ABSTRACT

KIM, KYUNGRIM. Novel Shear Mode Piezoelectric Sensors and Their Applications. (Under the direction of Dr. Xiaoning Jiang).

Piezoelectric sensors are widely applied in automotive, aerospace, energy, chemical, biomedical and electronic industries due to their high sensitivity, simple reading circuitry, low power consumption, and cost-effectiveness. Particularly, accelerometers for ultra-high temperature applications (> 1000 °C) are of great interest for monitoring of turbine engine and power plant. However, there are no accelerometers which can operate properly at such a high temperature. As another example, tactile sensors are in great demand for various biomedical applications, such as surgical tools and service robotics. However, it has been a challenge for both elasticity and force measurement using existing tactile sensing techniques. Inspired by the sensing technique needs for these applications, my dissertation research focuses on two main innovations: (1) a piezoelectric accelerometer for ultra-high temperature applications (> 1000 °C); and (2) a piezoelectric tactile sensor utilizing the acoustic wave sensing technique for elasticity and force sensing applications.

For high temperature accelerometers, YCa$_4$O(BO$_3$)$_3$ (YCOB) single crystal was identified as a sensing material due to its high temperature stability up to 1500 °C. The accelerometer was designed to operate in shear-mode which can offer the best overall sensor performance at high temperatures, minimizing the thermal transient and base bending effects. A clamping assembly was chosen, rather than an adhesive assembly, to prevent mechanical failures at high temperatures. In addition, the accelerometer was designed to operate without thin film electrodes, which can offer more stable sensing performance than the thin film electrode sensor, because that thin metal films tend to fail at elevated temperatures. The
A prototyped accelerometer was tested at temperatures ranging from room temperature to 1000 °C. The sensitivity of the prototype was measured to be 5.9 ± 0.06 pC/g throughout the tested frequency, temperature and acceleration ranges. The accelerometer retained the same sensitivity at 1000 °C for a dwell time of 9 hours, exhibiting high stability and reliability.

Acoustic wave sensing technique was deployed in the tactile sensor design, where the acoustic load impedance can be sensed by measuring the electrical impedance of the tactile sensor. Firstly, the effect of surface loads on Pb(Mg$_{1/3}$Nb$_{2/3}$)O$_3$-PbTiO$_3$ (PMN-PT) single crystal resonators in different vibration modes including thickness mode, thickness-shear mode, and face-shear mode was investigated. It was observed that the face-shear mode resonator possessed more than one order of magnitude higher sensitivity (ratio of electrical impedance change to acoustic impedance load change) than the other existing resonators. Secondly, using face-shear mode PMN-PT single crystal, a 6 × 6 piezoelectric tactile sensor array was designed, fabricated, and characterized for tissue elasticity and force measurements. Tissue mimicking phantoms with different elastic moduli and acoustic impedances were prepared for the elasticity sensing experiment. For the force measurement test, external forces were applied to the array’s sensing layer whose acoustic impedance varied with applied force. The prototyped sensor array exhibited an elasticity sensitivity of 23.52 Ohm/MPa with a resolution of 4.25 kPa and a force sensitivity of 19.27 Ohm/N with a resolution of 5.19 mN. Finally, the mapping of the phantom’s elastic modulus and applied force using the 6 × 6 tactile sensor array was successfully demonstrated.
Novel Shear Mode Piezoelectric Sensors and Their Applications

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DEDICATION

To my parents and my family who always support me.
Kyungrim Kim was born on July 19, 1983 in Seoul, South Korea. He received his Bachelor of Science in Mechanical and Automotive Engineering from Kook-min University, Korea in 2009. In August 2009, he started his work as a Ph. D. student in the Department of Mechanical & Aerospace Engineering, North Carolina State University. Currently, he is working as a Research Assistant at the Micro/Nano Engineering Laboratory under Dr. Xiaoning Jiang. His main research interests are high temperature piezoelectric sensors and ultrasonic sensors.

The following publications were authored or co-authored by Kyungrim Kim in peer-reviewed journals and conference proceedings during his study at North Carolina State University:
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CHAPTER 1

INTRODUCTION

1.1 Piezoelectricity

1.1.1 Piezoelectric Effect

Piezoelectric effect exists in certain materials such as crystals, ceramics, and biological materials which generate an electrical charge upon mechanical deformation in the material [1]. “Piezo” was derived from an ancient Greek word, “Piezein” which means squeeze or press. In the 1880’s, the piezoelectric effect was first discovered by the Pierre and Jacques Curie brothers [2]. They used various crystals such as tourmaline, zincblende, topaz and quartz, and found the phenomenon that an electrical charge was induced when a mechanical pressure was applied to these crystals. This effect is known as direct piezoelectric effect, as shown in Fig. 1.1(a) [3]. Within a year, Curie brothers also found that the piezoelectric crystals can generate the mechanical force or pressure when the electric field is applied to the crystal. This behavior of crystals is known as reverse or indirect piezoelectric effect, as can be seen in Fig. 1.1(b) [3]. Piezoelectric effect results from the asymmetric crystalline structure of the crystals. In this structure, the ions can only move along one axis in the crystal. When a mechanical stress is applied to the crystal, the opposite charges are created on each side of the crystal which generates the voltage across the crystal or vice versa [4]. The direct piezoelectric effect is widely used for sensing the dynamic changes in pressure, acceleration, and force or for energy harvesting applications. On the other hand, the reverse piezoelectric effect is utilized in actuation applications such as piezoelectric motors or in sound generation applications such as ultrasound transducers and resonator sensors [5].
1.1.2 Piezoelectric Constitutive Equation

The Curie brothers found that the electrical displacement generated from certain materials (quartz, tourmaline and zincblende) was proportional to the applied stress. This direct piezoelectric effect can be expressed as follows [6, 7]:

\[ D = dT \]  

(1.1)

where \( D \) is the electrical displacement, \( d \) is the piezoelectric strain coefficient and \( T \) is the applied stress. On the other hand, the reverse piezoelectric effect is related to the ratio between the generated strain and the applied electric field and can be expressed by:

\[ S = dE \]  

(1.2)

where \( S \) is the strain induced by the reverse piezoelectric effect and \( E \) is the magnitude of the applied electric field. These effects can be modified using the elastic properties of the material:

\[ D = dT = cdS = eS \]  

(1.3)

and

\[ T = cS = cdE = eE \]  

(1.4)

Here, \( c \) is the elastic modulus, \( s \) is the compliance coefficient (\( S = sT \)), and \( e \) is the piezoelectric stress constant. When the strain is applied to the piezoelectric material, it generates the elastic stress which is related to the mechanical strain. In addition, the strain generates the piezoelectric polarization \((D = eS)\) at the same time. The additional electric field is produced in the material from this polarization as following:

\[ E = \frac{D}{\varepsilon} = \frac{eS}{\varepsilon} \]  

(1.5)
where $\varepsilon$ is the permittivity of the material. The additional internal stress, which is against the deformation of materials, is produced by this electric field (i.e. $T = -eE$). As a result, the applied strain generates two stresses and the total stress can be modified:

$$T = -eE + c^E S$$

(1.6)

where the superscript $E$ refers to the constant electric field.

Similarly, the modified electric displacement can be obtained. When the electric field is applied to the piezoelectric material, the electric displacement is produced ($D = \varepsilon E$). The applied electric field also produces the additional strain ($S = dE$) and this strain generates the additional polarization of piezoelectric material ($D = eS$). Thus, when the electric field is constant, the total electrical displacement is:

$$D = \varepsilon^S E + eS$$

(1.7)

where the superscript $S$ refers to the constant strain. These piezoelectric constitutive equations can be described by three different ways using the relationship between the mechanical variables and the electrical variables (shown in Fig. 1.1) as follows:

$$D = \varepsilon^T E + dT$$

$$S = dE + s^E T$$

$$E = \beta^T E - gT$$

$$S = gD + s^D T$$

(1.8)

$$E = \beta^S D - hS$$

$$T = -hD + c^D S$$

(1.9)

(1.10)

The symbols’ meaning and their units are summarized in Table 1.1.
1.1.3 Piezoelectric Constants

Important constants including the frequency constant, electromechanical coupling coefficient, elastic constants, piezoelectric coefficients and mechanical quality factor can be calculated based on the resonance and anti-resonance frequencies for various electromechanical applications.

Frequency constants are one of the important parameters which determine the piezoelectric element dimensions to obtain the desired operational frequency. They are directly related to the resonance frequency, anti-resonance frequency, and the piezoelectric element’s dimensions known as the frequency determining factor. The frequency constants \( N \) can be calculated using the following equation [8]:

\[
N = f \times l
\]

where \( f \) and \( l \) are the resonance \( (f_r) \) or anti-resonance frequency \( (f_a) \), and the dimension of the crystal which determine the resonance and anti-resonance frequency, respectively. The electromechanical coupling coefficient implies the efficiency of the converting ratio between electric and mechanical energy. This factor determines the bandwidth of resonant devices, which affects the device performance significantly. Thus, the higher electromechanical coupling coefficient is usually favorable for efficient energy conversion. The electromechanical coupling is related to other piezoelectric coefficients as follows:

\[
k^2 = \frac{d^2}{\varepsilon^T s^E} = \frac{e^2}{\varepsilon^S c^D} = \frac{g^2}{\beta^T s^E} = \frac{h^2}{\beta^S c^D}
\]  

(1.12)

The electromechanical coupling coefficient \( (k) \) can be calculated using the resonance and anti-resonance frequencies of the piezoelectric element:
\[ k^2 = \frac{\pi f_r}{2 f_a} \cot \left( \frac{\pi f_r}{2 f_a} \right) \] (1.13)

where \( f_r \) is resonance frequency and \( f_a \) is anti-resonance frequency. The elastic compliance is directly related to the density and the sound velocity. This constant is given by [8]:

\[ s^E = \frac{1}{\rho v^2} \] (1.14)

Here, \( \rho \) is the density and \( v \) is the sound velocity of the piezoelectric element. The elastic compliance \( s^E \) also can be calculated using the density \( (\rho) \), element dimension \( (t) \), anti-resonance frequency \( (f_a) \) and electromechanical coupling coefficient \( (k) \) as follows:

\[ s^E = \frac{1}{4 \rho t^2 f_a^2 (1 - k^2)} \] (1.15)

where The piezoelectric strain constant \( (d) \) indicates the generated electric charge per unit of the applied mechanical stress or the achievable mechanical strain of piezoelectric element per unit of applied electric field. This constant is an important indicator especially for strain-dependent applications since the amount of induced strain of piezoelectric material is directly related to the piezoelectric strain constant. The piezoelectric strain constant can be calculated by [9]:

\[ d^2 = k^2 \times s^E \times e^T \] (1.16)

On the other hand, the piezoelectric voltage constant \( (g) \) can be defined by the ratio of the generated electric field to the applied mechanical stress or the ratio of the induced mechanical strain to the applied electrical displacement. The piezoelectric voltage constant is [9]:

\[ g = \frac{d^2}{e^T} \]
\[ g = \frac{d}{e^t} \]  

(1.17)

The mechanical quality factor \( Q_M \) refers to the power loss over the energy stored in the device, which is directly related to the sharpness of the resonance of piezoelectric element. For the resonator applications, the high quality factor is especially important because \( Q_M \) determines the amount of the power dissipation of the resonator while the electrical quality factor (inverse of dielectric loss) is important for non-resonant devices. The mechanical quality factor can be determined by the resonant frequency and the 3 dB bandwidth of the admittance as follows [8]:

\[ Q_M = \frac{f_r}{\Delta f} \]  

(1.18)

where \( \Delta f \) is the frequency difference at 3 dB of the maximum admittance.

1.2 Piezoelectric Device

1.2.1 Piezoelectric Sensing Device and Vibration Mode

Different types of piezoelectric sensing devices have been researched in the past few decades and used in various fields such as aerospace, automotive, chemical and bio-medical industries [10, 11]. The piezoelectric sensing devices can be divided into two categories, the non-resonant sensing and the resonant sensing, as shown in Fig. 1.3. The non-resonant method is an direct way since the device can generate electric charge or voltage directly responding to environmental input stimulations. This type of sensor utilizes the direct piezoelectric effect and is widely used as electromechanical sensors such as force, pressure, vibration sensors [12]. On the other hand, the resonant sensors are used to measure the input
response indirectly from the change in the output signal such as the frequency shift or the sound velocity change [13]. This type of sensor also requires the voltage source to oscillate the resonator at the specific frequency. The resonator sensors are widely used as the chemical and bio sensors due to their high sensitivity [14].

In order to design the piezoelectric devices for desired applications, specific vibration mode needs to be considered. The proper vibration mode can be selected considering the device’s operational frequency, the input and output types, device applications and its structural configuration requirements. In Table 1.2, the mode shapes, frequency range and application for various vibration modes are summarized.

1.2.2 Non-Resonant Piezoelectric Device

Non-resonant piezoelectric sensors use the direct piezoelectric effect. This type of sensors have advantages of small size, light weight, broad frequency range, simple signal conditioning, cost effective. Non-resonant piezoelectric accelerometers, vibration sensors and shock sensors are applied in a broad range of industries. For example, the piezoelectric accelerometer, which is the most popular piezo sensor and is introduced in detail in Chapter 2, is widely used in structural health monitoring to monitor acceleration or to detect the abnormal behavior of a structure. The accelerometer usually consists of a piezoelectric sensing element and a seismic mass. The seismic mass is the cause of the force under the influence of external accelerations and this force leads to the deflection of the piezoelectric material from its neutral position. The deflection of piezoelectric element generates an electric charge signal which is proportional to the acceleration. As a result, the acceleration
can be detected by measuring the output signal based on the direct piezoelectric effect. The piezoelectric accelerometer is usually configured with three types of vibration modes such as compression, shear, and flexural mode, as shown in Table 1.3. Each mode provides certain benefits in sensing performance, which can make the selected mode device more appropriate for the specific application than other mode devices.

The compression mode provides the merits of a simple design and high stiffness which leads to the accelerometer’s high natural frequency and broad usable frequency range. However, the compression mode can suffer from high susceptibility to base strain effects and thermal transient effects. In this design, the piezoelectric element is directly connected to the base of the sensor housing. Thus, the strain from the bending or thermal transient effects can produce undesired output signal, which affects the sensitivity of the accelerometer. For these reasons, the compression design is not desired for high temperature applications [15].

The shear mode accelerometer is the most common type of accelerometer which provides the best overall sensor performance. In this design, the piezoelectric crystal is clamped between a center post and seismic masses. The shear design offers the minimized base strain and thermal transient effects by isolating the piezoelectric element from the base. This design also exhibits a high operational frequency and a rigid structure compared to the flexural design.

Flexural designs utilize rectangular or disc shaped piezoelectric plates. The piezoelectric material’s own mass or additional weight bonded to the material can generate the bending of the material under the external acceleration. This type of device has the
benefit of a high output signal, low manufacturing cost, and light weight. However, the use of flexural mode accelerometers are limited by a narrow frequency range due to its low stiffness, compared to compression and shear designs. They also suffer from the low over-shock survivability and the effect from the thermal transient. This sensor is usually applied to applications that require the low profile of device.

1.2.3 Resonant Piezoelectric Device

Resonant piezoelectric devices such as acoustic wave sensors are very sensitive, rugged, reliable, and cost effective devices [16]. The acoustic wave commonly implies that the wave has a frequency well above the audible range, while the term acoustic cover a broad frequency range from $10^{-2}$ Hz to $10^{12}$ Hz [17]. Acoustic wave sensors including pressure, biochemical, vapor, humidity and temperature sensors are widely used in many applications such as automotive, medical and other industrial applications [18]. This type of sensors utilizes high frequency mechanical vibration based on the inverse piezoelectric effect. When the oscillating electric field is applied to the acoustic wave sensor, a synchronous mechanical deformation occurs in the piezoelectric element, which creates a mechanical or an acoustic wave. The created acoustic wave propagates through or on the surface of the piezoelectric element [17].

Various input stimuli such as a physical, chemical, and biological stimulus affect the characteristics of the propagation path. This small change in wave characteristics leads to the variations in the velocity and/or amplitude of the wave. Therefore, the physical quantity of the input stimulus can be expected by measuring the frequency or phase characteristics of the
acoustic wave sensor. This physical input can be pressure, temperature, or stress changes and the chemical input includes a certain gas concentration or the presence of a chemical agent. For the biological stimulus, it contains the concentration of different bacteria and the presence of a biological agent. Typically, the acoustic wave sensor can be divided into two types including the bulk acoustic wave (BAW) sensors and the surface acoustic (SAW) sensors [13].

In BAW sensors, the generated acoustic wave propagates through the bulk piezoelectric material. The most popular BAW sensor is an AT cut thickness-shear mode (TSM) sensor. The TSM is also known as a piezoelectric resonator since the device resonates creating electromechanical standing waves [16]. In TSMs, the resonator vibrates in the thickness mode and the particle displacement is maximized at the resonator faces. This makes the resonator highly sensitive to the surface loading. Furthermore, since there is no particle displacement in the vertical direction, this shear wave device can be suitable for liquid-based applications such as chemical or bio-sensors [16].

On the other hand, SAW sensors consist of a piezoelectric substrate, an input interdigitated transducer (IDT) and an output interdigitated transducer patterned on the substrate [19]. The electric field is applied to the input IDT to generate the acoustic wave and the generated wave propagates on the surface of the substrate between the IDTs known as the delay-line. Unlike TSM devices, the acoustic wave generated by the IDT is a surface-normal wave which can be completely damped out when the sensor is contacted by liquids. Thus, the SAW sensor is not desired for liquid sensing applications [18].
Shear horizontal acoustic plate mode (SH-APM) sensors use the combined design which can take advantage of both TSM and SAW. Since there is no surface-normal displacement in this type of sensor, it can be possible to use for liquid applications. Furthermore, the SH-APM sensor usually exhibits higher sensitivity compared to that of the TSM sensor [16]. The comparison of each acoustic wave sensor is summarized in Table 1.4.

1.3 Dissertation Outline

The main goal of this research is to investigate the design, fabrication, and testing methods of novel piezoelectric sensors for specific applications including high temperature acceleration sensing and acoustic wave tactile sensing. The dissertation consists of five chapters and references. Each chapter is briefly described as follows.

Chapter 1 reviews the fundamentals of piezoelectricity and piezoelectric sensing techniques, derivation of piezoelectric constitutive equations, and the concept of important piezoelectric constants. Chapter 2 presents a piezoelectric accelerometer using thickness-shear mode YCOB single crystal for ultra-high temperature applications (> 1000 °C). The state-of-the-art of high temperature sensing technologies is first reviewed, followed by discussions on their limitations, requirements and an overview of the project objectives. An analytical model of the dynamic response and thermal expansion of the prototyped accelerometer is next presented in this chapter. Furthermore, the sensor assembly design and experimental methods for high temperature sensor fabrication and tests are presented. Finally, the results for the analytical model, room and high temperature sensor tests are...
presented and discussed in detail. Chapter 3 introduces, for the first time, a surface acoustic load sensing technique using a piezoelectric resonator. Various piezoelectric resonators and their applications are reviewed and the motivation and objective of the project are presented. More importantly, an acoustic load sensing model for the piezoelectric resonator is derived using the KLM equivalent circuit model to compare resonators in different modes including thickness, thickness-shear, and face-shear mode. Experimental methods for fabrication and characterization of bulk piezoelectric resonators in three different modes and an $8 \times 8$ sensor array are next presented. Then, the modeling results and experimental results of each piezoelectric resonator are analyzed and discussed in order to verify the acoustic load sensitivity for tactile sensing applications. Chapter 4 investigates an acoustic wave tactile sensor array using the face-shear mode crystal for tissue elasticity and force sensing based on the acoustic load sensing technique introduced in Chapter 3. This chapter first reviews the current tactile sensing technologies and their applications including minimally invasive surgery. Also presented is a detailed analytical model for tissue elasticity sensing and force sensing utilizing the proposed acoustic wave sensing technique. A $6 \times 6$ tactile sensor array, which can measure elasticity and force effectively, are designed, fabricated and integrated with a switch and controller circuit. The experimental method for both elasticity sensing using tissue mimicking phantoms and force sensing are then designed. Finally, experimental results for both the tissue elasticity sensing test and force sensing test are analyzed and discussed. Chapter 5 summarizes the major conclusions of this dissertation, and recommendations for further research.
Figure 1.1 Working principles of (a) the direct piezoelectric effect and (b) the reverse piezoelectric effect.
Figure 1.2 The relationship between the mechanical variables and the electrical variables.
Figure 1.3 The piezoelectric sensing devices with the non-resonant method and the resonant method.
Table 1.1 Symbols and units for various constants.

