. Stemme, "The first surface micromachined pressure sensor for cardiovascular pressure measurements," presented at the IEEE MEMS 1998, 1998.
- [36] M. Tanimoto, F. Arai, T. Fukuda, H. Iwata, K. Itoigawa, Y. Gotoh, *et al.*, "Micro force sensor for intravascular neurosurgery and in vivo experiment," presented at the MEMS 1998. Proceedings, 1998.
- [37] T. Meiß, T. A. Kern, S. Sindlinger, and R. Werthschützky, "HapCath: Highly Miniaturized Piezoresistive Force Sensors for Interior Palpation of Vessels during Angioplasty," in *World Congress on Medical Physics and Biomedical Engineering, IFMBE Proceedings*, 2009, pp. 228-231

-
- [38] M. Kalantari, M. Ramezanifard, R. Ahmadi, J. Dargahi, and J. Kovecses, "A piezoresistive tactile sensor for tissue characterization during catheter-based cardiac surgery," *Int J Med Robot*, vol. 7, pp. 431-40, Dec 2011.
- [39] W. T. Park, R. K. Kotlanka, L. Lou, M. Hamidullah, and C. Lee, "MEMS tri-axial force sensor with an integrated mechanical stopper for guidewire applications," *Microsystem Technologies*, 2012.
- [40] B. Han, Y.-J. Yoon, M. Hamidullah, T.-H. Angel Lin, and W.-T. Park, "Silicon Nanowire Based Ring Shape Force Sensor For Sensorized Guidewires," presented at the The 17th International Conference on Solid-State Sensors, Actuators and Microsystems, Barcelona, Spain, 2013.
- [41] H. Rafii-Tari, C. J. Payne, and G.-Z. Yang, "Current and Emerging Robot-Assisted Endovascular Catheterization Technologies: A Review," *Annals of Biomedical Engineering*, vol. 42, pp. 697-715, Apr 2014.
- [42] B. Kuebler, U. Seibold, and G. Hirzinger, "Development of actuated and sensor integrated forceps for minimally invasive robotic surgery," *International Journal of Medical Robotics and Computer Assisted Surgery*, vol. 1, pp. 96-107, Sep 2005.
- [43] P. Puangmali, H. Liu, K. Althoefer, and L. D. Seneviratne, "Optical fiber sensor for soft tissue investigation during minimally invasive surgery," in *Robotics and Automation, 2008. ICRA 2008. IEEE International Conference on*, 2008, pp. 2934-2939.
- [44] P. Valdastrì, K. Harada, A. Menciassi, L. Beccai, C. Stefanini, M. Fujie, *et al.*, "Integration of a Miniaturised Triaxial Force Sensor in a Minimally Invasive Surgical Tool," *Biomedical Engineering, IEEE Transactions on*, vol. 53, pp. 2397-2400, 2006.
- [45] J. Peirs, J. Clijnen, D. Reynaerts, H. V. Brussel, P. Herijgers, B. Corteville, *et al.*, "A micro optical force sensor for force feedback during minimally invasive robotic surgery," *Sensors and Actuators A: Physical*, vol. 115, pp. 447-455, 9/21/ 2004.
- [46] Endosense: *The TactiCath Force-Sensing Ablation Catheter*. Available: <http://professional-intl.sjm.com/therapies/tacticath-quartz/home>
- [47] H. Medical: *Sensei X Robotic Catheter System*. Available: <http://www.hansenmedical.com/us/en/cardiac-arrhythmia/sensei-robotic-system/product-overview>
- [48] P. Polygerinos, D. Zbyszewski, T. Schaeffter, R. Razavi, L. D. Seneviratne, and K. Althoefer, "MRI-Compatible Fiber-Optic Force Sensors for Catheterization Procedures," *IEEE Sensors Journal*, vol. 10, pp. 1598-1608, Oct 2010.
- [49] P. Polygerinos, T. Schaeffter, L. Seneviratne, and K. Althoefer, "A Fibre-Optic Catheter-Tip Force Sensor with MRI Compatibility: A Feasibility Study," in *2009 Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Vols 1-20*, ed, 2009, pp. 1501-1504.
- [50] S. J. Medical: *PressureWire Aeris*. Available: <http://professional-intl.sjm.com/products/vas/intravascular-diagnostics-imaging/ffr/pressurewire-aeris>

