Design and development of broadband cMUT based ultrasonic transducer for medical imaging

Vinay, Sonu Sawant

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Design and Development of Broadband cMUT based Ultrasonic
Transducer for Medical Imaging

Vinay Sonu Sawant

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School of Mechanical & Aerospace Engineering

Design and Development of Broadband cMUT based
Ultrasonic Transducer for Medical Imaging

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A thesis submitted to the Nanyang Technological University in
fulfilment of the requirement for the degree of Doctor of
Philosophy

2007

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Abstract

The major objective of the current research is to develop ultrasonic transducers, which have a broader bandwidth and better sensitivity than conventional piezoelectric transducers, as they affect the axial resolution and image quality of the ultrasound system. As the transducer is targeted for use in Portable Home Ultrasound Image Scanners, it is also a pre-requisite that the transducer must integrate easily with the front-end electronics of the ultrasound system. This thesis describes the detailed study done on the conventional piezoelectric transducers and the recently emerging capacitive micro-machined ultrasonic transducers.

Finite Element Analysis is one of the major tools used in this work. Use of FE analysis on 1-D and 2-D PZT-5H control arrays and hybrid arrays has enabled the theoretical modeling of their behaviour. New material (PNN-PT-PZ) has been proposed for the transducer. Pulse echo results from both arrays show that the transducer arrays of new material (PNN-PT-PZ) give better results as far as bandwidth and badness factor is concerned. However, it is more expensive than conventional PZT material. Moreover, the fabrication of the 2-D array is more difficult and challenging.

In examining the alternative options, the newly developed capacitive micromachined ultrasonic transducer arrays were considered as a potential and promising alternative to the conventional transducer. These transducers besides providing a broader bandwidth do seem to overcome the fabrication difficulties which are present with piezoelectric

transducers. The cMUT is fabricated by using conventional CMOS processes and can be easily integrated to the front-end electronics of an ultrasound system.

As a consequence the capacitive micro-machined ultrasonic transducer has been studied in detail in this dissertation work. Analysis and design of the membrane of the cMUTs is critical, as far as the performance of the transducer is concerned. FEM analysis of the membrane dynamics is done at the cell level. The cross coupling between the transducer elements of the capacitive micro-machined ultrasonic transducer is one of the major factors, which affect the bandwidth of the transducer. The cross coupling is studied and analysed in detail using finite element analysis. A novel structural design has been proposed, which consists of an acoustical cavity filled with acoustically attenuating material to reduce the cross coupling between the transducer elements. The FEM modeling results show the bandwidth of the transducer is reduced considerably due to the cross coupling. The FEM results further illustrate that the proposed structure with appropriate design features can be adapted to reduce the cross coupling between the transducer elements and increase the bandwidth of the transducer. The fabrication of the proposed transducer was undertaken. The fabrication process was arrived at after careful study. The transducer is fabricated using the conventional MEMS fabrication processes. A 1-D array of 32 transducer elements is fabricated. The pulse-echo experiment has been carried out on the fabricated prototype of the cMUT to assess the bandwidth characteristics of the transducer. The experimental characterization results were compared with the FEM simulation results to establish the validity of the transducer design.

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List of Acronyms

cMUT	Capacitive Micromachined Ultrasonic transducer
CMOS	Complementary Metal-Oxide Semiconductor
FE	Finite Element
RIE	Reactive-Ion Etching
ATL	Agilent Technology Limited
DARPA	Defense Advanced Research Projects Agency
UW	University of Washington
HUS	Home Ultrasound System
PW	Pulse Wave
CW	Continuous Wave
PUIP	Programmable Ultrasound Image Processor
PZT	Leadzirconate-titanate
PVDF	Polyvinylidene difluoride
UBM	ultrasound backscatter microscopy
IVUS	Intravascular ultrasound imaging
ETWI	Electrical Through Wafer Interconnects
PDMS	Polydimethylsiloxane
LPCVD	Low Pressure Chemical Vapor Deposition
PECVD	Plasma Enhanced Chemical Vapor Deposition

Chapter 1 : Introduction

1.1 Background

It is the need of the future to transfer current centralized, and hospital-based medical diagnosis and healthcare delivery to a system, which shortens the time to deliver clinical information, as well as enables testing of a type and frequency not previously established outside the hospital. Keeping this as a major goal, the researchers at Nanyang Technological University (NTU) and University of Washington have taken up a challenging project called Distribute Diagnosis and Home Healthcare (D₂H₂). The major goal of D₂H₂ is to integrate technology and medicine to improve the delivery of healthcare to the people.

One of the three components of Distributed Diagnosis and Home Healthcare (D₂H₂) is to develop home-based imaging tools that would allow clinician to have access to the patient and be able to make decisions remotely. In making accurate diagnosis and precise decisions, clinicians typically rely on various imaging modalities, e.g., conventional X-rays, computed tomography, magnetic resonance, and nuclear medicine. Conventional X-rays, CT, MR, and nuclear medicine imaging devices are not portable and they are more suitable in hospitals and clinics due to their size, cost, and training required to utilize them. Ultrasound modality, on the other hand, is safe, non-invasive, inexpensive, easy to use, and capable of real-time imaging and is portable. An ultrasound machine can be wheeled to a patient for imaging instead of wheeling the patient to the ultrasound machine, which is required for most other imaging modalities. Basically there are three types of ultrasound systems based on the imaging modes, frame rate and various

diagnostic functions supported. The three types are referred to as High-end ultrasound systems, Mid-end ultrasound systems, and low-end ultrasound systems.

High-end ultrasound machines are bulky, heavy, expensive, and require a trained sonographer for imaging. Mid-end ultrasound systems are similar to High-end ultrasound system albeit with limited application. On the other hand, low end portable ultrasound systems, developed by several commercial enterprises are small, light, relatively inexpensive, and are mainly used in small hospitals and clinics. The University Of Washington (UW), with ATL, pioneered the research and development of a portable ultrasound device for battlefield trauma. The funding was granted by DARPA under the Technology Reinvestment Program. This UW initiative contributed to the development of SonoSite (an ATL spin-off company) and its five-pound hand-held device, which was commercialized in 1999. Even though the size and cost of the portable ultrasound machines have been reduced compared to high-end systems, they are application specific, difficult to operate, and still expensive for the home healthcare setting. The home ultrasound machines must be operable by less trained people, small in size, lightweight, durable for rough handling, flexible for multiple applications, and affordable. Even though this is not designed for home use, we believe that it will be introduced for home use in the next decade with decreasing costs and improving usability. In the beginning, it will be used for imaging the body and transmitting the resulting images to the clinic via telemedicine. However, it could be used eventually in both diagnosis and therapy.

Front end of these ultrasound systems consist of transducer, beamformer and RF modulator. The two components namely, the transducer and beamformer constitute about 30% of the ultrasound system in terms of cost as well as complexity. Also the electronic complexity of beamformer depends upon the type of transducer used. Therefore, it is necessary to put considerable effort in developing novel transducer that is low cost, and is able to give clinically accepted resolution and image quality for Home Ultrasound System (HUS). Further, the transducer used for the portable home ultrasound system, must be such that it can be integrated with the front-end circuitry of the ultrasound system.

The performance of the transducer is currently far from ideal, due to combinations of factors. It is crucial to realize that designing, and fabricating ultrasonic transducer is a process which requires a broad interdisciplinary approach. It requires knowledge from a variety of disciplines such as acoustics, mechanical vibration, electrical engineering, material science, signal processing, and medical imaging. Because of this involvement of multidisciplinary fields in design, the fabrication of ultrasonic transducers conventionally has been an empirical, trial-and-error approach. Comprehensive and accurate understanding of related physical problem is a prerequisite requirement for designers.

In this thesis, a few potential designs of the transducer for use with the Home Ultrasound System (HUS) are presented. The major goal while designing a transducer is to keep cost of the transducer as low as possible and increase fractional bandwidth of the transducer in order to improve resolution and image quality. The transducer has the potential of being

directly integrated with the front-end electronics of the ultrasound system. This thesis presents a detailed study of three material types namely, the piezoelectric transducers, hybrid arrays of piezoceramic transducers, and single crystal transducers. In the course of the study and design process, a novel single crystal material PNN-PT-PZ has been proposed [44] for the transducer and its effect on the bandwidth of the transducer studied in detailed. It will be shown later that the bandwidth of the transducer can be increased due to the use of this novel material. The cost of this material however, is higher than conventional piezoelectric materials.

As low cost is the driving parameter for the design, the PNN-PT-PZ based transducer could not qualify as an optimum solution. In search for alternatives, consideration was given to the possible use of silicon based Capacitive Micromachined Ultrasonic Transducers (cMUT). The Capacitive Micromachined Ultrasonic Transducers (cMUT), which provide broader bandwidth, are promising alternative to the conventional transducer. Moreover, the cost of these type transducers is very low; as they use CMOS/MEMS microfabrication processes, and are produced in batch production. The cMUTs can be easily integrated with the front-end electronics of ultrasound system. The cMUTs also have some potential problems such as crosstalk between the transducer elements which needs to be resolved. Cross-talk affects the bandwidth of the transducer and image quality of the ultrasound system. A novel structure is proposed for this Micromachined transducer, which reduces the cross-talk between the transducer elements and improves the fabrication process. The structure consists of an acoustical cavity filled

with an acoustically attenuating material in the substrate below the cMUT cells. The transducer is fabricated according to design specifications and characterized further.

1.2 Objectives of project

The major goal of this project is to design a transducer for a low cost, programmable and portable ultrasound system (Home Ultrasound System). In this perspective, following are the objectives of the project that the transducer array should fulfill.

1. Design a transducer of centre frequency between 2-10 MHz, which is useful for abdominal imaging.
2. Study the cross-talk between the transducer elements and its effect on the bandwidth of the transducer using finite element analysis.
3. Design a novel structure for the transducer to reduce the cross-talk between the transducer elements and to improve the bandwidth of the transducer. The targeted fractional bandwidth @ -6dB is 50 to 60% without backing and matching layer.
4. The transducer array shall have the potential to be integrated with front-end electronic circuit of the Ultrasound System.
5. Fabricate the first prototype of the cMUT transducer using conventional MEMS fabrication processes.
6. Characterise the fabricated transducer prototype using pitch-catch experiment to obtain bandwidth of the transducer.

1.3 Scope of research

This thesis focuses on the design and development of ultrasonic transducer, which gives broader bandwidth and can be utilized in the home ultrasound system. The research work evaluates comparison of different materials such as conventional piezoelectric material, hybrid piezoceramic materials and the single crystal materials using finite element analysis. The pulse echo response has been studied in detail using all types of materials for the transducer finite element models. The fractional bandwidth (@ -6dB) is been considered as a figure of merit for the comparison of different type of materials for the transducer. In this thesis the capacitive micromachined ultrasonic transducers has been studied in detail. The capacitive micromachined ultrasonic transducers are the promising alternatives to the conventional piezoelectric transducers due to broader bandwidth, low cost and they have potential to be integrated with the front-end electronics of the ultrasound system. The cMUT cell parameters have been designed initially using finite element analysis. The pulse echo response of the transducer is studied in detail at array level using finite element analysis. The acoustic cross coupling through the lamb waves in the transducer and its effect on the bandwidth of the transducer has been studied in detail using finite element analysis. The different structural modifications have been studied in the form of acoustic cavity below the cMUT cell filled with acoustically attenuating material. The effect of this acoustic cavity on the bandwidth of the transducer has been studied and compared. The transducer model is fabricated using the MEMS fabrication processes. In this thesis the pulse echo response of the transducer is characterised in the water tank.