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stress</td>
<td>T</td>
</tr>
<tr>
<td>Strain</td>
<td>S</td>
</tr>
<tr>
<td>Electric field strength</td>
<td>E</td>
</tr>
<tr>
<td>Electric displacement</td>
<td>D</td>
</tr>
<tr>
<td>Elastic compliance</td>
<td>s</td>
</tr>
<tr>
<td>Elastic stiffness</td>
<td>c</td>
</tr>
<tr>
<td>Permittivity</td>
<td>ε</td>
</tr>
<tr>
<td>Dielectric impermeability</td>
<td>β</td>
</tr>
<tr>
<td>Piezoelectric strain constant</td>
<td>d</td>
</tr>
<tr>
<td>Piezoelectric stress constant</td>
<td>e</td>
</tr>
<tr>
<td>Piezoelectric strain constant</td>
<td>g</td>
</tr>
<tr>
<td>Piezoelectric stress constant</td>
<td>h</td>
</tr>
<tr>
<td>Electromechanical coupling</td>
<td>k</td>
</tr>
<tr>
<td>Mechanical quality factor</td>
<td>Q_M</td>
</tr>
<tr>
<td>Electrical quality factor</td>
<td>Q_E</td>
</tr>
<tr>
<td>Frequency constant</td>
<td>N</td>
</tr>
<tr>
<td>Elastic velocity</td>
<td>ν</td>
</tr>
<tr>
<td>Density</td>
<td>ρ</td>
</tr>
</tbody>
</table>
Table 1.2 Various vibration modes and their properties.

<table>
<thead>
<tr>
<th>Vibration mode</th>
<th>Shape</th>
<th>Frequency (Hz)</th>
<th>Application</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>1k  10k  100k</td>
<td></td>
</tr>
<tr>
<td>Flexural mode</td>
<td></td>
<td>1M  10M  100M</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>1G</td>
<td></td>
</tr>
<tr>
<td>Length mode</td>
<td></td>
<td></td>
<td>kHz filter</td>
</tr>
<tr>
<td>Contour or face-</td>
<td></td>
<td></td>
<td>kHz resonator</td>
</tr>
<tr>
<td>shear mode</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thickness-shear</td>
<td></td>
<td></td>
<td>MHz filter</td>
</tr>
<tr>
<td>mode</td>
<td></td>
<td></td>
<td>MHz resonator</td>
</tr>
<tr>
<td>Thickness mode</td>
<td></td>
<td></td>
<td>MHz resonator</td>
</tr>
<tr>
<td>Surface acoustic</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>wave</td>
<td></td>
<td></td>
<td>SAW filter</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>SAW resonator</td>
</tr>
</tbody>
</table>
Table 1.3 The piezoelectric accelerometer with different vibration modes including compression, shear, and flexural mode.

<table>
<thead>
<tr>
<th></th>
<th>Compression design</th>
<th>Shear design</th>
<th>Flexural design</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Structure</strong></td>
<td><img src="image1" alt="Structure Image" /></td>
<td><img src="image2" alt="Structure Image" /></td>
<td><img src="image3" alt="Structure Image" /></td>
</tr>
<tr>
<td><strong>Advantage</strong></td>
<td>High rigidity;</td>
<td>Wide frequency range;</td>
<td>High output signal;</td>
</tr>
<tr>
<td></td>
<td>High operational</td>
<td>Low off axis sensitivity;</td>
<td>Low profile;</td>
</tr>
<tr>
<td></td>
<td>frequency;</td>
<td></td>
<td>Low manufacturing cost;</td>
</tr>
<tr>
<td></td>
<td>Broad useable</td>
<td>Less interruption from base</td>
<td>Light weight;</td>
</tr>
<tr>
<td></td>
<td>frequency range;</td>
<td>strain and thermal inputs;</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Simple structure;</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Small size</td>
<td></td>
</tr>
<tr>
<td><strong>Disadvantage</strong></td>
<td>High susceptibility to base strain effects and thermal transient effects</td>
<td>Less rigidity than compression mode;</td>
<td>Narrow frequency range;</td>
</tr>
<tr>
<td></td>
<td>Less sensitivity than flexural mode;</td>
<td></td>
<td>Low over-shock survivability;</td>
</tr>
</tbody>
</table>
Table 1.4 Different types of acoustic wave sensors.

<table>
<thead>
<tr>
<th>Structure</th>
<th>TSM</th>
<th>SAW</th>
<th>SH-APM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensitivity</td>
<td>Low</td>
<td>High</td>
<td>Low-Med</td>
</tr>
<tr>
<td>Sensitivity factors</td>
<td>Plate thickness</td>
<td>IDT finger spacing</td>
<td>Plate thickness, IDT finger spacing</td>
</tr>
<tr>
<td>Surface motion</td>
<td>Transverse</td>
<td>Normal and Transverse</td>
<td>Transverse</td>
</tr>
<tr>
<td>Medium</td>
<td>Gas and liquid</td>
<td>Gas</td>
<td>Gas and liquid</td>
</tr>
<tr>
<td>Operating frequency</td>
<td>Low (5-20 MHz)</td>
<td>High (30-500 MHz)</td>
<td>Med-High (25-200 MHz)</td>
</tr>
<tr>
<td>Mechanical strength</td>
<td>Med</td>
<td>High</td>
<td>Med</td>
</tr>
</tbody>
</table>
CHAPTER 2

HIGH TEMPERATURE PIEZOELECTRIC ACCELEROMETER

2.1 Background

2.1.1 High Temperature Sensing Applications

Sensing technologies for use in ultra-high temperatures (>1000 °C) are in great demand, particularly in the automotive, aerospace, and energy industries. In the aerospace propulsion system, high temperature (HT) sensors are necessary for intelligent propulsion system design, operation and for enhancement of system maintenance and safety [20-22]. These sensors need to be placed close to jet engines because of the reliability and noise requirements. Specifically, the sensor should be able to monitor propulsion component conditions, and these data will then be analyzed so that operating parameters of propulsion system can be optimized for operations under temperatures of 500 – 1000 °C, with lifetime up to 100000 hours [23]. However, conventional silicon-based and piezoelectric-based sensors are not able to function at such high temperatures, and thus, sensor devices are usually located in controlled environment [21]. This limitation leads to decreased fuel efficiency and reduced reliability. In space exploration, HT devices are also essentially required. For example, there are no commercially available piezoelectric devices that can operate in Venus due to its harsh environments including high temperatures (460 °C), high pressure (9 MPa) and corrosiveness [21, 24].

HT sensors are essential parts of automotive combustion systems for recording the system temperature, pressure, and vibration to improve the efficiency and reliability of internal combustion engines [23, 25]. As an example, combustion sensors or knock sensors
need to be located as close as possible to the high temperature source (e.g. combustion engine) for reliable monitoring of engine parameters [26, 27]. These sensors are usually required to work properly at high temperatures greater than 1000 °C with vibration sweeps up to 10 g for 30 hours [25]. Abnormal combustion such as a knock can cause decreasing of the engine efficiency, compression ratio and power output of internal combustion engines. Knocks also lead to damages of engine parts and uncomfortable metallic noises. Furthermore, exhaust pollution can be significantly increased by incomplete combustion. For these reasons, early detection of knock using HT sensors is necessary to resolve both economic and environmental concerns. The most common type of knock sensor is a vibration sensor or an accelerometer which can be directly screwed to the cylinder head gasket or engine block. Knocks can be detected by monitoring the vibrations in engine structures, generated by the abnormal combustion [20, 28-30].

Besides, various industries including energy and manufacturing industries very often require high temperature sensing technique. Nuclear power industries use ultrasonic transducers at high temperatures for various non-destructive testing (NDT) and nondestructive evaluation (NDE) applications. High temperature ultrasonic technique is an effective way to inspect the internal state of materials or structures in processing [24]. For example, in industrial plants, ultrasonic NDT of steel components is usually performed at temperatures up to 400°C. Ultrasonic monitoring can also be used to characterize composites and engineering materials during high temperature manufacturing such as polymer extrusion and curing processes of graphite/epoxy composites. In addition, high temperature ultrasonic
technique is used for ultrasound Doppler velocimetry of hot melt flows to monitor molten metal properties [24].

2.1.2 Review of Current Technologies for High Temperature Sensors

For high temperature applications, various types of sensors such as piezoresistive, capacitive, fiber optic and piezoelectric sensors have been researched [8, 31, 32]. Piezoresistive sensors are made of metal strain gauges or piezoresistive silicon structures. The resistive material is usually bonded to flexural structures, such as cantilever beams, which can be bended by an applied external force, pressure or acceleration. This bending can generate deformation of the resistive material configured with a Wheatstone bridge circuit, resulting in a change in output voltage. Compared to capacitive sensors, the piezoresistive sensors are less susceptible to electromagnetic interference (EMI). However, this type of sensors can suffer from the temperature dependence of the resistivity for ultra-high temperature applications [33-35].

Capacitive sensors usually consist of two separated capacitor plates. The distance between the plates changes according to the applied inputs such as force, pressure or acceleration. The capacitance of the sensor is related to the distance of two plates, and thus, the input change can be sensed by measuring capacitance of two plates. Capacitive sensors have the advantage of low thermal drift, high resolution and good noise performance. However, this type of sensor is not usually robust enough and it could suffer from the influence of parasitic capacitance which can have similar value as capacitance of the sensors [36-39].
For high temperature application, fiber optic sensor is widely used because of its immunity to electromagnetic interference and high operating temperature (< 2000 °C). For example, optical sensors made of single-crystal sapphire fiber have gained great attention due to their high melting point (2040 °C), robustness and chemical corrosion resistance. This type of sensor can be used at temperature up to 2000 °C without any high temperature degradation. However, challenges still remain in fiber optic sensors including a complicated fabrication process, and an expensive and complex signal processing system [40-44]. Important features for different types of HT sensor were summarized in Table 2.1.

2.1.3 HT Piezoelectric Sensors and Piezoelectric Materials

For high temperature applications, several piezoelectric materials have been extensively researched including quartz (SiO₂), lithium niobate (LiNbO₃, LN), gallium orthophosphate (GaPO₄), langasite (LGS) and aluminum nitride (AIN) [45]. The most popular HT piezoelectric material, Quartz, possesses high electrical resistivity (> 10¹⁷ Ω at room temperature), high mechanical quality factor, and excellent high temperature stability. Nevertheless, its low electromechanical and piezoelectric coefficients, high losses above about 450 °C, and α to β phase transition temperature at 573 °C limit the use of quartz for high temperature applications [23, 46]. LiNbO₃ has been reported that its Curie temperature is about 1100 °C with high electromechanical coefficients. However, practically, this material can only be used up to 600°C due to chemical decomposition (starts at 300 °C), increased attenuation and intergrowth transition, loss of oxygen to the environment, and limited resistivity. To make things worse, this material suffers from short lifetime, 10 days at
400 °C and 0.1 days at 450 °C due to its decomposition [47-49]. GaPO₄ shows similar features of quartz such as high electrical resistivity and mechanical quality factor, but it also exhibits high electromechanical coupling and greater piezoelectric sensitivity until α to β phase transition occurs (< 970 °C). However, decreased mechanical quality factor due to increased structural disorder at temperatures above 700 °C limits its usage [45, 50]. Langasite crystal, which has no phase transition prior to its melting point (1470 °C), also has been extensively studied for high temperature applications. However, the sensing performance of langasite sensors can be limited due to its low electrical resistivity and low quality factor at elevated temperatures resulted from oxygen ion transport and diffusion in the lattice [51-53]. Another popular HT piezoelectric material is aluminum nitride (AlN). AlN is a non-ferroelectric material which is intrinsically poled and its Curie temperature has not been reported yet. It has been reported that an AlN thin film can be used for high frequency ultrasonic transducer and can maintain its piezoelectric properties up to 1150 °C. However, it is difficult to get good quality and large size bulk AlN piezoelectric material [24, 54, 55].

Recently, rare earth calcium oxyborate single crystals ReCa₄O(BO₃)₃ (ReCOB, Re: rare earth elements such as Gd, La, and Y) has gained special attention for ultra-high temperature applications. Especially, YCa₄O(BO₃)₃ (YCOB) is known as a promising candidate for HT sensors due to its stable piezoelectric properties and no phase transformation before the melting temperature of 1500 °C as well as exceptional high resistivity [22, 56]. It has been reported that the limitation of YCOBs at elevated temperatures (> 1000 °C) was only the degradation of the platinum thin film electrodes
which was used during the test. The important properties of each HT piezoelectric material are shown in Table 2.2.

2.1.4 Thin Film Electrodes for High Temperature Application

As mentioned in the previous section 2.1.3, thin film electrodes (≈100 nm thickness) are usually needed for piezoelectric devices to apply the electric field effectively or to obtain generated charge signal. These thin film electrodes should be capable of withstanding high temperature operation for high temperature applications [57, 58]. Various thin film electrodes, including Pt, Pt-based alloys and other metallic alloys, have been researched for high temperature applications.

Platinum thin film electrodes are most widely used for both low and high temperature piezoelectric devices due to their high melting point and their oxidation resistance as well as excellent electrical properties [59]. Nevertheless, there are some limitations of the use at temperatures higher than 600 °C because of platinum thin film’s recrystallization and dewetting effects resulting in loss of electrical continuity and device failure [60]. Pt-based alloys such as Pt-Rh, Pt-Ir and Pt-Zr electrodes have been reported that they can be used without the surface degradation for temperature up to 750 °C [61].

Recently, Pt-Rh/ZrO₂ thin film, Pt/Rh alloy co-deposited with ZrO₂, has been developed for high temperature application and this electrode showed stable and consistent performance up to 850 °C. When the protective SiAlON ceramic coatings were applied to the device, improved electrode performance was obtained [57, 60, 62]. Ir based electrodes such
as Ir/TiAlN and IrO₂/Ti also have been studied for HT electrodes but their operating temperature was limited up to only 700 °C [63, 64].

Therefore, high temperature sensors without thin film electrodes are attractive for reliable high temperature applications since the use of thin film electrodes mentioned above is limited by the degradation at temperatures above 850 °C. Table 2.3 summarizes the important properties of thin film electrodes.

2.2 High Temperature Shear Mode Piezoelectric Accelerometer

2.2.1 Motivation and Objective

Accelerometers are sensing devices which can measure structure’s acceleration which have a lot of applications in various industry and science fields. To monitor vibration in internal combustion engines, jet engines, and rotating machinery, accelerometers are used. Navigation technologies for air vehicles and various missile control systems also require highly sensitive accelerometers. In addition, micro-machined accelerometers are essential components for small electronic devices such as smart phones, tablet computers, and video game controllers in order to sense the motion or orientation of devices. Besides, single-axis and multi-axis accelerometers can be used to detect device’s acceleration changes induced by shocks and falling in a resistive medium. On the other hand, for high temperature applications, the accelerometer featured with high operational temperature, simple structures, high sensitivity and stability, low power consumption and long life time is great demand in various industrial fields [27]. For this purpose, piezoelectric accelerometers for ultra-high temperature application (> 1000 °C) are of particular interest because there are no
commercially available HT accelerometers which can operate properly at such high temperature [65]. In this chapter, the design, fabrication and testing of a shear mode YCOB high temperature vibration sensor were studied. The proposed accelerometer was specially designed to withstand ultra-high temperature. The prototyped accelerometer was successfully tested at the temperature ranging from room temperature to 1000 °C and at frequency ranging from 50 Hz to 350 Hz. The sensor stability and reliability was also measured for dwelling test at 1000 °C.

2.2.2 Proposed High Temperature Accelerometer

The proposed accelerometer was specially designed for ultra-high temperature application (> 1000 °C). Fig. 2.1 shows the schematic of a shear-mode piezoelectric accelerometer design. Firstly, this piezoelectric sensor used YCOB single crystal as a sensing material to take advantage of its high temperature stability of up to 1500 °C. It has already been reported that YCOBs can be used at high temperatures up to 1000 °C [50, 66]. Furthermore, this accelerometer was designed to operate in shear-mode which can offer the best overall sensor performance for high temperature application when compared to other types of accelerometers. In shear-mode design, the sensing crystals, YCOBs, are clamped or bonded between a center post and seismic masses. Under acceleration, a charge signal is generated by a shear stress that occurs between the seismic mass and the sensing crystal. The crystals are able to be separated from the base and housing, and that can leads to reduced thermal transient and base bending effects at high temperature. More importantly, the accelerometer was designed to operate without thin film electrodes such as Pt or Pt based alloy films. This
electrodeless design can guarantee stable performance of the accelerometer without electrode degradation at elevated temperatures compared to devices with thin film electrodes [67-69].

