Flexible ultrasound transducers have significant advantages over the non-conformal transducer probes available commercially for applications like medical ultrasonography, NDT, and other similar applications requiring contact-based imaging. They can be enveloped over curved surfaces and non-planar objects, giving a better volumetric image. In this study, a flexible 2D array was designed, fabricated and tested to demonstrate its effectiveness.

In the 3MHz array we fabricated, there were 64 elements arranged as an $8 \times 8$ array with a mean pitch of 2mm. The array was obtained by mechanically dicing a $20mm \times 20mm$ PZT-5H 1-3 composite wafer. The array was then epoxy-bonded to a thin gold foil on one side, which acted as the common ground electrode, while the other side with sputtered gold acted as another electrode. 64 coax cables were soldered on the sputtered side of each element, and 2 more were soldered on the common ground electrode. This assembly was held fixed under a press, and was potted using a rubber solution (Sylgard 170) that demonstrated the required flexibility and electrical insulation requirements. The array was housed in an aluminum tube. The free ends of the wires were soldered onto a PCB which was mounted over the housing, and a cover was screwed on the top. A compact and modular flexible transducer package with thickness 15mm and mean diameter of 57mm was fabricated.

The fabricated 3MHz 2D array was successfully tested for conformability by deforming it using spheres of different diameters ($3'''$). The average capacitance was over 90pF and the impedance value below 750Ω at resonance frequency. Pulse-echo experiments using a pulser-receiver demonstrated a peak-peak voltage ($V_{p-p}$) of 25mV for an input signal with energy $2\mu J$.  

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Flexible Two-dimensional Ultrasonic Transducer Array: Design, Fabrication and Characterization

by

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A thesis submitted to the Graduate Faculty of North Carolina State University in partial fulfillment of the requirements for the degree of Master of Science

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Dedication

To my parents, family, friends and God.
Biography

Vishal Kulkarni did most of his schooling in Mumbai, India. He was interested in science and technology since a very young age. He completed his Bachelor of Engineering degree in Electronics and Telecommunication from University of Mumbai. During his undergraduate years, he participated in and won many robotics competitions. After graduation, he decided to pursue higher education in the Department of Mechanical and Aerospace Engineering at North Carolina State University where he was fortunate enough to work with some of the most knowledgeable and wise people. During the second year, he began working for Dr. Xiaoning Jiang, from whom he has gained a lot of knowledge and understanding of the field of ultrasonics and micro-engineering. He is interested in ultrasonic transducer design, microfabrication, and device testing and characterization.
Acknowledgments

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I am grateful to my friends who have always been there for me. You all are awesome! I’m eternally indebted to my parents, Meena and Ashok Kulkarni without whom I would not be the man I am. I love you both with all my heart.

Finally, I thank God for everything.
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Chapter 1

Introduction

Ultrasound has been used as an effective diagnostic tool in the manufacturing industry and medicine for years. Depending on the application, different kinds of ultrasonic transducer are designed and manufactured. The development of two-dimensional ultrasound transducers has enabled the generation of real time volumetric imaging. This has enabled a better visual representation of the target of interest. For example, the faults and cracks inside a metal structure can be better illustrated in three dimensions by using a 2D ultrasonic transducer array. However, a lot of structures to be evaluated have a certain curvature. Traditional, rigid transducers do not work efficiently on such curved surfaces. In such cases, flexible transducer arrays can be used to conform to the targets curvature and give better images. Flexible linear arrays have been used for NDE of pipes and other curved structures but fabricating a flexible two-dimensional transducer array presents certain challenges like complex interconnections or bulky controlling mechanisms. There is a need for a cost-effective, simple yet effective method to fabricate flexible two-dimensional ultrasonic transducers.

1.1 Summary of Thesis Contribution

This thesis will present the design, fabrication and characterization of an 8x8 3.3 MHz flexible 2D array. The transducer array was designed first, followed by experimentation of different structural and bonding materials. Then, the actual device prototype was fabricated and tested. Electrical and acoustic characterization was performed for all 64
elements. Pulse-echo tests were carried out in the 100 kHz to 10 MHz range. The average pulse-echo output for a 100 V input pulse was 231.4 mV while the average loop sensitivity was -53.1 dB. The pulse-echo tests were also performed for a flat and curved surface to demonstrate the differences in pulse-echo strengths and the times of flight. Finally, conformity and structural strength tests were performed using 3” and 4” diameter steel spheres.

1.2 Outline of the Thesis

A brief summary of the results and the outline of the thesis is included in Chapter 1. The second chapter contains ultrasound fundamentals, applications, imaging fundamentals, basic structure of an ultrasound imaging system and general description of a typical transducer. It also discusses transducer arrays and the importance of conformable transducers for curved surface imaging. It also contains a brief review of related work, highlighting the merits and demerits of existing transducer array technologies. Three kinds of transducer arrays will be discussed - rigid 2D, flexible 1D and flexible 2D transducer arrays.

The third chapter explains the design process used in the development of the transducer. Design criteria which impacted the final layout are explained. The selection of materials is justified and the KLM model of the device is constructed.

The fourth chapter presents the actual prototyping process and the problems faced during the prototyping stage. It also lists the modifications that were carried out to overcome these problems.

The fifth chapter describes the characterization techniques and the significance of the parameters measured. The test setup and procedure is also explained in this chapter. Finally, the results are organized with the help of graphs and tables.

Finally, the sixth chapter summarizes the capabilities of the device and its functionality. Potential future research areas are discussed which could improve the device specifications.
Chapter 2

Background

Ultrasonics is defined as the band of sound above 20KHz. Ultrasonics has applications in various disciplines like chemistry, physics, engineering, biology, food industry, medicine, oceanography, seismology, etc. Nearly all of these applications are based on two unique features of ultrasonic waves:

- Ultrasonic waves travel about 100,000 times slower than electromagnetic waves. This provides a way to display information in time, create variable delay, etc.

- Ultrasonic waves can easily penetrate opaque materials, unlike many other types of radiation. This provides a highly desirable way to probe and image the interior of opaque objects.

2.1 Applications of Ultrasound

The typical applications of ultrasonic waves over the full spectrum of frequencies are shown in Figure 2.1.
2.1.1 Medical Applications

Medical sonography, also called ultrasonography is a diagnostic technique used to visualize muscles, tendons and other internal organs and determine their size and structure or locate any lesions with real-time images. It is also used to visualize the fetus during prenatal care. Other areas of medicine where non-invasive acoustic imaging of the body is useful are cardiac, urological, and ocular imaging. Focused high-energy ultrasound pulses can be used to break kidney stones and gall bladder stones into small fragments that pass through the body without much difficulty. High Intensity Focused Ultrasound (HIFU) is used to ablate tumors or other tissue non-invasively. Ultrasound at various intensities is used to deliver chemotherapy to brain cancer cells and various drugs to other tissues in a procedure called acoustic targeted drug delivery (ATDD). Low intensity ultrasound is useful for therapeutic bone regeneration.

2.1.2 Non-Destructive Evaluation

Nondestructive evaluation refers to a technology which allows the in situ inspection of a component without adversely affecting its usefulness or structural integrity. This is especially important in applications in the aerospace, mechanical, civil and energy industry to discover defects in key structures due to internal changes or wear. Ultrasonic waves can be used to determine the location, size and shape of the internal defects without harming the structure. Most metals, plastics and aerospace composites can be examined
using ultrasound NDE. Ultrasonic waves at lower frequencies (e.g. 30-500KHz) can also be used to inspect materials with less density, like wood, concrete and cement.

2.1.3 Other Applications

Other applications of ultrasound include underwater sensing, cleaning, drilling, cavitation, humidification, welding etc. Real time structural health monitoring and material processing by monitoring material and structural properties are the relatively modern applications of ultrasound [1].

2.2 Ultrasound Imaging

Ultrasound imaging is a non-invasive technique which utilizes high-frequency sound waves to image internal and external features of suitable targets. It is based on the principle of reflection of sound. When an acoustic wave traveling through a medium encounters a change in material properties like sound velocity, elasticity or density, there occurs reflection, refraction and scattering of the waveform, resulting in a backwards propagating echo.