References

- [51] T. Meiss, T. Rossner, C. M. Faria, S. Voellmeke, and T. Opitz, "Performance of a novel micro force vector sensor and outlook into its biomedical applications," *Bioelectronics, Biomedical, and Bioinspired Systems V and Nanotechnology V*, vol. 8068, 2011.
- [52] L. Beccai, S. Roccella, A. Arena, F. Valvo, P. Valdastri, A. Menciassi, *et al.*, "Design and fabrication of a hybrid silicon three-axial force sensor for biomechanical applications," *Sensors and Actuators A: Physical*, vol. 120, pp. 370-382 2005.
- [53] O. Hammarstrm, P. Benkowski, P. v. Malmborg, and L. Tenerz, "Sensor and guide wire assembly," 6336906, Jan 8, 2002, 2002.
- [54] Z. Chu, P. M. Sarro, and S. Middelhoek, "Silicon three-axial tactile sensor," *Sensors and Actuators A: Physical*, vol. 54, pp. 505-510 1996.
- [55] B. J. Kane, M. R. Cutkosky, and G. T. A. Kovacs, "A traction stress sensor array for use in high-resolution robotic tactile imaging," *Microelectromechanical Systems, Journal of*, vol. 9, pp. 425-434, 2000.
- [56] L. Wang and D. J. Beebe, "A silicon-based shear force sensor: development and characterization," *Sensors and Actuators A: Physical*, vol. 84, pp. 33-44 2000.
- [57] S. Büttefisch, S. Büttgenbach, T. Kleine-Besten, and U. Brand, "Micromechanical three-axial tactile force sensor for micromaterial characterisation," *Microsystem Technologies*, vol. 7, pp. 171-174 2001.
- [58] É. Vázsonyi, M. Ádám, C. Dücső, Z. Vízváry, A. L. Tóth, and I. Bársony, "Three-dimensional force sensor by novel alkaline etching technique," *Sensors and Actuators A: Physical*, vol. 123-124, pp. 620-626, 2005.
- [59] G. Vasarhelyi, M. Adam, E. Vazsonyi, Z. Vizvary, A. Kis, I. Barsony, *et al.*, "Characterization of an Integrable Single-Crystalline 3-D Tactile Sensor," *Sensors Journal, IEEE*, vol. 6, pp. 928 -934, 2006.
- [60] Y. Hu, R. B. Katragadda, H. Tu, Q. Zheng, Y. Li, and Y. Xu, "Bioinspired 3-D Tactile Sensor for Minimally Invasive Surgery," *Journal of Microelectromechanical Systems*, vol. 19, pp. 1400 -1408 2010.
- [61] Y. Zhang, "Sensitivity enhancement of a micro-scale biomimetic tactile sensor with epidermal ridges," *Journal of Micromechanics and Microengineering*, vol. 20, p. 085012, 2010.
- [62] R. R. He and P. D. Yang, "Giant piezoresistance effect in silicon nanowires," *Nature Nanotechnology*, vol. 1, pp. 42-46, Oct 2006.
- [63] A. C. H. Rowe, "Silicon nanowires feel the pinch," *Nature Nanotechnology*, vol. 3, pp. 311-312, Jun 2008.
- [64] P. Singh, W.-T. Park, J. Miao, L. Shao, R. Krishna Kotlanka, and D.-L. Kwong, "Tunable piezoresistance and noise in gate-all-around nanowire field-effect-transistor," *Applied Physics Letters*, vol. 100, p. 063106, 2012.
- [65] A. A. Barlian, W. T. Park, J. R. Mallon, A. J. Rastegar, and B. L. Pruitt, "Review: Semiconductor Piezoresistance for Microsystems," *Proceedings of the IEEE*, vol. 97, pp. 513 -552, 2009.