1.4 Outline of the report

Design and development of broadband cMUT based ultrasonic transducer for use in Home Ultrasound System forms the problem statement for this thesis. The flow of information in this thesis is organized as follows.

Chapter 1: Following a brief description of the introduction and objectives of the report, a literature review is presented in detailed in Chapter 2.

Chapter 2: This starts with the basic background of ultrasound, ultrasound systems (High-end, low end, and portable ultrasound systems), and ultrasonic transducers. This chapter gives the detailed literature review of the ultrasonic transducers including conventional piezoelectric transducers and capacitive Micromachined ultrasonic transducers. In case of piezoelectric transducers, the various topics on which the literature review is based are transducer materials, array of the transducer elements, array element configurations, and fabrication process. In the separate section of this chapter, Capacitive Micromachined Ultrasonic Transducers are explained in detailed. The topics covered in this section are basic configuration of capacitive Micromachined ultrasonic transducers, membrane materials, array of transducer elements, different types of fabrication processes, Finite Element analysis, and acoustic coupling in the capacitive Micromachined ultrasonic transducer.

Chapter 3: describes basic theoretical foundation in piezoelectric transducers as well as capacitive Micromachined ultrasonic transducers. It starts with the basic piezoelectricity in the conventional piezoelectric transducers. In the separated section the basic principle

of cMUT is presented. It also explains the equivalent circuit of the transducer based on Mason's theory.

Chapter 4: presents detailed finite element analysis of piezoelectric and piezoceramic transducers. The finite element models have been constructed for the 1-D and 2-D array of the transducer. The time domain impulse response analysis has been performed the finite element models of the transducer. The material properties of piezoelectric, hybrid piezoceramic materials and PNN-PT-PZ material have been incorporated in the model. The fractional bandwidth (@ -6dB) for each material is compared and discussed in detail. Chapter 5: presents detailed finite element analysis of cMUT at cell level. The 2-D finite element model has been constructed for single cMUT cell. The various configurations of the cMUT cell have been studied and the collapse voltage is obtained and compared. The pulse echo response of the cMUT cell has been done to obtain the bandwidth of the transducer.

Chapter 6: presents the cross-talk study in capacitive Micromachined ultrasonic transducer. The 2-D finite element models of the transducer array have been constructed. The pulse echo response has been carried out for the transducer array. It gives the proposed design, which consists of an acoustic cavity filled with an acoustically attenuating material to reduce the cross-talk between the transducer elements. The acoustic cavity filled with the acoustically attenuating material has been model. Various shapes of the acoustic cavity have been studied to observe their effect on the bandwidth of the transducer.

Chapter 7: presents the fabrication process and characterization of the transducer in detailed. The fabrication process of the transducer has been explained in detail. The characterisation for pulse echo response has been done and explained in detail.

Chapter 8: concludes the thesis and describes various issues which can be part of future research work.

Chapter 2 : Literature Review

2.1 Background of Ultrasound

An exploding firecracker is a good way to start thinking about ultrasound. The explosion is a result of the following process. The powder inside the firecracker ignites and undergoes a very rapid combustion that quickly increases the pressure inside the paper casing of the cracker. The casing yields and the hot gases inside the cracker are quickly released into the atmosphere. This causes a local pressure wave to be formed at the cracker, which propagates out into the surrounding air. As the propagation continues, energy losses in the air eventually cause the pressure wave to be dissipated. The firecracker explosion is a quite good approximation to an impulse and, as we know from system theory, this impulse will contain components of all frequencies. Many of these frequency components will be transmitted very efficiently by the air. These are in the audio frequency range, 20 to 15,000 Hz. However, the firecracker will also produce many higher frequency components. These higher frequency components are not efficiently transmitted through the air and are severely attenuated by the time they reach our ears. Because of this severe attenuation of high-frequency sounds, our auditory sense cannot detect these high-frequency sounds (even though they may be present). Sounds in these frequency regions have been given a special name, *ultrasound*. Although ultrasound refers to any frequency beyond our auditory sensibility range (i.e., >20,000 Hz), clinically used ultrasound usually refers to frequencies around one megahertz [1].

Clearly, producing ultrasound waves by exploding firecrackers does not sound like a very useful approach in the clinical setting. However, this is exactly the approach used by geologists to explore the structure of the earth's crust; a series of explosive charges are set off on the surface of the earth and the resulting ultrasonic waves are recorded by an array of sensitive geophones (rugged but sensitive microphones) placed at various distances away from the charges. Each geophone will pick up the initial explosion and subsequent signals, which are reflections from the different layers (or strata) under the ground. The patterns of signals in the network of geophones can be used to map these underground structures.

Clinical ultrasound is produced in a much simpler manner using *lead zirconate titanate crystals* that are excited with impulses of voltage [68, 72]. These crystals are piezoelectric and are usually formed into discs with conducting metal electrodes deposited on the surfaces of the crystals. These electrodes are excited with an impulse of voltage. The crystal responds to this impulse by either constricting or expanding in thickness in a very short time. The crystal then relaxes back toward its original dimensions, but it overshoots, and we observe second-order under damped mechanical oscillations. The frequency of the oscillations is given by the mechanical resonance of the crystal and is typically from 1 to 10 MHz. These mechanical oscillations are radiated out from the surface of the crystal and enter the tissues against which the transducer is directed. The circuitry required to produce this electrical impulse that drives the quartz crystal is quite complex; it is capable of producing an output pulse of 100 to 200 V with extremely fast rise times. Although the radiating beam is closely approximated by plane

waves for short distances away from the crystal, the beam pattern starts to diverge as the distance away from the crystal increases. The region where the plane wave approximation is valid is given by the relation

$$d = \frac{D^2}{4L}$$

Where D is the diameter of the crystal and L is the wavelength of the radiation. Some ultrasound transducers have an acoustic lens mounted on their surfaces that is used to focus the radiation into an hourglass-shaped beam. When used in an imaging application, beams with small structures. It should be noted that these transducers are piezoelectric, and if such an ultrasonic transducer is irradiated with the beam of ultrasonic pressure waves, it will mechanically resonate with this radiation and as it does, it will produce an electrical signal. Therefore, these transducers can be used not only as transmitters of ultrasonic radiation but as receivers and transducers of this radiation [1].

2.2 Basic Ultrasound Instrumentation

There are three requirements for diagnostic medical ultrasound such as generation of a beam, reception of the returning echo, and processing of the signal for the display. All types of ultrasonic scanning equipment have these basic features. Changes in the reception and in the processing, analysis, and display of returning echo signals differentiate one scanning mode from another.

Fig 2.1 Basic Ultrasound Instrumentation

The basic ultrasonic scanner is electronically complex; a block diagram Fig. 2.1 describes the main system components and operation of the unit. The block diagram shows and electroacoustic conversion device connected to the transmitter-receiver section of the

scanner. This component, called the transducer, is responsible for the generation of an ultrasound beam and the detection of returning echoes. Once an echo is received, the induced signal is electronically processed to enhance the signal for viewing purposes. The displayed signal must be related to interaction that has taken place in the body.

All the subsystems are controlled by a master synchroniser. An electronic signal from the master synchronizer to the transmitter initiates the process. A pulsed ultrasound beam is generated and sent into the body. It undergoes various interactions and when an interface is encountered, a fraction of the beam's energy is reflected back toward the transducer. The transducer, acting as a receiver rather than a transmitter converts the ultrasound wave (returning echo) to an electronic signal that is processed and displayed. To correlate the amplitude of the induced signal with the depth of the interface, the time interval between the time that an ultrasound pulse leaves the transducer and the time the echo returns must be known. Any special application based on elapsed time, such as time gain compensation (TGC) and gating applied to the detected signals, also requires knowledge of the exact time of travel for the pulse. All these subsystems must be activated simultaneously by the master synchronizer to ensure proper timing. For each system the clock in the master synchronizer measures the elapsed time from generation of the ultrasound pulse to reception of the echoes. The ultrasound image formation can be explained in the following steps.

2.2.1 Transducer Excitation

When voltage is applied to the conducting plates, the molecules twist to align themselves with the electrical field) positive molecules toward the negative electrode, negative molecules toward the positive electrode), thereby thickening the crystal. If the plates are reversed in polarity, the molecules will twist back in the opposite direction, creating a decrease in the crystal thickness. The flipping back and forth of the polarity causes expansion and contraction of the crystal, which creates mechanical vibrations. When the expanding and contracting crystal is placed on the body, sound waves (mechanical vibrations or pressure waves) are passed into the body. Thus a voltage applied across the piezoelectric material creates mechanical motion (sound waves). In the crystal this phenomenon (the converse piezoelectric effect) allows an ultrasound beam to be generated by the transducer. Alternatively, the piezoelectric effect enables the same transducer to receive an ultrasonic echo. Ultrasound waves returning from interactions with interfaces in the body strike the crystal and induce electrical signals. These are processed and ultimately displayed. In these echo-ranging systems the transducer must pause after transmitting the sound wave to listen for the returning echoes. Because the crystal cannot send and receive signal simultaneously, it must be pulsed for transmission after an appropriate listening time has elapsed. The master synchronizer sends an electronic signal to a pulsar or transmitter to initiate the process. The start command is, in turn, relayed to the transducer to generate short burst of ultrasound.

2.2.2 Beam Focusing

Divergence of the ultrasonic beam causes a decrease in the intensity with increasing distance from the transducer. This is easily understood by considering a constant amount of energy spread over a larger and larger area. The amount of energy incident on unit area reflector will decrease as the distance between the reflector and the transducer increases. The reflected beam will also diverge, returning to the transducer and partially missing the ultrasound wave reaching the transducer Fig. 2.2. Thus divergence is responsible to intensity loss for both the transmitted and the reflected beam.

Fig 2.2 Energy loss due to divergence, Energy is lost in both transmitted and received direction

The ultrasound beam can be focused, which concentrates the acoustic energy over a small area at a specific distance from the transducer. Compared with a nonfocused beam, the focused beam produces a stronger echo reflected to the transducer for detection. This is true only for structures within the focal zone of the transducer, however, where less divergence occurs.

2.2.3 Interaction with tissue

Several nontransducer related variables influences the overall sensitivity of the system. The intensity of the sound beam is not spatially uniform as the beam leaves the face of the transducer. The intensity of the beam incident on an interface on the incident energy is reflected, the position of the reflecting interface affects the intensity of the returning echo. In addition, the beam is attenuated with depth; that is, the medium transmitting the sound removes acoustic energy from the beam. The composition, shape, and size of structures in the medium also contribute to the reflectivity and thus to the sensitivity. When scanning human subjects, the operator cannot control characteristics of the medium. A transducer can be selected, however, that will have properties conducive to the diagnostic information desired from the portion of the body to be scanned.