Acoustic imaging is well suited for certain materials due to uniformity of sound velocity through them. Given a constant speed of sound, the distance from the target can be calculated by measuring the time interval between the transmitted and reflected sound and using the simple kinematic equation

\[ distance = \frac{velocity \times time}{2} \]  \hspace{1cm} (2.1)

The factor of a half is due to the fact that the time taken for the transmitted pulse to reach the target is the same as the time taken for the echo to be received.

2.3 Ultrasound Imaging System

The typical components of an ultrasonic imaging system are given in Figure 2.2.
Figure 2.2: Components of a typical ultrasound imaging system . [2]

The core component is the ultrasonic transducer, which converts electrical energy from the pulser to mechanical energy, creating sound waves, and vice versa. The received electric signals are amplified and filtered at the analog front-end and then converted to digital data using the Analog to Digital Converters (ADCs). The beamformers are used for transducer arrays to compensate for time of flight variation between different array elements and the focal zone. The digital signals are then processed and an image is generated and displayed.

2.4 Ultrasonic Transducer

A transducer is a device that converts energy from one form to another. An ultrasonic transducer converts electrical energy to mechanical energy which creates the sound waves, and vice versa. They operate at frequencies greater than 20KHz [3]. Based on the physical principles which permit the radiation of ultrasound, some types of transducers are described below

- electrostatic
  It consists of a fixed electrode and a movable electrode, charged electrostatically in opposite polarity; motion of the movable electrode changes the capacitance between the electrodes and thereby makes the applied voltage change in proportion to the amplitude of the electrode’s motion. [4]

- electrodynamical
  It contains at least one electric coil placed in a static magnetic field and which can
move about a rest position in a vertical free space. It is widely used in loudspeakers. [5]

- magnetostrictive
  It consists of a large number of plates or laminations of a magnetostrictive material (like nickel) arranged in parallel with one edge of each laminate attached to the surface to be actuated or sensed. [6]

- piezoelectric
  It typically involves the use of a piezoelectric ceramic material (like PZT) sandwiched between two electrodes that provide attachment points for electrical contact. [6]

- capacitive
  It comprises of two parallel metallic plates separated by a dielectric material. The value of the capacitance changes due to change in the value of input quantity to be measured. [7]

2.4.1 The Piezoelectric Effect

Of the above transducer types, piezoelectric transducers are of main interest for medical and NDE applications. Piezoelectric transducers typically use special materials called piezoceramics, piezocomposites or piezoelectric polymers. Typical advantages of piezoceramics are their compact, rugged mechanical design, high efficiency and great range of operation temperature. The standard piezoceramic material is lead zirconium titanate (PZT). Piezocomposites generally have the advantage of lower acoustic impedance and higher electromechanical coupling coefficient as compared to piezoceramics. Piezopolymers have the benefits of higher elasticity and better strain bearing ability as compared to piezoceramics and piezocomposites.

When a piezoelectric material is subjected to an external stress, voltage potential difference is produced between its surfaces. This is called the piezoelectric effect. Conversely, a voltage applied across two particular surfaces of a piezoceramic results in mechanical strain, which is called the inverse piezoelectric effect.

Here we use piezoceramic as an example to introduce piezoelectric effect in detail. Any polycrystalline ceramic consists of many randomly oriented crystals, forming do-
mains which cumulatively account for the bulk properties of the material. In certain ceramics called ferroelectric ceramics, most of these domains can be aligned by application of a strong direct current (DC) field, thus turning the ferroelectric ceramic into a piezoelectric ceramic. This process, usually carried out at elevated temperatures, is called poling of the ceramic. More domains aligned result in higher piezoelectric effect. An externally applied DC voltage causes mechanical deformation, but an applied AC voltage causes the ceramic to alternate in size. The ceramic has its natural resonant frequency. If an AC field is applied at this resonant frequency, it oscillates with a higher efficiency resulting in acoustic waves with more intensity.

2.4.2 Basic Transducer Structure

A schematic of a typical piezoelectric transducer is shown in Figure 2.3 [8].

![Figure 2.3: Basic structure of a piezoelectric transducer](image)

The main components are the active element, backing and front layer. A brief overview of these components is given below and each component will be discussed in
1. Active Element

The active element, which is a piezoelectric structure, converts electrical energy to ultrasonic sound waves as a transmitter. It can also act as a receiver converting the received ultrasound waves to electrical signals. Commonly used materials are polarized ceramics cut in a variety of manners to produce different wave modes. Other materials and structures such as piezopolymers, relaxor single crystals and composites are being employed for certain applications.

2. Backing Layer

The backing is usually used to control the vibration of the transducer by absorbing the energy radiating from the back face of the element. Typically, it is a layer of highly attenuative, high density material.

3. Front Layer

The front layer serves as an interface between the transducer element and the testing environment. It acts as an acoustic transformer between the high acoustic impedance of the active element and the low acoustic impedance of the medium (like water) or wedge. This is often called the matching layer.

4. Interconnection

The front and back electrodes need to be connected to an external cable or circuit. Miniature cables are bonded to these electrodes using bonding agents like conductive silver epoxy or solder [9]. However, in transducer arrays with many elements, flexible circuits are usually used to reduce physical bulk and avoid the inherent difficulties with wire-bonding numerous elements.

2.5 Transducer Arrays

The configuration shown in Figure 2.3 is for single element transducers. However, single element transducers are used infrequently except for some high frequency applications where transducers arrays are difficult to fabricate. For lower frequency applications,
transducer arrays are widely used because each element is individually pulsed to create an acoustic wavefront and a large number of elements increases focusing and steering capability and can increase coverage area as well. Phased array transducers can have as many as thousands of elements. The common types of phased array transducers are summarized in Table 2.1. The schematic for each is shown in Figure 2.4.

Table 2.1: Transducer Array Element Patterns [10].

<table>
<thead>
<tr>
<th>No.</th>
<th>Element Patterns</th>
<th>Advantages</th>
<th>Disadvantages</th>
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<tr>
<td>1</td>
<td>1D Linear</td>
<td>Easily manufactured, focuses in one axis at different depths and angles</td>
<td>Requires large size for deep focusing, beam divergence increases with angle and depth, no skewing ability</td>
</tr>
<tr>
<td>2</td>
<td>2D Square</td>
<td>Steering capability in 3D, spherical or single-axis focus</td>
<td>Complex design and manufacturing process, requires large number of pulser/receivers and complex programming</td>
</tr>
<tr>
<td>3</td>
<td>1.5D Square</td>
<td>Excellent steering capability, focuses in one axis at different depth and angles, reduces unwanted beam artifacts</td>
<td>Complex design and manufacturing process, requires large number of pulser/receivers and complex programming</td>
</tr>
<tr>
<td>4</td>
<td>1D Annular</td>
<td>Spherical focusing at different depths, very good for detecting small reflectors</td>
<td>No steering capability, requires large aperture for sharp focusing, requires complex programming</td>
</tr>
<tr>
<td>5</td>
<td>2D Annular</td>
<td>Elliptical or spherical beam steering capability at different depths and angles</td>
<td>Complex design and manufacturing process, requires large number of pulser/receivers and complex programming, generates longitudinal waves only</td>
</tr>
<tr>
<td>6</td>
<td>1D Circular</td>
<td>Elliptical or spherical beam with steering capability at different depths, typically used for curved targets</td>
<td>Complex design and manufacturing</td>
</tr>
</tbody>
</table>
In addition to the above general types, customized arrays offer many possibilities for adaptation to certain demanding applications. Examples include curved arrays and flexible arrays.

2.6 Flexibility and Conformability

A lot of structures to be examined by ultrasonic imaging are not completely flat. They can be irregular shaped and have curved surfaces. The traditional rigid ultrasonic transducers have wedges which are mismatched to the irregular surface of the component. This mismatch creates an irregular, thick coupling layer between the transducer wedge and the surface under consideration, leading to distortions and sensitivity loss. Dispersion of acoustic energy also occurs because of the sound wave not being incident onto the surface. All these factors cause a degradation of transducer performance. Mechanical scanning using a small aperture transducer can be used to get high resolution images but real-time imaging is impossible in this case.
A thin conformable array can be wrapped around the extremities and curved surfaces of the component. This configuration results in a thin coupling layer of uniform thickness and incident acoustic waves with less dispersion loss. The flexible transducer can also provide multiple, unique points of visualization of the internal objects or flaws. No mechanical scanning is needed for obtaining high resolution images and thus, good images can be obtained even by inexperienced users. Therefore, the use of conformable transducers has many advantages over rigid transducers when obtaining images of components with curved or irregular surfaces.