-
- [66] T. Toriyama, Y. Tanimoto, and S. Sugiyama, "Single crystal silicon nano-wire piezoresistors for mechanical sensors," *Journal of Microelectromechanical Systems*, vol. 11, pp. 605-611, Oct 2002.
- [67] Y. Paska and H. Haick, "Interactive Effect of Hysteresis and Surface Chemistry on Gated Silicon Nanowire Gas Sensors," *Acs Applied Materials & Interfaces*, vol. 4, pp. 2604-2617, May 2012.
- [68] L. Lou, K. Ramakrishna, L. Shao, W.-T. Park, D. Yu, L. Lim, *et al.*, "Sensorized guidewires with MEMS tri-axial force sensor for minimally invasive surgical applications," in *Engineering in Medicine and Biology Society (EMBC), 2010 Annual International Conference of the IEEE*, ed, 2010, pp. 646 - 6464.
- [69] P. Singh, J. Miao, W.-T. Park, and D.-L. Kwong, "Gate-bias-controlled sensitivity and SNR enhancement in a nanowire FET pressure sensor," *Journal of Micromechanics and Microengineering*, vol. 21, p. 105007, 2011.
- [70] L. Lou, S. Zhang, W.-T. Park, J. M. Tsai, D.-L. Kwong, and C. Lee, "Optimization of NEMS pressure sensors with a multilayered diaphragm using silicon nanowires as piezoresistive sensing elements," *Journal of Micromechanics and Microengineering*, vol. 22, p. 055012 2012.
- [71] S. Zhang, L. Lou, and C. Lee, "Piezoresistive silicon nanowire based nanoelectromechanical system cantilever air flow sensor," *Applied Physics Letters*, vol. 100, pp. 023111-023111-3 2012.
- [72] W. Thomson, "On the electro-dynamic qualities of metals:--effects of magnetization on the electric conductivity of nickel and of iron," *Proceedings of the Royal Society of London*, vol. 8, pp. 546-550, 1856.
- [73] P. W. Bridgman, "Some properties of single metal crystals," *Proceedings of the National Academy of Sciences of the United States of America*, vol. 10, pp. 411-415, 1924.
- [74] P. W. Bridgman, "The effect of homogeneous mechanical stress on the electrical resistance of crystals," *Physical Review*, vol. 42, pp. 0858-0863, Dec 1932.
- [75] J. W. Cookson, "Theory of the piezo-resistive effect," *Physical Review*, vol. 47, pp. 194-195, Jan 1935.
- [76] C. S. Smith, "Piezoresistance Effect in Germanium and Silicon," *Physical Review*, vol. 94, pp. 42-49 1954.
- [77] W. P. Mason and R. N. Thurston, "Use of piezoresistive materials in the measurement of displacement, force, and torque," *Journal of the Acoustical Society of America*, vol. 29, pp. 1096-1101, 1957.
- [78] H. Rolnick, "Tension coefficient of resistance of metals," *Physical Review*, vol. 36, pp. 0506-0512, Aug 1930.
- [79] S. D. Senturia, *Microsystem design* Boston : Kluwer Academic Publishers, 2001.
- [80] W. Lu, P. Xie, and C. M. Lieber, "Nanowire Transistor Performance Limits and Applications," *IEEE Transactions on Electron Devices*, vol. 55, pp. 2859-2876, Nov 2008.

References

- [81] K. Nishiguchi, Y. Ono, A. Fujiwara, H. Yamaguchi, H. Inokawa, and Y. Takahashi, "Infrared detection with silicon nano-field-effect transistors," *Applied Physics Letters*, vol. 90, May 28 2007.
- [82] K. Reck, J. Richter, O. Hansen, and E. V. Thomsen, "Piezoresistive effect in top-down fabricated silicon nanowires," in *Mems 2008: 21st IEEE International Conference on Micro Electro Mechanical Systems, Technical Digest*, ed, 2008, pp. 717-720.
- [83] I. Park, Z. Li, A. P. Pisano, and R. S. Williams, "Top-down fabricated silicon nanowire sensors for real-time chemical detection," *Nanotechnology*, vol. 21, Jan 8 2010.
- [84] W.-T. Park, R. Kotlanka, L. Lou, M. Hamidullah, and C. Lee, "MEMS tri-axial force sensor with an integrated mechanical stopper for guidewire applications," *Microsystem Technologies*, vol. 19, pp. 1005-1015, 2013.
- [85] B. Han, Y.-J. Yoon, M. Hamidullah, A. T.-H. Lin, and W.-T. Park, "Silicon nanowire-based ring-shaped tri-axial force sensor for smart integration on guidewire," *Journal of Micromechanics and Microengineering*, vol. 24, p. 065002, 2014.
- [86] J. Kim, K. Park, H. Kim, and K. Chun, "Fabrication of a piezoresistive pressure sensor for enhancing sensitivity using silicon nanowire," in *Solid-State Sensors, Actuators and Microsystems Conference, 2009. TRANSDUCERS 2009. International*, 2009, pp. 1936-1939.
- [87] Y. Zhang, X. Y. Liu, C. H. Ru, Y. L. Zhang, L. X. Dong, and Y. Sun, "Piezoresistivity Characterization of Synthetic Silicon Nanowires Using a MEMS Device," *Journal of Microelectromechanical Systems*, vol. 20, pp. 959-967, Aug 2011.
- [88] N. Singh, K. D. Buddharaju, S. Manhas, A. Agarwal, S. C. Rustagi, G. Lo, *et al.*, "Si, SiGe nanowire devices by top-down technology and their applications," *Electron Devices, IEEE Transactions on*, vol. 55, pp. 3107-3118, 2008.
- [89] S. Phipps, T. H. J. Yang, F. K. Habib, R. L. Reuben, and S. A. McNeill, "Measurement of tissue mechanical characteristics to distinguish between benign and malignant prostatic disease," *Urology*, vol. 66, pp. 447-450, Aug 2005.
- [90] A. M. Davis, S. Dische, L. Gerber, M. Saunders, S. F. Leung, and B. O'Sullivan, "Measuring postirradiation subcutaneous soft-tissue fibrosis: State-of-the-art and future directions," *Seminars in Radiation Oncology*, vol. 13, pp. 203-213, Jul 2003.
- [91] J. D. Brown, J. Rosen, Y. S. Kim, L. Chang, M. N. Sinanan, and B. Hannaford, "In-vivo and in-situ compressive properties of porcine abdominal soft tissues," in *Medicine Meets Virtual Reality 11: Nextmed: Health Horizon*. vol. 94, J. D. Westwood, H. M. Hoffman, G. T. Mogel, R. Phillips, R. A. Robb, and D. Stredney, Eds., ed, 2003, pp. 26-32.
- [92] B. S. Garra, E. I. Cespedes, J. Ophir, S. R. Spratt, R. A. Zuurbier, C. M. Magnant, *et al.*, "Elastography of breast lesions: Initial clinical results," *Radiology*, vol. 202, pp. 79-86, Jan 1997.