2.2.4 Reception

A system has been devised in which a transducer is used to generate an ultrasound beam via the converse piezoelectric effect and then to direct that beam into the body. As the ultrasound wave strikes various interfaces in the body, some of the energy is transmitted and some is reflected. The reflected echo returns toward the transducer.

Transmit gain

Most of the ultrasound units include an output, power, or transmit gain control that adjusts the voltage spike to the transducer to produce an acoustic pulse of higher intensity. This results in a stronger echo. The typical excitation voltage ranges from 300 to 600V but may be as high as 900 V. Frequently the control is adjustable in 3 dB (50%) increments.

Increased sound wave intensity enhances the detectability of weak reflectors. However, higher intensity levels increase patient exposure and unlimited power gain does not continue to improve sensitivity. In addition, high intensity distorts the sinusoidal pressure wave, altering both the propagation and the reception of sound waves.

Amplification

The echo, in the form of waves, strikes the crystal and induces a radiofrequency (RF) signal that mimics the ultrasound waveform, since the voltage variation is in response to pressure-induced thickness changes in the crystal. The microvolt or millivolt RF signal is amplified to 1 V to 10 V for processing and display purposes. Combinations of linear, exponential, logarithmic, and variable amplification can be used.

The most common type of receiver gain is logarithmic. The amount of amplification or gain is adjustable by the operator. Weak signals undergo greater amplification than do strong signals. The disparity between weak and strong reflectors is diminished. Dynamic

range (usually expressed in decibels) is the ratio of the largest to the smallest signal that can be accommodated by a system component. Logarithmic amplification reduces the dynamic range of the induced RF signals. Decreasing the dynamic range is called compression.

Most gain controls adjust only the amplification of the received signals and have no effect on the intensity of the generated ultrasound beam. The improved sensitivity in these cases is the result of extra amplification only and possible tuning of the receiver to allow detection of weaker signals.

Time Gain Compensation

One problem that must be considered is the attenuation of an ultrasonic beam with depth. Equally reflective interfaces produce different signal levels, depending on their relative distances from the transducer. It is often advantageous to display reflectors of similar size, shape, and reflection coefficients with equal signal strengths or brightness levels. Exponential amplification is used to correct the signals for attenuation. This can be easily demonstrated.

Assume that a phantom is composed of alternating layers of water and gelatin as shown in Fig 2.3 rather than signals of equal amplitude. To compensate for this attenuation, a time-gain compensation (TGC) control is used to increase the amplitude of processed signals with time or depth. The amplification could be a reverse exponential function, because the signals decrease exponentially. TGC also contributes to signal compression.

When the liver is scanned with TGC, all echoes should be of nearly the same amplitude, unless an abnormality is present or vessel is observed. Other terms for TGC include *depth gain compensation* (DGC), swept gain control, depth-varied gain, distance-attenuation compensation, and sensitivity-time control.

Fig 2.3 Phantom composed of alternating layers of water and gelatin

Automatic TGC (also called as adaptive gain control, autogain control, and auto-sensitivity control) adjusts the time varying gain without operator intervention. The amount of amplification is determined by the general decrease in echo-induced signal amplitude. This assumes that the detected structures have similar reflectivities.

Different tissues have varying rates of attenuation. The frequency response of the attenuation rate depends on the tissue type. This, along with the capability to enhance a

particular area of interest, makes a variable gain control desirable. Most diagnostic ultrasound units include a combination of TGC controls. The near gain or mean gain adjusts the level at which the initial signals are amplified. The delay regulates the depth at which the TGC begins. Often it is convenient not to amplify signals originating close to the transducer because they have high amplitude levels. The slope of the TGC indicates the amount of compensation that is applied with depth. Some ultrasound units have a completely variable gain control, permitting a particular area of interest to be enhanced beyond that provided by TGC. The far gain represents the maximum amount that signals can be amplified. In this zone the signals are amplified by a constant amount, but they exhibit an exponential decrease because of attenuation.

The variable TGC controls permit adjustable compensation for different frequency transducers, allowing greater amplification when high frequencies are used. Flexibility in the TGC controls enables higher quality images to be obtained. Nevertheless, the sonographer must be more aware of the functions of the different controls, furthermore, he or she must also interact more with the system during scanning. A scanner with selectable TGC controls can produce poor quality images when operated by an inexperienced individual. Automation of the TGC controls provides more consistent images, but they are often poorer in overall quality.

2.2.5 Signal Processing

The radiofrequency (RF) signals induced in the transducer by the returning echoes could be amplified and then displayed, but it would then appear similar to the acoustic pulse

sent out Fig 2.4. If numerous interfaces are present, the interpretation of the received signals becomes confusing. For, viewing ease, the goal is to process the signal before displaying it. The processing lowers the information content while, at the same time facilitating the association of final output with physical structures.

Fig 2.4 Comparison of radiofrequency and acoustic waveforms (A: Induced RF signal in a crystal, B: The acoustic wave)

The normal processing procedure involves rectification (converting the negative portion of the RF signal to positive) after the signal has been amplified and has undergone TGC amplification Fig. 2.5. Alternatively, the negative component of the RF wave can be eliminated rather than flipped to positive. The peaks of the RF waves are electronically surrounded or enveloped, resulting in a processed signal that could be further amplified for display purposes Fig 2.6. Enveloping is often referred to as demodulation and is generally accomplished by passing the signal through a circuit with a slow time response. The overall outline of the pulse is retained, but the internal fast oscillations are lost. This is not the true meaning of demodulation, but it shows how the term is used in this

context. Demodulation normally involves removing a signal that has undergone interference with a carrier wave, which is important in Doppler ultrasound.

Fig 2.5 Rectification (A: generation of radiofrequency due to incident sound wave, B: rectified signal negative components are flipped to become positive)

Fig 2.6 Enveloping (A is an induced radiofrequency signal, B is the enveloping i.e. an electronic surrounding of the peaks, C is the resultant enveloped signal)

Usually the area under the enveloped signal is electronically measured. Determination of the area is called integration. This area is then represented as a spike for A-mode scanning or as a dot for other scanning techniques. An increase in the amplitude of the induced RF signal results in a larger area under the curve and therefore an increase in the height of the spike or brightness of the dot. A reject control (also called threshold or suppression control) may be added to eliminate peaks below or above a certain level, as selected by the operator. The reject control is similar to the lower level (or upper level) discriminator used in radiation counting systems. Rejection of signals decreased the dynamic range.

Other techniques can be incorporated into signal processing. These include (1) leading edge detection- representing the beginning of the RF pulse as a spike or dot; (2) peak detection- representing the maximum height of the enveloped signal as a spike or dot; and

(3) differentiation- representing the slope of the enveloped signal as a spike or dot. Usually integration is employed in conjunction with leading edge detection.

2.2.6 Display

A cathode ray tube (CRT) is the simplest output device used for display purposes. It is a large evacuated (low-pressure) glass tube with a potential difference of 10,000 V from back to front. This voltage difference accelerates electrons emitted from the cathode (filament) at the back of the CRT toward the screen. A special focusing cup near the filament confines the electrons to a narrow beam and prevents them from spreading out over a wide area. The glass envelope maintains the vacuum. The glass screen on the front of the CRT is covered with a phosphorescent material (usually zinc sulfide, zinc sulfate, or potassium iodide) that produces light when struck with electrons. Two sets of special electrical or magnetic plates control the position of the electron beam on the X (across the screen) and Y (up and down the screen) axes of the tube face. Electrical fields are produced when a voltage difference is applied to one set of plates. If no voltage is applied, no electrical field is present and the electrons travel along a straight-line path to the center of the screen. Applying a positive voltage to one plate in the set, however, and a negative voltage to the other (i.e., creating a potential difference across the plates) deflects the beam toward the positive plate; and reversing the voltage deflects it in the opposite direction. The two sets of plates allow the beam to be positioned anywhere on the CRT screen.

For simplest scanning system (A-mode) a voltage to the X plates moves the time base sweep across the display at a constant rate that correlates with the speed of ultrasound in tissue. The sweep, which must advance the equivalent of 1cm on the distance scale for every 13 μ s, is initiated by a pulse from the master synchronizer; and as it progresses across the display, the signal detected for each interface is amplified and sent to the Y plates to control the electron beam's vertical position.

2.2.7 A-Mode Scanning

A-mode (amplitude-mode) scanning is based on the echo-ranging principle, similar to sonar. A pulsed ultrasound wave is directed into the patient's body, and the echoes generated at various interfaces are detected. Only structures that lie along the direction of propagation are interrogated. This sampling according to the beam path is called the line of sight or scan line.

The amplitude refers to the strength of a detected echo signal. A-mode displays the amplitude of the signal as a spike in the vertical dimension versus depth or time of the signal in the horizontal dimension. (Depth and time are interchangeable because they are directly proportional.) An increase in the amplitude or strength of the signal gives rise to an increase in the height of the spike. This variation in signal strength is due to the reflectivities of different interfaces and the attenuation of the beam as the ultrasound wave travels to and from the various interfaces. Time gain compensation attempts to correct for attenuation loss. Strong reflectors far from the transducer may produce

greater-amplitude signals than weak reflectors nearer the transducer. The A-mode scan also contains spatial information; that is, it registers the distance between interfaces.

Multiple interfaces encountered along the sampling direction are detected by a series of echoes. Fig 2.7 shows three interfaces as three separate spikes. Interface *I* generates a larger signal than interface *II* or interface *III*. This A-mode scan also preserves the spatial relationship of the observed structures. Interface *II* is located closer to interface *III* than to interface *I*.

Fig 2.7 A-mode scan and display (A, Three interfaces (I, II, and III), corresponding display of the interfaces

2.2.8 B-mode scanning

In B-mode imaging the amplitude of the signal (detected echo strength) is represented by the brightness of a dot. The A-mode spike on the CRT display is converted into a dot,

which can be demonstrated by rotating the spike 90 degrees out of the plane of the paper. (i.e., the axis of rotation is the horizontal axis). The amplitude or strength of the signal is designated by the brightness of the dot. The position of the dot represents the depth (time) of the interface from the transducer Fig. 2.8. Normally we are interested in constructing a two-dimensional image of the area of interest rather than acquiring data from a single scan line or line of sight. This is accomplished by compound B-scanning, where by multiple sets of dot are combined to delineate the echo pattern from internal structures with the body. The patient is scanned from many different directions. The superimposition of multiple scan lines creates a composite two-dimensional image, which has the advantage of portraying the general contour of the patient and the internal organs Fig. 3.9. Compound B-mode scanning produces a static image that can be envisioned as a stop-action photograph of the reflecting surfaces.