2.7 Previous Research

There has been research on many different kinds of ultrasonic transducers consisting of piezoelectric materials like piezoceramics, piezoelectric single crystals, piezopolymers, and piezocomposites in various array configurations. The design and fabrication processes vary greatly depending on the kind of transducer to be manufactured for specific applications. As such, some of the research relevant to the topic has been discussed below.

2.7.1 Rigid Arrays

Hurrell and Duck [11] fabricated a 2D hydrophone array using piezo-electric PVDF for underwater applications. They used a 9m thick prepoled polyvinylidene fluoride PVDF film as the piezoelectric material with a rubber silicone backing. One side of the
PVDF was left without an electrode so that the electrical charge would be coupled only at the interconnection ends. However, this meant that there were no physical kerfs in between the coupled areas. PVDF also has a low dielectric constant, low Curie temperature, and low acoustic impedance which makes it unsuitable for many NDE applications.

Ratsimandresy et al. [12] fabricated a 3 MHz 64 × 64 element array by partial dicing of the piezoceramic array elements. Even though he mentioned that the ideal design would comprise complete dicing of the elements, the idea was not used due to inherent complications with dicing such a large number of elements. Matching layers, made of epoxy and polyurethane, were used for acoustic impedance matching. There also existed 53 dead elements in the prototype due to difficulties in processing.

![Diagram of partial dicing](image)

Figure 2.6: Partial dicing

The problem of complete dicing of elements was addressed by Light et al. to fabricate various NxN transducer arrays with different array elements count and different center frequencies [13][14]. A flexible polyimide circuit was used for interconnection and curvilinear and linear rigid transducers were fabricated. A limited number of array elements in each transducer were wired. A lossy epoxy was used as the backing and for the stability of the electrical connections. The entire transducer was enclosed in an aluminum shell for protection. The problem of dead elements (10
2.7.2 Flexible Arrays

Flexible 1D Arrays

In order to produce a flexible transducer a variety of methods have been attempted. Bowen et al. [15] fabricated a transducer, flexible in one dimension by using lapped PZT-5A fibers. Air gaps between the fibers provided the flexibility. Kobayashi et al. developed a flexible transducer for high temperatures up to 160°C. It consisted of a film created by dispersing PZT particles in a sol-gel (PZT) creating a PZT/PZT composite film [16][17]. The stainless steel foil acted as a flexible electrode as well as the substrate for the sol-gel spray. The other electrode was conductive silver paste on top of the PZT thick film. Two dimensional arrays could be created by patterning the top electrodes but there would be excessive crosstalk due to absence of kerfs and low sensitivity due to dispersed PZT particles instead of a solid piezoceramic. Hayward et al. [18] developed a 1D flexible array using PVDF as the receiver part.

![Flexible 1D array using PVDF as receiver](image)

Flexible 2D Arrays

2D arrays combine the advantages of conformability and real-time volumetric imaging. Hayward et al. [18] proposed a theoretical modeling strategy for the design of flexible ultrasonic arrays. It involved embedding piezoelectric platelets in a polymer matrix so that flexibility in two-dimensions could be attained. Harvey et. al. [19] used a novel method to fabricated 2D arrays by replacing the platelets by piezoelectric fibers with a 250m diameter. The place-and-fill technique was used in which the piezoceramic fibers were manually positioned and then filled with epoxy to arrange the PZT fibers in periodic and random arrangements as shown in Figure 2.5.
The frequency had an inverse relation with the fiber thickness and had to be limited to 3 MHz due to difficulty in handling very thin fibers.

2.8 Discussion

Thus, we see that different designs and manufacturing techniques are utilized based on the applications and critical properties under consideration. It was inferred that the complete dicing of the array elements holds an advantage over partial dicing or kerfless arrays. The use of a thin flexible foil as a substrate and an electrode provided the necessary flexibility and electrical connectivity. The various interconnection methods used were useful for a large number of elements but posed certain problems such as reduction of flexibility due to addition of an extra layer, dead elements due to improper alignment, long procurement time and extra cost. The use of lossy epoxy for the dual purpose of transducer backing and interconnection stability was explored. The platelet-polymer design showed a lot of promise.
Chapter 3

Design

Several crucial characteristics have to be considered when laying down the design guidelines for flexible 2D ultrasonic transducers. First of all, it is important to understand the physics behind ultrasound wave propagation associated with the design requirements which is included in section 3.1.

Specific requirements for the components of the transducer structure are discussed in section 3.2. Finally, the analysis of key transducer characteristics like center frequency, bandwidth and loop sensitivity are included in section 3.3.

3.1 Ultrasound Wave Propagation

3.1.1 Fundamentals

All materials are comprised of atoms, which can be externally forced into vibrational motion about their resting, or equilibrium position. In solids, sound waves can propagate in four principal modes creating different types of waves: longitudinal, shear, surface and plate. Based on the directions of vibrations, some different waves are mentioned in table 3.1 [20].
Table 3.1: Examples of wave types in solids and corresponding particle vibration direction [20].

<table>
<thead>
<tr>
<th>Wave Type in Solid</th>
<th>Particle Vibrations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Longitudinal</td>
<td>Parallel to wave direction</td>
</tr>
<tr>
<td>Transverse (shear)</td>
<td>Perpendicular to wave direction</td>
</tr>
<tr>
<td>Surface (Rayleigh)</td>
<td>Elliptical orbit (symmetrical mode)</td>
</tr>
<tr>
<td>Plate Wave (Lamb)</td>
<td>Component perpendicular to surface</td>
</tr>
<tr>
<td>Plate Wave (Love)</td>
<td>Parallel to plane layer, perpendicular to wave direction</td>
</tr>
<tr>
<td>Stoneley (Leaky Rayleigh waves)</td>
<td>Wave guided along interface</td>
</tr>
<tr>
<td>Sezawa</td>
<td>Antisymmetric mode</td>
</tr>
</tbody>
</table>

Longitudinal and shear modes are the two modes of propagation commonly used in ultrasonic testing. The particle movements of the medium, which are responsible for longitudinal and shear wave propagation are illustrated in Figure 3.1.

![Figure 3.1: Particle motion and wave propagation in longitudinal and transverse waves](image-url)
As can be seen, the longitudinal waves travel through a medium in the form of alternate compressions and rarefactions. Hence, they are also called pressure or compressional waves. In transverse waves, the particles oscillate perpendicular or transverse to the direction of wave propagation. The use of longitudinal waves for ultrasonic applications involves a thickness mode piezoelectric material while the use of transverse waves involves a shear mode piezoelectric. Shear waves, which are generally weaker than longitudinal waves, require an acoustically solid medium to propagate and hence, cannot propagate effectively through liquids and gases [21]. As penetration through a medium is a crucial requirement for common ultrasonic applications, longitudinal waves, which are known to experience less attenuation during propagation, are utilized in most cases.

3.1.2 Attenuation

As sound travels through a medium, its intensity decreases with distance traveled. In an ideal material medium, the signal amplitude is reduced only by the spreading of the wave. In natural materials, however, there is absorption as well as scattering which further diminish the sound intensity. Scattering is the reflection of sound waves in directions other than the direction of propagation. Absorption is the conversion of sound energy into other forms of energy. This combined effect of scattering and absorption is called attenuation of sound. The amplitude of a decaying plane wave is given by the formula below [22].

\[ P(x) = P_0 e^{\alpha x} \]  

(3.1)

where \( P_0 \) is the reference wave pressure (amplitude), \( P(x) \) is the wave pressure at distance \( x \) from the reference point and \( \alpha \) is the attenuation coefficient and is a characteristic of the medium and the wave frequency. The unit for the attenuation at certain frequency is Nepers per meter (Np/m), but it is more commonly expressed in decibels/length. The relation between the attenuation constant \( \alpha \) and wave frequency \( f \) can be expressed by the formula below [22].

\[ \frac{\alpha}{f^2} = \frac{A_1 \times f_r^2}{f_r^2 + f^2} + B \]  

(3.2)
Here, $A_1$ and $B$ are empirical constants, $f_r$ is the characteristic relaxation frequency. Thus we see that the attenuation is proportional to the square of the wave frequency. There is higher attenuation and consequently, lower penetration at higher frequencies. A correct choice of operating frequency ensures adequate wave propagation and penetration for a particular application but effective sound transmission from one media to another also needs to be considered. Acoustic impedance of the mediums is a vital factor in ensuring maximum sound transmission.