-
- [93] A. L. McKnight, J. L. Kugel, P. J. Rossman, A. Manduca, L. C. Hartmann, and R. L. Ehman, "MR elastography of breast cancer: Preliminary results," *American Journal of Roentgenology*, vol. 178, Jun 2002.
- [94] N. Tanaka, M. Higashimori, M. Kaneko, and I. Kao, "Noncontact active sensing for viscoelastic parameters of tissue with coupling effect," *Biomedical Engineering, IEEE Transactions on*, vol. 58, pp. 509-520, 2011.
- [95] R. Damadian, "TUMOR DETECTION BY NUCLEAR MAGNETIC RESONANCE," *Science*, vol. 171, pp. 1151-&, 1971.
- [96] A. G. Sorensen, "Magnetic resonance as a cancer imaging biomarker," *Journal of Clinical Oncology*, vol. 24, pp. 3274-3281, Jul 2006.
- [97] I. Khalkhali, I. Mena, and L. Diggles, "REVIEW OF IMAGING TECHNIQUES FOR THE DIAGNOSIS OF BREAST-CANCER - A NEW ROLE OF PRONE SCINTIMAMMOGRAPHY USING TC-99M SESTAMIBI," *European Journal of Nuclear Medicine*, vol. 21, pp. 357-362, Apr 1994.
- [98] P. Peng, A. S. Sezen, R. Rajamani, and A. G. Erdman, "Novel MEMS stiffness sensor for force and elasticity measurements," *Sensors and Actuators a-Physical*, vol. 158, pp. 10-17, Mar 2010.
- [99] G. Tholey, J. P. Desai, and A. E. Castellanos, "Force feedback plays a significant role in minimally invasive surgery - Results and analysis," *Annals of Surgery*, vol. 241, pp. 102-109, Jan 2005.
- [100] D. Kalanovic, M. P. Ottensmeyer, J. Gross, G. Buess, and S. L. Dawson, "Independent testing of soft tissue visco-elasticity using indentation and rotary shear deformations," in *Medicine Meets Virtual Reality 11: Nextmed: Health Horizon*. vol. 94, J. D. Westwood, H. M. Hoffman, G. T. Mogel, R. Phillips, R. A. Robb, and D. Stredney, Eds., ed, 2003, pp. 137-143.
- [101] P. Wellman, R. D. Howe, E. Dalton, and K. A. Kern, "Breast tissue stiffness in compression is correlated to histological diagnosis," *Harvard BioRobotics Laboratory Technical Report*, 1999.
- [102] A. Samani, J. Bishop, C. Luginbuhl, and D. B. Plewes, "Measuring the elastic modulus of ex vivo small tissue samples," *Physics in Medicine and Biology*, vol. 48, pp. 2183-2198, Jul 2003.
- [103] R. Q. Erkamp, P. Wiggins, A. R. Skovoroda, S. Y. Emelianov, and M. O'Donnell, "Measuring the elastic modulus of small tissue samples," *Ultrasonic Imaging*, vol. 20, pp. 17-28, Jan 1998.
- [104] R. Erkamp, P. Wiggins, A. Skovoroda, S. Y. Emelianov, and M. O'Donnell, "Measuring the elastic modulus of small tissue samples," *Ultrasonic Imaging*, vol. 20, pp. 17-28, 1998.
- [105] S. Omata and Y. Terunuma, "New tactile sensor like the human hand and its applications," *Sensors and Actuators a-Physical*, vol. 35, pp. 9-15, Oct 1992.