2.3 Analytical Modeling

2.3.1 Dynamic Response Model

In order to obtain the dynamic response model, the accelerometer system can be divided into two systems, a mechanical and electrical system [70]. Firstly, the mechanical system can be expressed by the differential equation derived from the free body diagram of the accelerometer, as shown in Fig. 2.2:

\[ m\ddot{x}_i = m\ddot{x}_o + c\dot{x}_o + kx_o \]  \hspace{1cm} (2.1)

where \( m \) is the seismic mass, \( c \) is the damping coefficient, \( k \) is the spring constant, \( x_i \) is the shaker displacement, \( x_o \) is the relative displacement between the seismic masses and the shaker. To solve Eq. (2.1), the initial condition can be assumed by:

\[ x_o(0) = \dot{x}_o(0) = 0 \]  \hspace{1cm} (2.2)

By substituting the initial condition into Eq. (2.1) and using the Laplace transformation, the mechanical transfer function, \( H_m \), can be obtained. The transfer function represents the relationship between the input acceleration and the relative displacement:

\[ H_m(s) = \frac{x_o(s)}{\dot{x}_i(s)} = \frac{m}{k} \frac{\omega_n^2}{s^2 + 2\xi \omega_n s + \omega_n^2} \]  \hspace{1cm} (2.3)
Here, $\omega_n$ is the resonant frequency of the accelerometer and $\xi$ is the damping ratio of the sensor structure. The resonant frequency ($f_n$) of the accelerometer can be calculated using the spring constant and the seismic mass:

$$\omega_n = \sqrt{\frac{k}{m}}$$  \hspace{1cm} (2.4)

$$f_n = \frac{1}{2\pi} \sqrt{\frac{k}{m}}$$  \hspace{1cm} (2.5)

In electrical system, the charge output ($Q$) is a function of the relative displacement:

$$Q = k_q \dot{x}_o$$  \hspace{1cm} (2.6)

Here, $k_q$ is the charge output of unit displacement and can be expressed by:

$$k_q = \frac{AG}{l} d_{26}$$  \hspace{1cm} (2.7)

where $A$ is the area of the crystal, $G$ is the shear modulus and $l$ is the length of the crystal, respectively. Fig 2.3(a) and (b) show the equivalent circuit of the accelerometer system and equivalent resistance ($R$) and capacitance ($C$) can be obtained as follows:

$$R = \frac{R_{\text{amp}} R_{\text{leak}}}{R_{\text{amp}} + R_{\text{leak}}} \cong R_{\text{amp}}$$  \hspace{1cm} (2.8)

$$C = C_{\text{crys}} + C_{\text{wire}} + C_{\text{amp}}$$  \hspace{1cm} (2.9)

From Eqs. (2.6-2.9), the governing equation of the electric system can be derived:

$$\dot{\dot{x}}_o = \frac{k_q}{C} \dot{x}_o - \frac{v_o}{\tau}$$  \hspace{1cm} (2.10)

where $R_{\text{amp}}$ and $R_{\text{leak}}$ represent resistance of the amplifier and crystal, $C_{\text{crys}}$, $C_{\text{wire}}$ and $C_{\text{amp}}$ are capacitance of the crystals, wires and the amplifier, respectively. $v_o$ is output voltage from the
sensor and $\tau$ is RC time constant. The electrical transfer function ($H_e$) can be obtained by applying the Laplace transformation to Eq. (2.10).

$$H_e(s) = \frac{v_o(s)}{x_o(s)} = \frac{k_q}{C} \frac{\tau s}{\tau s + 1}$$  \hspace{1cm} (2.11)

The transfer function represents the relationship between the relative displacement and output voltage. Finally, the completed dynamic response model of the accelerometer can be obtained by merging the mechanical transfer function ($H_m$) and the electrical transfer function ($H_e$) as follows:

$$\frac{v_o(s)}{\dot{x}_i(s)} = S_T \frac{\tau s}{\tau s + 1} \frac{\omega_n^2}{s^2 + 2\xi\omega_n s + \omega_n^2}$$  \hspace{1cm} (2.12)

By substituting the frequency operator $j\omega$ for $s$ in Eq. (2.12), the magnitude ratio can be obtained. Eq. (2.12) can be simply modified to approximate the voltage sensitivity ($S_T$) and the charge sensitivity ($S_Q$) at low frequencies since $\omega >> 1/\tau$ and $\omega_h >> \omega$.

$$S_T = \frac{k_q m}{C k}$$  \hspace{1cm} (2.13)

$$S_Q = S_T \times C = k_q m$$  \hspace{1cm} (2.14)

2.3.2 Thermal Expansion Model

As mentioned in section 2.2.1, the thermal expansion issue of HT sensors is one of the most critical problem which limits sensors operational temperature. Thus, allowable temperature and thermal expansion effect of each component should be considered when the sensor is designed to achieve stable and reliable HT sensor performance. In the proposed sensor design,
suitable materials, whose allowable temperatures were well above the tested temperature range (25–1000 °C), were chosen as can be seen in Table 2.4. The coefficient of thermal expansion (CTE) of YCOB crystals, $\alpha_c (K^{-1})$, can be calculated using the following equation [71]:

$$\alpha_c(T) = 1.13 \times 10^{-8} \times T + 9.18 \times 10^{-6}$$ (2.15)

Similarly, the thermal expansion coefficient of Inconel 601, $\alpha_i (°C^{-1})$, can be calculated using the following equation obtained from the data sheet (Special Metals, Corp.):

$$\alpha_i(T) = 4.33 \times 10^{-8} \times T + 13.75 \times 10^{-6}$$ (2.16)

All sensor components underwent thermal expansion at the same time when the sensor was exposed to the high temperature environment. The amount of thermal expansion depends on the materials thermal expansion coefficient (Inconel and YCOB crystal in this case). This difference in thermal expansion leads to undesirable compression or micro-gaps between connected parts, which can be the reason of sensor malfunction at the specific temperature. Thermal expansion ($\Delta L$) of sensor components along the axis of the bolt was calculated using following equation:

$$\Delta L = \alpha \times \Delta T \times L$$ (2.17)

On the other hand, by applying axial force ($F$) to the bolt, the thermal expansion difference between Inconel and YCOB crystal can be compensated. Optimized torque can be calculated using the follows:

$$T_b = k_n \times d \times F$$ (2.18)

$$\delta = \frac{FL}{AE}$$ (2.19)
Here, $T_B$ is applied torque to the bolt, $k_n$ is a nut factor, $d$ is diameter of the bolt, $\delta$ is deformation of the bolt by the applied force, $L$ is the length of all sensor components along the axis of bolt, and $E$ is Young’s modulus.

The micro-gap was calculated to be 4.1 $\mu$m at 1000 °C using Eqs. (2.15) and (2.16). The optimized clamping torque for tight contact between crystal and mass was calculated to be 0.6 N·m, as shown in Fig. 2.4 and this torque compressed the seismic masses and crystals by 5$\mu$m. Thus, applied clamping torque can compensate the thermal expansion of sensor components, maintaining physical and electrical contact at 1000 °C.

2.4 Experimental Methods

2.4.1 YCOB Crystal

The YCOB compound including Y$_2$O$_3$ (PIDC, 4 N-99.99% purity), CaCO$_3$ (Alfa Aesar, 4 N) and H$_3$BO$_3$ (Alfa Aesar, 4 N) was synthesized by calcining the mixed powder at 1000 °C for 10 h. The compound materials were sintered at 1250 °C and were loaded into an iridium crucible at 1530 °C. The Czochralski pulling technique was used to grow the crystals. The grown crystals were slowly cooled down to room temperature over a 24-h period. YCOB is a monoclinic biaxial crystal which belongs to the space group $C_m$. According to the IEEE Piezoelectric Standard [72], the crystal’s physical axes $X$, $Y$, $Z$ can be determined, where the $Y$ and $Z$ axes are parallel to the $b$ and $c$ axes, respectively, and the $X$ axis is perpendicular to the $Y$ and $Z$ axes. Thickness shear mode ($d_{26}$) YCOB crystals were used for the HT accelerometer in this study design. (YXt)-30° cut samples with dimensions of 20×10×1 mm$^3$ were prepared since (YXt)-30° cut samples showed the highest piezoelectric and
electromechanical properties compared to different cut samples (θ = 90°, 60°, 45°, 20°, 10°, 0°, and 33°) [73]. All the samples were vacuum sputtered with platinum thin films. The capacitance of the sample and the resonance and anti-resonance frequencies of the thickness shear vibration were measured using an impedance phase-gain analyzer (HP 4294A). Based on measured data, the coupling coefficient $k_{26}$, elastic compliance $s_{66}$ and piezoelectric coefficients $d_{26}$ and $g_{26}$ were calculated using the following formulae [22, 56]:

$$k_{26}^2 = \frac{\pi}{2} \frac{f_r}{f_a} \cot \left( \frac{\pi}{2} \frac{f_r}{f_a} \right) \quad (2.20)$$

$$s_{66}^E = \frac{1}{4 \rho f_a^2 f_r^2 (1 - k_{26}^2)} \quad (2.21)$$

$$d_{26}^2 = k_{26}^2 \times s_{66}^E \times \left( \frac{\varepsilon_{22}^T}{\varepsilon_0} \right) \quad (2.22)$$

$$g_{26} = \frac{d_{26}}{\varepsilon_{22}^T} \quad (2.23)$$

Here, $f_r$ is the resonance frequency, $f_a$ is the anti-resonance frequency, $\varepsilon_{22}^T$ is the absolute permittivity of the piezoelectric crystals and $\varepsilon_0$ is the permittivity of free space, respectively. Electrical properties of (YXt)−30° cut and non-rotated YX cut thickness shear mode YCOB crystals were shown in Table 2.5 [74]. The electrical resistivity was measured at 950 °C and other properties such as the dielectric constant, electromechanical coupling factor, piezoelectric strain coefficient, piezoelectric voltage coefficient, and elastic constant were measured at room temperature. The dielectric constant was measured to be 12.3 for different crystal cuts. The electromechanical coupling factor ($k_{26}$) was 0.19 for YXt 0° cut sample while the increased coupling (= 0.22) was obtained from YXt -30° cut. For YXt 0° cut
samples, the piezoelectric strain coefficient \((d_{26})\) and piezoelectric voltage coefficient \((g_{26})\) were calculated to be 8 pC/N and 0.07 Vm/N. The \(d_{26}\) and \(g_{26}\) for YXt-30° cut sample were 10 pC/N and 0.09 Vm/N, respectively. More importantly, the high mechanical quality factor (9000–16,000) and the high electrical resistivity \(2 \times 10^7 \ \Omega \cdot \text{cm}\) at elevated temperature of 950 °C supported that the YCOB single crystal is a promising candidate for high temperature sensing applications [50, 75]. In this research, (YXt)–30° cut YCOB crystals with dimensions of 12 mm × 10 mm × 1 mm were used as sensing materials for proposed HT accelerometer, as shown in Fig. 2.5, which exhibited the highest thickness shear piezoelectric and electromechanical properties [75].

2.4.2 Assembly Design

An assembly design of the prototyped shear-mode accelerometer was shown in Fig. 2.6. The overall dimensions were measured to be 30 mm × 26 mm × 17 mm. Four pieces of YCOB crystals (6) were arranged symmetrically on both sides of the center post (3). The crystals were rigidly secured between the center post and seismic masses (1) by only a bolt (4) and a nut (6). Adhesive parts were not used in the sensing parts to prevent the adhesive failures at high temperatures. The seismic masses and the center post were made of Inconel 601. They were arranged to be electrically and mechanically isolated from each other, and thus, they were able to be used as electrical connections directly. Other sensor components including housing and signal wires were also made of Inconel 601 due to its great resistance to high temperature oxidation and corrosion, and good electrical conductivity [67-69]. Since the thin film electrodes such as gold and platinum were not used in the assembly, the induced
electrical charge from each crystal was directly collected through the center post and seismic masses. The elimination of these films can lead to greatly reduced thin film degradation at high temperatures [60]. However, in this case, the sensor performance can be affected by the contact condition of surfaces and the applied contact forces between the crystals and Inconel. The optimized force (0.6 Nm) was applied to the bolt and nut using the torque control driver (model 285-50, Wiha Quality Tools). One Inconel signal wire was welded to the side of the base while another wire was clamped between the nut and the seismic mass directly. The bottom of the sensor was bonded to the top of an alumina rod (7) using a high temperature adhesive (Resbond 989, Cotronics Corp.). Since thin film electrodes and adhesives were not used in the sensor assembly, the accelerometer was expected to work stably at high temperatures without failures such as the oxidation and corrosion of electrodes.

2.4.3 Experimental Setup

Figs. 2.7 and 2.8 show a schematic and a picture of an experimental setup for high temperature accelerometer tests. A vertical tube furnace (Model GSL 1100X, MTI Corporation) was used to obtain the high temperature environment and the accelerometer was located in the tube through an alumina rod. The alumina rod was bonded to an aluminum bolt, and the bolt was screwed to a vibration exciter whose maximum force was 178 N (VG 100, Vibration Test Systems, Inc.). Input signal was generated and amplified by a function generator (Model AFG3101, Tectronix) and a power amplifier (Type 2706, Brue & Kjaer), respectively. The amplified signal was transmitted to the vibration exciter to generate the desired frequency and amplitude of vibration. Generated output signal from the sensing
crystals was also amplified through a charge amplifier (Type 2635, Bruel & Kjaer). Amplified output signal was recorded by both a lock-in amplifier (Model SR830, Stanford Research Systems) to filter 60 Hz power line interference and an oscilloscope (Model DSO7104B, Agilent Technologies) for transient signal analysis. The reference acceleration of the vibration exciter was measured by a commercial accelerometer (Model 352C22, PCB Piezotronics) and recorded on the oscilloscope through a signal conditioner (Model 482A16, PCB Piezotronics). The shaker was wrapped by an aluminum foil to minimize the electromagnetic interference to the accelerometer. The output charge of the accelerometer was measured with different temperatures (25–1000 °C), frequencies (50–350 Hz), accelerations (0.5–5 g) and dwell time at 1000 °C.

2.5 Results and Discussion
2.5.1 Analytical Modeling Result

Fig. 2.9 shows a frequency response model results for an ideal sensor (red line) and a whole sensor system (blue line). The low-frequency response was limited by the RC time constant ($\tau$) which determines the length of time that the crystal can hold generated charge. For this accelerometer, YCOB’s resistance ($R_{\text{leak}}$) was much larger than that of the amplifier. Thus, equivalent $R$ can be assumed to be the same as $R_{\text{amp}}$ and the low-frequency response can be determined by only the time constant of the charge amplifier (1 Hz). On the other hand, the high-frequency response was limited by the resonance frequency of the sensor and all other components. The resonance frequency of the accelerometer itself was calculated using Eq. (2.5) and found to be 120 kHz ($f_{n1}$). Usable frequency range for the ideal accelerometer was 1
Hz to 24 kHz. This calculation included only the seismic mass of the sensor \((m_1)\) and spring constant of crystals \((k_1)\). To calculate the resonance frequency of the whole sensor system, the effect of all other components such as the rod, mounting bolt and armature dynamic weight of the shaker should be considered. The effective spring constant \((k_2)\) was determined dominantly by the rod since the spring constant of the rod was much lower than others. The effective mass \((m_2)\) should include the mass of the rod and the armature mass of the shaker. Using the effective spring constant and mass, the resonance frequency of the sensor system was calculated to be 1.6 kHz \((f_{n2})\). Usable frequency range was determined to be 1 Hz to 335 Hz, which was narrower than the ideal sensor. The accelerometer parameters used in modeling were shown in Table 2.6. Charge sensitivity \((S_Q)\) of the accelerometer at flat frequency response was calculated to be 4.7 pC/g (=0.48 pC s^2/m). The seismic masses and the piezoelectric coefficient mainly affect the sensitivity of accelerometers, and thus, proper sensitivity can be selected for a specific application by using different design and sensing crystals.

### 2.5.2 Room Temperature Test Result

Fig. 2.10 shows the room temperature accelerometer test result with frequencies ranging from 1 Hz to 3 kHz. The prototyped accelerometer’s resonance frequency and usable frequency range were 1.75 kHz and 1–350 Hz, respectively. Sensitivity at the room temperature was found to be 5.6 ± 0.11 pC/g. This sensitivity was a little higher than the sensitivity from the modeling result (4.7 pC/g). Possible reason could be electromagnetic interference (EMI) induced by magnetic field change in the electromagnetic shaker. EMI can
affect the output charge of the accelerometer and the sensitivity. Fig. 2.11 presents the output voltage signal from the sensor with different driving voltage signal from the function generator. There was a 180° phase difference between two signals and this was because of the inverting action of the integrator circuit in the charge amplifier.

2.5.3 High Temperature Test Result

Fig. 2.12 indicates the charge output of the sensor at 1000 °C with different acceleration in frequencies ranging from 50 Hz to 350 Hz. The charge sensitivity at 1000 °C was found to be 6.3 ± 0.10 pC/g, which was a little higher than the room temperature value ~5.6 pC/g. This was because of the temperature dependence of piezoelectric coefficient of YCOB crystals. For instance, the piezoelectric coefficient of YCOB crystals is 10 pC/N at room temperature and 12 pC/N at 1000 °C [48]. Fig. 2.13 shows measured charge sensitivity with the tested temperatures (25–1000 °C) and frequency range (50–350 Hz). The charge sensitivity was found to be reasonably stable and the average sensitivity was determined to be 5.9 ± 0.06 pC/g over the tested frequency and temperature ranges. The measured sensitivity for 9-hour dwelling time at 1000 °C was shown in Fig. 2.14. The average sensitivity of 9-h dwelling time test was found to be 6.0 ± 0.12 pC/g.

Table 2.7 shows the comparison between the prototyped HT accelerometer and the commercial HT accelerometer (Model 357C91, PCB Piezotronics). The same sensing geometry and output type were used for both prototyped and commercial sensors. The impressive feature of the prototyped one was its much higher operating temperature than that of the commercial one. Sensor performances such as the charge sensitivity and capacitance
can be improved by using thinner YCOB crystal layers. The lower resonant frequency of the prototyped sensor can be achieved by reducing the sensor size and weight. During the dwelling test, a small crack was found in the ceramic bonding that connected the base and the alumina rod. This was because of thermal expansion difference of ceramic bonding and was one of the main reasons of unstable sensor performance after the 9-h dwelling time. Reduced electrical conductivity between Inconel and YCOBs due to the increased oxidation of the Inconel parts could be another reason.

2.6 Conclusion
In conclusion, a shear mode piezoelectric accelerometer using YCOB single crystals for high temperature applications was, for the first time, successfully demonstrated with high performance at temperatures up to 1000 °C. The dynamic response model and the thermal expansion model for the high temperature accelerometer were performed to expect the sensor’s performance at high temperatures. The prototyped sensor was tested in the frequency ranging from 50 Hz to 350 Hz and the temperature ranging from 25 °C to 1000 °C. It was found that the sensitivity of the accelerometer was 5.9 ± 0.06 pC/g throughout the tested frequency and temperature range. Stable sensitivity (6.0 ± 0.12 pC/g) was also observed from the 9-h dwelling test at 1000 °C. The experimental results agreed reasonably well with the modeling results. These results reveal high stability and reliability of the accelerometer prototype at high temperatures.
Figure 2.1 A schematic of a shear-mode piezoelectric accelerometer design.
Figure 2.2 The free body diagram of the accelerometer.
Figure 2.3 The equivalent circuit of the accelerometer system.
Figure 2.4 The calculated clamping torque as a function of temperature for tight contact between crystal and mass.
Figure 2.5 Prepared (YXt)–30° cut YCOB crystals with dimensions of 12 mm × 10 mm × 1 mm.
Figure 2.6  (a) An assembly design and (b) a photograph of the prototyped shear-mode accelerometer.
Figure 2.7 A schematic of an experimental setup for high temperature accelerometer tests.
Figure 2.8 A photograph of an experimental setup for high temperature accelerometer tests.
Figure 2.9 Frequency response model results for an ideal sensor (red line) and a whole sensor system (blue line).
Figure 2.10 The room temperature accelerometer test result with frequencies ranging from 1 Hz to 3 kHz.

Sensitivity at 25 °C = 5.6 pC/g
Figure 2.11 The output voltage signal from the sensor with different driving voltage signal from the function generator.
Figure 2.12 The charge output of the sensor at 1000 °C with different acceleration in frequencies ranging from 50 Hz to 350 Hz.
Figure 2.13 The measured charge sensitivity with the tested temperatures (25–1000 °C) and frequency range (50–350 Hz).
Figure 2.14 The measured sensitivity for 9-hour dwelling time at 1000 °C. The average sensitivity of 9-h dwelling time test was found to be $6.0 \pm 0.12 \text{ pC/g}$. 
Table 2.1 The important features for different types of high temperature sensors.

<table>
<thead>
<tr>
<th></th>
<th>Piezoresistive</th>
<th>Capacitive</th>
<th>Piezoelectric</th>
<th>Fiber optic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Measured parameters</td>
<td>Resistance</td>
<td>Capacitance</td>
<td>Charge and voltage</td>
<td>Intensity and phase</td>
</tr>
<tr>
<td>Resolution (strain)</td>
<td>$10^{-6}$</td>
<td>$10^{-6}$</td>
<td>$10^{-12}$</td>
<td>$10^{-10}$</td>
</tr>
<tr>
<td>Frequency range</td>
<td>DC – kHz</td>
<td>DC– kHz</td>
<td>MHz</td>
<td>DC– 100 kHz</td>
</tr>
<tr>
<td>Temperature range (°C)</td>
<td>&lt; 600</td>
<td>&lt; 400</td>
<td>&lt; 1000</td>
<td>&lt; 2000</td>
</tr>
<tr>
<td>Temperature stability</td>
<td>Poor</td>
<td>Good</td>
<td>Good</td>
<td>Good</td>
</tr>
<tr>
<td>Integration</td>
<td>Easy</td>
<td>Easy</td>
<td>Easy</td>
<td>Hard</td>
</tr>
<tr>
<td>Power consumption</td>
<td>Medium</td>
<td>Medium</td>
<td>Low</td>
<td>High</td>
</tr>
<tr>
<td>Cost &amp; complexity</td>
<td>Low</td>
<td>Low</td>
<td>Low</td>
<td>High</td>
</tr>
<tr>
<td>Electrode degradation</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
</tr>
</tbody>
</table>
Table 2.2 The important properties of high temperature piezoelectric materials.