Fig 2.8 Comparison of B-mode and A-mode displays

Fig 2.9 Compound B-mode scanning (A-The transducer is moved to probe the patient along several lines of sight. B-In the display the interfaces are formed as a composite of dots along the lines of sight

2.3 Proposed Ultrasound Image Scanner

Ultrasound imaging has become a very widely used medical imaging modality because it is safe, noninvasive, portable, easy to use, relatively inexpensive, and displays images in real time. Ultrasound imaging allows the operator to move a transducer probe over the patient's body or inside and select the desired view to visualize tissue structures, perfusion, and blood flow patterns in real time. Tremendous computational power for real-time ultrasound processing is needed to support the capabilities and features available in modern ultrasound machines. The high data rates and computing power required to have restricted designs of ultrasound machines to algorithm-specific hardware with limited programmability.

Researchers and clinicians in medical ultrasound have been exploring exciting concepts and new applications, e.g., tissue characterization, speckle reduction, three-dimensional (3-D) acquisition and display, and image enhancement. Unfortunately, most of these ideas have not been realized into ultrasound machines because of the inflexibility of the algorithm-specific hardware found in most ultrasound machines. New applications cannot be easily implemented on the algorithm-specific hardware because many of these applications require various image processing algorithms like convolution, Fast Fourier Transform (FFT), median filtering, and more complex operations such as motion estimation, image registration, and image segmentation. Depending on the target application and its implementation strategy, a different set of algorithms is needed. Furthermore, the input–output and computational aspects of these algorithms vary widely

and change with time as the algorithms are refined and better implemented based on the results from clinical trials.

To resolve this problem, a research team at University of Washington has developed a new programmable and portable ultrasound machine [3]. The basic aim of this research is to develop programmable a computing system inside the ultrasound machine, which is capable of supporting the data flow and computational needs for a variety of algorithms. This will enable new clinical applications and innovative concepts in medical ultrasound to be developed, clinically tested, and routinely used natively within the ultrasound machine itself. Extensive research has been carried out to design a flexible and high performance programmable ultrasound processing system, which is termed as Programmable Ultrasound Image Processor (PUIP). Various clinical applications were developed on it to expand the diagnostic capability and efficiency of medical ultrasound machines.

Fig. 2.10 shows the schematic diagram of a next-generation ultrasound scanner with dual image processors. Current generation ultrasound scanners only have the Base Ultrasound Image Processor, not the PUIP. By processing the data from the beamformer, the Signal Processor produces pre-scan converted image information. The Base Ultrasound Image Processor further processes the pre-scan converted data into basic B-mode, color flow, and Doppler images. Both the Signal Processor and the Base Ultrasound Image Processor operate in a hardwired fashion. On the other hand, the PUIP processes in a programmable fashion the data coming from the Signal Processor directly or the processed signal from

the output of the Base Ultrasound Image Processor for advanced ultrasound image processing.

Fig 2.10 Next Generation Ultrasound Machine with dual image processor

2.4 Ultrasonic Transducers

Ultrasonic transducer is the most critical part, and the front of the front end of the ultrasound system. The ultrasound waves are generated in transducer and transmitted to the human body. The reflected ultrasound signals from the tissues are received by transducer, and then sent to ultrasound system through coaxial cable for further processing and image formation. The material used for the ultrasonic transducer is one of the critical parameters, which affects the bandwidth of the transducer and further affects the image quality of the ultrasound systems. Various researchers have done a lot of work to find suitable material for the transducer fabrication.

2.4.1 Transducer Selection

The following critical parameters have to be taken into consideration, while selecting the transducer for the ultrasound system.

Frequency

The depth of the structures of interest is a primary consideration when selecting the optimum transducer. The objective is to enhance lateral resolution by scanning the area of interest with a narrow beam. Evaluation of the near field depth and far-field divergence indicates that transducers with high frequencies and large diameters maintain beam shape to greater depths. Although large-diameters have a large near-field beam width and thus provide poor lateral resolution at shallow depths, the near field extends to greater depths in the body compared with a smaller diameter transducer. The divergence of the beam in the far field is less which makes the larger transducer preferable for deep-lying structures Fig 2.11.

Fig 2.11 Effect of transducer diameter on lateral resolution

Fig 2.12 Effect of frequency on lateral resolution, A: High Frequency Transducer and B: Low frequency transducer

A similar effect is observed with frequency; that is, with other factors constant, an increased frequency results in a deeper near field and a less diverging far field. This may compensate for the loss in penetration Fig 2.12.

If shallow structures and irregular surfaces are of interest, a small diameter high-frequency transducer should be used.

Q-value

The Q value, or mechanical coefficient, assesses two essential characteristics of the pulsed ultrasound beam-axial resolution and bandwidth. The Q value can thus be thought of as having two separate definitions relating to the two affected beam characteristics. These definitions are

$$Q = \frac{\text{Energy Stored per cycle}}{\text{Energy lost per cycle}}$$

$$Q = \frac{\text{Centrefrequency}}{\text{Bandwidth}}$$

A high-Q transducer stores energy in the crystal and therefore loses very little each cycle. After being stimulated by the voltage pulse, it vibrates (rings) for an extended duration, producing a long pulse. A low-Q transducer, on the other hand generates a short pulse after excitation because most of its energy is lost during the first few vibrations Fig. 2.13. Diagnostic pulsed-wave ultrasound uses low-Q transducer, the Q value being typically 2 to 3. High-Q transducers (700 or greater) are good for continuous-wave ultrasound. Quartz, with a Q value of 25000 is advantageous for therapeutic applications.

Fig 2.13 Pulse length versus ringing, A, 6-cycle pulse (High Q, long ringing); B, 2.5 cycle pulse (low Q, short ringing)

Axial Resolution

The Q of a transducer in conjunction with the backing material used for damping and the frequency of the transducer determines the pulse length, which in turn limits the axial resolution of the transducer. The axial resolution specifies how close together two objects can be along the axis of the beam and yet still be detected as two distinct entities. Axial resolution also specifies the smallest object detectable along the axis of the beam. For a constant wavelength, as the pulse is shortened the axial resolution improves.

For diagnostic ultrasound imaging, the resolution (minimum object size detectable) should be on the order of 1 mm. Commonly, the spatial pulse length is used as a specifier

for the axial resolution. The best possible axial resolution is the spatial pulse length divided by 2.

Bandwidth

From Equation, it can be seen that the Q value is indirectly proportional to the bandwidth or frequency spread of a transducer. The narrow bandwidth decreases the sensitivity of the transducer. For a transducer operating with a continuous output, only a single frequency is generated. For a pulsed system, however, because of imperfections in the crystal and the complex nature of the damped pulse, a range of frequencies is generated. Many waves of differing frequency combine to form the pulse. Fourier analysis enables a complex waveform to be broken into its various frequency components. An algebraic summation of sine waves of varying frequency yields the original waveform, in other words, any complex waveform can be considered a series of sine waves with different frequencies and amplitude.

A frequency spectrum can be analysed by measuring the amplitude versus the time signal induced by an echo returning from flat steel target located a fixed distance from the transducer. With Fourier analysis the signal wave form is converted from the time domain to the frequency domain. Frequency domain is a mathematical term meaning that the various frequency components of the signal are identified. Amplitude plotted against frequency shows a spectrum of values, with the centre frequency (f_0) having the greatest amplitude Fig 2.14. Amplitude in this instance indicates the relative importance of each

frequency in the ultrasound pulse. The amplitude represents the fraction of the pulse composed of waves of a particular frequency.

Fig 2.14 Relationship between the pulse length and the frequency spectrum

The centre frequency is the primary operating frequency or the natural resonance frequency of the transducer, which depends on the crystal thickness.

$$Q = \frac{f_0}{\Delta f}$$

Where Δf is the bandwidth between the 6 dB points on each side of the centre frequency (f_0).

A 6 dB loss occurs if the amplitude is reduced to half its original value. From the frequency spectrum, the frequency corresponding to a 6 dB loss is marked on each side of the centre frequency. The bandwidth is calculated by subtracting the lower frequency from the higher frequency Fig 2.15. Sometimes the bandwidth is specified as the half-power bandwidth. The half-power bandwidth corresponds to a 3 dB loss, or the amplitude is reduced by a factor of $2^{1/2}$. The half-power bandwidth is always narrower than the half-amplitude bandwidth.

Fig 2.15 Determination of bandwidth by subtracting f_1 from f_2 , Limits f_1 and f_2 correspond to one half the centre frequency amplitude

Fig 2.16 Comparison of frequency spectra generated by low-Q and high-Q transducers: low Q, broad bandwidth ($f_1''-f_1'$) versus high Q, narrow bandwidth ($f_h''-f_h'$)

When the second definition of Q value for a transducer is used, it becomes evident that a high-Q transducer has a very narrow bandwidth whereas a low-Q transducer has a broad bandwidth Fig 2.16.

As the beam is transmitted through matter, ultrasound waves of higher frequency are absorbed more rapidly than those of lower frequency, causing a shift in the frequency spectrum toward a lower range. This is similar to the shift toward higher energies as an x-ray beam penetrates the body, because low energies are absorbed more readily. The preferential absorption of the high-frequency components influences the sensitivity of the

system. A broadband transducer (as opposed to one with a narrow bandwidth) has greater sensitivity as the sound beam penetrates tissue. A transducer with a broad frequency spectrum is also easier to match electronically with the rest of the ultrasound system, which allows the receiver to be tuned to variable frequencies to maximize the sensitivity at various depths.

Acoustic Impedance

Another transducer problem is the poor acoustical match between tissue and PZT. In ultrasound the quantity analogous to momentum is acoustic impedance. In classical mechanics “momentum” is equal to mass times velocity, in ultrasound “density” replaces the mass and is then multiplied by the velocity. The velocity used is that of sound in the medium. The product of density (ρ) times velocity (c) is called the acoustic impedance (Z).

$$Z = \rho c$$

This quantity is a measure of the resistance to sound passing through the medium. It is similar to electrical resistance, which is the degree difficulty experienced by electrons in traversing a specific type of material. Acoustic impedance is expressed as kilograms per square meter per second ($\text{Kg/m}^2/\text{s}$) and is the product of the units of density times velocity. The combination of units in the meter-kilogram-second (mks) system is given a special name the Ryal. The bulk acoustical impedance of PZT-5H is $Z = 34.2 \text{ MRayls}$ and for water is $Z_{\text{eff}} = 1.5 \text{ MRayls}$. Tissue is mostly water and therefore the impedance of water is a good approximation for most soft tissue. The reflection coefficient for this interface is given by

$$R = \frac{Z_{eff} - Z_{PZT-5H}}{Z_{eff} + Z_{PZT-5H}} = -0.916$$

Which means that $R = 91.6\%$ of the sound is reflected back towards the transducer and does not enter the water due to the acoustical impedance mismatch. To overcome this problem, a quarter wavelength thick matching layer, with the acoustic impedance of the geometric mean of the impedances of water (or tissue) and PZT-5H, is used to match the two layer at the operating frequency. The effective acoustic impedance seen by the PZT is

$$Z_{eff} = Z_{ML} \frac{Z_{water} \cos \beta l + jZ_{ML} \sin \beta l}{Z_{ML} \cos \beta l + jZ_{water} \sin \beta l}$$

Where Z_{ML} is the acoustic impedance of the matching layer ($Z_{ML} = 7.2$ MRayls for water and PZT-5H). $\beta = 2\pi l/\lambda$, and l is the thickness of the matching layer. This is not broadband solution and provides only moderate matching. In practice, two matching layers are often used for more efficient and broadband transmission of the sound into tissue.