References

- [106] Y. Hasegawa, M. Shikida, H. Sasaki, K. Itoigawa, and K. Sato, "An active tactile sensor for detecting mechanical characteristics of contacted objects," *Journal of Micromechanics and Microengineering*, vol. 16, pp. 1625-1632, Aug 2006.
- [107] M. H. Araghi and S. P. Salisbury, "Improved Evaluation of Dynamic Mechanical Properties of Soft Materials With Applications to Minimally Invasive Surgery," *IEEE-ASME Transactions on Mechatronics*, vol. 18, pp. 973-980, Jun 2013.
- [108] Y. Murayama and S. Omata, "Fabrication of micro tactile sensor for the measurement of micro-scale local elasticity," *Sensors and Actuators a-Physical*, vol. 109, pp. 202-207, Jan 1 2004.
- [109] P. Brunetto, L. Fortuna, P. Giannone, S. Graziani, and F. Pagano, "A Resonant Vibrating Tactile Probe for Biomedical Applications Based on IPMC," *IEEE Transactions on Instrumentation and Measurement*, vol. 59, pp. 1453-1462, May 2010.
- [110] K. Azgin, T. Akin, and L. Valdevit, "Ultrahigh-Dynamic-Range Resonant MEMS Load Cells for Micromechanical Test Frames," *Journal of Microelectromechanical Systems*, vol. 21, pp. 1519-1529, Dec 2012.
- [111] P. Peng, A. S. Sezen, R. Rajamani, and A. G. Erdman, "Novel MEMS stiffness sensor for force and elasticity measurements," *Sensors and Actuators A: Physical*, vol. 158, pp. 10-17, 2010.
- [112] K. Miller, K. Chinzei, G. Orsengo, and P. Bednarz, "Mechanical properties of brain tissue in-vivo: experiment and computer simulation," *Journal of Biomechanics*, vol. 33, pp. 1369-1376, Nov 2000.
- [113] A. Nava, E. Mazza, M. Furrer, P. Villiger, and W. H. Reinhart, "In vivo mechanical characterization of human liver," *Medical Image Analysis*, vol. 12, pp. 203-216, Apr 2008.
- [114] Y. Murayama, M. Haruta, Y. Hatakeyama, T. Shiina, H. Sakuma, S. Takenoshita, *et al.*, "Development of a new instrument for examination of stiffness in the breast using haptic sensor technology," *Sensors and Actuators a-Physical*, vol. 143, pp. 430-438, May 16 2008.
- [115] Y. Murayama, J. Mizuno, H. Kamakura, Y. Fueta, H. Nakamura, K. Akaishi, *et al.*, "Mouse zona pellucida dynamically changes its elasticity during oocyte maturation, fertilization and early embryo development," *Human cell*, vol. 19, pp. 119-125, 2006.
- [116] Y. Murayama, C. E. Constantinou, and S. Omata, "Micro-mechanical sensing platform for the characterization of the elastic properties of the ovum via uniaxial measurement," *Journal of Biomechanics*, vol. 37, pp. 67-72, Jan 2004.
- [117] V. Jalkanen, B. M. Andersson, A. Bergh, B. Ljungberg, and O. A. Lindahl, "Prostate tissue stiffness as measured with a resonance sensor system: a study on silicone and human prostate tissue in vitro," *Medical & Biological Engineering & Computing*, vol. 44, pp. 593-603, Jul 2006.
- [118] S. Kudo, "A Study of Vibratory Tactile Sensor Using a Horn-Type Longitudinal Bar Resonator," *Japanese Journal of Applied Physics*, vol. 49, p. 07HD03, 2010.