### 3.1.3 Acoustic Impedance

When a wave traveling in a medium encounters a change in acoustic properties (usually at the boundary between two different media), a part of its energy is reflected and a part is transmitted. The reflected wave travels back through the first medium while the transmitted wave continues propagating through the second medium. The angle of the reflected wave is equal to the incident angle but the angle of the transmitted is deflected. This effect is similar to the optical refraction and follows the Snells Law [23].

\[
\frac{C_1}{\sin \theta_i} = \frac{C_2}{\sin \theta_T} \tag{3.3}
\]

where $C_1$ and $C_2$ are the wave velocities in the two mediums and $\theta_i$ and $\theta_T$ are the angles of incidence and transmission, respectively. This is schematically explained in Figure 3.2. Thus, we can see that if the wave velocities are equal, there will be no deflection in the angle of transmitted wave. The wave velocities are directly related by the acoustic impedances of the media.
Acoustic impedance is defined as the resistance of a medium to acoustic pressure. It is the product of a material's density \( \rho \) and acoustic velocity \( c \).

\[
Z = \rho c 
\]  

(3.4)

Acoustic impedance is important to determine [20]

- the acoustic transmission and reflection at the boundary of two materials with different acoustic impedances
- the materials to be used for fabrication of an ultrasonic transducer
- the absorption of sound in a particular medium

For good transmission of sound energy from the transducer to the medium, there should be a good match of the acoustic impedances of the two materials. The transmission coefficient between two materials having acoustic impedances \( Z_1 \) and \( Z_2 \) can be calculated using the following formula [22]
transmitted energy = \left[ \frac{4Z_1Z_2}{(Z_1 + Z_2)^2} \right] \times 100(\%) \quad (3.5)

From the equation, it is clear that at \( Z_1 = Z_2 \), the transmission is 100%. There usually exists an acoustic impedance mismatch between the medium (commonly water) and the piezoelectric material used in the active element in a transducer. This necessitates the use of one or more extra layers called the matching layer between the active element and the medium. The specific requirements for the matching layer will be discussed in section 3.3.4. Acoustic impedance matching and mismatch should also be taken into account while selecting a suitable backing layer as discussed in section 3.3.3.

3.2 Transducer Characteristics

After gaining an understanding of the effects of the medium on the acoustic wave, it is imperative to establish the acceptable values for crucial transducer characteristics. The center frequency, bandwidth, and loop sensitivity are the parameters for which required acceptable values will be determined. The design and fabrication process will be directed toward achieving the transducer characteristics near these values.

3.2.1 Center Frequency

The ideal center frequency or resonance frequency is selected based on the trade-off between penetration and resolution. The center frequency of the transducer depends on the thickness of the piezoelectric material and the speed of sound in the material. Basic vibration theory states that even though electrical impulses of many frequencies will induce physical vibrations in the transducer, exciting the transducer at its resonant frequency will result in a much stronger response. The fundamental resonance frequency of a transducer is given by

\[ f_r = \frac{c}{2L} \quad (3.6) \]

where \( c \) is the wave velocity in the material and \( L \) is the thickness of the material [25].
The wavelength of a wave in a material is equal to the wave velocity divided by frequency

\[ \lambda = \frac{c}{f} \]  

which results in the wavelength being equal to twice the thickness of the piezoelectric material. A frequency of 3.3 MHz was selected as it is commonly used for both medical imaging and non-destructive evaluation and gives an acceptable penetration and image resolution.

### 3.2.2 Bandwidth

A transducer emits waves at a range of frequencies, the strongest being the center frequency. This range of frequencies is called the frequency bandwidth of the transducer.
The bandwidth primarily depends on the electromechanical coupling coefficient of active material and backing material used. For a particular resonance frequency, the pulse duration and bandwidth are inversely proportional. Thus, heavily damped transducers have a much broader bandwidth. The broad frequency range implies that the transducer can be excited by more than a single frequency and has a higher resolving power. However, bandwidth and penetration related sensitivity usually contradict with each other. Thus, depending on the application, the right bandwidth should be determined. For this transducer, a -6 dB bandwidth ratio greater than 60% was required.

### 3.2.3 Loop Sensitivity

Loop sensitivity, also called loop gain is the measure of the conversion of input voltage to the received pulse echo voltage which is given by $S = 20\log\frac{V_r}{V_e}$ where $V_e$ is the excitation pulse in volts and $V_r$ is the received signal in volts. It is usually a negative number expressed in decibels (dB). For the particular application, a loop sensitivity greater than -65 dB was required.
Having established the desired values for key characteristics, the design process can be started with the final goal of achieving these values.

### 3.3 Transducer Structure

The basic ultrasound transducer structure is already introduced in Chapter 1. This fundamental structure remains about the same for any application. However, a lot of modifications are possible to enhance or suppress certain characteristics of the transducer depending on the application. For a conformable two-dimensional ultrasound transducer, the following basic structure was envisioned.

![Frontal cross section of basic transducer array structure](image)

The device structure design was kept as simple as possible without compromising any of the requirements. The design consists of the two-dimensional array bonded to
a flexible conductive substrate. The flexibility can be attained by having the thickness of the substrate sufficiently small. This conductive substrate also acts as the common ground for all the 64 elements in the array.

The cross-section of the layers for each transducer element is given in Figure 3.4.

Figure 3.7: Layers for a single transducer element

The specific design and material requirements for each layer are discussed in detail in the following subsections.

### 3.3.1 Piezoelectric Material

Lead zirconate titanate (PZT) is arguably the most popular piezoelectric material for manufacturing transducers. However, the characteristic acoustic impedance for PZT is 30-40 MRayls while the acoustic impedance of water is 1.5 MRayls causing a significant impedance mismatch. Additionally, PZT does not absorb ultrasound effectively, giving it a poor damping and thus, narrow bandwidth and long ringing times. There are certain advantages of combining the PZT with a polymer phase like lower density and permittivity and higher elastic compliance. Such ceramic-polymer composite materials
are called piezocomposites. The table compares the parameters of ceramics, polymers and composites.

Table 3.2: Advantages (+) and Disadvantages (−) of Piezoelectric Ceramics, Polymers and Composites [26]

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Ceramic</th>
<th>Polymer</th>
<th>Ceramic-polymer composite</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acoustic Impedance</td>
<td>High (-)</td>
<td>Low (+)</td>
<td>Medium (+)</td>
</tr>
<tr>
<td>Coupling Factor</td>
<td>High (+)</td>
<td>Low (-)</td>
<td>High (+)</td>
</tr>
<tr>
<td>Spurious Modes</td>
<td>Many(-)</td>
<td>Few (+)</td>
<td>Few (+)</td>
</tr>
<tr>
<td>Dielectric Constant</td>
<td>High (+)</td>
<td>Low (-)</td>
<td>Medium (+)</td>
</tr>
<tr>
<td>Flexibility</td>
<td>Stiff (-)</td>
<td>Flexible (+)</td>
<td>Medium (+)</td>
</tr>
<tr>
<td>Cost</td>
<td>Cheap (+)</td>
<td>Expensive (-)</td>
<td>Medium (+)</td>
</tr>
</tbody>
</table>

Connectivity Patterns for piezocomposites The arrangement of the phases within a composite is critical for the electromechanical properties of composites. The concept of connectivity or microstructural arrangement of component phases in the composite was first introduced by Newnham et al. in 1978 [27] and later amended by Pilgrim et al. in 1987 [28]. Connectivity is the number of dimensions in which a phase is self-connected. For a biphasic composite, sixteen connectivity patterns are possible, ranging from (0-0), where neither phase is self-connected to (3-3), where each phase is self-connected in three dimensions. Conventionally, the first digit refers to the piezoelectrically active phase while the second digit refers to the soft polymer or inactive phase. Figure 3.5 represents ten such microstructural agreements. Only ten are represented because (0-1), (0-2), (0-3), (1-2), (1-3), and (2-3) correspond to two structures each. For example, (1-3) represents both 1-3 and 3-1 connectivities.
Piezocomposites offer several key advantages: high electromechanical coupling constant ($k_t$), higher sensitivity, and a low acoustic impedance. Figure 3.6 represents the behavior of coupling coefficient $k_t$ and acoustic impedance $Z_{PZT}$ with the ceramic volume fraction in a 1-3 piezoelectric/polymer composite.
It is interesting to note that the coupling coefficient $k_t$ is higher for the ceramic-polymer composite as compared to the ceramic alone. $k_t$ is defined as the ratio of mechanical energy stored in a thickness mode transducer to the electrical energy supplied or vice-versa [31]. Thus, it would be expected that the energy conversion would be decreased as the volume of active material is decreased. However, a soft polymer provides a more compliant environment for the partially clamped ceramic. This relief of lateral clamping accounts for increased thickness mode coupling constant.