2.4.2 Selection of Transducer Materials

Ferro electric materials strongly exhibit the piezoelectric effect, and they are ideal materials for medical ultrasound. For many years, the ferroelectric ceramic leadzirconate-titanate (PZT) has been the standard transducer material for medical ultrasound, in part because of its high electromechanical conversion efficiency and low intrinsic losses. The properties of PZT can be adjusted by modifying the ratio of zirconium to titanium and introducing small amounts of other substances, such as lanthanum [4]. PZT has a high dielectric constant compared with many piezoelectric materials, resulting is favorable

electrical characteristics. Ceramic is mechanically strong, and it can be machined to various shapes and sizes. PZT can operate at temperatures up to 100°C or higher, and it is stable over long periods of time. The disadvantages of PZT include its high acoustic impedance ($Z = 30 \text{ MRayls}$) compared with body tissue ($Z = 1.5 \text{ MRayls}$) and the presence of lateral modes in array elements. One or more acoustic matching layers can largely compensate for the acoustic impedance mismatch. The effect of lateral modes can be diminished by choosing the appropriate element dimensions or by subsidizing the elements. Other piezoelectric materials are used for various applications. Composites are made from PZT interspersed in an epoxy matrix [5]. Lateral modes are reduced in a composite because of its inhomogeneous structure. By combining the PZT and epoxy in different ratios and spatial distributions, one can tailor the composite's properties for different applications.

As 2-D array transducers have become essential to implement focusing and phase-correction in the elevation dimension as well as real time volumetric imaging. Unfortunately, the small size of 2-D array element results in a small clamped capacitance and a large electrical impedance near resonance. These elements have poor sensitivity because their impedance is much higher than the electrical impedance of the transmit and receive circuitry. Sensitivity can be improved by using an N layer structure of PZT ceramic with the layers connected acoustically in series and electrically in parallel [6]. The multiplayer ceramic (MLC) was manufactured using thick film technology with plated through vias to electrically interconnect the electrode layers. The resulted MLC is

having lower impedance as compared to single layer. It has lower insertion loss and images clearly showed a higher signal-to-noise ratio for the MLC array.

Polyvinylidene difluoride (PVDF) is a ferroelectric polymer that has been used effectively in high-frequency transducers [7]. The copolymer of PVDF with trifluoroethylene has improved electromechanical conversion efficiency.

Relaxor ferroelectric materials, with general formula, $\text{Pb}(\text{B}_1\text{B}_2)\text{O}_3$ ($\text{B}_1: \text{Zn}^{2+}, \text{Mg}^{2+}, \text{Sc}^{3+}, \text{Ni}^{2+} \dots, \text{B}_2: \text{Nb}^{5+}, \text{Ta}^{5+} \dots$) are some materials of superior dielectric and piezoelectric properties as compared to PZT. Lead-magnesium-niobate (PMN) is most common among them. It becomes piezoelectric when a large direct-current (DC) bias voltage is applied [8]. They have a very large dielectric constant ($\epsilon > 20,000 \epsilon_0$), resulting in higher transducer capacitance and lower electrical impedance.

The use of high frequency focused single element ultrasonic transducers is becoming increasingly widespread due to the importance of ultrasound backscatter microscopy (UBM) as an ophthalmologic imaging tool. These kinds of transducers are used to resolve major structures of the eye such as cornea, iris, lens, and retina. Lithium niobate (LiNbO_3) was chosen as the active element because a highly sensitive device is desired for the UBM application [8].

This paper gives the fabrication technique to produce focused single element transducers in the range of 25MHz- 100MHz with bandwidths exceeding 60% and insertion loss level

in the -15dB to -30dB range. Lithium niobate (LiNbO_3) was chosen as the active element because a highly sensitive device is desired for the UBM application. A high thickness mode coupling coefficient ($k_t=0.49$) and low clamped dielectric constant ($\epsilon^S=28$) make LiNbO_3 a good choice in producing transducers that are efficient and easily matched to the electrical load. The major problem with this material is its' high acoustic impedance ($Z_{\text{eff}}=34\text{MRayls}$) when compared to the water load. In order to better match the active element to water, a $7.3\text{ MRayls } \lambda/4$ matching layer was used in this design.

2.4.3 Scanning with Array Transducers

Array transducers use the same principles as acoustic lenses to focus an acoustic beam. In both cases, variable delays are applied across the transducer aperture. With a sequential or phased array, however, the delays are electronically controlled and can be changed instantaneously to focus the beam in different regions. Linear arrays were first developed for radar, sonar, and radio astronomy, and they were implemented in a medical ultrasound system by Somer in 1968. Linear-array transducers have increased versatility over piston transducers. Electronic scanning involves no moving parts, and the focal point can be changed dynamically to any location in the scanning plane. The system can generate a wide variety of scan formats, and it can process the received echoes for other applications, such as dynamic receive focusing, correction for phase aberrations [9], and synthetic aperture imaging [10]. The disadvantages of linear arrays are due to the increased complexity and higher cost of the transducers and scanners. For high-quality ultrasound images, many identical array elements are required (currently 128 and rising).

The array elements are typically less than a millimeter on one side, and each has a separate connection to its own transmitter and receiver electronics.

The widespread use of array transducers for many applications indicates that the advantages often outweigh the disadvantages. In addition, improvement in transducer fabrication techniques and integrated circuit technology has led to more advanced array transducers and scanners.

2.6 Capacitive Micromachined Ultrasonic Transducer

The capacitive Micromachined ultrasonic Transducers (cMUTs) are novel transducers, which are having potential to replace conventional ultrasonic transducers in the commercial ultrasound systems used in hospitals, as they offer the advantage of increased bandwidth with comparable sensitivity to piezoelectric transducers. These transducers are easy to fabricate, as they use conventional CMOS processes. The cost of the transducer is considerably reduced due to batch production of the transducer, which is one of the major advantages over conventional piezoelectric transducer. As the cMUT is fabricated by using conventional COMS process, it can be easily integrated to the front-end electronics of an ultrasound system. This section gives review of cMUTs and some of the fabrication processes used in fabricating capacitive micromachined ultrasonic transducers

Since it has been pioneered in the early 1990s at Stanford University [11], [12], extensive research has been conducted on modeling, fabrication and characterization of this new

type of transducers. The fabrication process of the transducer has been reported earlier [13], which is based on surface micromachining techniques. Further research in this area of fabrication of cMUT found one more promising fabrication technique using wafer bonding technology [14]. During early years of research on cMUTs, the transducer was designed using simple equivalent circuit, which is based on Mason's model [15], and the validity of the model has been confirmed by experimental results [16]. One dimensional linear cMUT arrays have been characterised, including the acoustical cross-talk in this array.

2.6.1 Review on Finite element analysis of cMUTs

Finite element analysis is an important tool, which has been used by many researchers to understand the transducer characteristics such as cross-coupling phenomenon in the transducer and to design and optimize some of the critical parameters of the transducer. Finite element analysis has been used by Yongrae Roh and Butrus T. Khuri-Yakub [17] to study the cross talk between the array elements using commercial code ANSYS. The analysis shows origin and level of crosstalk between array elements through Stoneley and Lamb waves. They investigated various structural variations of the wafer such as change of wafer thickness, etched trenches in the wafer and the wall structure between array elements to reduce the cross talk between array elements. Alessandro Caronti et al also studied cross talk phenomenon using finite element analysis [18]. They also studied several variations of the array design to reduce cross talk. Bayram et al studied the dynamic behaviour of immersion cMUT transducer using the time domain, nonlinear finite element package, LS-DYNA [19]. They compared the pressure generated by the cMUT during conventional and collapse-snapback operations. Eccardt et al used finite

element analysis to derive linear and non linear transducer equivalent circuits. By using these models the nonlinear behaviour of the cMUT can be investigated much easier and faster compared to FEM simulations alone [20]. Massimo Pappalardo et al studied the cross talk phenomenon through 3-D finite element models using ANSYS [21]. They validated the FEM results with optical displacement measurements. Ladabaum et al designed time domain models of cMUT array using commercial software PZflex to study the cross-talk in silicon substrate [22] and substrate ringing in cMUTs [23].

2.6.2 Review on Fabrication and characterization

The capacitive micromachined ultrasonic transducer was first fabricated in Stanford University by Jin X. C. et al by using surface micromachining technique [12]. This is the conventional fabrication process used for building the cMUTs. The main principle of building membranes that are supported on all sides above a substrate is first depositing or growing a sacrificial layer on the substrate. Then, this sacrificial layer is covered with the membrane material, and removed with wet etch through holes in the membrane. This principle stayed the same throughout the evolution of the conventional cMUT fabrication, but the way of defining the sacrificial layer changed. Various researchers tried different methods for cavity formation and cavity sealing. To facilitate the cavity formation in cMUT and to reduce the number of steps involved in the fabrication process, wafer bonding technology has been first used to fabrication transducer in Stanford University [14]. Fabricating cMUTs by the wafer-bonding technique in which cMUT cell cavity is formed in one wafer and membrane of the transducer is formed on the other transducer. The two wafers are then bonded together by using conventional wafer bonding

techniques such as fusion bonding or eutectic bonding techniques. This process provides more uniform cell membrane. It doesn't require to deposit and etch sacrificial layer. The number of steps involved in the transducer fabrication is reduced due to this technique.

The characterisation of the cMUT transducer has been done by various researchers to measure various parameters such as output pressure of the transmitted ultrasound waveforms, receive sensitivity of the transducer and the fractional bandwidth, and to study the cross coupling [24-32]. Jin et al did the characterization of 1-D and 2-D transducer to find out the receive sensitivity and the fractional bandwidth of the transducer [16], [17], [25]. They also did radiation pattern measurements. They also studied the radiation patterns to confirm the cross talk phenomenon due to stoneley waves and lamb waves. McIntosh et al did the characterisation of the cMUT transducer in air, when driven from a standard ultrasonic voltage source [26]. Experiments were conducted using 1mm square devices in air operating in both pitch-catch and pulse echo modes. The dependence on DC bias voltage was examined, together with calibration measurement using microphones. The radiated beam profile, and far-field directivity pattern, has been measured for both broad bandwidth and one-burst excitation, using a scanned miniature receiver.