-
- [119] R. Aoyagi and T. Yoshida, "Frequency Equations of an Ultrasonic Vibrator for the Elastic Sensor Using a Contact Impedance Method," *Japanese Journal of Applied Physics*, vol. 43, pp. 3204-3209, 2004.
- [120] T. Hemsell, R. Stroop, D. O. Uribe, and J. Wallaschek, "Resonant vibrating sensors for tactile tissue differentiation," *Journal of Sound and Vibration*, vol. 308, pp. 441-446, Dec 4 2007.
- [121] S. Omata, 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*, vol. 112, pp. 278-285, May 1 2004.
- [122] A. T. H. Lin, Y. Jize, and A. A. Seshia, "Electrostatically transduced face-shear mode silicon MEMS microresonator," in *Frequency Control Symposium (FCS), 2010 IEEE International*, 2010, pp. 534-538.
- [123] S. Kudo, "Sensitivity of Frequency Change of Piezoelectric Vibratory Tactile Sensor Using Longitudinal-Bar-Type Resonator," *Japanese Journal of Applied Physics*, vol. 46, pp. 4704-4708, 2007.
- [124] S. Kudo, "Vibratory tactile sensor integrated a longitudinal resonator and a force sensor," *Jsmc International Journal Series C-Mechanical Systems Machine Elements and Manufacturing*, vol. 49, pp. 675-680, Sep 2006.
- [125] A. Erglis, I. Narbutė, D. Sondore, A. Grave, and S. Jegere, "Tools & Techniques: coronary guidewires," *EuroIntervention*, vol. 6:1-8, 2010.
- [126] J. B. Simpson, D. S. Baim, E. W. Robert, and D. C. Harrison, "A new catheter system for coronary angioplasty," *American Journal of Cardiology*, vol. 49, pp. 1216-1222, 1982.
- [127] K. L. Chan, K.-T. Tiew, A. Astuti Lee, J. Luo, S. S. Y. Ng, and M. Je, "A tactile sensor ASIC for a sensorized guidewire in minimally invasive surgical operations," in *Solid State Circuits Conference (A-SSCC), 2010 IEEE Asian*, ed, 2010, pp. 1-4.
- [128] S. D. Senturia, *Microsystem Design*: Springer 2000.
- [129] E. Bassous, "Fabrication of novel three-dimensional microstructures by the anisotropic etching of," *IEEE Transactions on Electron Devices*, vol. 25, pp. 1178 - 1185 1978.
- [130] J. M. Gere and B. J. Goodno, *Mechanics of materials*: Toronto, ON : Cengage Learning, c2009. 7th ed. / James M. Gere, Barry J. Goodno., 2009.
- [131] M. A. Hopcroft, W. D. Nix, and T. W. Kenny, "What is the Young's Modulus of Silicon?," *Journal of Microelectromechanical Systems*, vol. 19, pp. 229 -238, 2010.
- [132] M. J. Madou, *Fundamentals of Microfabrication: The Science of Miniaturization, Second Edition: The Science of Miniturization*: CRC Press 2002.
- [133] L. H. Shu, K. Ueda, I. Chiu, and H. Cheong, "Biologically inspired design," *CIRP Annals - Manufacturing Technology*, vol. 60, pp. 673-693, 2011.
- [134] T. Someya, T. Sekitani, S. Iba, Y. Kato, H. Kawaguchi, and T. Sakurai, "A large-area, flexible pressure sensor matrix with organic field-effect transistors for artificial skin