The impedance decreases with the reduction of ceramic and increase of polymer fraction because the density of polymer is much less compared to the density of ceramic. Lower acoustic impedance is a desirable feature when the transmission medium is water or tissue. A material's acoustic impedance can also be given by

$$Z = \sqrt{\rho E} \quad (3.8)$$

where $\rho$ is the density and $E$ is the Young's modulus of elasticity. Thus, acoustic
impedance is reduced by using a soft, light polymer instead of a stiff, heavy ceramic. High volume fraction of polymer reduces the coupling coefficient, thus, decreasing the sensitivity and the bandwidth.

Other practical factors that needed consideration were price, availability, shipping time etc. The piezocomposite used was PZT 5H/epoxy 1-3 piezocomposite. Some important properties of this composite are given in the table below.

<table>
<thead>
<tr>
<th>Property</th>
<th>Symbol</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Curie temperature</td>
<td>$T_c$</td>
<td>210°C</td>
</tr>
<tr>
<td>Electromechanical coupling coefficient</td>
<td>$k_t$</td>
<td>0.64</td>
</tr>
<tr>
<td>Density</td>
<td>$\rho$</td>
<td>4820 kg/m$^3$</td>
</tr>
<tr>
<td>Youngs Modulus</td>
<td>E</td>
<td>$47GPa = 4.7 \times 10^{10} N/m^2$</td>
</tr>
</tbody>
</table>

Using these values, we can calculate some parameters. The velocity of sound through the PZT 5H 1-3 piezocomposite can then be calculated as

$$
   c = \sqrt{\frac{E}{\rho}} = \sqrt{\frac{4.7 \times 10^{10}}{4.82 \times 10^5}} = 3123 \text{ m/s}
$$

The acoustic impedance is

$$
   Z = 4.87 \times 10^3 \times 3123 = 15.21 \text{ MRayls}
$$

For a resonance frequency of 3.3 MHz, the wavelength will be $\lambda = 946 \mu m$

It can be inferred that the thickness is half of the wavelength. As the wavelength required was calculated as 946 $\mu m$, the thickness of the piezocomposite element can be about 473 $\mu m$.

**Electrode on Piezocomposite** The main properties necessary for the electrodes were chemical inertness (corrosion and oxidation resistance), high electrical conductivity for small thicknesses and good adhesion to the ceramic. The resistivity of some common metals used as electrodes are given in Table 3-3 [34].
Galvanic corrosion is a localized mechanism by which metals can be preferentially corroded [35]. The Galvanic Corrosion Chart gives an estimate of the propensity of a particular metal to get corroded. Higher nobility of the metal implies better corrosion resistance.

![Galvanic Corrosion Chart](image)

As can be seen, gold is a very noble metal and hence, combined with its good electrical conductivity, is an ideal choice as an electrode. However, an adhesion layer is required to attach gold to PZT substrate [36]. Chromium is a commonly used and
effective adhesion layer for gold. The common combination of 300Å Cr as the adhesion layer and 2000Å gold as the conductive electrode was used.

**Common Electrode** Gold was an obvious choice again as far as electrical conductivity was concerned. The main concern here was the strength and thickness of the film. It was necessary to keep the gold foil as thin as possible to prevent losses in the transmission side of the transducer. According to the design, the foil is supported by the rubber, and the strength and flexibility of the foil depends on the structure as a whole as opposed to just the gold foil. This enabled the use of a very thin 36m gold foil as the substrate.

### 3.3.2 Matching Layer

The acoustic impedance of the piezocomposite is still pretty high (15.21 MRayls) as compared to the transmission medium, which is water (1.5 MRayls). This large impedance mismatch causes a large portion of the available energy to be reflected back into the backing material where it must be absorbed to avoid false echoes. This problem can be solved by using a matching layer of proper thickness and acoustic impedance, which acts as a mechanical transformer, transmitting the energy into the medium and improving the sensitivity. Multiple matching layers can also be used [37] but they increase the overall thickness of the matching layer which is undesirable in this design as the matching layer acts as a dielectric between one face of the piezocomposite and the common electrode. According to the transmission line Krimholtz-Leedom-Matthaei model [38], the ideal thickness of the matching layer is quarter of the wavelength of the acoustic wave. The bandwidth can be increased by using a quarter-wave matching layer as shown by experiments by McSkimin [39]. For a wavelength of 946m, the thickness, $t$ of an ideal matching layer will be

$$ t = \frac{946}{4} = 473\mu m \quad (3.11) $$

However, the actual thickness can vary slightly and is usually determined by experimentation and simulations. The matching layer for the piezocomposite was approximately 260μm.

The acoustic impedance of the matching layer is given by the geometric mean of
the acoustic impedances of the piezoelectric element and the transmission medium [40]. In this case,

\[ Z_m = \sqrt{(Z_P Z T Z_{\text{water}})} = \sqrt{(15.21 \times 1.5)} = 4.78 \text{ MRayls} \quad (3.12) \]

This is the theoretical value for acoustic impedance of the matching layer.

### 3.3.3 Backing Layer

If the back wave is reflected back into the transducer it can cause unnecessary ringing or interference with the incoming and outgoing signals. This can even cause overheating of the transducer leading to associated problems. The function of the backing layer is to absorb energy transmitted to the back side and thus prevent unnecessary reflections from the back interface. For that purpose, a suitable material with proper acoustic impedance and attenuation coefficient is used to avoid reverberations of the piezoelectric element by mechanically damping the oscillations. The backing layer is usually a dosed epoxy or polymer with a high attenuation coefficient and impedance close to that of the piezoelectric material. However, this causes most of the energy to be lost and results in lower loop sensitivity of the transducer. A large impedance mismatch (i.e. a low acoustic impedance backing layer) will reflect most of the energy towards the front face and increase the loop sensitivity. However, the matching layer must be designed properly in this case to transmit all the back reflected energy towards the target. A silicone elastomer, Sylgard 170 was selected based on favorable experimental results. It has a specific gravity of 1.37 which gives a density of 1370 kg/m³ [41]. The speed of sound through Sylgard 170 is 1000 m/s [42]. Therefore, its acoustic impedance is

\[ Z_{\text{sylgard}} = 1.37 \text{ MRayl} \quad (3.13) \]

It has low impedance as compared to the piezocomposite which makes it suitable for high loop sensitivity applications.


3.4 KLM Model

The most common models in case of piezoelectric transducers are the Mason [43], the Krimholtz-Leedoem-Matthaei (KLM) [38], and the Redwood [44] models. Each model has its own advantages and disadvantages, and areas of application. The KLM model has a feature that it can represent the acoustic part of the piezoelectric element as a transmission line. Additional layers like matching layers, backing layers, other electronic circuitry can also be included in this model.

The general structure of a KLM model of a basic ultrasonic transducer is shown in Figure 3.8

![KLM Model Diagram](image)

Figure 3.11: The KLM Model of the piezoelectric transducer [45]

3.4.1 Impedance based on KLM Model

In this model, $V_{in}$ and $I_{in}$ are the respective voltage and current applied to the piezoelectric crystal which produce the resulting acoustic forces $F_1$ and $F_2$ and the particle velocities $U_1$ and $U_2$ at the respective faces of the piezoelectric element. The input parameters include the thickness of the crystal $d$, the area of the crystal $A$, and the
properties of the matching and the backing layers. The model also consists of a capacitor $C_0$, impedance $jX_1$, and a transformer with the turns ratio $(1 : \phi)$ that converts the electrical signal into the appropriate acoustic values. The values for these parameters are given below.

$$Z_0 = \rho c A$$

$$C_0 = \frac{\epsilon_0 \epsilon_s A}{d}$$

$$X_1 = \frac{h_{33}^2}{\omega^2 Z_0} \sin\left(\frac{\omega d}{c}\right)$$

$$\phi = \frac{(\omega Z_0)}{(2h_{33})} \sin^{-1}\left(\frac{\omega d}{2c}\right)$$

where

- $\epsilon_0$ : permittivity of free space ($8.854 \times 10^{-12}$)
- $\epsilon_s$ : clamped relative permittivity of the piezoelectric
- $\rho$ : density of the piezocomposite
- $c$ : speed of longitudinal sound waves in the piezocomposite
- $A$ : effective aperture area
- $d$ : thickness of the piezo layer
- $\omega = 2\pi f$ and changes with the frequency
- $h_{33}$ : piezoelectric pressure constant for the piezocomposite

$$h_{33} = k_t \sqrt{\frac{E}{\epsilon_0 \epsilon_s}}$$

where $k_t$ is the electromechanical coupling coefficient of the piezoelectric resonator.