Oralkan et al presented first pulse-echo phased array B-scan sector images using a 128-element, 1-D linear cMUT array to demonstrate the viability of cMUTs for ultrasound imaging, especially for medical and underwater applications [27]. They fabricated 128 elements 1-D cMUT arrays with uniform response across the arrays. The arrays have

been operated in immersion with no failure or degradation in performance over the time. For doing imaging experiments, they constructed a resolution test phantom roughly mimicking the attenuation properties of soft tissue. The PC-based experimental system, including custom-designed electronic circuits to acquire the complete set of 128 x 128 RF A-scans from all transmit-receive elements combinations. The pulse echo response analysis has been conducted. They obtained the centre frequency and the fractional bandwidth of the transducer. They constructed B-scan images with a sector angle of 90 degrees. They also performed preliminary investigations on the effects of crosstalk among array elements on the image quality. Oralkan et al also presented first volumetric images obtained using 2-D cMUT array. They fabricated 128 x 128 element 2-D cMUT array with through wafer interconnects [28]. They used PC-based experimental system for further processing of RF data from the transducer. They conducted the pulse echo response of the transducer to find out the fractional bandwidth and the centre frequency of the transducer. The images were constructed using RF beamforming and synthetic phased array approaches and visualized by surface rendering and multiplanar slicing techniques. Their study experimentally demonstrated that 2-D cMUT array can be fabricated with high yield using silicon IC fabrication processes, individual electrical connections can be provided using through wafer vias, and flip-chip boning can be used to integrate these dense 2-D arrays with electronic circuits for practical 3-D imaging applications. David Mills et al presented real time in-vivo images from a cMUT array and compares them to PZT array images that were made with GE LOGIQ 9 ultrasound imager [29]. The cMUT images show equivalence in most cases to PZT images. They found that in some cases anatomical borders and texture are more clearly resolved in the

cMUT array images. They measured the fractional bandwidth and compared with the PZT transducers. It was also found that the sensitivity of the cMUT transducer is lesser as compared to that of the PZT arrays. They also found some existing challenges in the present cMUT technology such as the reliability of the transducer, cross talk between the array elements, packaging and spectral notches.

In this thesis the research work is focused on the cross talk between the elements of the transducer and with in the elements of the transducer. In this research work the effect of cross talk on the fractional bandwidth of the transducer has been studied in detailed. It also provides novel design to reduce the crosstalk between the elements.

2.6.3 Comparison with conventional transducers

Throughout the history of ultrasound imaging, piezoelectric crystals, ceramics, polymers, and recently piezo-composite materials have been used to generate and detect ultrasound. Although the idea of capacitive ultrasound transducers is as old as the early piezoelectric transducers, piezoelectric materials have dominated ultrasonic transducer technology. The reason why capacitive transducers have not been popular is that electric field strength is less as compared to piezoelectric transducers. However, recent advances in micro fabrication technology have made it possible to build capacitive ultrasound transducers competing with piezoelectric transducers. Moreover, cMUTs offer advantages of improved bandwidth, ease of fabrication of large arrays with individual electrical connections, and integration with electronics [27].

CMUT technology is not simply a low-cost replacement of piezoelectric transducer technology. Many features inherent in cMUT technology enable revolutionary advances in ultrasound imaging. Currently, real time volumetric imaging is the focus of extensive research in ultrasound. The realization of such systems depends on design and fabrication of 2-D transducer arrays. There are difficulties in fabricating these arrays due to limitations in the existing transducer array and interconnect technologies. CMUTs are fabricated using standard silicon integrated circuit (IC) fabrication technology. This technology makes it possible to fabricate large arrays using simple photolithography. Individual electrical connections to transducer elements are provided by through-wafer interconnect. Two-dimensional CMUT arrays with as many as 128x128 elements already have been successfully fabricated and characterized [25]. These 2-D arrays can be integrated with electronics in the form of a 3-D multichip module by flip-chip bonding [28].

Another enabling feature inherent to CMUT technology is wide bandwidth. A wideband transducer does not simply increase the resolution, but it also enables the design of new image modalities and analysis tools. A prominent area of research in medical ultrasound is tissue harmonic imaging, in which energy is transmitted at a fundamental frequency and an image is formed from the energy at the second harmonic [30]. In current harmonic imaging systems, the transmit frequency is set to $2/3$ of the center frequency, and the transmit frequency is set to $2/3$ of the center frequency and the receive frequency is set to $4/3$ of the center frequency of the transducer, resulting in sub optimal operation both in transmit and receive [31]. CMUTs provide a flat response over a wide frequency range, enabling optimal tissue harmonic imaging.

CMUTs also are promising for high-frequency applications such as intravascular ultrasound imaging (IVUS), in which high-frequency operation using miniature probes is vital. CMUTs operating at frequencies as high as 60MHz have been fabricated and tested successfully [32]. Experimental front looking and side-looking IVUS arrays also have been designed and fabricated. Another area of extensive research is ultrasonic tissue characterization, often based on spectral analysis [33] and sub band processing [34] of backscattered signals, in which wide bandwidth is crucial.

CMUTs have many promising applications other than medical and underwater imaging as well. Air-coupled, non-destructive evaluation [35], microphones with RF [36] and optical detection schemes [37], surface and bulk acoustic wave devices, and smart micro fluidic channels [32] are among these applications.

Chapter 3 : Basic Theoretical Foundation

3.1 Basic structure of Linear Transducer array

Designing an ultrasound transducer array involves many compromises. Ideally, a transducer has high sensitivity or SNR, good spatial resolution, and no artifacts. The individual array elements should have wide angular response in the steering dimensions, low cross coupling, and electrical impedance matched to the transmitter. Fig. 3.1 illustrates the connections to the transducer assembly. The transmitter and receiver circuits are located in the ultrasound scanner and are connected to the array elements through 1 to 2 m of coaxial cable. Electrical matching networks can be added to tune out the capacitance of the coaxial cable and/or the transducer element and increase the signal-to-noise ratio (SNR). Acoustically, the array elements are loaded on the front side by one or two quarter-wave matching layers and the tissue medium. The matching layers may be made from glass or epoxy. A backing material, such as epoxy, loads the backside of the array elements. The faceplate protects the transducer assembly and also may act as an acoustic lens. Faceplates are often made from silicone or polyurethane.

Fig 3.1 General Components of Piezoelectric Ultrasonic Transducer

3.2 Mathematical description of piezoelectricity

In a solid material, longitudinal waves normal to plates can be described by a scalar compressibility as for fluids. For a dielectric, non-piezoelectric material, which is elastic with compressibility κ and dielectric constant ϵ , we have the following Equations describing the material,

$$p(z,t) = -\kappa^{-1} \frac{\delta\psi(z,t)}{\delta z} \quad \text{Mechanical} \quad (3.1a)$$

$$E(z,t) = \epsilon^{-1} D(z,t) \quad \text{Electrical} \quad (3.1b)$$

Where E is electrical field and D is the electric displacement. The electrostriction is so weak that in these equations we have approximated the coupling between the electric field and mechanical deformation to zero. Mechanically we can have pressure waves in

the material and electrically the plate will be a capacitor without coupling between the mechanical and electric fields. Inside the material there is no free charge, and we can calculate the electric displacement from the electrostatic equation

Fig 3.2 Cross section of piezoelectric plate with load and backing material used as ultrasonic bulkwave transducer

$$\text{div } D = \frac{\partial D}{\partial z} = 0 \text{ which gives } D(z, t) = q(t) \quad (3.2)$$

for $z \in [0, L]$ and 0 else $q(t)$ is the electric charge per surface area on the left electrode in Fig. 3.2. The charge on the right electrode is $-q(t)$. The electric field is then found from Eq. (3.1b) and the voltage between the left and the right electrode is

$$V_3(t) = \int_0^L dz E(z, t) = \int_0^L dz \frac{q(t)}{\epsilon} = \frac{q(t)L}{\epsilon} \quad (3.3)$$

The plate of such a material behaves electrically as a capacitor with capacitance per unit area gives as

$$C_0 = q(t) / V_3(t) = \varepsilon / L \quad (3.4)$$

For ferroelectric material that is polarized along the z -axis, the material equations for transversally clamped compressions in the z -direction take the form

$$p(z, t) = -\kappa^{-1} \frac{\partial \psi(z, t)}{\partial z} + hD(z, t) \quad (3.5a)$$

$$E(z, t) = -h \frac{\partial \psi(z, t)}{\partial z} + \varepsilon^{-1} D(z, t) \quad (3.5b)$$

Where the electric field $E(z, t)$ and the electric displacement $D(z, t)$ are directed along the z -axis, ψ is particle displacement, h is the piezoelectric constant and ε is the dielectric constant of the material. We now have a coupling between the mechanical and electric material equations, where the coupling is described by the piezoelectric constant h .

To get a rough insight into the physical aspects of Eq. (3.5a, 3.5b) we integrate the electric field to find the voltage between the electrodes in Fig. 3.2. In analogy with Eq. (3.3) we get

$$V_3(t) = \frac{q(t)}{C_0} - h \{ \psi(L, t) - \psi(0, t) \} = \frac{q(t)}{C_0} - h\psi_T(t) \quad (3.6)$$

Where $\psi_T(t) = \psi(L, t) - \psi(0, t)$ is the increase in plate thickness, and C_0 is the clamped capacitance per unit area of the plate when ψ_T is clamped to zero. Hence, an increase in plate thickness develops a voltage $-h\psi_T(t)$ across the electrodes. This could also be viewed as an extra charge $-hC_0\psi_T(t)$ on the electrodes, and if we short the electrodes

during this change in thickness, a charge $q = hC_0\psi_T(t)$ will flow between the electrodes to keep $V_3=0$

3.3 Standard piezoelectric material constants

The piezoelectric materials are generally anisotropic and describe a transversally clamped material. We note that it is two linear equations, relating 4-field variable: The pressure p , the strain $\partial\psi/\partial z$, the electric displacement D , and the electric field E .

In solid materials it is common to use the mechanical tension $T=-p$ as a variable. The strain, S , is defined as $S = \frac{\partial\psi}{\partial z}$. The state of the piezoelectric material can thus be described by four variables:

Mechanical variables:	Electrical Variables:	
Strain: S	Field: E	
Tension: T	Displacement: D	(3.7)

The material equations are thus two equations relating these four variables and can be linearized when the amplitude of the variables is small, to take the common forms:

Piezoelectric const:	d	e
Direct effect:	$D = dT + \varepsilon^T E$	$D = eS + \varepsilon E$
Indirect effect:	$S = S^E T + dE$	$T = c^E S - eE$

(3.8)

Piezoelectric const:	g	h
Direct effect:	$E = -gT + D / \varepsilon^T$	$E = -hS + D / \varepsilon^T$
Indirect effect:	$S = s^D T + gD$	$T = c^D S - hD$

Where c is the mechanical stiffness constant, s is the mechanical compliance constant, and ε is the dielectric constant. The superscript indicates the variable that is constant in the measurement. Eqs. (3.5a, b) are the h-equations with

$$p = -T \quad \kappa = 1 / c^D \quad \varepsilon = \varepsilon^S \quad (3.9)$$

All the constants are related, and the relations can be found by rearrangement of the Equations. For example, the e-equations can be brought into the form of the h-equations

$$E = -e / \varepsilon^S S + D / \varepsilon^S$$

$$T = (c^E + e^2 / \varepsilon^S) S - e / \varepsilon^S D$$

giving the relations

$$h = e / \varepsilon^S \quad c^D = c^E + e^2 / \varepsilon^S$$

$$e = h \varepsilon^S \quad c^E = c^D - h^2 \varepsilon^S \quad (3.10)$$

It thus appears as if the material with D constant is stiffer than the material with E constant. D constant is achieved in the transducer with open electric port, since $q=D$ is constant. The wave velocity as we have introduced it is therefore the one obtained from this constant

$$c = v_1^D = 1 / \sqrt{\rho \kappa} = \sqrt{c^D / \rho} \quad (3.11)$$

The situation with E constant can only be achieved for the DC situation, since external electrodes can only influence the displacement D . With vibrations in the plate thickness, shorting the electrodes will not give zero E -field because it also depends on the strain in the material. The voltage is the integral of the E -field across the plate, and is zero with shorted plates, but still the E -field is different from zero and varies with z so that its integral is zero. By rearranging the e- and h-equations, we find the d-constants and the g-constants as:

$$d = e / c^E \quad \varepsilon^T = \varepsilon^S + e^2 / c^E \quad s^E = 1 / c^E \quad (3.12)$$

$$g = h / c^D \quad \varepsilon^T = \varepsilon^S (1 + h^2 \varepsilon^S / c^E) \quad s^D = 1 / c^D$$