<table>
<thead>
<tr>
<th>Material</th>
<th>GaPO$_4$</th>
<th>LN</th>
<th>LGS</th>
<th>YCOB</th>
<th>GdCOB</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T_{\text{max}}$ (°C)</td>
<td>970</td>
<td>1150</td>
<td>1450</td>
<td>1510</td>
<td>1470</td>
</tr>
<tr>
<td>$T_{\text{use}}$ (°C)</td>
<td>700</td>
<td>600</td>
<td>800</td>
<td>~1250</td>
<td>&lt; 1200</td>
</tr>
<tr>
<td>Temperature limited by</td>
<td>Phase transition</td>
<td>Resistivity</td>
<td>Resistivity</td>
<td>Resistivity</td>
<td>Resistivity</td>
</tr>
<tr>
<td>Mechanical quality factor</td>
<td>10000</td>
<td>2000</td>
<td>15000</td>
<td>9000</td>
<td>5000</td>
</tr>
<tr>
<td>Dielectric loss</td>
<td>0.15</td>
<td>5.5</td>
<td>0.5-0.2</td>
<td>0.3-0.1</td>
<td>0.3-0.2</td>
</tr>
<tr>
<td>Dielectric permittivity variation (%)</td>
<td>5</td>
<td>~40</td>
<td>25-15</td>
<td>15-10</td>
<td>10-9</td>
</tr>
<tr>
<td>Resistivity (Ω·cm)</td>
<td>$2.3\times10^7$</td>
<td>$6.6\times10^5$</td>
<td>$2\times10^6$-$1\times10^7$</td>
<td>$4\times10^6$-$2\times10^8$</td>
<td>$5\times10^6$-$8\times10^6$</td>
</tr>
<tr>
<td>Measured temperature (°C)</td>
<td>@ 800</td>
<td>@ 500</td>
<td>@ 500 −</td>
<td>@ 800 −</td>
<td>@ 800 −</td>
</tr>
<tr>
<td></td>
<td>600</td>
<td>1000</td>
<td>1000</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Operating temperature (°C)</td>
<td>Properties</td>
<td>Adhesion layer</td>
<td>Applications</td>
<td></td>
<td></td>
</tr>
<tr>
<td>---------------------------</td>
<td>------------</td>
<td>----------------</td>
<td>---------------</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pt</td>
<td>600</td>
<td>Surface degradation at temperatures below 600 °C</td>
<td>N/A</td>
<td>Piezoelectric Sensors</td>
<td></td>
</tr>
<tr>
<td>IrO₂/Ti</td>
<td>650</td>
<td>Ir oxide electrode enhance the organic destruction yield</td>
<td>Ti</td>
<td>Anode applications</td>
<td></td>
</tr>
<tr>
<td>Ir/TiAlN</td>
<td>700</td>
<td>Better O₂ diffusion barrier than that of Pt electrode</td>
<td>TiAlN</td>
<td>Ferroelectric capacitors</td>
<td></td>
</tr>
<tr>
<td>Pt/Ir, Pt/Rh</td>
<td>750</td>
<td>Excellent fatigue resistance and stable up to 750 °C</td>
<td>N/A</td>
<td>Ferroelectric capacitors</td>
<td></td>
</tr>
<tr>
<td>Pt/Zr</td>
<td>750</td>
<td>No metal surface degradation up to 750 °C</td>
<td>Zr</td>
<td>SAW devices</td>
<td></td>
</tr>
<tr>
<td>Pt-10%Rh/ZrO₂</td>
<td>850</td>
<td>Film recrystallization, agglomeration, and dewetting (700-1000 °C)</td>
<td>Zr</td>
<td>SAW devices</td>
<td></td>
</tr>
</tbody>
</table>
Table 2.4 Temperature limits and thermal expansion coefficients of the sensor components.

<table>
<thead>
<tr>
<th>Material</th>
<th>Temperature limit</th>
<th>Thermal expansion coefficient (ppm/°C)</th>
<th>Young’s modulus (GPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>YCa₄O(BO₃)₃</td>
<td>1500 °C (melting temp.)</td>
<td>11.2 @ 1000 °C</td>
<td>55.6 (Shear modulus)</td>
</tr>
<tr>
<td>Inconel 601</td>
<td>1411 °C (melting temp.)</td>
<td>17.8 @ 1000 °C</td>
<td>124.7</td>
</tr>
<tr>
<td>Adhesive</td>
<td>1648 °C (maximum temp.)</td>
<td>4.5 @ 25 °C</td>
<td>N/A</td>
</tr>
<tr>
<td>Alumina</td>
<td>2072 °C (melting temp.)</td>
<td>8.2 @ 25-1000 °C</td>
<td>370.0</td>
</tr>
</tbody>
</table>

Table 2.5 Properties of YCOB crystals with different crystal cuts.

<table>
<thead>
<tr>
<th>Crystal cut</th>
<th>εₑ</th>
<th>kₑ⁰</th>
<th>dₑ⁰ (pC/N)</th>
<th>gₑ⁰ (Vm/N)</th>
<th>sₑₑ (m²/N)</th>
<th>Qₑ⁰</th>
<th>ρ (Ohm.cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>YXt 0°</td>
<td>12.3</td>
<td>0.19</td>
<td>8.0</td>
<td>0.07</td>
<td>0.17×10⁻¹⁰</td>
<td>9000</td>
<td>2×10⁷</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>(@ 950 °C)</td>
</tr>
<tr>
<td>YXt -30°</td>
<td>12.3</td>
<td>0.22</td>
<td>10</td>
<td>0.09</td>
<td>0.18×10⁻¹⁰</td>
<td>16000</td>
<td></td>
</tr>
</tbody>
</table>
Table 2.6 The accelerometer parameters used in modeling.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Seismic mass ( (m_1) ) (kg)</td>
<td>( 47.68 \times 10^{-3} )</td>
</tr>
<tr>
<td>Mass of sensor system ( (m_2) ) (kg)</td>
<td>( 407.53 \times 10^{-3} )</td>
</tr>
<tr>
<td>Spring constant of four crystals ( (k_1) ) (N/m)</td>
<td>( 2.67 \times 10^{10} )</td>
</tr>
<tr>
<td>Spring constant of sensor system ( (k_2) ) (N/m)</td>
<td>( 4.26 \times 10^{7} )</td>
</tr>
<tr>
<td>Charge output of unit displacement ( (k_q) ) (C/m)</td>
<td>0.27</td>
</tr>
<tr>
<td>Capacitance ( (C) ) (nF)</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Prototyped HT accelerometer</td>
</tr>
<tr>
<td>--------------------------------</td>
<td>-----------------------------</td>
</tr>
<tr>
<td>Sensitivity (pC/g)</td>
<td>5.9</td>
</tr>
<tr>
<td>Frequency range (Hz)</td>
<td>&lt; 350</td>
</tr>
<tr>
<td>Resonant frequency (kHz)</td>
<td>1.75</td>
</tr>
<tr>
<td>Temperature limit (°C)</td>
<td>&gt; 1000</td>
</tr>
<tr>
<td>Capacitance (pF)</td>
<td>50</td>
</tr>
<tr>
<td>Sensing geometry</td>
<td>Shear</td>
</tr>
<tr>
<td>Output type</td>
<td>Charge</td>
</tr>
<tr>
<td>Housing material</td>
<td>Inconel</td>
</tr>
<tr>
<td>Size (mm³)</td>
<td>30.0 × 26.0 × 17.0</td>
</tr>
<tr>
<td>Weight (g)</td>
<td>164.0</td>
</tr>
</tbody>
</table>

* Model 357C91, PCB Piezotronics
CHAPTER 3
SURFACE ACOUSTIC LOAD SENSING

3.1 Background

3.1.1 Piezoelectric Resonators and Applications

Piezoelectric resonators including bulk acoustic wave (BAW) resonators and surface acoustic wave (SAW) resonators are widely used in various applications such as chemical, biomedical, information and service industries [76, 77]. Piezoelectric resonators can be used for sensing of different types of loading material such as liquids, gases and solids [13, 78]. They are highly sensitive to external load variations including force, pressure, temperature, viscosity, mass and viscoelasticity changes [13]. For chemical applications, resonator sensors have been developed for detection of pH level, mercury, corrosion, and humidity with specially designed thin sensing film. For example, a pH sensor using a piezoelectric AT-cut quartz resonator coated with cross-linked copolymer films has been studied [79]. The sensor is able to measure the frequency shift upon the change in the viscoelastic properties of the polymer film, which is dependent of the pH level of the polymer film. Corrosion sensors which are made of quartz or other piezoelectric material have been developed for sensing the degradation of fuel cells by measuring the mass change induced by corrosion on the surface area of the electrodes [80]. The resonator sensors have also been used as humidity sensors since they are able to detect small mass changes (less than a nanogram) of vapor deposition [81]. Similarly, the mercury sensor with gold thin film can be used for environmental pollution control and detection of mercury vapor concentrations in the air [82]. Piezoelectric biosensors consist of piezoelectric resonators and biological sensing films designed with
specific requirements. The biosensor can recognize different types of biological materials such as proteins (enzymes and antibodies), organelles, cells and tissue, and can sense any biochemical changes in these materials [83]. For example, piezoelectric-based immunosensors have a thin film made of the antibodies at the surface of the resonator. When the sensor is exposed to the target antigen, the film properties such as the elasticity, density and viscosity change. These changes can be detected by measuring the sensor parameters such as frequency and electrical impedance shift [84]. For information and service industry, there are several applications including fingerprint identification, touch screens and other touch-based device applications. Recently, a finger printer which uses acoustic impedance measurement with the piezoelectric resonator has been developed by Sonavation [85]. Piezoelectric composite (lead zirconate titanate (PZT)/epoxy 1–3 composite) was used for the micro-machined 2-D resonator array. The acoustic load impedance distribution on the sensor surface can be detected by measuring the electrical impedance of each element of the array in a rapid fashion. As a result, the pattern of the ridge and valley of the fingerprint can be mapped by their acoustic impedance difference (air and tissue). [86, 87].

3.1.2 Review of Piezoelectric Materials for Resonators

Quartz is a commonly used material in piezoelectric resonators due to its very high quality factor [88]. AT-cut quartz is the most popular type of crystal which vibrates in the thickness shear mode. They can be used to measure the mass deposition rates of additional thin layers on the resonator electrode surface. Using Sauerbrey’s equation, the linear relationship between deposited mass on the surface and the resonance frequency of the resonator can be
obtained [89]. The resonance frequency shift of the loaded quartz resonator can be measured with high resolution, since the quality factor of the loaded resonator is still high enough [90, 91]. However, quartz crystal resonators can suffer from their low electromechanical factor and low piezoelectric coefficients [90].

Lead zirconate titanate (Pb(Zr, Ti)O$_3$, PZT) is also widely used in various electromechanical devices such as resonators, ultrasonic transducers, sensors, and actuators [92]. PZTs have the advantage of having excellent piezoelectric characteristics including high piezoelectric coefficient (the order of hundreds of pC/N), relatively high dielectric constants, large electromechanical coupling coefficients, high response rate, low energy consumption and large displacement and ease of manufacturing [93-95]. Since the manufacturing of PZT thin films is relatively easy, they have been used in microelectromechanical systems (MEMS) for various integrated piezoelectric devices such as sensors, and actuators energy-harvesting devices [95].

Advanced piezoelectric materials such as lead magnesium niobate–lead titanate (PMN–PT) single crystals have also been intensively investigated [8, 96]. Its elastic compliance is about 6 times higher than that of PZT–5H and this offers decreased resonator size for a given operating frequency. The electromechanical coupling coefficient ($k_{33}$) of PZT-5H is about 0.75 while that of PMN–PT is greater than 0.9, which makes it possible to have a broader operating bandwidth [97, 98]. The piezoelectric coefficient of PMN-PT is 3 to 5 times higher than that of PZT–5H and this provides enhanced performance of piezoelectric resonators [8, 99]. However, the use of binary PMN-PT crystals can be limited by its low coercive field (2.5 kV/cm) for the application which requires a high excitation signal [8, 100].
In addition, the resonator performance reduction due to the low depoling temperature (TR/T = 75 °C to 95 °C) could be an issue [101]. On the other hand, ternary PIN–PMN–PT crystal resonator has recently gained attention since it has a significantly improved coercive field (5 kV/cm) and TR/T (117 °C) as well as high electromechanical couplings ($k_{33} > 0.9$) and piezoelectric coefficients ($d_{33} > 1500$ pC/N) compared to PMN-PT crystal resonator [102, 103]. In the thickness-shear mode, the PIN-PMN-PT crystal also shows the reasonably high shear piezoelectric coefficients ($d_{15} > 2000$ pC/N) and electromechanical coupling coefficients ($k_{15} > 0.85$) [104].

3.1.3 Face-Shear Mode PIN-PMN-PT Resonators

Recently, the face-shear mode PIN–PMN–PT resonator has drawn special attention due to its ultralow frequency constant (500 Hz·m), high piezoelectric coefficient ($d_{36}$ of 1600-2800 pC/N) and ultrahigh elastic compliances ($s_{66} > 120$ pm²/N) [105]. It also has much higher mechanical quality factor $Q_M$ (>120, compared to 30 for thickness shear mode crystals) and its low frequency constant can provide a high sensitivity to the acoustic load, which are interesting features for acoustic load sensing applications (those specified in section 3.3.2) [8, 105]. In addition, since the poling electrode is the same as the active electrode, the resonator can be easily repolarized. The face-shear mode PIN-PMN-PT resonator with these unique properties is preferred over other thickness or thickness shear for advanced acoustic load sensing applications [106]. The poling direction, electrodes, and vibration mode for three different mode resonators ((a) thickness mode, (b) thickness-shear mode, and (c) face-shear
mode) are shown in Fig. 3.1. The unique vibration shape of the face-shear mode resonator which occurs along the diagonal axis on the top surface was shown in Fig. 3.2.

3.2 Acoustic Load Sensing
3.2.1 Motivation and Objective
In this chapter, a new acoustic load sensing technique, which uses the relationship between electrical impedance of the piezoelectric resonator and acoustic load impedance, was introduced for bio-medical applications. For example, in the minimally invasive surgery or robotic surgery, it is necessary to use sensing devices such as a force sensor or a tactile sensor in order to improve user sensation. The acoustic load sensing technique can be highly sensitive as a biometric sensor due to the extremely high contrast ratio of acoustic impedance between air and tissue (4000:1) compared with capacitive (dielectric permittivity, 32:1) and thermal method (thermal conductivity, 8:1) [85]. Thus, the developed acoustic impedance sensing technique is promising for various applications in biomedical industry (minimally invasive surgery, robotic surgery and artificial skin sensor) and service robotics (touch screen, fingerprint reader and other touch based devices) [107]. The acoustic load sensing ability of the PMN-PT single-crystal resonators with different vibration modes including the thickness-mode, thickness-shear-mode, and face-shear-mode were investigated. Different acoustic loads were applied to each crystal resonator and their sensitivities to electrical impedance change induced by the applied acoustic loads were compared. Krimholtz, Leedom, and Matthaei (KLM) model simulation was used to verify the experimental results.
3.2.2 Principle of Proposed Sensing Technique

Piezoelectric single crystal resonators such as PMN-PT, PIN-PMN-PT, PZN-PT, etc. with different vibration modes can be used for the proposed sensing technique. This technique uses the relationship between the electric impedance of the piezoelectric resonator and the acoustic impedance of the surface load. Fig. 3.3 (a) shows the equivalent circuit for the unloaded piezoelectric resonator. The electrical impedance ($Z_{AB}$) is:

$$\frac{1}{Z_{AB}} = \frac{I}{V} = \left( \frac{1}{j\omega C_0} \parallel Z_1 \right)^{-1} = \frac{1}{Z_1} + j\omega C_0$$  \hspace{1cm} (3.1)

where

$$Z_1 = R_1 + j\omega L_1 + \frac{1}{j\omega C_1}$$  \hspace{1cm} (3.2)

Here, $C_0$, $R_1$, $L_1$ and $C_1$ is the static capacitance, motional resistance, motional inductance and motional capacitance, respectively. When the surface load is applied to the front surface of the piezoelectric resonator, the electrical impedance ($Z_{AB}$) of piezoelectric element changes accordingly. The surface load can be any material including air, water, tissue, rubber, etc. The applied load ($Z_L$) to the front surface of the sensor acts as an additional acoustic load layer. This acoustic load can be considered as an additional resistance or radiation resistance ($R_a$) which affects the electrical impedance ($Z_{AB}$) in the equivalent circuit of the piezoelectric resonator. Fig. 3.3 (b) represents the equivalent circuit for the loaded piezoelectric resonator. The electrical impedance of loaded resonator is:

$$\frac{1}{Z'_{AB}} = \frac{I}{V} = \left( \frac{1}{j\omega C_0} \parallel Z'_1 \right)^{-1} = \frac{1}{Z'_1} + j\omega C_0$$  \hspace{1cm} (3.3)
where

\[ Z'_1 = R_i + j\omega L_i + \frac{1}{j\omega C_1} + R_a(Z_L) \quad (3.4) \]

Here, \( R_a \) is the radiation resistance and is a function of acoustic load \( (Z_L) \). As a result, the acoustic load impedance can be sensed by measuring the electrical impedance of the piezoelectric resonator. The schematics of (a) the unloaded resonator and (b) the loaded resonators are shown in Fig. 3.4.