The acoustic impedances for the piezoelectric element, the backing layer and the matching layer are given by $Z_0, Z_b$ and $Z_f$ respectively. The input impedances for the layers can be calculated recursively as follows

$$Z_{in,b} = \frac{Z_0 Z_b + jZ_0\tan\left(\frac{\omega d}{2c}\right)}{Z_0 + jZ_0\tan\left(\frac{\omega d}{2c}\right)}$$

34
\[ Z_{\text{in},f} = Z_0 \frac{Z_f + jZ_0 \tan\left(\frac{\omega d}{2c}\right)}{Z_0 + jZ_0 \tan\left(\frac{\omega d}{2c}\right)} \]  

(3.20)

The input impedance of the transmission is then given by

\[ Z_a = \frac{Z_{\text{in},b} Z_{\text{in},f}}{Z_{\text{in},b} + Z_{\text{in},f}} \]  

(3.21)

Based on transmission line theory by Pozar [47], the input impedance of the electrical port is given by

\[ Z_e = \frac{1}{j\omega C_0} + jX_1 + \frac{Z_a}{\phi^2} \]  

(3.22)

This is the impedance as measured by the impedance analyzer.

3.4.2 Pressure based on the KLM Model

An analysis will be performed to determine the pressure radiated and received by the transducer when it is excited by a voltage \( V_i n = |V_i n(\omega)| e^{j\theta} \). The current transmission coefficients \( \Gamma_b \) and \( \Gamma_f \) are given by

\[ \Gamma_b = \frac{Z_0 - Z_b}{Z_0 + Z_b} \]  

(3.23)

\[ \Gamma_f = \frac{Z_0 - Z_f}{Z_0 + Z_2} \]  

(3.24)

Transmission line theory can be used to solve for particle velocities \( U_1 \) and \( U_2 \) yielding

\[ U_2 = -\frac{I_{in}}{\phi} \frac{\left(\Gamma_b e^{-jkd} - e^{-jkd}\right)}{e^{jkd} - \Gamma_b \Gamma_f e^{-jkd}} (1 + \Gamma_f) \]  

(3.25)

\[ U_1 = -\frac{I_{in}}{\phi} \frac{\left(e^{jkd} - \Gamma_f e^{-jkd}\right)}{e^{jkd} - \Gamma_b \Gamma_f e^{-jkd}} (1 + \Gamma_b) \]  

(3.26)

Replacing \( I_{in} \) by \( V_i n \) and solving for the pressure wave leaving the surface of the piezoelectric material gives
\[ P_2(\omega) = \frac{Z_2 V_3(\omega)}{\phi Z e} \left( \Gamma_b e^{-jkd} - e^{jkd} \right) (1 + \Gamma_f) \] (3.27)

\[ P_1(\omega) = \frac{Z_1 V_3(\omega)}{\phi Z e} \left( e^{jkd} - \Gamma_f e^{-jkd} \right) (1 + \Gamma_b) \] (3.28)

where \( P_2(\omega) \) and \( P_1(\omega) \) are the pressure waves radiated from the front and back faces of the piezoelectric element respectively.

A pressure transfer function could be used to calculate the pressure propagated from the piezoelectric element to the matching layer and subsequently to the medium. This can be expressed as a product of the pressure transfer functions through each of the front layers

\[ \frac{P_m}{P_{in,front}} = \prod_{n=1}^{n=N} \frac{Z_{in,n+1}}{Z_{in,n+1} \cos \left( \frac{\omega d}{2c} \right) + j \sin Z_n \sin \frac{\omega d}{2c}} \] (3.29)

if \( N \geq 1 \) where
\( P_m \): pressure propagated in the medium
\( P_{in,front} \): input pressure from the first front layer
\( Z_n \): acoustic impedance of the front layer under consideration
\( Z_{in,n+1} \): input impedance of following layer

The Pressure to Generator Voltage (PGV) transfer function is obtained by the product of \( P_2(\omega) \) and the pressure transfer function. The pulse-echo response can be obtained by the inverse Fourier Transform of the PGV.

Commercial software like PiezoCAD (Sonic Concepts) and PZFlex (Weidlinger Associates Inc) are available which allow the generation of the KLM model and computation of the pulse-echo graph after the parameters are entered in the user interface. A free online software program [47] was used to obtain the impedance and pulse-echo graphs used in this thesis.
Chapter 4

Fabrication

4.1 Manufacturing of components

The manufacturing processes for some of the components of the transducer are detailed below.

4.1.1 PZT 5H 1-3 Piezocomposite

The PZT 5H piezocomposite was fabricated using the dice-and-fill technique [48]. A typical dice-and-fill process involves bonding the piezoceramic to a metal block and mounting it on a high-speed saw. The saw blades are used to make two sets of closely spaced, parallel cuts such that the two sets are perpendicular to each other. The width of the saw blade depends on the width of the gap or kerf needed. For kerf widths up to 50µm, mechanical saws are quite effective but rod fragility and blade wear become important factors for smaller kerfs [49]. In this case, the kerf width was required to be about 23µm and laser machining was used to reach the desired kerf width [50]. The PZT was then removed from the block and cleaned ultrasonically to remove residual particles from the cut surfaces. The kerfs were filled with a high viscosity copolymer poured over the diced PZT disk. This copolymer became quite hard after curing, thus enabling easy machining at room temperature to get the required thickness of 0.77 mm. The electrode, consisting of chromium (500Å) as the adhesion layer and gold (2000Å) as the conductive layer, was deposited on one face of the piezocomposite. Electrode deposition was done
using a physical vapor deposition technique called sputtering. The matching layer with suitable acoustic impedance was spin-coated onto the face opposite to the electrode, and cured. Finally, poling was carried out for five minutes in a 75°C oil bath with an electric field of 22 kV/cm to align the domains along the desired axis [51]. The final result was a 8 × 8 rectangular array with an aperture size of 15.7 mm × 15.7 mm and an element pitch of 2 mm with Cr/Au electrode deposited on one face and a matching layer on the other.

4.1.2 Sylgard 170

Sylgard 170 was purchased as a two part elastomer kit. Calibrated syringes were used to draw the two parts in a 1 : 1 ratio by volume. These two parts were thoroughly mixed until the mixture had a uniform color. The viscosity value of Sylgard 170 at 23°C after 2 minutes is 3000 mPa.s, but after 30 minutes, the viscosity goes up to 11,000 mPa.s [52]. Thus, the Sylgard 170 was used immediately after mixing and fresh batches of mixture were made for each use.

4.1.3 Wires

66 coaxial wire pieces (each approximately 2 in length) of a total 38 AWG (7 wire strands each of 40 AWG) thickness were used as the electrical interconnects for the 64 elements and the common ground electrode. About 2 mm of insulation was stripped from either end of the wire to expose the metal wire strands, which were then tinned using solder to facilitate soldering.

4.2 Assembly

The transducer assembly was carried out after extensive experimentation and analysis to determine the feasibility of all the process steps and their effects on the critical required final device parameters.
4.2.1 Common Electrode

A 53µm thick gold foil measuring 18mm × 18mm was used as the common electrode. This conductive foil also acts as the substrate for the 64 piezocomposite elements. The piezocomposite elements were aligned on the substrate and bonded using conductive silver epoxy [53]. A conductive epoxy was chosen because metals have a less damping energy loss than non-metallic epoxy bonds [54]. Also, the use of a non-conductive epoxy would have meant the addition of an insulating dielectric between the piezocomposite and the common ground, reducing the connectivity and lowering the loop sensitivity. It was necessary to minimize the bond thickness of the epoxy to reduce the energy losses in the piezocomposite vibrations. Slight clamping was used to ensure a uniform thin layer of silver epoxy. Excessive clamping and elevated curing temperatures were avoided as they could depole the piezocomposite. Room temperature curing (23°C) was carried out for 8 hours to avoid the risk of depoling and obtain a lesser electrical resistance as compared to heat curing [53].