The piezoelectric coupling factor, K , is often used to characterize the piezoelectric strength of a material. The square of it is defined as the ratio of the electric, W_E , to the mechanical, W_M , potential energy in the material when it is strained. The piezoelectric factor is then given by

$$K^2 = \frac{W_E}{W_M} = \frac{\varepsilon^S E^2}{c^E S^2} \quad (3.13)$$

The ratio of the electric potential energy to the total potential energy is called the electromechanical coupling factor, k_t

$$k_t^2 = \frac{K^2}{1 + K^2} = \frac{e^2}{(c^E + e^2 / \varepsilon^S) \varepsilon^S} = \frac{e^2}{c^D \varepsilon^S} = \frac{h^2 \varepsilon^S}{c^D} = h^2 \varepsilon K = g \cdot e \quad (3.14)$$

3.7 Capacitive Micromachined Ultrasonic Transducers (cMUTs)

Capacitive micromachined ultrasonic transducers (cMUTs) have been recently emerging as an attractive alternative to conventional piezoceramic transducers, since their first introduction [31-34]. They have been shown to have broad frequency bandwidth, very good sensitivity, and having low cost. Besides, cMUTs are built on silicon using standard surface micromachining techniques, and therefore have all the advantages of IC processing, such as parallel production, batch fabrication and very high level of integration. All these qualities made cMUTs and cMUT arrays an alternative to Piezoceramic transducers. Apart from this currently real-time volumetric imaging is the focus of extensive research in ultrasound. The realization of such system depends on design and fabrication of 2-D transducer arrays. There are difficulties in fabricating these arrays and interconnect technologies because of small element size. The fabrication of 2D cMUTs arrays involves the use of electrical through-wafer interconnects (ETWI) which brings electrical connection of each element from transducer side to the backside of the wafer.

3.7.1 Basic Principle

CMUTs are made of silicon nitride (Si_3N_4) membranes that are supported on all sides of above a conductive silicon substrate. The gap between the membrane and the silicon substrate is vacuum-sealed for immersion applications. Another electrode is deposited on top of the Si_3N_4 membrane, which creates a parallel plate capacitor with the conductive silicon substrate as shown in Fig.3.3.

Fig 3.3 A schematic cross-section of a cMUT

The principle of operation of cMUTs in transmit relies on the electrostatic attraction force between charges of opposite polarities [35]. When a DC Voltage is applied to the cMUT cell between the top and bottom electrode, the membrane of the cMUT cell gets deflected towards bottom substrate, which is ground electrode. The electrostatic attraction force is balanced by the restoring force of the mechanical spring formed by the membrane. If the DC bias voltage is removed, the cMUT membrane restores back its initial condition. If DC bias voltage is increased further, the deflection of cMUT membrane will increase. At some bias voltage there is sudden collapse of cMUT membrane. At this stage, the cMUT membrane can not restore back its initial position due to plastic deformation of membrane. The particular bias voltage is called collapse voltage. The expression for the collapse voltage is derived in the previous literature [12]. It is necessary to apply DC voltage just below the collapse voltage for the maximum efficiency of the transducer [59, 62]. If AC voltage is applied, membrane vibrates and sends mechanical energy in surrounding medium. On receive the membrane vibrations are converted into electrical current by applying a DC voltage across the cMUT. The capacitance variation of the cMUT caused by the membrane vibrations generates a current in external circuit. This

current is converted into voltage across a resistor and then amplified for further processing.

3.7.2 Theoretical Background

In both the analysis and design of cMUTs, it is important to differentiate between a cMUT as a receiver and a cMUT as an emitter. The purpose of a receiving cMUT is to detect the small ultrasonic signals that result after insonification of a sample. Thus, a linearized small signal analysis is appropriate. In contrast, the aim of an emitting cMUT is to launch ultrasonic waves of sufficient magnitude to enable applications of interest. Hence, the displacement of the emitting cMUT's membrane can be large, and a more involved nonlinear analysis is necessary to quantitatively predict a collapse voltage and an electromechanical coupling factor. The first order analysis of an emitting cMUT, derivation for collapse voltage, equivalent circuit are presented in detail by research group at Stanford university [12]. Another accurate model has been suggested by Alessandro Caronti *et. al.* [70]. The linear and nonlinear circuits for cMUTs are presented in detail by Khuri-Yakub *et. al.* [12] and Peter-Christian Eccardt *et. al.* [21]. Alessandro Caronti presented the acoustic coupling models for the cMUTs [22]. Greg Wojcik *et. al.* studied the time domain models of the cross-talk in the cMUTs [23]. The influence of the electrode size on the transducer performance and efficiency has been studied by Baris Bayram *et. al.* [59]. Goksen G. Yaralioglu *et. al.* calculated the residual stress and Young's modulus of the membrane of the cMUTs [60]. He further calculated and measured the electromechanical coupling coefficient of cMUTs.

3.7.5 Cross talk

Lamb waves refer to elastic perturbations propagating in a solid plate with free boundaries, for which displacements occur both in the direction of wave propagation and perpendicular to the plane of the plate. Lamb wave represent one of the types of normal or plate modes in an elastic waveguides, in this case a plate with free boundaries.

Consider a plane harmonic Lamb wave propagating in a plate of thickness $2d$ in the positive x direction as shown in Fig. 3.4. For the region occupied by the plate, the scalar potential ϕ and vector ψ of the displacements satisfying wave equations assuming isotropic solid plate:

Fig 3.4 Lamb wave propagation in a solid plate with free boundaries

$$\frac{\partial^2 \phi}{\partial^2 x^2} + \frac{\partial^2 \phi}{\partial^2 z^2} + k_l^2 \phi = 0, \quad (3.15)$$

$$\frac{\partial^2 \psi}{\partial^2 x^2} + \frac{\partial^2 \psi}{\partial^2 z^2} + k_t^2 \psi = 0 \quad (3.16)$$

Here $k_l = \omega \sqrt{\rho / (\lambda + 2\mu)}$ and $k_t = \omega \sqrt{\rho / \mu}$ are the wave numbers for longitudinal and transverse modes, respectively, ω is the circular frequency, λ and μ are the elastic Lamé constants, and ρ is the density of the medium. The components U and W of the particle displacement along the x and z axes, respectively, and the stress components σ_{xx} , σ_{zz} and σ_{xz} may be represented in terms of ϕ and ψ according to the equations assuming isotropic media.

$$U = \frac{\partial \phi}{\partial x} - \frac{\partial \psi}{\partial z}$$

$$W = \frac{\partial \phi}{\partial z} - \frac{\partial \psi}{\partial x}$$

$$\sigma_{xx} = \lambda \left(\frac{\partial^2 \phi}{\partial^2 x^2} + \frac{\partial^2 \phi}{\partial^2 z^2} \right) + 2\mu \left(\frac{\partial^2 \phi}{\partial^2 x^2} - \frac{\partial^2 \psi}{\partial x \partial z} \right),$$

$$\sigma_{zz} = \lambda \left(\frac{\partial^2 \phi}{\partial^2 x^2} + \frac{\partial^2 \phi}{\partial^2 z^2} \right) + 2\mu \left(\frac{\partial^2 \phi}{\partial^2 z^2} + \frac{\partial^2 \psi}{\partial x \partial z} \right),$$

$$\sigma_{xz} = \mu \left(2 \frac{\partial^2 \phi}{\partial x \partial z} + \frac{\partial^2 \psi}{\partial^2 x^2} - \frac{\partial^2 \psi}{\partial^2 z^2} \right).$$

For the case of lamb wave, we represent ϕ and ψ in the form of

$$\phi = A_s chqze^{ikx} + B_a shqze^{ikx},$$

$$\psi = D_s shsze^{ikx} + C_a chsze^{ikx}$$

Where A_s, B_a, C_a , and D_s are arbitrary constants, k is the Lamb wave

number, $q = \sqrt{k^2 - k_l^2}$; $s = \sqrt{k^2 - k_t^2}$; the factor $e^{i\omega t}$ is dropped for brevity.

Form the boundary conditions, the displacement components U and W are arrived at

$$U = U_s + U_a;$$

$$W = W_s + W_a,$$

Where

$$U_s = Ak_s \left(\frac{chq_s z}{shq_s d} + \frac{2q_s s_s}{k_s^2 + s_s^2} \cdot \frac{chs_s z}{shs_s d} \right) e^{i(k_s x - \omega t - \Pi/2)}$$

$$W_s = -Aq_s \left(\frac{shq_s z}{shq_s d} - \frac{2k_s^2}{k_s^2 + s_s^2} \cdot \frac{shs_s z}{shs_s d} \right) e^{i(k_s x - \omega t)};$$

$$U_a = Bk_a \left(\frac{shq_a z}{chq_a d} - \frac{2q_a s_a}{k_a^2 + s_a^2} \cdot \frac{shs_a z}{chs_a d} \right) e^{i(k_a x - \omega t - \Pi/2)},$$

$$W_a = -Bq_a \left(\frac{chq_a z}{chq_a d} - \frac{2k_a^2}{k_a^2 + s_a^2} \cdot \frac{chs_a z}{chs_a d} \right) e^{i(k_a x - \omega t)}$$

Where A and B are new arbitrary constants.

Fig 3.5 Symmetrical Lamb waves

Fig 3.6 Antisymmetrical Lamb waves

The displacement equations describe two groups of waves both propagating in the plate independently of one another. The first group of waves, indicated by the subscript s , describes waves in which the motion occurs symmetrically with respect to the plane $z = 0$ (the displacement U has the same signs, the displacement W opposite signs in the upper and lower halves of the plate). The second group of waves, indicated by the subscript a , describes waves in which the motion occurs antisymmetrically with respect to the plane $z = 0$ (the displacement U has opposite signs, the displacement W the same signs in the upper and lower halves of the plate). They are symmetric and anti-symmetric Lamb waves as shown in Fig. 3.5 and Fig. 3.6, both of which have zeroth, first, second and higher order modes, and have phase velocity dispersion with their dependence on the plate thickness and frequency.