### 3.3 Analytical Modeling

#### 3.3.1 Piezoelectric Properties of Face Shear Mode Crystal

According to the IRE standards [108, 109], piezoelectric properties of the face-shear mode crystal can be determined. The frequency constant \( (N_{36}) \) and the elastic compliance \( (s_{66}^E) \) can be calculated using Eqs. (3.5) and (3.6):

\[
N_{36} = f_r \cdot h \quad (3.5)
\]

\[
s_{66}^E = \frac{F^2}{4\rho(N_{36})^2} \quad (3.6)
\]

where \( f_r \) is the resonance frequency, \( h \) is the crystal length, \( F \) is a correction constant, and \( \rho \) is the crystal density. The correction constant for the face-shear mode can be written by:

\[
F = \frac{2\kappa_0 \alpha}{\pi} \quad (3.7)
\]
where

\[
\alpha = 1 - 0.05015 \times \sqrt{\frac{s_{11}^E + s_{22}^E}{2s_{66}^E}} \tag{3.8}
\]

\[
\tan \kappa_0 + \kappa_0 = 0 \tag{3.9}
\]

Here, \(\kappa_0\) is the Eigen value of the shear mode and \(\alpha\) is a correction constant. Since the crystal was rotated by Zt \(\pm 45^\circ\), the value of \(s_{11}\) and \(s_{22}\) are much smaller than that of \(s_{66}\). Thus, we can assume that \(\alpha\) is equal to 1. \(\kappa_0\) can be calculated from the first root of Eq. (3.9) and found to be 2.0288. On the other hand, the electromechanical coupling coefficient \((k_{36})\) can be calculated using the following equations:

\[
r = \frac{f_r^2}{f_a^2 - f_r^2} \tag{3.10}
\]

\[
k_{36} = \frac{1}{\sqrt{1 + rp}} \tag{3.11}
\]

\[
p = \frac{4\alpha^2 b}{\kappa_0^2 + 2} \tag{3.12}
\]

\[
b = 1 - 0.0691 \times \frac{s_{11}^E + s_{22}^E}{2s_{66}^E} \tag{3.13}
\]

where \(r\) is a capacitance ratio, \(p\) is a factor characteristic of the face shear mode, and \(b\) is a correction constant. In this case, we can also assume that \(b\) is equal to 1. Finally, the piezoelectric coefficient \((d_{36})\) can be determined from Eq. (3.14).

\[
d_{36}^2 = \frac{\varepsilon_{33}^T s_{66}^E}{1 + rp} = k_{36}^2 \varepsilon_{33}^T s_{66}^E \tag{3.14}
\]
3.3.2 Surface Load Sensing Model

Fig. 3.5 shows a schematic diagram of the cross section of a piezoelectric with two driving electrodes and an acoustic load layer on one surface. This resonator can be considered as a three-port device which has one electrical port and two mechanical ports, as can be seen in Fig. 3.6 (a). Here, $F_1$, $F_2$, $v_1$ and $v_2$ indicate force on the back face, shear force on the front face (radiating surface), particle velocity at the back face and particle velocity at the front face, respectively. In addition, $V$ and $I$ represent the voltage and the current across the electrodes of the resonator. The relationship between the force, voltage, velocity, and current can be described through a $3 \times 3$ impedance matrix, as shown in Fig. 3.6 (b):

$$\begin{pmatrix}
F_1 \\
F_2 \\
V
\end{pmatrix} =
\begin{pmatrix}
Z_{11} & Z_{12} & Z_{13} \\
Z_{21} & Z_{22} & Z_{23} \\
Z_{31} & Z_{32} & Z_{33}
\end{pmatrix}
\begin{pmatrix}
u_1 \\
u_2 \\
I
\end{pmatrix}
$$

(3.15)

The piezoelectric constitutive equations are related to the electric displacement $D$, the electric field $E$, the shear stress $T$, and the elastic strain $S$. The piezoelectric equations of piezoelectric resonator vibrating in the face-shear are:

$$T_6 = c_{66}^D S_6 - h_{36} D_3$$

(3.16)

$$E_3 = -h_{36} S_6 + \frac{D_3}{\varepsilon_{33}^s}$$

(3.17)

where $c_{66}^D$ is the shear elastic stiffened constant, $h_{36}$ is the piezoelectric coefficient ($= e_{33} / \varepsilon_{33}^s$, $e_{33}$ is the piezoelectric stress constant), and $\varepsilon_{33}^s$ is the clamped dielectric constant. The piezoelectric resonator can be assumed to be a perfect insulator, and thus, electrical charge $D_3$ can be expressed by:
\[
\frac{\partial D_z}{\partial z} = 0
\]  

(3.18)

From the Newton’s second law, the stress equation of motion can be derived:

\[
\frac{\partial T_3}{\partial z} = \rho \frac{\partial^2 \xi}{\partial t^2}
\]  

(3.19)

By combining above equations, the plane-wave equation can be easily obtained as follows:

\[
\varepsilon_0 \frac{\partial^2 \xi}{\partial z^2} = \rho \frac{\partial^2 \xi}{\partial t^2}
\]  

(3.20)

The solution of the plane wave second order differential equation can be defined by:

\[
\xi = (C_1e^{-j\beta z} + C_2e^{j\beta z})e^{j\omega t}
\]  

(3.21)

Here, the constants \( C_1 \) and \( C_2 \) can be determined from the boundary conditions \( (u = \partial \xi / \partial t \) and \( F=TA) \) at the surfaces of the resonators. The current \( I \) and the voltage \( V \) across the electrodes can be expressed by:

\[
I = j\omega AD_3
\]  

(3.22)

\[
V_3 = \int_0^t E_s dz
\]  

(3.23)

Using Eqs. (3.16-3.23), the \( 3 \times 3 \) impedance matrix (3.15) which specifies the relations among the terminal variables can be solved. The solved electromechanical matrix can be written by:

\[
\begin{pmatrix}
F_1 \\
F_2 \\
V
\end{pmatrix} =
\begin{pmatrix}
Z_c A / j \tan \beta l & Z_c A / j \sin \beta l & h_{36} / j\omega \\
Z_c A / j \sin \beta l & Z_c A / j \tan \beta l & h_{36} / j\omega \\
h_{36} / j\omega & h_{36} / j\omega & 1 / j \omega C_0
\end{pmatrix}
\begin{pmatrix}
v_1 \\
v_2 \\
I
\end{pmatrix}
\]  

(3.24)
where \( Z_C \) is the characteristic stiffened acoustic impedance of the piezoelectric material \( (= (c_{\omega_0}^0 / \rho)^{1/2}) \), \( C_0 \) is the clamped (zero strain, high frequency) capacitance of the resonator, \( l \) is the crystal length, \( \rho \) is the density, and \( \beta \) is the wave number \( (= \omega l(c_{\omega_0}^0 / \rho)) \). The characteristic impedances of the backing and the load can be written by:

\[
Z_1 = (\rho_1 c_1)^{1/2} = F_1 / (A v_1)
\]

and

\[
Z_2 = (\rho_2 c_2)^{1/2} = F_2 / (A v_2)
\]

In the 3 \times 3 impedance matrix from Eq. (3.24), the first 2 \times 2 sub-matrix represents the ultrasonic propagation in the piezoelectric medium with the equations of a mechanical transmission line, characteristic impedance, and phase velocity. The piezoelectric coefficient \( h_{33} \) terms determine the electromechanical coupling of the piezoelectric resonator. The last matrix element is the electrical impedance of the piezoelectric resonator. The three ports model can be expressed as an electrical equivalent circuit for more convenience. In this equivalent circuit, the mechanical force can be considered as the electrical voltage and the particle velocity as the electrical current using the fundamental electromechanical analogies. Using Eqs. (3.24-3.26), a KLM equivalent circuit model which has two acoustic ports and one electrical port can be obtained, as shown in Fig. 3.7 [110]. The acoustic impedance from the right side \( (Z_r) \) and the acoustic impedance from the left side \( (Z_l) \) are given by [111],

\[
Z_r = Z_c Z_{EF} + Z_c \tanh(\gamma l_c / 2) \\
Z_l = Z_c Z_{GH} + Z_c \tanh(\gamma l_c / 2)
\]

\[
Z_1 = Z_c Z_{EF} + Z_c \tanh(\gamma l_c / 2) \\
Z_2 = Z_c Z_{GH} + Z_c \tanh(\gamma l_c / 2)
\]
where \( \gamma_c \) is the complex wave propagation factor \( (= j\omega (\rho_c / c_{66})^{1/2}) \), and \( l_c \) is the length of the resonator crystal. \( Z_{EF} \) and \( Z_{GH} \) are the acoustic load impedance at the port EF and GH, respectively. The acoustic impedance at the port CD is combination of \( Z_r \) and \( Z_l \) with the parallel arrangement. Thus, \( Z_{CD} \) is:

\[
Z_{CD} = \frac{1}{Z_r + \frac{1}{Z_l}} \tag{3.29}
\]

The total electrical impedance \( (Z_{AB}) \) at port AB can be described by:

\[
Z_{AB} = \frac{1}{j\omega C_0} + jX_1 + \frac{1}{N^2} Z_{CD} \tag{3.30}
\]

where

\[
jX_1 = \frac{1}{j\omega C_0} \frac{k_{36}^2}{\alpha} \sin \alpha \tag{3.31}
\]

and

\[
\frac{1}{N^2} = \frac{1}{\omega C_0} \frac{4k_{36}^2}{\alpha} \frac{1}{Z_c} \frac{\sin^2 \alpha}{2} \tag{3.32}
\]

and

\[
\alpha = \omega l_c \sqrt{\frac{\rho_c}{v_c}} \tag{3.33}
\]

\( N, X_1, C_0, \alpha, \rho_c \) and \( v_c \) are the turn ratio of a transformer, additional reactance of the equivalent circuit, clamped capacitance of the crystal, and the complex acoustic wave phase shift \( (= \pi \omega / \omega_0) \), the crystal density and the speed of shear sound waves in the crystal, respectively. Finally, the input electrical impedance \( (Z_{AB}) \) with acoustic loads can be obtained using Eqs. (3.27-3.33) [112, 113]:
\[ Z_{AB} = \frac{1}{j \omega C_0} \left\{ 1 + \frac{e_{56}^2 / e_{56}^6}{\omega l} \left[ j(Z_{EF} + Z_{GH})Z_c \sin \omega l \frac{\rho}{c_{66}} - 2Z_c^2 (1 - \cos \omega l \frac{\rho}{c_{66}}) \right] \right\} \]

Eq. (3.34) indicates that the electrical impedance \( Z_{AB} \) depends on only the external load impedance \( Z_{EF} \) and \( Z_{GH} \) for the constant initial resonator conditions. It can be assumed that the load at the port GH \( Z_{GH} \) is zero and at the port EF \( Z_{EF} \) is \( Z_L \) since the acoustic load \( Z_L \) was applied to one side of the crystal. Thus, \( Z_{AB} \) is:

\[ Z_{AB} = \frac{1}{j \omega C_0} \left\{ 1 - \frac{k_{36}^2}{\alpha} \frac{2 \tan \frac{\alpha}{2} - j \frac{Z_L}{Z_c}}{1 - j \frac{Z_L}{Z_c} \cot \alpha} \right\} \]

### 3.4 Experimental Method

#### 3.4.1 Piezoelectric Bulk Resonator Tests

Piezoelectric crystal resonators operating in three different modes, including thickness mode, thickness-shear mode and face-shear mode for the acoustic impedance sensing test were prepared. The crystal types and dimensions of resonators used in the experiments are summarized in Table 3.1. Cr/Au electrodes were deposited onto 10 × 10 mm surfaces using E-beam evaporation. 10-cm co-axial wires (AWG 25, Hitachi Cable Ltd., Tokyo, Japan) were bonded to electrode surfaces of each resonator using silver epoxy for the electrical connection. The capacitance, impedance, resonant frequency and anti-resonant frequency were measured using an impedance analyzer (HP4294A, Agilent). The piezoelectric
properties were calculated using equations shown in section 3.3.1. The calculated properties of the face-shear PIN-PMN-PT single crystal are summarized in Table 3.2.

The acoustic surface loads were fabricated using the silicone rubber (Sylgard 170, Dow Corning Corp., Midland, MI) and aluminum oxide (Al₂O₃) powder mixtures. Different percentages of alumina powders (0%, 10%, 20%, 30%, and 40%) were mixed with the silicone rubber to control the acoustic load impedance of the mixtures. Prepared mixtures samples (10 × 10 × 2 mm³) were applied to electrode surface of each piezoelectric resonator and cured for 24 hours in a vacuum desiccator. The schematic of the single-surface loaded piezoelectric resonator was presented in Fig. 3.8.

The impedance analyzer was connected to each wire on both sides of piezoelectric crystals. The electrical impedance spectrum of the resonator with acoustic loads was measured at the tested frequency range. In addition, the longitudinal sound velocities of rubber mixtures were measured using the pulse–echo method to obtain the shear sound velocities of rubber mixtures. A 30-MHz transducer was used for pulse–echo tests. The transducer was placed 5 mm away from the top surface of the rubber mixture sample which was immersed in a water tank. The ultrasound waves propagate through the sample targets and the time of flight of the pulse-echo waves between the top and bottom surfaces of the targets were measured [114]. Then, the shear velocity of the rubber was calculated by assuming that the Poisson’s ratio of rubbers was 0.5 the same as that of tissues. A digital caliper and a micro balance were used to measure the dimension, the weight, and the density of the rubber mixture. Finally, the shear acoustic impedance (Zₘ) of rubber mixtures was calculated by the equation:
\[ Z_s = \rho v_s \]  \hspace{1cm} (3.36)

where \( \rho \) is the density and \( v_s \) is the shear sound velocity of the rubber mixtures.

### 3.4.2 Sensor Array Tests

An 8 × 8 array was designed, fabricated and tested for the acoustic load sensing. A face-shear mode PMN-PT was chosen for the sensing crystal to take advantage of its high sensitivity to the surface load. Firstly, a piezoelectric crystal plate (10 × 10 × 1 mm\(^3\)) was lapped to 300 μm and Cr/Au electrodes were deposited on both large surfaces. The crystal plate which was bonded to a silicon wafer using bonding wax was diced into 64 sensing elements (820 μm × 820 μm × 300 μm). To obtain the clear face-shear mode of sensing elements, the crystal thickness was controlled to be 300 μm for certain side length and thickness ratios (< 0.5) [109].

Electrical wires (38 AWG) were connected to the top surface of the array for the row electrode connection. PDMS (Sylgard 184, Dow Corning Corp.) was applied to the array for a flexible and transparent substrate. PDMS substrate has an advantage of low acoustic impedance which was essential to reduce unwanted damping effects. The 8 × 8 array was detached from the silicon wafer and electrical wires were connected to the bottom surface of the array for the column. PDMS was also applied to the bottom surface of the array. The top and bottom electrode wires were soldered to the PCB board which was connected to the impedance analyzer. Fig. 3.9 shows the fabricated 8 × 8 array using face-shear mode PMN-PT crystal.
For the sensor array test, different liquid such as water and isopropyl alcohol (IPA) was used whose acoustic impedance was approximately 0.9 Mrayl and 1.5 Mrayl, respectively. The sample droplet was applied to the top surface of the array and the electric impedance shift of selected element was measured 5 times using the impedance analyzer. For the array imaging test, a water droplet was dropped onto the top surface of the sensor array and the electrical impedance shift of 64 elements were measured. The experimental setup for acoustic impedance load sensing using the 8 × 8 array was shown in Fig. 3.10.

3.5 Results and Discussion

3.5.1 Modeling Results

The calculated electrical impedance spectrum of the air loaded (blue) and water loaded (red dot) face-shear-mode resonator using Eq. (3.35) is shown in Fig. 3.11. The electrical impedance at the resonant frequency increased when the water was loaded while the electrical impedance at anti-resonant frequency increased. This result shows the electrical impedance dependence on the acoustic load. The Input parameters used in the KLM model are shown in Table 3.3.

Fig. 3.12 presents the calculated sensitivity of different resonator as a function of the surface loading. The sensitivity (SZ) is the ratio between electrical impedance (ZAB) shift and the applied surface loads (ZL) difference. Thus, S_{Z} can be calculated by using:

\[ S_{Z} = \left| \frac{dZ_{AB}}{dZ_{L}} \right| \]  

(3.37)
The face-shear mode showed the highest sensitivity of 63 Ω/Mrayl at the resonant frequency, which was 55 times higher than other modes. The reason of the high sensitivity of face-shear mode can be explained by the nature of advanced properties of face-shear mode crystals. The important determining factors for the electrical impedance characteristic of piezoelectric resonators are the resonance frequency ($\omega$), clamped capacitance ($C_0$), acoustic load ($Z_L$) and acoustic impedance of resonators ($Z_C$), as can be seen in Eq. (3.35). The resonator dimension ($t$ and $l$) determines the resonant frequency ($\omega_0 = \pi v/t$ or $\pi v/l$) and the static capacitance ($C_0 = \varepsilon A/t$) of the resonator. The product of the resonance frequency and the capacitance ($=\omega_0C_0$) mainly affects the electrical impedance at resonance and anti-resonance. Figs. 3.13 (a) and (b) show the effect of $\omega_0C_0$ on the electrical impedance responses to the acoustic load. The slope ($|Z_{AB}/Z_L|$) increases as the magnitude of $\omega_0C_0$ decreases from 0.1 to 0.001. The relationship between $\omega_0C_0$ and the crystal aspect ratio for three different modes was shown in the small inset in Fig. 13 (b). The frequency determining factor of thickness and thickness-shear mode is the crystal’s thickness ($t$) for square plate (dimensions: $l \times l \times t$) resonators. $\omega_0C_0$ of those resonators can be expressed by $\pi\varepsilon v(l/t)^2$. For face-shear mode, however, the frequency determining factor is the lateral length ($l$) and $\omega_0C_0$ is $\pi\varepsilon v(l/t)$. Thus, for different modes with the same aspect ratio ($l : t$), face-shear mode resonator is featured with lower $\omega_0C_0$ than other modes. As a result, the electrical impedance of the face-shear mode can shift more rapidly under surface loads. These results revealed that the face shear mode resonators can be highly sensitive to surface loads over other two modes resonators. The acoustic impedance of the crystal is also a critical factor which determines the load sensitivity. The low acoustic impedance can lead to a significantly increased slope between the electrical impedance and
the acoustic load, as can be seen in Figs. 3.14 (a) and (b). Since the acoustic impedance of the face-shear mode resonator is much lower than that of other mode resonators, the face-shear can have a higher load sensitivity compared to the others. Nonetheless, the load sensitivity of face-shear mode resonator is more dependent on the product of the angular frequency and the static capacitance. This is because $\omega_0 C_0$ of the face-shear mode is more than two orders lower than that of other modes. The electromechanical coupling coefficient (k) does not affect the magnitude of electrical impedance change. It affects only the resonant frequency and the anti-resonant frequency shifts. Since $\alpha$ is constant near the resonant frequency ($\alpha = \pi \omega_r / \omega_0$) and the anti-resonance frequency ($\alpha = \pi \omega_0 / \omega_0$), it is not a critical factor that determines the load sensitivity of the piezoelectric resonator. On the other hand, the lateral length of the face-shear mode resonator should be considered in order to obtain the desired load sensitivity. Fig. 3.15 shows the electrical impedance shift under surface loads with different lateral size of the face-shear resonator. The lateral length of face shear mode resonator is an important factor that determines the resonant frequency and the load sensitivity as mentioned earlier. The amount of impedance shift and the load sensitivity to the surface load increased as the resonant frequency increases. The sensitivity is proportional to the resonant frequency of the resonator and is inversely proportional to the resonator length, as can be seen in Table 3.4. For instance, the 1-mm-length resonator can have 10 times higher sensitivity compared to the sensitivity of the 10-mm-length resonator ideally. This result shows the possibility of the miniaturized face-shear resonator arrays as highly sensitive surface load sensing devices.
3.5.2 Bulk Resonator Test Results

Measured shear acoustic impedance and shear sound velocity of the rubber mixtures was shown in Fig. 3.16. It was found that the both values increased as the composition ratio of aluminum oxide powders increased. Figs. 3.17-3.19 present the measured electrical impedance under different acoustic loads for thickness, thickness-shear and face-shear mode resonator, respectively. The small insets indicate close-ups of the impedance changing at the resonant frequency. As the acoustic load impedance increased, the resonator’s electrical impedance increased at the resonant frequency. The larger electrical impedance change was found in the face-shear mode resonator, compared to thickness and thickness-shear mode resonators. It was found that the thickness-shear mode resonator’s electrical impedance shift was too small and not clear.

Figs. 3.20 and 3.21 show the measured electrical impedance under acoustic load impedance for thickness, thickness-shear and face-shear mode resonators at resonant frequency and anti-resonant frequency, respectively. For the face-shear mode resonator, the electrical impedance at the resonant frequency and anti-resonant frequency increases and decreases rapidly as the acoustic load increases. The sensitivity which indicates the ratio of electrical impedance change to the applied acoustic impedance was found to be at least 20 times higher for the face-shear mode than those for thickness-shear mode and thickness mode resonators. Table 3.5 summarizes the sensitivities of face-shear mode, thickness-shear mode, and thickness mode at the resonance and anti-resonance. From these results, it was revealed that the face-shear mode resonators possess much higher sensitivity to the surface acoustic loads at both resonance and anti-resonance than the thickness-shear and thickness mode
resonators. However, measured sensitivities were slightly different from calculated sensitivity values. Since the electrical wires were used to connect the resonator and the impedance analyzer, wires resistance should be considered as an additional resistance. In addition, the silver epoxy was used for the wire bonding on the resonator surface and this also could be considered as an additional surface load. These factors could be the reasons of the sensitivity difference between measured and calculated values. Fig. 3.22 shows the measured electrical impedance with various thicknesses of rubber layers (1 to 4 mm). The impedance change was found to be less than 5 % which was very small compared to the change based on the acoustic loading.