References

- applications," *Proceedings of the National Academy of Sciences of the United States of America*, vol. 101, pp. 9966-9970, July 6, 2004 2004.
- [135] J. Dargahi and S. Najarian, "Human tactile perception as a standard for artificial tactile sensing - a review," *International Journal of Medical Robotics and Computer Assisted Surgery*, vol. 1, pp. 23-35, Jun 2004.
- [136] J. Tegin and J. Wikander, "Tactile sensing in intelligent robotic manipulation—a review," *Industrial Robot: An International Journal*, vol. 32, pp. 64-70, 2005.
- [137] H. B. Muhammad, C. M. Oddo, L. Beccai, C. Recchiuto, C. J. Anthony, M. J. Adams, *et al.*, "Development of a bioinspired MEMS based capacitive tactile sensor for a robotic finger," *Sensors and Actuators a-Physical*, vol. 165, pp. 221-229, Feb 2011.
- [138] N. Wettels, V. J. Santos, R. S. Johansson, and G. E. Loeb, "Biomimetic Tactile Sensor Array," *Advanced Robotics*, vol. 22, pp. 829-849, 2008/01/01 2008.
- [139] D. P. J. Cotton, P. H. Chappell, A. Cranny, N. M. White, and S. P. Beeby, "A Novel Thick-Film Piezoelectric Slip Sensor for a Prosthetic Hand," *Sensors Journal, IEEE*, vol. 7, pp. 752-761, 2007.
- [140] A. B. Vallbo and R. S. Johansson, "Properties of cutaneous mechanoreceptors in the human hand related to touch sensation," *Human Neurobiology*, vol. 3, pp. 3-14, 1984 1984.
- [141] P. Delmas, J. Z. Hao, and L. Rodat-Despoix, "Molecular mechanisms of mechanotransduction in mammalian sensory neurons," *Nature Reviews Neuroscience*, vol. 12, pp. 139-153, Mar 2011.
- [142] K. O. Johnson, "The roles and functions of cutaneous mechanoreceptors," *Current Opinion in Neurobiology*, vol. 11, pp. 455-461, Aug 2001.
- [143] E. Goldstein, *Sensation and perception*: Cengage Learning, 2013.
- [144] M. Pare, R. Elde, J. E. Mazurkiewicz, A. M. Smith, and F. L. Rice, "The Meissner corpuscle revised: a multiafferented mechanoreceptor with nociceptor immunochemical properties," *The Journal of Neuroscience*, vol. 21, pp. 7236-7246, 2001.
- [145] T. Maeno, K. Kobayashi, and N. Yamazaki, "Relationship between the structure of human finger tissue and the location of tactile receptors," *Jsme International Journal Series C-Mechanical Systems Machine Elements and Manufacturing*, vol. 41, pp. 94-100, Mar 1998.
- [146] H. Takahashi-Iwanaga and H. Shimoda, "The three-dimensional microanatomy of Meissner corpuscles in monkey palmar skin," *Journal of Neurocytology*, vol. 32, pp. 363-371, 2003.
- [147] S. Kuroki, H. Kajimoto, H. Nii, N. Kawakami, and S. Tachi, "Proposal of the Stretch Detection Hypothesis of the Meissner Corpuscle," in *Haptics: Perception, Devices and Scenarios*. vol. 5024, M. Ferre, Ed., ed: Springer Berlin Heidelberg, 2008, pp. 245-254.

-
- [148] K. Hosoda, Y. Tada, and M. Asada, "Anthropomorphic robotic soft fingertip with randomly distributed receptors," *Robotics and Autonomous Systems*, vol. 54, pp. 104-109, 2006.
- [149] P. Dario, D. De Rossi, C. Domenici, and R. Francesconi, "Ferroelectric polymer tactile sensors with anthropomorphic features," in *Robotics and Automation. Proceedings. 1984 IEEE International Conference on*, 1984, pp. 332-340.
- [150] D. Yamada, T. Maeno, and Y. Yamada, "Artificial finger skin having ridges and distributed tactile sensors used for grasp force control," in *Intelligent Robots and Systems, 2001. Proceedings. 2001 IEEE/RSJ International Conference on*, 2001, pp. 686-691 vol.2.
- [151] X. Chen, J. Sakai, S. Yang, and S. Motojima, "Biomimetic Tactile Sensors with Fingerprint-Type Surface Made of Carbon Microcoils/Polysilicone," *Japanese Journal of Applied Physics*, vol. 45, p. L1019, 2006.
- [152] K. Nishikawa, Y.-K. Park, M. Aizuddin, K. Yoshinaga, K. Ogura, M. Umezu, *et al.*, "Development of carbon microcoils (CMC) sensor system with high sensitivity for effective acquisition of tactile information," in *Intelligent Robots and Systems, 2005.(IROS 2005). 2005 IEEE/RSJ International Conference on*, 2005, pp. 2061-2066.
- [153] T. Kawamura, K. Nejigane, K. Tani, and H. Yamada, "Force Characteristics for Fine Deformation of CMC Touch Sensor and Estimation of Force Variance Using Hybrid Tactile Sensor System," *Journal of Robotics and Mechatronics*, vol. 24, p. 423, 2012.
- [154] K. O. Johnson, T. Yoshioka, and F. Vega-Bermudez, "Tactile functions of mechanoreceptive afferents innervating the hand," *Journal of Clinical Neurophysiology*, vol. 17, pp. 539-558, 2000.
- [155] I. Birznieks, P. Jenmalm, A. W. Goodwin, and R. S. Johansson, "Encoding of direction of fingertip forces by human tactile afferents," *The Journal of Neuroscience*, vol. 21, pp. 8222-8237, 2001.
- [156] Wbensmith. (2007). *100x light micrograph of Meissner's corpuscle (or tactile corpuscle) at the tip of a dermal papillus*. Available: http://en.wikipedia.org/wiki/File:WVSOM_Meissner%27s_corpuslce.JPG
- [157] J. M. Stellman, *Encyclopaedia of Occupational Health and Safety*: International Labour Organization 1998.
- [158] F. H. Silver, J. W. Freeman, and D. DeVore, "Viscoelastic properties of human skin and processed dermis," *Skin Research and Technology*, vol. 7, pp. 18-23, Feb 2001.
- [159] T. R. Tilleman, M. M. Tilleman, and M. H. A. Neumann, "The elastic properties of cancerous skin: Poisson's ratio and Young's modulus," *The Israel Medical Association journal: IMAJ*, vol. 6, pp. 753-755, 2004.
- [160] R. Aoyagi and T. Yoshida, "Frequency equations of an ultrasonic vibrator for the elastic sensor using a contact impedance method," *Japanese Journal of Applied Physics Part 1-Regular Papers Short Notes & Review Papers*, vol. 43, pp. 3204-3209, May 2004.