![Silver Epoxy on non-electrode face of transducer elements](image)

Figure 4.1: Silver epoxy on non-electrode face of transducer elements
### 4.2.2 Partial Potting

Based on empirical observations, the common electrode was found to lack the necessary strength to support the elements during wire bonding. Thus, it was necessary to reinforce the structural strength of the common electrode. A process called partial potting was developed for this purpose. A small amount of Sylgard 170 was poured into the gaps between the elements ensuring that there was no spillage on top of the electrodes. The low viscosity of Sylgard 170 (3000 mPa.s after 2 minutes at 23°C [52]) ensured uniform flow into the small gaps between the elements. Sylgard 170 has the tear strength value of 3.5 kN/m, which was helpful in achieving the necessary strength for the further processes. The ground wires were then bonded to the common electrode using the conductive silver epoxy.

![Partial potting using Sylgard 170](image)

Figure 4.2: Partial potting using Sylgard 170

### 4.2.3 Wire Bonding

Tin lead soldering was considered as the wire bonding method of choice due to its high bond strength and good conductivity. Solder flow is also easier to control compared to other epoxies which eliminates the need to use masks for isolation and short circuit prevention. Another key advantage is the elimination of long-term life concerns as with the organic epoxy bond. Soldering, however, presents with a potential disadvantage that
the PZT elements could experience depoling due to elevated temperatures necessary for soldering. An evaluation was performed to determine the feasibility of tin lead soldering the wires to the PZT elements by using dummy elements and varied temperatures and heat exposure times. The final wire soldering was carried out at a temperature of 440°C with a heat exposure window of 1-2 seconds.

Figure 4.3: Wires soldered to electrode of each element

4.2.4 Potting

The potting or encapsulation of components is a common engineering design tool to protect them from the environmental exposure to moisture, chemicals, dirt or other contaminants. It involves covering a device in an epoxy, urethane or silicone and curing it to get an embedded device. Some advantages of potting using silicone are [52]

- excellent electric insulating properties
- flexibility
- resistance to humidity, water and fire
- no or low toxicity
- resistance to thermal shock and vibration
The potting was carried out after placing the device in aluminum housing. Sylgard 170 was poured carefully to get a uniform thickness of 4 mm. The setup was cured for 24 hours at 23°C to get the encapsulated device.

![Potting setup](image)

**Figure 4.4: Potting setup**

### 4.2.5 Printed Circuit Board Connection

A 4 layer printed circuit board was designed for easy electrical connection to the transducer elements. A unique, proprietary interconnection layout, shown in figure 4.5, was developed for minimum wire overlap, ease of soldering and a good, clean appearance.
The wires were cut to the appropriate length leaving some room for flexing and were soldered on to the PCB. The connections were then covered with a layer of wax as an electrical insulator to prevent short circuits.
Finally, a protective aluminum cover was affixed to give the final device as in figure 4.7.
Chapter 5

Characterization

To quantitatively describe the effectiveness of the transducer for a particular application, it is necessary to characterize the parameters affecting its transmitting, receiving and physical properties of a transducer. The testing phase consisted of three types of characterization: electrical properties, mechanical properties, acoustic properties.

5.1 Electrical Properties

The electrical properties of interest were the capacitance and resonant frequency impedance for each element.

5.1.1 Test Setup

The capacitance and impedance of each array element was measured using the Agilent 4294A Impedance Analyzer. The effective test setup for electrical characterization is shown in figure 5.1. Only one wire is shown for one element for easy understanding of the diagram.
5.1.2 Results and Discussion

The impedance graph is obtained by plotting the impedance curves and considering the values at the resonant frequency of each element. This graph can be compared to the graph obtained by the KLM model.
Figure 5.2: Comparison of impedance graphs obtained from theoretical and measured values

The minimum impedance values that are obtained at the resonance frequency for all the 64 elements are shown in the graph below.
Table 5.1: Summary of impedance results

<table>
<thead>
<tr>
<th>Impedance</th>
<th>Value (Ω)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minimum (theoretical)</td>
<td>364.18</td>
</tr>
<tr>
<td>Average Minimum (measured)</td>
<td>414.73</td>
</tr>
<tr>
<td>Maximum (measured)</td>
<td>518</td>
</tr>
<tr>
<td>Minimum (measured)</td>
<td>222</td>
</tr>
</tbody>
</table>

It is observed that there is reasonable agreement between the theoretical and average measured values of the minimum impedance. The slight variation is due to the fact that the theoretical model does not consider losses due to wires, silver epoxy and solder. The difference in the maximum and minimum values of impedance are a result of inherent variations in the piezoelectric elements, thickness of epoxy and solder and resistance of wires.
5.2 Mechanical Properties

The transducer was required to conform to 3” and 4” diameter spherical steel balls. This was confirmed by manually applying force to press the device against the steel ball surface.

5.2.1 Test Setup

The array was pressed onto the 3” and 4” diameter steel spheres to test the strength, flexibility, and conformity of the flexible transducer face. The steel spheres were fixed so that there would be no wobbling or unwanted displacement. Consistent pressure was applied till the entire transducer flexible face was in contact with the steel surface.

![Testing of mechanical properties of the flexible transducer array](image)

Figure 5.4: Testing of mechanical properties of the flexible transducer array

5.2.2 Results and Discussion

The transducer flexibility was sufficient that the entire transducer face (including the entire array) was in contact with the 3 and 4 steel spheres. The transducer did not experience any cracking, breakage or other physical damage during the course of this testing. Hence, it was concluded that the mechanical properties of the transducer were adequately good for regular use.
5.3 Acoustic Properties

The acoustic characteristics of the device were analyzed using pulse-echo testing.

5.3.1 Test Setup

The setup used for pulse echo testing is shown below.

![Pulse echo test setup and equipment](image)

Figure 5.5: Pulse echo test setup and equipment

The transducer was connected to an Olympus 5077PR Square Wave Pulser/Receiver. The following settings were used for the pulser/receiver.

Table 5.2: Pulser/receiver settings

<table>
<thead>
<tr>
<th>Setting</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Input voltage</td>
<td>100V (peak-peak)</td>
</tr>
<tr>
<td>Pulse repetition frequency</td>
<td>200 Hz</td>
</tr>
<tr>
<td>Transducer frequency</td>
<td>3.5 - 4.0 MHz</td>
</tr>
<tr>
<td>High pass filter threshold</td>
<td>1 MHz</td>
</tr>
<tr>
<td>Low pass filter threshold</td>
<td>10 MHz</td>
</tr>
</tbody>
</table>
The pulser/receiver was connected to an Agilent DSO 7104B Digital Storage Oscilloscope to obtain the pulse echo waveforms. A stainless steel block with acoustic impedance of 46 MRayl was used as the reflective target. The transducer was fixed parallel to the target at a distance of 4.5 mm. The medium of acoustic wave propagation was water at room temperature (20°C).

5.3.2 Results and Discussion

Pulse Echo Voltage

The pulse echo voltage graphs were obtained for each element for an input voltage impulse of 100 V (peak to peak) amplitude. The pulse echo waveforms from the KLM Model software and one of the actual transducer elements are shown below.

![Pulse Echo Waveform (KLM Model)](image)

Figure 5.6: Pulse echo waveform from the KLM model
As can be seen, the ringing time for the actual element is longer than the model. This is because the model does not consider the effect of silver epoxy, gold common electrode, the solder, and the wires. The peak to peak amplitude of the model is about 275 mV while the peak to peak amplitude for the measured pulse echo is about 240 mV. This difference is due to some damping losses in the silver epoxy and the gold common electrode which are not considered in the KLM model.