3.5.3 Sensor Array Test Results

Fig. 3.23 presents the result of the array sensing test with different liquid droplets, water and IPA. The electric impedance of tested element increased by 11.2 ± 0.59 % for IPA while it increased by 14.9 ± 0.28 % for H₂O from the initial electrical impedance value at resonance frequency. This difference in the electric impedance shift was due to the acoustic impedance difference of the loads, i.e. higher acoustic impedance of water (1.5 Mrayl) compared to IPA’s acoustic impedance (0.9 Mrayl). When the face-shear mode crystal is resonating, the particle vibration occurs along the diagonal axis on the top or bottom surfaces, as can be seen in Fig. 3.2. The vibration at the surface can be damped if the crystal’s surfaces are loaded by certain materials (e.g. rubber or liquid). Because of this damping effect, there will be the vibrational energy loss and suppression of mechanical resonance peaks which lead to the electrical impedance shift at the resonance. When the loaded material’s acoustic impedance
is higher, it is able to impede the vibration and reduce the resonance sharpness more effectively. This effect can explain the reason of the larger electrical impedance shift from the water loaded element compared to the IPA loaded element. It was also found that the slope of the calculated result was higher than that of the experimental result. The reason of this difference could be the loss effect including the mechanical and dielectric loss which was not considered in the calculation.

For the image test, water droplet was applied to the sensor array and the electric impedance shift of each element was measured. The droplet shape can be mapped by unloaded and loaded element’s acoustic impedance difference, as shown in Fig. 3.24. The element in darker color showed larger electric impedance shift while the bright part of the image indicated small impedance shift. The variations of electrical impedance shift due to the combined effect of the surface force distribution on the sensing array and the droplet thickness variation were observed.

3.6 Conclusion
In conclusion, a face-shear mode PIN-PMN-PT resonator for surface acoustic load sensing was studied. Different mode resonators including thickness mode, thickness-shear mode, and face-shear mode resonators were used and the sensitivities to the acoustic loads of each resonator were obtained. The experimental results were verified by the resonator equivalent circuit using KLM model. It was found that the face-shear mode resonator possesses more than one order of magnitude higher sensitivity, compared to other mode resonators. It was also found that surface load sensitivity increased as lateral sizes of resonators decreased,
which can be a merit for large area surface sensing using an array of face-shear resonators. Finally, a 8 × 8 sensor array using face-shear mode crystals was designed, fabricated and tested. Different acoustic impedance loads (IPA and H₂O) were applied to the sensor array and the electric impedance shift of the array was measured. The electric impedance shift upon surface load changes was successfully demonstrated. In addition, a droplet shape was effectively mapped by the acoustic impedance difference of loaded and unloaded elements on the sensor array. The revealed unique properties of face-shear mode piezoelectric resonators can be favorable for a broad range of applications where high sensitivity to the surface load is necessary such as artificial skins, bio-chemical sensors, touch screens, and other tactile-based sensors.
Figure 3.1 The poling direction, electrodes, and vibration mode for three different mode resonators; (a) thickness mode, (b) thickness-shear mode, and (c) face-shear mode.

Figure 3.2 The unique vibration shape of the face-shear mode resonator which occurs along the diagonal axis on top surface.
Figure 3.3 (a) The equivalent circuit for the unloaded piezoelectric resonator and (b) The equivalent circuit for the loaded piezoelectric resonator.

Figure 3.4 The schematics of (a) the unloaded resonator and (b) the loaded resonator with the object.
Figure 3.5 A schematic diagram of the cross section of a piezoelectric with two driving electrodes and an acoustic load layer on one surface.
Figure 3.6 (a) The resonator with three ports, one electrical port and two mechanical ports, and (b) the resonator with a $3 \times 3$ impedance matrix.
Figure 3.7 The equivalent circuit model of loaded piezoelectric resonator using KLM transmission line model.
Figure 3.8 The schematic of the single-surface loaded piezoelectric resonator with electrical wire connections.
Figure 3.9 The fabricated $8 \times 8$ array using face-shear mode PMN-PT crystal.
Figure 3.10 The experimental setup for acoustic impedance load sensing using the $8 \times 8$ array.
Figure 3.11 The calculated electrical impedance spectrum of the air loaded (blue) and water loaded (red dot) face-shear mode resonator.
Figure 3.12 The calculated sensitivity of different resonator as a function of the surface loading.
Figure 3.13 (a) The effect of $\omega_0 C_0$ on the electrical impedance responses to the acoustic load at the resonant frequency and (b) at the anti-resonant frequency. The small inset in Fig. 3.11 (b) shows the relationship between $\omega_0 C_0$ and the crystal aspect ratio for three different modes.

Figure 3.14 The electrical impedance as a function of acoustic load with different acoustic impedance of the piezoelectric resonator (a) at the resonant frequency and (b) at the anti-resonant frequency.
Figure 3.15 The electrical impedance shift under surface loads with different lateral size of the face-shear resonator.
Figure 3.16 The measured shear acoustic impedance and shear sound velocity of the rubber mixtures with different alumina compositions.
Figure 3.17 The measured electrical impedance under different acoustic load for thickness mode resonator.
Figure 3.18 The measured electrical impedance under different acoustic load for thickness-shear mode resonator.
Figure 3.19 The measured electrical impedance under different acoustic load for face-shear mode resonator.
Figure 3.20 The measured electrical impedance under acoustic load impedance for thickness, thickness-shear and face-shear mode resonators at resonant frequency.

$S_{z, FS} = 47 \text{ Ohm/Mrayl}$

$S_{z, TS} = 2 \text{ Ohm/Mrayl}$

$S_{z, T} = 0.5 \text{ Ohm/Mrayl}$
Figure 3.21 The measured electrical impedance under acoustic load impedance for thickness, thickness-shear and face-shear mode resonators at anti-resonant frequency.
Figure 3.22 The measured electrical impedance with various thicknesses of rubber layers (1 to 4 mm).
Figure 3.23 The array sensing test result with different liquid droplets (water and IPA).
Figure 3.24 (a) Sensor array with applied water drop; (b) Mapped image for water drop.
Table 3.1 The crystal types and dimensions of resonators used in the experiments.

<table>
<thead>
<tr>
<th>Vibration mode</th>
<th>Crystal</th>
<th>Dimensions (mm$^3$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness ($d_{33}$)</td>
<td>PMN-PT</td>
<td>10 ×10 × 0.5</td>
</tr>
<tr>
<td>Thickness-shear ($d_{15}$)</td>
<td>PMN-PT</td>
<td>10 ×10 × 1</td>
</tr>
<tr>
<td>Face-shear ($d_{36}$)</td>
<td>PIN-PMN-PT</td>
<td>10 ×10 × 1</td>
</tr>
</tbody>
</table>

Table 3.2 Calculated properties of the face-shear mode single crystal.

<table>
<thead>
<tr>
<th>$f_r$ (kHz)</th>
<th>$f_a$ (kHz)</th>
<th>$\varepsilon_{33}^T/\varepsilon_0$</th>
<th>$s_{66}^e$ (m$^2$/N)</th>
<th>$k_{36}$</th>
<th>$d_{36}$ (pC/N)</th>
<th>$N_{36}$ (Hz·m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Values</td>
<td>68.25</td>
<td>93.45</td>
<td>2940</td>
<td>111×10$^{-12}$</td>
<td>0.76</td>
<td>1280</td>
</tr>
</tbody>
</table>

Table 3.3 The Input parameters used in the KLM model.

<table>
<thead>
<tr>
<th>$l_c$ (mm)</th>
<th>$v_c$ (m/s)</th>
<th>$\rho_c$ (kg/m$^3$)</th>
<th>$C_0$ (nF)</th>
<th>$Z_c$ (Rayl)</th>
<th>$Z_{L,\text{air}}$ (Rayl)</th>
<th>$Z_{L,\text{water}}$ (Rayl)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Values</td>
<td>10</td>
<td>1632</td>
<td>8100</td>
<td>2.6</td>
<td>13.2</td>
<td>420</td>
</tr>
</tbody>
</table>
Table 3.4 The sensitivity of the face-shear mode resonators with different lateral sizes and resonant frequencies.

<table>
<thead>
<tr>
<th>Crystal length (mm)</th>
<th>Sensitivity (Ohm/Mrayl)</th>
<th>Resonant frequency (kHZ)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>63</td>
<td>68.25</td>
</tr>
<tr>
<td>5</td>
<td>126</td>
<td>136.50</td>
</tr>
<tr>
<td>2</td>
<td>315</td>
<td>341.25</td>
</tr>
<tr>
<td>1</td>
<td>631</td>
<td>682.50</td>
</tr>
</tbody>
</table>

Table 3.5 The sensitivities of face-shear mode, thickness-shear mode, and thickness mode at the resonance and anti-resonance.

<table>
<thead>
<tr>
<th></th>
<th>Face-shear mode</th>
<th>Thickness-shear mode</th>
<th>Thickness mode</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensitivity at resonance</td>
<td>47</td>
<td>2</td>
<td>0.5</td>
</tr>
<tr>
<td>Sensitivity at anti-resonance</td>
<td>74624</td>
<td>785</td>
<td>53</td>
</tr>
</tbody>
</table>

(Ohm/Mrayl)
CHAPTER 4

ACOUSTIC WAVE TACTILE SENSOR

4.1 Background

4.1.1 Tactile Sensors and Applications

Touch is a basic means of perception that human beings use to interact with the environment. Through this interaction, information about the different physical properties of an object or its environment can be obtained. Such information includes shape, texture, and temperature as well as the presence of friction or sharp objects [115]. Tactile sensor can be used to perceive these properties and defined as “a device or system that can measure a given property of an object or contact event through physical contact between the sensor and the object” [116]. Over the past few decades, numerous tactile sensing technologies have been researched for a broad range of applications, including touch-based electronics, surgical tools, health care and service robotics, processing of natural products, and mobile devices for personal security. In the surgery field, tactile sensors for minimally invasive surgery have been intensively researched, where they can be used to detect tissue’s softness and hardness, the grasping force exerted by surgical instruments, and the pressure of blood vessels and ducts [117-119]. In the medical diagnosis field, tactile sensors have been developed [120] for characterizing the stiffness of human ovum [121] and also for sensing the intraocular pressure (IOP) in human eye [122]. For health care and service robotics applications, tactile sensors can be used for an artificial skin installed on humanoid robots and other health care devices or a tool handle and controller for game consoles [123-125]. Tactile sensing technology is also important in the agricultural and food production industries. Since natural
products such as fruit and vegetables are soft, delicate and different in their shape, it is clear that the highly sensitive tactile sensing is greatly needed [116]. For automated factories handling natural products, the robotic handling system with tactile sensors can then be used for reducing hygiene risks, eliminating contamination, and preventing human errors.

4.1.2 Minimally Invasive Surgery (MIS)

Minimally invasive surgery (MIS, also known as “keyhole surgery”) is a revolutionary surgical technique and has been intensively researched in the last few decades. This technique combines engineering and medical technology using special surgical tools such as indenting, grasping, and rolling devices and laparoscopic cameras, as shown in Fig. 4.1. MIS has a lot of benefits for patients, including a minimized incision size (3 – 12 mm), which is much smaller than that of conventional surgery (> 50 mm). The small incision leads to reduced blood loss, faster recovery time, less pain during the recovery period, shorter hospitalization periods, fewer complications due to infection, better cosmetic results, and reduced overall cost [117, 119, 126].

On the other hand, the open surgery has advantages of allowing the surgeon to see the target tissue structures directly. Thus, it is possible for the surgeon to palpate tissues in order to perceive the applied grasping force and/or to differentiate tissue types. In this case, the elasticity information is important since tissue’s elastic properties are closely related to the tissue's biochemical composition [127, 128]. Based on the elasticity feedback, surgeons control the required force to grab tissues safely without damaging them and also separate the target tissue from other tissues [129]. For example, tumor or cancer breast tissues can be
differentiated from normal breast fat tissues because that abnormal tissues tend to be much stiffer (~ 560 kPa) than normal tissues (~24 kPa) [128]. Fig. 4.2 illustrates the elastic modulus of different type of tissues.

In MIS, due to the small incision, the surgeon is not permitted to access the operating area directly. This drawback can cause the loss of tactile sensation and reduced location perception, which leads to unexpected errors and endangers the patient’s safety [130, 131]. For example, MIS surgeons are not able to feel tissues’ hardness or softness by palpation, and hence, types of tissues or structures such as nerves, vessels, and ducts cannot be distinguished. To make things worse, invisible targets covered by other tissues cannot be easily detected even by using an advanced laparoscopic camera, but they can be detected by a simple sense of touch [129]. For these reasons, it is in critical need of a novel tactile sensor capable of sensing both applied force and soft tissue stiffness for better surgical results in MIS.

4.1.3 Review of Current Tactile Sensing Technologies

Various electromechanical tactile sensors have been researched using different sensing mechanisms. Piezoresistive-type tactile sensors are widely used due to their compatibility with silicon micromachining technology, which enables the fabrication of high resolution sensor arrays at a relatively low cost [115]. Beebe et al. proposed a piezoresistive tactile sensor on a silicon substrate [132], where the local strain generated by external force can be estimated by measuring the resistance of piezoresistive strain gauges [133]. The relationship between the resistance and the strain can be written by:
\[
\frac{\Delta R}{R_0} = G \frac{\Delta L}{L_0}
\]  

(4.1)

where \(\Delta R\), \(R_0\), \(G\), \(\Delta L\), \(L_0\) are the resistance shift induced by strain, original resistance of the strain gauge, the gauge factor, the length change induced by the applied force, and original length of the strain gauge. Based on this relationship, the force distribution can be measured using an array of piezoresistive strain gauges. The capacitive-type tactile sensor is another popular device, consisting of two electrode plates, an upper plate and a lower plate. This type of sensor measures the capacitance change in the gap between two electrodes. The gap decreases as applied force on the surface of the upper plate increases, and thus, the capacitance increases until the gap is closed [134]. The capacitance change generated by the change in the gap between two electrodes can be simply calculated by:

\[
\Delta C = \frac{\varepsilon A}{\Delta d}
\]  

(4.2)

where \(\Delta C\), \(\varepsilon\), \(A\), \(\Delta d\) represent the capacitance change, dielectric permittivity of medium, electrode areas, and the gap height (distance between two electrodes) change. Lee et al. developed a capacitive tactile sensor with high sensitivity [135] under normal loads since the capacitance can be measured with very high resolution (e.g. the capacitance change of 2.9 \% per 1 mN). This sensor is also suitable for large-area sensing because of its simple sensing mechanism [135, 136]. The piezoelectric tactile sensors are widely used in the biomedical field because of their simple structure. When the external force is applied to the piezoelectric material such as PZT or PVDF, the electrical charge is induced according to the direct
piezoelectric effect. Therefore, the force variation can be detected directly by measuring the amount of induced electrical charge from the piezoelectric materials as shown in Eq. 4.3.

\[ Q = dF \]  (4.3)

Here, \( Q \) is generated electrical charge, \( d \) is piezoelectric coefficient, and \( F \) is applied force.

Dargahi and Qasaimeh et al. developed the piezoelectric-based tactile sensor using polyvinylidene fluoride (PVDF) thin films for endoscopic applications [118, 126]. The sensor was able to measure both the force applied onto the object and the object’s compliance using the relative deformation information of contacted objects.

Electromechanical sensing technique mentioned above can be used for elasticity measurement as well. Typically, they consist of a sensing film and a flexible beam structure which can be deformed by the applied force, as shown in Fig. 4.3 [137]. When a hard object is placed on the beam as a target and the force is applied onto it, little deformation will be induced in the beam (see Fig. 4.3(a)). On the other hand, if the target is a soft object, the beam will deform by the same applied force (see Fig. 4.3(b)). As the stiffness of target object decreases, the deflection of the beam increases under the same force. A sensing material such as piezoresistive and piezoelectric material, which is firmly attached to the beam, deforms and generates the output signal corresponding to the amount of deflection according to Eqs. (4.1-4.3). The induced output signal is dependent on the stiffness of the object, and hence, the target’s elasticity can be estimated. In electromechanical tactile sensing, two quantities, force and elasticity, are measured separately and indirectly. In summary, existing tactile sensing techniques are either complicated and time consuming, or cannot be used to measure both
force and elasticity concurrently [117-119, 126, 138]. Table 4.1 shows the comparison of different types of electromechanical tactile sensor technologies.

Elasticity imaging methods using the excitation technique, also known as elastography, have also been developed to characterize soft tissue’s elastic properties [127, 128, 139]. These imaging methods can be divided into two types: static and dynamic elastography. In the static elastography or compression elastography, slow and repetitive compression is applied to tissues using an ultrasonic transducer in order to generate strain as shown in Fig. 4.4 [128]. Under external pressures, the larger strain (A – B in Fig. 4.4(b)) is induced from soft tissue than from hard tissue (C – D in Fig. 4.4(b)) since soft tissue can easily deform compared to hard tissue. The generated strain in the tissue can be measured by using the pulse-echo method. However, this method offers only qualitative information which means that the absolute Young’s modulus cannot be obtained directly because that the stress induced in the tissues is unknown. Instead, strain ratios between the target tissue and the surrounding normal tissues can be calculated, and thus, the elasticity of the target can be estimated.

In the dynamic elastography method, shear-wave elastography (SWE) is popular and widely used method due to its fast imaging process [140-142]. This elastography uses ultrasound transducers to generate ultrasound waves which can induce the acoustic radiation force in tissue. This force (also called “acoustic wind”) pushes underlying tissues along the propagating direction and produces transversely oriented shear-waves within tissue, as illustrated in Fig. 4.5. An ultrasound system is usually used to measure the sound speed of the shear-waves. In this case, quantitative information can be obtained (i.e. elasticity in kPa)
since the wave propagation speed is directly proportional to tissue’s stiffness and Young’s modulus. For example, shear-waves travel faster in stiff tissue than in soft tissue due to the difference in the elastic modulus. Despite their efficacy, all of the elastography methods mentioned above are complicated and/or time consuming because they need highly technical equipment such as an ultrasound imaging system which is less applicable in MIS. In addition, plenty of practice for the transducer control is also required since the image quality is user-skill dependent [130]. Table 4.2 summarizes advantages, disadvantage and applications of different types of elasticity measurement methods.

4.2 Acoustic Wave Tactile Sensor

4.2.1 Motivation and Objective

As reviewed in the previous section (4.1.2), a tactile sensor which can measure tissue’s elastic properties and applied forces is necessary for minimally invasive surgery. However, existing tactile sensing methods discussed in the previous section can be complicated and time consuming and/or not real-time process, or they may not be able to measure both force and elasticity at once. Moreover, elasticity imaging methods require expensive and highly technical medical equipment and operation expertise as well. Consequently, there is no tactile sensor being integrated in tools used for MIS. Tactile sensors in this purpose should be able to measure the applied force and the elastic modulus of tissues in a simple and rapid manner. Easy integration with MIS tools is also essentially required.