References

- [161] S. Kudo, "Sensitivity of frequency change of piezoelectric vibratory tactile sensor using longitudinal-bar-type resonator," *Japanese Journal of Applied Physics Part 1- Regular Papers Brief Communications & Review Papers*, vol. 46, pp. 4704-4708, Jul 2007.
- [162] J. J. Wortman and R. A. Evans, "Youngs modulus shear modulus and poissons ratio in silicon and germanium," *Journal of Applied Physics*, vol. 36, pp. 153-&, 1965.
- [163] V. Kaajakari, T. Mattila, A. Oja, J. Kiihamaki, H. Kattelus, M. Koskenvuori, *et al.*, "Square-extensional mode single-crystal silicon micromechanical RF-resonator," in *TRANSDUCERS, Solid-State Sensors, Actuators and Microsystems, 12th International Conference on, 2003*, 2003, pp. 951-954.
- [164] V. Kaajakari, T. Mattila, A. Qja, J. Kiihamaki, H. Kattelus, M. Koskenvuori, *et al.*, *Square-extensional mode single-crystal silicon micromechanical RF-resonator*, 2003.
- [165] A. T. H. Lin, J. E. Y. Lee, J. Yan, and A. A. Seshia, "Methods for enhanced electrical transduction and characterization of micromechanical resonators," *Sensors and Actuators a-Physical*, vol. 158, pp. 263-272, Mar 2010.
- [166] K. Tanaka, R. Kihara, A. Sanchez-Amores, J. Montserrat, and J. Esteve, "Parasitic effect on silicon MEMS resonator model parameters," *Microelectronic Engineering*, vol. 84, pp. 1363-1368, May-Aug 2007.
- [167] J. E. Y. Lee and A. A. Seshia, "Parasitic feedthrough cancellation techniques for enhanced electrical characterization of electrostatic microresonators," *Sensors and Actuators a-Physical*, vol. 156, pp. 36-42, Nov 2009.
- [168] R. C. Jaeger, J. C. Suhling, and R. Ramani, "Errors associated with the design, calibration and application of piezoresistive stress sensors in (100) silicon," *Components, Packaging, and Manufacturing Technology, Part B: Advanced Packaging, IEEE Transactions on*, vol. 17, pp. 97-107, 1994.
- [169] R. C. Jaeger, J. C. Suhling, M. T. Carey, and R. W. Johnson, "Off-axis sensor rosettes for measurement of the piezoresistive coefficients of silicon," *Components, Hybrids, and Manufacturing Technology, IEEE Transactions on*, vol. 16, pp. 925-931, 1993.
- [170] J. C. Suhling and R. C. Jaeger, "Silicon piezoresistive stress sensors and their application in electronic packaging," *IEEE Sensors Journal*, vol. 1, pp. 14-30, 2001.
- [171] Z. Y. Wang, K. Tian, Y. Z. Zhou, L. Y. Pan, C. H. Hu, and L. T. Liu, "A high-temperature silicon-on-insulator stress sensor," *Journal of Micromechanics and Microengineering*, vol. 18, Apr 2008.