The graph of the obtained pulse echo wave amplitude for all the 64 elements is shown below.
The KLM model considers an ideal transducer without any acoustic losses due to epoxy damping, acoustic energy transfer to housing, backing and matching layer irregularities, friction etc. Thus, the theoretical value of the pulse echo amplitude is much higher than the average measured value. The elements, themselves, show variations in the pulse-echo amplitude due to variations in piezoelectric properties, silver epoxy thickness, backing rubber layer properties, wire length and testing conditions. However, each element showed a pulse echo of significant amplitude.
Resonance Frequency

The resonance frequency obtained from the oscilloscope and the graph was obtained as below.

![Resonance Frequency Graph](image)

Figure 5.9: Measured resonance frequency for all transducer elements

As can be seen from the graph, there are slight variations in the resonance frequency. The factors that could affect the resonance frequency are the variations in the thickness of the piezocomposite layer and the damping effect of epoxy and other materials. The thickness variations could be due to the fabrication process used for the piezocomposite. It could also be due to reduction in element thickness by clamping while curing the silver epoxy and the polymer.

Distance from Target

The distance from the target was calculated from the Time of Flight measurements.

\[
\text{distance from target} = \frac{t_{\text{interval}} \cdot c}{2}
\]

54
where $t_{\text{interval}}$ is the time between the transmission of the sound wave and reception of the echo and $c$ is the sound velocity in water.

![Distance from target](image)

**Figure 5.10:** Distance of transducer elements from the target

<table>
<thead>
<tr>
<th>Distance</th>
<th>Value (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum</td>
<td>4.8</td>
</tr>
<tr>
<td>Minimum</td>
<td>4.3</td>
</tr>
<tr>
<td>Average</td>
<td>4.49</td>
</tr>
</tbody>
</table>

The maximum variation in distance is 0.5 mm. The transducer was held parallel to the target in the water with the help of a fixture. The alignment was done manually using a bubble level level indicator.
6dB Bandwidth Ratio

The Fast Fourier Transform (FFT) of the pulse-echo waveform was obtained and compared to the FFT waveform obtained from the KLM model.

Figure 5.11: FFT waveform from KLM model
The 6 dB bandwidth ratio was obtained from the pulse-echo waveform by measuring the gap between the two frequencies along the -6 dB line. The following formula was used

\[ B(\%) = 2 \times \frac{f_u - f_l}{f_u + f_l} \times 100 \quad (5.1) \]

where \( f_u \) and \( f_l \) are the upper and lower frequencies respectively at the -6 dB line.
Table 5.5: 6dB Bandwidth Results

<table>
<thead>
<tr>
<th>6dB Bandwidth Ratio</th>
<th>Value (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Theoretical</td>
<td>66.9</td>
</tr>
<tr>
<td>Average (measured)</td>
<td>65.86</td>
</tr>
<tr>
<td>Maximum (measured)</td>
<td>84.09</td>
</tr>
<tr>
<td>Minimum (measured)</td>
<td>53.89</td>
</tr>
</tbody>
</table>

Variations in piezoelectric element and backing layer properties cause the differences in bandwidth values for the transducer elements. However, the theoretical and average measured values are very close to each other.

**Loop Sensitivity**

Loop Sensitivity calculations were also performed using the measured output voltage. The loop sensitivity is given by the following equation. \( S = 20 \log \frac{V}{V_c} \)
where $V_e$ is the amplitude of the wave emitted by the transducer and $V_r$ is the amplitude of the reflected wave received by the transducer.

![Diagram of Loop Sensitivity](image)

**Figure 5.14: Loop sensitivity of the transducer elements**

<table>
<thead>
<tr>
<th>Loop Sensitivity</th>
<th>Value (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Theoretical</td>
<td>-47.72</td>
</tr>
<tr>
<td>Average (measured)</td>
<td>-53.06</td>
</tr>
<tr>
<td>Maximum (measured)</td>
<td>-48.9</td>
</tr>
<tr>
<td>Minimum (measured)</td>
<td>-60.53</td>
</tr>
</tbody>
</table>

The measured and theoretical values of the loop sensitivity are close to each other. The ideal case KLM model has a higher loop sensitivity than the lossy actual transducer. The loop sensitivity is directly related to the pulse echo voltage and thus, the factors...
that affect the pulse echo also affect the loop sensitivity. These include variations in the piezocomposite elements, the differences in silver epoxy thickness causing variable damping, density variations or small defects in backing rubber layer.

5.4 Summary of Results

The results are summarized in the table below:

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Theoretical Value</th>
<th>Measured Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average Capacitance at 100KHz</td>
<td>90.22 pF</td>
<td>86.33 pF</td>
</tr>
<tr>
<td>Average Impedance at resonance frequency</td>
<td>364.18 Ω</td>
<td>414.73 Ω</td>
</tr>
<tr>
<td>Average Pulse-Echo Voltage amplitude</td>
<td>410.89 mV</td>
<td>231.4 mV</td>
</tr>
<tr>
<td>Average Loop Sensitivity</td>
<td>-47.72 dB</td>
<td>-53.1 dB</td>
</tr>
<tr>
<td>Average Resonance Frequency</td>
<td>3.3 MHz</td>
<td>3.29 MHz</td>
</tr>
<tr>
<td>Average Bandwidth Ratio</td>
<td>66.9%</td>
<td>65.9%</td>
</tr>
<tr>
<td>Average Distance from Target</td>
<td>4.5mm</td>
<td>4.49 mm</td>
</tr>
</tbody>
</table>

It can be seen that there is good conformity between the theoretical and measured values. All the parameters are within the acceptable limits mentioned in Chapter 3. There were no dead elements, which gives a manufacturing yield of 100%.
Chapter 6

Conclusion

Two dimensional flexible ultrasound transducers have significant advantages over non-conformal transducer probes available commercially for applications involving inspection of irregular or curved target surfaces. A 64 element compact, conformable 2D ultrasonic transducer array with center frequency of 3.3 MHz was designed, fabricated and characterized successfully for NDE applications. The measured element -6dB bandwidth ratio is 65.9

6.1 Inferences

A 8×8 two dimensional conformable ultrasonic transducer array was designed. It is a layered structure consisting of the common electrode substrate, the conductive bonding epoxy, the matching layer, the piezocomposite, sputtered Cr/Au electrodes, backing layer made of a silicone elastomer, metallic solder and coaxial cable interconnects to the printed circuit board.

Extensive experimentation was carried out to develop the fabrication process. The transducer array was fabricated by bonding 64 PZT-5H elements onto a common conductive substrate. Simple coaxial wire interconnects were used for electrical contact. The entire structure was encapsulated in a silicone elastomer and was housed in an aluminum case.

The measured electrical properties indicate an average capacitance of 86.33 pF and an average resonant frequency impedance of 414.73 ohms. The acoustic characterization
was carried out using the pulse echo testing method to give a loop sensitivity of -53.1 dB, average resonance of 3.29 MHz and average bandwidth ratio of 65.9%.

The design and fabrication processes developed in this thesis enable the manufacturing of a two-dimensional flexible ultrasonic transducer. The main advantages of this transducer are the low manufacturing costs, simplicity of structure, and ease of operation. It is expected that the approach adopted in this thesis could enable the fabrication of low-cost, simple, efficient two-dimensional ultrasonic transducers for real time volumetric imaging of curved, irregular surfaces.

6.2 Future Work

There is a scope for more research and improvements to the transducer array fabricated in this thesis.

An automated setup could be developed for bonding of the elements onto the substrate. This would enable the use of a fast curing conductive epoxy and significantly reduce the manufacturing time. It would also mean perfect alignment of the elements and uniform element kerf widths, which could give a much better image. A fast-curing elastomer with equal or superior self-leveling and degassing properties could mean a further reduction in manufacturing time. A room temperature automated metallic bonding method could be designed for the wire interconnects. A flexible polyimide structure could also be designed for electrical interconnection in case of large arrays.

Improvements or modifications could be made to the transducer structure for various applications. For contact probes, the face of the transducer could be covered with an acoustically matched layer for protection. The size of the array could be increased or decreased depending on the surface area to be scanned. Corresponding changes would have to be made to the aluminum housing in such a case.

Higher frequency transducers could also be developed for use in high resolution medical imaging. The transducer features would be very small for high frequency applications, which would necessitate the use of automated manufacturing processes. There would also need to be certain modifications in the materials or manufacturing processes of the piezoelectric material, encapsulating elastomer, matching layer, interconnections etc.
Bibliography


[33] Bushberg et al. The Essential Physics of Medical Imaging. Williams and Wilkins, 1994


[47] www.biosono.com