In this chapter, a novel tactile sensor for MIS applications using the acoustic wave sensing technique was introduced for the first time. The acoustic wave tactile sensor array
was designed, fabricated, and tested for simple and quick tactile sensing. For the elasticity sensing, the relationship between the target tissue’s acoustic impedance and the sensor array’s electrical impedance was used. The prototyped tactile sensor array was also capable of measuring the external force. For the force sensing, the nonlinear elastic effect (also known as the strain hardening effect) of the sensing layer was used which leads to changes in the acoustic impedance of the sensing layer [143].

4.2.2 Face-Shear Acoustic Tactile Sensing Concept

The proposed acoustic wave tactile sensor array was designed for both elasticity and force sensing. As discussed in the Chapter 3, a face-shear mode PMN-PT crystal resonator was selected for a sensing element due to its exceptionally high sensitivity to the acoustic load impedance. It is also known that the elastic properties of tissue such as the Young’s modulus and shear modulus are closely related to the tissue's characteristic acoustic impedance. Hence, measuring acoustic impedance of tissue can result in tissue elastic properties.

The characteristic acoustic impedance can be expressed by:

\[ Z = \rho v \]  \hspace{1cm} (4.4)

or

\[ Z = \sqrt{\rho E} \]  \hspace{1cm} (4.5)

Here, \( Z \) is the characteristic acoustic impedance, \( \rho \) is the density, \( v \) is the sound speed and \( E \) is the elastic modulus. One can observe that different types of tissue have different acoustic impedances because of their elasticity difference. For example, the acoustic impedance of muscle tissues (1.75 Mrayl) is higher than that of fat tissues (1.38 Mrayl) [142]. In acoustic
wave tactile sensing, the target tissue’s acoustic impedance acts as an acoustic load impedance \((Z_L)\), and thus, the elastic modulus of tissue can be calculated by:

\[
E = \rho v^2 = \frac{Z_L^2}{\rho}
\]

As can be seen in Eq. (4.6), the elastic modulus \((E)\) depends on the square of the acoustic load impedance \((Z_L)\). Furthermore, it has been demonstrated that the tissue’s acoustic impedance is dominantly affected by its elasticity rather than its density [144]. This is because there is only a small variation in density of different tissues \((1000 \pm 8 \text{ kg/m}^3)\) compared with the acoustic impedance difference [142]. These facts support the hypothesis that the acoustic impedance can be a sensitive parameter in determining the elasticity of tissues by a simple acoustic impedance measurement. The schematic of the elastic sensing method was presented in Fig. 4.6. The details for elasticity sensing will be discussed in the Chapter (4.3.1).

In force sensing, an elastic layer (e.g. PDMS) was designed as a sensing layer and attached to one surface of the face-shear resonator as shown in Fig. 4.7. It is noteworthy that soft solid materials such as rubbers and polymers exhibit nonlinear elasticity due to their cross linking system [145], meaning that their elastic modulus changes with external load. Therefore, when the compressive force \((F)\) is applied to the PDMS sensing layer, the effective Young’s modulus of the PDMS layer increases due to its nonlinear elasticity. This change in the elastic modulus \((\Delta E)\) leads to the change in the acoustic impedance of the sensing layer \((\Delta Z_L)\) [146]. The density can remain constant by assuming that the material (PDMS) behavior is incompressible [147]. The acoustic impedance variation due to the
elastic modulus change can be detected by measuring the electrical impedance of the sensor according to the acoustic load sensing model introduced in Chapter 3.2. As a result, external force can be estimated by the sensing layer’s acoustic impedance change resulting from the sensing layer’s nonlinear elasticity. The details for force sensing will be discussed in the Chapter (4.3.2).

4.3 Analytical Modeling

4.3.1 Tissue Elasticity Sensing Model

The surface load sensing model derived from the Chapter (3.3.2) can be used for the tissue stiffness sensing model. The electrical impedance of a sensor is:

$$Z_{AB} = \frac{1}{j\omega C_0} \left(1 + \frac{e_{36}^2 e_{33}^2 c_{66}}{\rho \sqrt{c_{66}^2}} \frac{j(Z_{EF} + Z_{GH}) Z_c \sin \omega l \sqrt{\frac{\rho}{c_{66}}} - 2Z_c^2 (1 - \cos \omega l) \sqrt{\frac{\rho}{c_{66}}}}{Z_c^2 + Z_{EF} Z_{GH}} \sin \omega l \sqrt{\frac{\rho}{c_{66}}} - j(Z_{EF} + Z_{GH}) Z_c \cos \omega l \sqrt{\frac{\rho}{c_{66}}}} \right)$$  \hspace{1cm} (4.7)

In the case of the tissue elasticity sensing, the acoustic impedance at the port EF (see Fig. 3.7) can be considered as an acoustic impedance of the backing layer. The acoustic impedance at the port GH can be analogous to the acoustic load impedance at the front surface of the sensor. Thus, Eq. (4.7) can be simplified by:

$$Z_{AB} = \frac{1}{j\omega C_0} \left(1 + \frac{k_{36}^2}{\beta l} \frac{j(Z_B + Z_L) Z_c \sin \beta l - 2Z_c^2 (1 - \cos \beta l)}{Z_c^2 + Z_B Z_L} \sin \beta l - j(Z_B + Z_L) Z_c \cos \beta l} \right)$$  \hspace{1cm} (4.8)

where $\omega$ is the angular frequency, $k$ is the electromechanical coupling coefficient, $\beta$ is the wave number, $l$ is the crystal length and $C_0$ is the clamped capacitance of the sensing element. $Z_c$, $Z_B$ and $Z_L$ represent the acoustic impedance of the piezoelectric element, the
acoustic load impedance of the backing and the acoustic load impedance at the front surface, respectively. As can be seen in Eq. (4.8), the electrical impedance \( Z_{AB} \) of the sensing element depends only on the external acoustic load impedance \( Z_L \) for constant initial conditions of the sensor array. As a result, acoustic load impedance \( Z_L \) can be obtained by measuring the electrical impedance of the piezoelectric element \( Z_{AB} \), and hence, the elastic modulus of targets \( E \) can be calculated by obtained acoustic load impedance \( Z_L \) using an equation:

\[
E = \frac{Z_L^2}{\rho}
\]  

(4.9)

where \( \rho \) is the density which can be assumed a constant in this case. Therefore, by combining Eq. (4.8) and Eq. (4.9), the resulted electrical impedance \( Z_{AB} \) of the piezoelectric element can be expressed as a function of the elastic modulus of targets as follows:

\[
Z_{AB} = \frac{1}{j\omega C_0} \left( 1 + \frac{k_{36}^2}{\beta l} \frac{j(Z_B + \sqrt{E\rho}Z_C) \sin \beta l - 2Z_C^2(1 - \cos \beta l)}{Z_C^2 + Z_B \sqrt{E\rho} \sin \beta l - j(Z_B + \sqrt{E\rho}Z_C \cos \beta l)} \right) \]  

(4.10)

or

\[
Z_{AB} = f(E)
\]  

(4.11)

4.3.2 Force Sensing Model

In order to obtain the force sensing model, the polymer (PDMS) sensing layer’s nonlinear elasticity should be characterized first. There are several ways to simulate the nonlinear elasticity model of polymer materials, including the neo-Hookean model [148], second-order Ogden model [149] and third-order Mooney-Rivlin model [150]. In this study, the Mooney-
Rivlin model was chosen due to its simplicity and good agreement with experimental data for rubber or polymer materials.

The strain energy density in rubber-like material can be expressed as [137, 147]:

\[ W = C_1(I_1 - 3) + C_2(I_2 - 3) \]  \hspace{1cm} (4.12)

where \( W \) is the stored strain energy density, \( I_1 \) and \( I_2 \) are the first and second strain invariants, and \( C_1 \) and \( C_2 \) are the constants whose unit is the same as that of stress. The strain invariants can be defined as:

\[ I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \]  \hspace{1cm} (4.13)

\[ I_2 = \lambda_1^2 \lambda_2^2 + \lambda_2^2 \lambda_3^2 + \lambda_3^2 \lambda_1^2 \]  \hspace{1cm} (4.14)

Here, \( \lambda_1, \lambda_2 \) and \( \lambda_3 \) are the principal draw ratios along the edges of the material and can be defined by:

\[ \lambda_i = (L_i + u_i)/L_i \]  \hspace{1cm} (4.15)

where \( L \) is original length and \( u \) is deformation along the axis \( (i = 1, 2, 3) \) induced by applied force. By substituting Eqs (4.13 and 4.14) into Eq. (4.12), the equation of the strain energy \( (W) \) is

\[ W = C_1(\lambda_1^2 + \lambda_2^2 + \lambda_3^2 - 3) + C_2(\lambda_1^2 \lambda_2^2 + \lambda_2^2 \lambda_3^2 + \lambda_3^2 \lambda_1^2 - 3) \]  \hspace{1cm} (4.16)

Assuming that the material is non-compressive and the volume change is considered, the draw ratio can be expressed as:

\[ \lambda_1 = \lambda = 1 - \varepsilon \]  \hspace{1cm} (4.17)

and
\[ \lambda_2^2 = \lambda_3^2 = 1 - 2\nu(1 - 1/\lambda) \] \hspace{1cm} (4.18)

where \( \nu \) is the Poisson ratio and \( \varepsilon \) is the strain. Thus, the Mooney-Rivlin equation is:

\[ W = C_1[\lambda^2 + 2(1-\nu(\lambda-1))^2 - 3] + C_2[2\lambda^2(1-\nu(\lambda-1))^2 + (1-\nu(\lambda-1))^4 - 3] \] \hspace{1cm} (4.19)

The strain energy can be written by:

\[ W = \int \sigma d\varepsilon \] \hspace{1cm} (4.20)

Therefore, the stress can be expressed as:

\[
\sigma = \frac{dW}{d\varepsilon} = C_1\left(-2(1-\varepsilon) + \frac{4\nu}{(1-\varepsilon)^2}\right) + \\
C_2 \left( 4\nu - 4(1-\varepsilon)(1 - 2\left(1 - \frac{1}{1-\varepsilon}\right)\nu) + \frac{4\nu\left(1 - 2\left(\frac{1}{1-\varepsilon}\right)\nu\right)}{(1-\varepsilon)^2} \right)
\] \hspace{1cm} (4.21)

or

\[ \sigma = f(\varepsilon) \] \hspace{1cm} (4.22)

where \( \sigma \) is the stress and \( \varepsilon \) is the strain. The constants \( C_1 \) and \( C_2 \) can be estimated from a Mooney–Rivlin plot, (\( \sigma/\lambda - 1/\lambda_2 \) versus \( 1/\lambda \)) \hspace{1cm} [137] and the stress-strain diagram can be obtained by calculating Eq. (4.21). The elastic modulus (\( E \)) can be defined as:

\[ E = \frac{\partial \sigma}{\partial \varepsilon} \] \hspace{1cm} (4.23)

From Eqs. (4.21 and 4.23), \( E \) can be rewritten by:
\[ E = C_1 \left(2 + \frac{8\nu}{(1-\nu)^3}\right) + C_2 \left(-\frac{8\nu}{1-\nu} + \frac{8\nu^2}{(1-\nu)^2} + 4\left(1 - 2\left(1 - \frac{1}{1-\nu}\right)\nu\right) + \frac{8\nu\left(1 - 2\left(1 - \frac{1}{1-\nu}\right)\nu\right)}{(1-\nu)^3}\right) \] (4.24)

or

\[ E = f(\varepsilon) \] (4.25)

Strain (\varepsilon) is directly related to stress (\sigma) and stress is defined as the average force (F) per unit area (A). Thus,

\[ E = f(\sigma) \] (4.26)

and

\[ \sigma = \frac{F}{A} \] (4.27)

Therefore, the elastic modulus (E) is:

\[ E = f(F) \] (4.28)

The elasticity sensing model (see Eq. 4.10) can be modified using Eqs. (4.24). The electrical impedance of the piezoelectric element (\(Z_{AB}\)) can be represented as a function of elastic modulus (E):

\[ Z_{AB} = \frac{1}{j\omega C_0} \left(1 + \frac{k_{36}^2 j(Z_B + \sqrt{E\rho})Z_C \sin \beta l - 2Z_C^2 (1 - \cos \beta l)}{\beta l (Z_C^2 + Z_B \sqrt{E\rho}) \sin \beta l - j(Z_B + \sqrt{E\rho})Z_C \cos \beta l}\right) \] (4.29)
Finally, by substituting Eq.(4.28) into Eq. (4.29), the resulted electrical impedance can be expressed as a function of force:

\[
Z_{AB} = \frac{1}{j\omega C_0} \left( 1 + \frac{k_{36}^2}{\beta l} \left( Z_B^2 + \sqrt{E(F)\rho} Z_C \sin \beta l - 2Z_C^2(1 - \cos \beta l) \right) \right) \sin \beta l - j \left( Z_B^2 + \sqrt{E(F)\rho} Z_C \cos \beta l \right) (4.30)
\]

or

\[
Z_{AB} = f(F) \quad (4.31)
\]

4.4 Tactile Sensor Array
4.4.1 Array Design and Fabrication

A 6 × 6 tactile sensor array was designed and fabricated. The sensor array consisted of total 36 face-shear mode PMN-PT sensing elements whose operational frequency was about 1 MHz. For the electrical connection, a 6 × 6 row-column electrode array was designed in order to reduce the number of electric wires. PDMS was used as a sensing layer and a composite filling material due to its transparency, simple fabrication, and biocompatibility. The design specification for the fabricated sensor array is summarized in Table 4.3.

The array fabrication process was described in Fig. 4.8. The face-shear mode PMN-PT crystal plate (10 mm × 10 mm × 0.5 mm) with electrodes (Ti/Au, 10/50 nm) on both large surfaces (10 mm × 10 mm) were prepared first (a). The crystal plate was then bonded to a silicon wafer using bonding wax and diced into 36 sensing elements (element dimension: 800 μm × 800 μm × 500 μm) with a pitch of 1.3 mm using a dicing saw (DAD-321, Disco) (b). A
silicon wafer with Ti/Au electrode on the top surface was prepared (c) and diced to 0.3 mm in depth to form the bottom (column) electrode (d). The 6 × 6 array bonded to the silicon wafer was attached to the bottom electrodes with epoxy (Epo-tek 301, Epoxy Technology, Inc.). The bonding wax was melted at 70 °C on the hot plate and the silicon wafer was removed to expose the top surface of the array (e). Then, gold coated thin ribbons (0.4 mm in width) were bonded to the top surface of the array using silver epoxy to form top (row) electrodes (f). Thin electric wires (38 AWG) were attached to each top and bottom electrode as electric connections. Finally, a PDMS (Sylgard 184 at 10:1 ratio between pre-polymer and curing agent) layer was applied to the top of the array through a circular-shaped mold as a sensing and protecting layer (g).

4.4.2 Switch and Microcontroller Circuit

Each element of the 6 × 6 array is connected to two electrode lines, the row (top) electrode and the column (bottom) electrode line. The row electrode runs perpendicular to the column electrode, and thus, any single element can be selected by connecting two electrodes, reducing the total number of electrode connections. For example, the element located in the 3rd row and 3rd column can be addressed by selecting the top electrode T3 and the bottom electrode B3, as shown in Fig. 4.9. Both the top and bottom electrodes of the array were wired to the input of a 8-channel relay board (5V relay module, SainSmart) whose outputs were connected to an impedance analyzer (HP 4294A, Agilent). The relays were controlled using a microcontroller (PIC18F2550, Electronics-DIY) which was connected to a personal computer through a USB connector for scanning all 36 elements. Fig. 4.10 illustrates the
block diagram of the array impedance measurement system and the switch circuit. The photograph picture of the fabricated 6 × 6 array with the switch and microcontroller circuit was shown in Fig. 4.11.

4.5 Experimental Method

4.5.1 Tissue Elasticity Sensing Test Using Tissue Mimicking Phantoms

Tissue mimicking phantoms were prepared using beef gelatin powders (Great Lakes Gelatin Company, IL). The elastic stiffness of the tissue phantom can be controlled by the weight ratio of gelatin powder in the prepared phantoms. In this study, weight ratios (WR) of gelatin powder ranging from 5 wt % to 30 wt % were used. To prepare phantoms, gelatin powders were blended with warm water (65 °C) in Petri dishes and the mixture was degassed at vacuum pressure using a vacuum desiccator. The degassed mixtures were stored in a refrigerator for 3 hours [151]. The fabricated tissue mimicking phantoms with different WR are shown in Fig. 4.12.

The force-deformation method [152] was used to characterize the tissue phantom’s shear modulus as reference data. Strain (ε) and stress (σ) of each tissue phantom were obtained and the Young’s modulus was calculated using Eq. (4.20):

\[ E = \frac{\sigma}{\varepsilon} = \frac{F/A_0}{\Delta L/L_0} \]  (4.32)

where \( E \), \( F \), \( A_0 \), \( \Delta L \) and \( L_0 \) are the Young’s modulus, the applied normal force, the initial area on which the force is applied, the change in thickness of the phantom and the original
thickness of the phantom, respectively. The shear modulus of the tissue phantom can be calculated using Eq. (4.21):

$$G = \frac{E}{2(1+\nu)}$$  \hspace{1cm} (4.33)

where $G$ and $\nu$ are the shear modulus and the Poisson’s ratio of the tissue phantom. In this study, a Poisson’s ratio of 0.5 was used for phantom samples, which was the same as that of tissues in published work [127].

The elasticity sensing test using the bulk face-shear mode resonator without the sensing layer was conducted first in order to verify the elasticity sensing ability of the acoustic wave sensing technique. A face-shear mode bulk crystal resonator (10 mm × 10 mm × 1 mm) was prepared and tissue phantom samples were attached to the resonator, followed by electrical impedance measurement.

For the array sensing test, it is important to maintain a good contact between the phantoms and the sensing layer, without an air gap. Thus, phantom mixtures in liquid phase were applied to the center of the array and then cured. The electrical impedances of 36 sensor elements were measured after the phantoms were completely gelled.

4.5.2 Force Sensing Test with PDMS Sensing Layer

For the force sensing test, a stress-strain curve of the sensing layer should be obtained first in order to confirm its nonlinear elasticity. A PDMS sample (5 mm × 5 mm × 1.5 mm) with 10:1 mixture ratio was prepared and the normal force was applied to the top of the PDMS through the tip of a force gauge (HF-10, ALIYIQI). The force resolution of the force
gauge was 0.01 N and the gauge was rigidly fixed on a 3-axis stage (460A Series, Newport). The deformation of the PDMS sample induced by the applied force was read through the scale on the y-axis of the stage with a displacement resolution of 1 μm. The stress-strain diagram of PDMS was then obtained based on the force and deformation measurement data.

Secondly, the force sensing test using the bulk crystal with the sensing layer were performed prior to the array test in order to verify the feasibility of force sensing of the acoustic wave technique. A bulk face-shear mode PMN-PT single crystal resonator (5 mm × 5 mm × 1 mm) was prepared and tested with a PDMS sensing layer. Copper electrode tapes were attached to the top and bottom surfaces of the resonator and connected to the impedance analyzer. The electrical impedance change (ΔZ_{AB}) at resonance was measured 4 times using the impedance analyzer. The photograph picture of the force sensing test setup using the bulk crystal resonator was shown in Fig. 4.13.

Finally, the force sensing test using the sensor array was conducted with a similar experimental setup. A 2 × 2 mm² tip with the force gauge was used to apply the forces to the array. The electrical impedance shift (ΔZ_{AB}) for all 36 elements was obtained. Fig. 4.14 shows the experimental setup for the force sensing test using the sensor array.

4.6 Results and Discussion
4.6.1 Tissue Elasticity Sensing Test Results

Through conventional stress/strain measurements, the stress-strain relationship of gelatin tissue phantoms with different WR was shown in Fig. 4.15. As the applied stress increased, the strain increased almost linearly. The slope indicated the Young’s modulus of each sample,
and it increased with gelatin weight ratio. The shear modulus of tissue phantoms can be calculated using Eqs. (4.20) and (4.21) and the result was shown in Fig. 4.16. It was found that the shear modulus increased from 124 kPa to 432 kPa when the gelatin weight ratio increased from 5 wt % to 30 wt %.

From the acoustic wave tactile sensor tests, the acoustic impedance changes for phantom with different weight ratios were measured 3 times. The averaged data was used for comparison with the simulation results. Fig. 4.17 shows the measured and calculated relative electric impedance shift of the bulk resonator as a function of shear modulus of tissue phantoms. The Modeling result was obtained using Eq. (4.8), with input parameters given in Table 1. It was observed that 45 ± 2.5 % of electrical impedance shift of the resonator was resulted, with the shear modulus of phantoms changing from 120 kPa to 430 kPa. Measured and calculated results were in a good agreement.

The electric impedance shift (ΔZ_{AB}) of 36 elements loaded with gelatin phantoms (5 wt % and 30 wt %) was also measured 3 times for averaging using the impedance analyzer. The varied acoustic load distribution on the array can be mapped by the measured electrical impedance shift, as shown in Fig. 4.18. Red dotted circles indicate the loaded position of gelatin samples. The larger electrical impedance shift of loaded elements was found from 30 wt % gelatin phantom than that from 5 wt % gelatin phantom, suggesting that the acoustic impedance difference of phantoms can be resulted from their different elastic properties. Specifically, 0.63 ± 0.04 % and 1.50 ± 0.04 % of electrical impedance changes were measured from elements loaded with 5 wt % and 30 wt % phantoms, respectively. The corresponded shear modulus was 124 kPa and 432 kPa for 5 wt % and 30 wt % phantoms,
respectively. The results reveal that the relationship between the resonator’s electrical impedance shift and the tissue’s acoustic impedance variation can be simply and quickly detected using proposed tactile sensor array for tissue elasticity measurement. However, compared to the bulk crystal sensing results, the array sensing result showed small electrical impedance variation ratio under same acoustic loads. This is because all surfaces of the sensing element in the array are enclosed by PDMS while surfaces of the bulk crystal sensor are exposed to the air. The enclosed PDMS acts as additional acoustic loads which can dampen surface vibrations of sensing elements, and thus, reduce the acoustic load sensitivity of the sensor array. Nonetheless, it is still noteworthy that more than two times higher electrical impedance shift was observed from 30 wt % phantom (1.5 %) compared with 5 wt % phantom (0.63 %), showing the possibility of proposed sensor array for elasticity sensing applications.

4.6.2 Force Sensing Test Results
The stress-strain diagram of PDMS was obtained based on the force and deformation measurement data, as shown in Fig. 4.19. The slope of the curve indicated the Young’s modulus and it increased when the stress and strain increased. This nonlinear elasticity of the sensing layer can be used for the force measurement. When the external force is applied to the sensing layer, its elastic modulus changes with the applied force. This change leads to the variation in the acoustic impedance, and thus, the force can be sensed by means of the sensor’s electrical impedance shift. For the model calculation, the constants $C_1 (= -6.6271)$ and $C_2 (= 6.4852)$ were obtained from a Mooney–Rivlin plot, as shown in Fig. 4.20. In Fig.
4.21, the calculated stress-strain diagram of PDMS using the third-order Mooney model was presented. The model result was found to agree well with the measured stress-strain diagram (Fig. 4.19). The calculated elastic modulus of PDMS was shown in Fig. 4.22. The elastic modulus increased as the strain increased. The increased elastic modulus leads to the increased acoustic impedance (Eq. 4.9), and thus, the shift of the output electrical impedance (Eqs. 4.28 and 4.31). The degree of the slope in the Fig. 4.22 is directly related to the softness of the sensing layer. The soft sensing layer with low elastic modulus can offer larger strain under a relatively small force, resulting in the high force sensitivity (i.e. the amount of shift in the electrical impedance of sensing elements under the applied unit force). This slope degree and softness of sensing layers can be controlled by the ratio of the PDMS mixture. As mentioned in the Chapter (4.4.1), for the PDMS sensing layer in prototyped tactile sensor array, a 10:1 mixture ratio (recommended ratio for general purposes) of pre-polymer and curing agent (or cross-linking agent) was applied. However, different mixture ratios can be used in order to obtain softer or stiffer PDMS layers. For example, PDMS with 30:1 ratio can have much lower elastic modulus ($E < 0.42$ MPa) than that of PDMS with 10:1 ratio ($E < 2.04$ MPa) [153]. On the other hand, the maximum force sensing limit can be obtained theoretically when the strain induced by compression is closed to 1, as can be seen in Fig. 4.23. In practice, however, applying such high forces to the sensing layer is not possible. Furthermore, the stiffness of the sensing element also limits the applicable force range. Based on these considerations, the maximum force sensing limit of the prototyped tactile sensor was determined by 4 N.
The force sensing test results using bulk resonators were shown in Fig. 4.24. The normal force ranging from 0.1 N to 5 N was applied to the sensing layer on the resonator. Electrical impedance shifts for 1 N, 2 N, 3 N and 4 N forces were measured to be 18.10 ± 0.86 %, 37.48 ± 0.55 %, 45.89 ± 0.51 %, and 47.44 ± 0.52 %, respectively. It was found that the electrical impedance rapidly increased up to 2 N due to the increased acoustic impedance of the sensing layer. This was caused by the fact that the PDMS sensing layer was hardened or saturated by the external force. As a result, the sensing layer’s stiffness and acoustic impedance were no longer able to change upon external force input. The measured results were compared with the modeling results in Fig. 4.25 and both results were observed to be in a good agreement.

Fig. 4.26 shows the force sensing test result using the 6 × 6 sensor array. Normal forces were applied to the array and the electrical impedance of the 36 elements was scanned. Red dotted squares indicate the areas over which forces were applied. The amount of the electrical impedance shift of loaded elements increased as with the applied force. 0.36 ± 0.01 %, 0.43 ± 0.07 %, 0.87 ± 0.02 %, and 1.56 ± 0.10 % of electrical impedance shifts were found for 1 N, 2 N, 3 N and 4 N loaded sensor arrays, respectively. These shifts were smaller than those of the bulk crystal sensing results. The damping effect of the sensing element induced by enclosed PDMS can explain this difference between the bulk crystal sensing and array sensing results similar to the elasticity sensing results (Chapter. 4.6.1). The force sensing test results suggests that the prototyped sensor array can be promising for the external force sensing as well as elasticity sensing. The sensing performance specification
based on the test results for the proposed acoustic wave tactile sensor array is summarized in Table 4.4.

4.7 Conclusion

In conclusion, an acoustic wave tactile sensing array using face-shear mode PMN-PT crystals was designed, fabricated, and tested for tissue elasticity and external force sensing. Phantoms with a variety of Young’s moduli and normal forces were applied to the center of the sensor array. The changes in the electric impedance of the loaded sensing elements for all 36 elements were measured. It was found that the loaded phantom’s elasticity and the magnitude of the applied force can be measured by collecting the electric impedance of the sensor array. The simulation results obtained using the KLM model and the Mooney-Rivlin model showed a good agreement with the experimental results. The proposed tactile sensor array has a great potential for a number of advanced biomedical applications and robotics applications such as elasticity sensors for MIS tools, novel artificial skin for surgical robotics, and bio-imaging systems.
Figure 4.1 Minimally invasive surgery with specially designed tools [Intuitive Surgical, Inc., 2009].
Figure 4.2 Elastic moduli of different type of tissues.
Figure 4.3 Elasticity sensing using typical electromechanical sensors. (a) A beam with a hard object and (b) deformed beam with a soft object.
Figure 4.4 (a) Principle of static elastography; (b) Slow and repetitive compression is applied to tissues using an ultrasound transducer in order to generate strain for hard and soft masses.
Figure 4.5 Principle of shear wave elastography. Radiation force is induced by a focused ultrasound beam and this force produces transversely oriented shear-waves within tissue.
Figure 4.6 The schematic of the elastic sensing method.
Figure 4.7 The schematic of the force sensing method.
Figure 4.8 Array fabrication process of the $6 \times 6$ array using face-shear mode PMN-PT single crystal resonators.
Figure 4.9 The row (top) electrode and the column (bottom) electrode lines connected to $6 \times 6$ array. The element located in the 3$\text{rd}$ row and 3$\text{rd}$ column can be addressed by selecting the top electrode T3 and the bottom electrode B3.
Figure 4.10 The block diagram of the array impedance measurement system and the switch circuit.
Figure 4.11 The fabricated $6 \times 6$ array with the switch and microcontroller circuits.
Figure 4.12 The fabricated tissue mimicking phantoms with different weight ratios.
Figure 4.13 The photograph picture of the force sensing test setup using the bulk crystal resonator.
Figure 4.14 The experimental setup for the force sensing test using the sensor array.
Figure 4.15 The stress-strain measurement result for gelatin tissue phantoms with different weight ratio.
Figure 4.16 The shear modulus of tissue phantoms as a function of gelatin concentration (wt %).
Figure 4.17 The measured and calculated relative electric impedance shift of the bulk resonator as a function of tissue phantom’s shear modulus.
Figure 4.18 Tissue phantom elasticity sensing test result using the $6 \times 6$ sensor array; (a) 5 wt % gelatin phantom (b) 30 wt % gelatin phantom. Red dotted circles represent the position of each gelatin sample.
Figure 4.19 The measured stress-strain diagram of PDMS sensing layer.
Figure 4.20 Calculated Mooney–Rivlin plot ($\sigma/\lambda-1/\lambda^2$ verses $1/\lambda$); constants $C_1$ (= -6.6271) and $C_2$ (= 6.4852) can be obtained from the linear fitted curve in the plot.
Figure 4.21 The calculated stress-strain diagram of PDMS using the third-order Mooney model.
Figure 4.22 The calculated elastic modulus of PDMS as a function of strain.
Figure 4.23 The calculated elastic moduli of PDMS with strain range from zero to one.
Figure 4.24 Force sensing test results for bulk resonators with external force ranging from 0.1 N to 5 N.
Figure 4.25 The measured and calculated relative electric impedance shift of the bulk resonator as a function of applied force.
Figure 4.26 Force sensing test result using the 6 × 6 sensor array; (a) 2 N (b) 3 N and (c) 4 N forces were applied. Red dotted squares indicate the area over which forces were applied.
Table 4.1 Comparisons of different types of electromechanical tactile sensor technologies.

<table>
<thead>
<tr>
<th></th>
<th>Resistive</th>
<th>Capacitive</th>
<th>Piezoelectric</th>
</tr>
</thead>
<tbody>
<tr>
<td>Principle</td>
<td>Piezo-resistive effect</td>
<td>Electrical capacitance variation</td>
<td>Direct piezoelectric effect</td>
</tr>
<tr>
<td>Sensitivity</td>
<td>Fair</td>
<td>Excellent</td>
<td>Good</td>
</tr>
<tr>
<td>Resolution</td>
<td>Fair</td>
<td>Good</td>
<td>Good</td>
</tr>
<tr>
<td>Repeatability</td>
<td>Poor</td>
<td>Good</td>
<td>Good</td>
</tr>
<tr>
<td>Advantage</td>
<td>Easy integration; Cost-effectiveness</td>
<td>Immunity to temperature variation; Simple internal structure</td>
<td>Low cost; Simple structure; Simple signal process</td>
</tr>
<tr>
<td>Disadvantage</td>
<td>Low sensitivity; Hysteresis</td>
<td>Susceptibility to external influence (electromagnetic noise)</td>
<td>Lack of DC response due to the nature of the piezoelectric effect</td>
</tr>
<tr>
<td>Elasticity</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>measurement</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Elasticity</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>measurement</td>
<td></td>
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Table 4.2 Advantages, disadvantage and applications of different types of elasticity imaging methods.

<table>
<thead>
<tr>
<th>Advantage</th>
<th>Static elastography</th>
<th>Dynamic elastography</th>
<th>MR elastography</th>
<th>Shear-wave elastography</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Easy implementation; Simple mechanism;</td>
<td>Quantitative approach; Noninvasive stiffness assessment;</td>
<td>Quantitative approach;</td>
<td>Real-time process; User-skill independent;</td>
</tr>
<tr>
<td>Disadvantage</td>
<td>Qualitative approach; Difficult interpretation; Noisy or misleading images; Pressure and movement of the transducer control needed; Plenty of practice is required;</td>
<td>High cost; Limited availability; Lack of real time imaging; External vibrators needed; Long acquisition time (10 to 15 min);</td>
<td>Weak shear waves (few microns of displacement); Over-heating and over acoustic power problems;</td>
<td></td>
</tr>
<tr>
<td>Application</td>
<td>Measurement of the relative stiffness and size of the lesion, lymph nodes, and thyroid nodules</td>
<td>Non-invasive assessment of breast cancer, liver fibrosis, liver tumors, and skeletal muscle</td>
<td>Non-invasive diagnosis of cirrhosis, severe fibrosis, and diffuse organ diseases.</td>
<td></td>
</tr>
</tbody>
</table>
Table 4.3 Design specification for the fabricated 6 × 6 sensor array.

<table>
<thead>
<tr>
<th>Sensing element #</th>
<th>Array dimensions</th>
<th>Element dimensions</th>
<th>Pitch</th>
</tr>
</thead>
<tbody>
<tr>
<td>36 (6 × 6)</td>
<td>9 × 9× 1 mm³</td>
<td>800 × 800 × 500 μm³</td>
<td>1.3 mm</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Crystal type</th>
<th>Crystal mode</th>
<th>Support material</th>
<th>Operational frequency</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMN-PT</td>
<td>Face-shear mode</td>
<td>PDMS</td>
<td>1 MHz</td>
</tr>
</tbody>
</table>

Table 4.4 Sensing performance specification for the proposed acoustic wave tactile sensor array.

<table>
<thead>
<tr>
<th></th>
<th>Shear modulus sensing</th>
<th>Force sensing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Range</td>
<td>~ 432 kPa</td>
<td>~ 1 N</td>
</tr>
<tr>
<td>Sensitivity</td>
<td>23.52 Ohm/MPa</td>
<td>19.27 Ohm/N</td>
</tr>
<tr>
<td>Resolution</td>
<td>4.25 kPa</td>
<td>5.19 mN</td>
</tr>
</tbody>
</table>
CHAPTER 5
CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE WORK

5.1 Conclusions
The primary goal of this dissertation was to investigate the design, fabrication and testing methods of novel shear mode piezoelectric sensors for specific applications: (1) high temperature acceleration or vibration sensing; and (2) acoustic wave tactile sensing. For the first time, this research demonstrated a piezoelectric accelerometer using shear mode YCa$_4$O(BO$_3$)$_3$ (YCOB) crystals for vibration sensing at 1000 °C. This is also the first study to utilize the acoustic wave sensing technique with face-shear mode Pb(Mg$_{1/3}$Nb$_{2/3}$)O$_3$-PbTiO$_3$ (PMN-PT) crystals for novel tactile sensing applications including elasticity sensing and force sensing. In the previous chapters, simulation and experimental results for each sensing technique were presented. Based on the presented results, the following conclusions can be drawn from the high temperature accelerometer:

- YCOB single crystals were used for the sensing elements in the prototyped accelerometer because YCOB crystal can possess high resistivity without any phase transition at high temperatures up to 1500 °C. As expected, stable performance of the sensor was observed at the tested temperature range. However, the charge sensitivity at 1000 °C was measured to be a little higher than the room temperature value. This can be attributed to the temperature dependence of the piezoelectric coefficients of YCOB crystals.
• The proposed sensor was designed to operate in shear mode. In this case, the crystals were separated from the base and housing and thus thermal transient and base bending effects at high temperature could be reduced. It is notable that the sensor maintained good physical and electrical contact at high temperatures since a bolt and nut assembly was used instead of an adhesive assembly.

• It was observed that bulk metal Inconel as electrodes has great resistance to high temperature oxidation and corrosion, resulting in high temperature piezoelectric accelerometers with excellent sensing performance without degradation at high temperatures when compared with accelerometers with thin film electrodes such as gold and platinum films.

• Overall stable sensitivity for the prototyped sensor was obtained throughout the tested frequency, temperature and acceleration ranges. Dynamic response model and thermal expansion model were performed and were successfully verified by experimental results. In addition, the dwelling test revealed that the prototype sensor can work at 1000 °C with reasonably stable sensitivity for more than 9 hours.

For the acoustic wave tactile sensor, the following conclusions can be drawn:

• The tactile sensing technique studied in this dissertation utilizes the relationship between the electrical impedance of the piezoelectric resonator and the acoustic impedance of the surface load. It is noteworthy that this technique can be applied effectively for elasticity sensing and force sensing since the acoustic impedance is
directly related to the material’s elastic properties. KLM model and Mooney-Rivlin model were successfully established to predict the force and elasticity sensing performance of the proposed tactile sensor.

- The acoustic load sensitivity, which indicates the ratio of electrical impedance change to the applied acoustic load impedance, was compared for piezoelectric resonators in different modes including thickness, thickness-shear and face-shear mode. It was found that the face-shear mode resonator possessed at least 20 times higher sensitivity than other mode resonators. The revealed unique performance of face-shear mode piezoelectric resonators can be favorable for an acoustic wave device.

- For elasticity measurement tests the electric impedance shift of resonators loaded with gelatin phantoms was obtained using the impedance analyzer. The elasticity difference of tissue phantoms leads to acoustic load impedance variation. This varied acoustic load can be detected by the electrical impedance shift of resonators. Hence, tissue’s acoustic impedance sensing can be a simple and rapid method for tissue elasticity measurement.

- The force measurement using the prototyped tactile sensor was achieved by utilizing the nonlinear elasticity of a sensing layer on the top of the sensor. When the external force is applied to the sensing layer, its elastic modulus varies with the applied force. This change leads to the variation in the acoustic impedance, and thus, the force can be sensed by means of the sensor’s electrical impedance shift measurement. The results showed that the prototyped tactile sensor can be used to measure applied force effectively.
• Load distribution mapping tests were carried out using the $6 \times 6$ tactile sensor array. The distributions of the elasticity of tissue phantoms and the applied pressures on the sensor array were successfully mapped by the electrical impedance shift measurement.

5.2 Recommendations for Future Work

Recommendations for future research directions and strategies can be offered to each research topic. In terms of the high temperature accelerometer, the sensor size can be reduced to the micro-scale for low profile sensor mounting in industrial applications. Since the use of devices in aerospace applications is restricted by space and weight issues, the miniaturized device will be more suitable for this purpose. In addition, small sensors can lead to great dimensional stability due to low thermal expansion, which is preferred for higher temperature applications. Future work could also incorporate the sensor packaging with the passivation coating to reduce environmental noise such as thermal effects and electromagnetic interference noise (EMI), which can affect the charge sensitivity significantly. Further research should also address thermal and surface effects of YCOB single crystal and Inconel alloy for higher temperature applications (> 1300 °C).

For the acoustic wave tactile sensor array, the switching circuit, such as a multiplexer configuration, needs to be further developed and integrated with the sensor array to achieve rapid, precise, and real-time scanning using the sensor array. The sensor array developed in this work is not able to be bent or stretched since it consists of crystals, metal wires, and/or
hard substrates. In future work, the sensor array can be integrated with flexible circuitry for advanced applications such as artificial or smart skins for humanoid robots. Finally, the experiments have been performed only for a tissue mimicking phantom. In vitro or in vivo experiments need to be performed to verify the feasibility of tactile sensing using the prototyped sensor array in actual medical applications.
REFERENCES


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