Chapter 4 : Transducer Finite Element Models

4.1 Introduction

In the past, various research groups have used different materials in hybrid configurations to improve the bandwidth of the transducers, which is a major precondition for improving the image quality. Widely used materials for transducers are piezoelectric crystals of various types, piezoceramics, polymers (PVDF), and recently piezocomposites materials. The researchers studied the performance of the transducer in terms of bandwidth and signal to noise ratio using finite element analysis as well as the experimental methods. In this chapter, a new material PNN-PT-PZ has been considered for the fabrication of medical ultrasonic transducers to improve the bandwidth of the transducer. The performance of this material has been studied in detail using finite element analysis, which is presented in this chapter.

In medical ultrasound imaging, bandwidth and Signal-to-Noise Ratio (SNR) are the two fundamental parameters which govern the performance and quality of ultrasound images. These parameters govern the spatial resolution and the extent of tissue penetration. Considerable effort has been put by previous investigators using multilayer hybrid composite arrays [6, 39-41]. David M. Mills *et al.* has [39-42] investigated 1-3 PZT/polymer composites to improve SNR and bandwidth of such a transducer by mathematical analysis using Finite Element Analysis (FEA) with a follow up experimental validation to the FE analysis. Though the multilayer composite arrays have been found to increase SNR and the bandwidth, the fabrication of such a composite

transducer arrays is difficult [41, 42] and expensive as compared to single crystal arrays. The single crystal materials used for high efficiency actuators and sensors are the solid solutions of materials such as Lead Zirconium Niobate PZN/PT and Lead Magnesium Niobate (PMN)/PT. The improvements in the SNR and bandwidth of the single crystal transducer arrays have been reported. Essentially this enhanced performance is due to the increase in the electromechanical coupling (k_{33}). A typical case is that of in the literature Saitoh *et al.* [43], who developed a single crystal transducer array (PZN—9%PT) that resulted in a 5.2 dB increase in sensitivity over PZT-5A and has a fractional bandwidth (@-6dB) of 82%.

An addition to the family of single crystal materials is a new piezo-ceramic PNN-PT-PZ ($\text{PbNi}_{1/3}\text{Nb}_{2/3}\text{O}_3\text{-PbTiO}_3\text{-PbZrO}_3$) has been developed by Kondo *et. al.* in Fujitsu Labo [45]. Its first application is expected to be for use in single or multi-layer, high frequency, piezoelectric actuators. In this chapter, the performance of this novel material is studied and compared with the conventional piezoelectric materials, piezoceramics and other single crystal materials. Because of the high coupling constant ($k=+0.8$) of PNN-PT-PZ material, it can be potentially used for medical and other wide-band sonar applications. This material can be sintered or diced in similar way to conventional PZT. The basic transducer technology developed with PZT could also be applied with this material. Thus, PNN-PT-PZ is expected to be an alternative material for medical ultrasonic transducer.

In this chapter, PNN-PT-PZ material has been investigated as an active material for ultrasonic transducers. FEM comparisons of PNN-PT-PZ vs. PZT-5H, piezoceramic

hybrid material, and other single crystal transducer arrays have been described in detail. Finite element transient analysis for all types of transducers has been carried out in the present work in order to establish the pulse echo responses. In the following sections the finite element model used in the study is described in detail. The pulse echo responses of the transducer model have been presented in order to establish comparison between PNN-PT-PZ material, PZT-5H material and hybrid piezoceramic materials.

4.2 Finite element modeling and computer simulation

Finite element modeling has been adopted in the design of ultrasonic transducers and imaging arrays. Design procedure of the transducer can be accelerated by using finite element modeling and the analysis of transducer array. Existing designs are being optimized and new concepts are being explored. This recent acceptance follows the convergence of improvements on many fronts: necessary computer resources are more accessible, lean, specialized algorithms replacing general-purpose approaches, and better material characterization.

The FEM simulations have been performed using commercial software package PZflex (Weidlinger Assoc.; New York, NY and Los Altos, CA). The finite element method is numerical technique and the results of the finite element analysis are subjected to deviate from the experimental results and exact results. It is essential to validate the Finite element software package before it is used effectively and extensively for transducer design and analysis. The key issues such as material characterization, model configuration and model implementation are the primary sources of the errors in the finite

element analysis. These issues are addressed and subsequent validation exercise based on incremental model-build-test is presented in detailed in various documents [48-53].

The present chapter deals with the detailed comparative study of pulse echo response of the transducer array with different materials. In the pulse echo response a flat reflector is kept in the water tank at a distance of 5cm. As human body has high percentage of water and tissue is mostly water and therefore the impedance of water is a good approximation for most soft tissue. Hence, water has been used in these finite element models as well as instead of human tissue. The ultrasound waveform, which are transmitted by the transducer are reflected back from the flat reflector to the transducer. In the finite element model the pressure of the reflected waveform at the transducer surface is measured in time domain. The pressure field in time domain of the reflected waveforms and their corresponding frequency spectrums are plotted for different transducers with different materials.

The 3D finite element models of the part of the 1-D and 2-D arrays of the transducer have been used in this study. The azimuth direction, the elevation direction and the length of transducer array have been taken along X, Y and Z-axis respectively as shown in Fig 4.1, 4.2 and 4.3. Planar symmetry has been used in the finite element model that allows modeling of one half of the width of the transducer in azimuth direction so that the representative part can produce the complete array. Different materials such as PZT-5H control material and piezoceramic materials with hybrid array, single crystal material have been used for the analysis. The finite element models of the 1-D array of PZT-5H

control material and Hybrid composite material are shown in Fig. 4.1 and Fig. 4.2 respectively. Finite element models of 2-D array of PZT-5H control, and hybrid composite material are shown in Fig. 4.3 and Fig. 4.4 respectively. The Fig. 4.1 is also used for showing the 1-D array of PNN-PT-PZ material. The thickness of the PNN-PT-PZ material is different from the PZT-5H material as shown in Table 4.3. An epoxy backing (lightly loaded with Al_2O_3 and phenolic micro-balloons), a single matching layer and two active transducer elements have been considered at the center with one neighboring element on either side of the active element. It should be noted that the same matching layer is used for both PZT-5H control and hybrid array. Material properties used for PZT-5H Control material and PNN-PT-PZ material in the finite element modeling and analysis are listed in Table 4.1 [46] and Table 4.2 shows the properties of backing and matching material used in the Finite Element analysis. Table 4.3 contains details of the array models of PZT-5H control material, hybrid material, and PNN-PT-PZ material. In the finite element model the pitch represents the distance between the mid-point of the first element to the mid-point of the second element. The gap distance between the two elements is represented by W_{kerf} .

Table 4.1 Material properties of PZT-5H control and PNN-PT-PZ used for modeling

Properties	Materials	
	PZT-5H	PNN-PT-PZ
ρ (kg/m ³)	7820	8150
C_{11}^D (N/m ²)	1.37×10^{11}	1.27×10^{11}
C_{33}^D (N/m ²)	1.26×10^{11}	1.02×10^{11}
C_{44}^D (N/m ²)	2.23×10^{10}	5.79×10^{10}
C_{12}^D (N/m ²)	8.79×10^{10}	9.26×10^{10}
C_{13}^D (N/m ²)	9.23×10^{10}	9.48×10^{10}
C_{66}^D (N/m ²)	2.48×10^{10}	7.26×10^{10}
e_{x5} (C/m ²)	16.1	11.5
e_{z1} (C/m ²)	-9.44	-8.20
e_{z3} (C/m ²)	22.5	15.8
ϵ_{11}^S (F/m)	1310	2480
ϵ_{33}^S (F/m)	1200	820
Q (mech.)	90	90

Table 4.2 Polymer material properties used for modeling

Material	Ablefilm 5025E	Epo-Tek 301-2	Backing Epoxy
ρ (g/cm ³)	Varies*	1.08	1.03
ν_1 (m/s)	1.50	2.64	2.68
ν_s (m/s)	0.85	1.15	1.18

α_1 (dB/mm)	5.6	3.6	3.5
α_2 (dB/mm)	42.0	8.3	3.5
$(\epsilon_{33})^S / \epsilon_0$	1	1	1

* see Table 4.5

Table 4.3 FEM optimised array dimensions and properties

	PZT-5H	Hybrid	PNN-PT-PZ
Pitch (μ m)	160	160	160
w_{kerf} (μ m)	50	50	50
t (μ m)	438	438	237
L (mm)	8	8	8
t_{ML} (μ m)	110	110	110
ρ (g/cm ³)	2.6	1.9	2.7

The input function to the transducer is sine function with driving frequency 6.0 MHz. The input function and corresponding frequency spectrum is shown in Fig. 4.5. To maintain the accuracy; 8 to 20 mesh elements per wavelength at the highest frequency (12.5 MHz) has been used. For the improvement of computational efficiency, a preliminary investigation on time steps shows that the time steps should be 6.5 nS in water and 3.7 nS elsewhere for the PZT-5H control model and for PNN-PT-PZ model as well. While maintaining the accuracy, variation in mesh size and time steps in relation to the longitudinal velocity of sound in each material also have been taken into account. These simulations have been performed on Windows 2000 on a 1400MHz Intel ®

Pentium ® III CPU and the estimated time was found to be 2-5 minutes.

Comparative study has been done on the FE model in terms of pressure field, bandwidth, sensitivity and signal to noise ratio. The electrical impedances (amplitude and phase) of the transducer arrays have been plotted to obtain resonant frequencies of the transducer model. The pulse echo response functions are shown. The bandwidth and Selfridges's badness parameter (see sect 4.3) are calculated for the optimized transducer arrays. Finite element transient analysis for the transducer models has been considered in the present work to study the pulse echo responses. Effect of the hybrid arrays on the bandwidth of the transducer has been studied on the above parameters. The new material PNN-PT-PZ used in the FE analysis and its effect on the bandwidth has been studied. The results of FE analysis for different materials are described in detail in following sections.

Fig 4.1 Finite element model of 1-D array of PZT-5H Control Material

Table 4.4 Colour code and dimensions of various materials used in Finite Element Model

Colour	Material	Dimensions in μm in x direction	Dimensions in μm in y direction
Purple	Backing material	400	150
Light-green	Matching material	110	150
Pink	Silver electrode	14	1000
Green	Water	440	1000
Blue	PZT-5H	438	150

Fig 4.2 Finite element models of 1-D array of Hybrid Composite material

Table 4.5 Colour code and dimensions of various materials used in Finite element model of 1-D transducer array

Colour	Material	Dimensions in μm in x direction	Dimensions in μm in y direction
Purple	Backing material	400	150
Light-green	Matching	110	150
Pink	Silver electrode	14	1000
Pink-2	PZT-5A	145	150
Green	Water	440	1000
Blue	PZT-5H	145	150
Red	Epo-Tek 301-2	130	150

Fig 4.3 Finite element model of 2-D array of PZT-5H Control material

Table 4.6 Colour code and dimensions of various materials used in Finite element model of 2-D transducer array of PZT-5H and PNN-PT-PZ material

Colour	Material	Dimensions in μm in x direction	Dimensions in μm in y direction	Dimensions in μm in z direction
Purple	Backing material	400	150	150
Light-green	Matching material Ablefilm 5025E	110	150	150
Pink	Silver electrode	14	1000	1000
Green	Water	440	1000	1000
Blue	PZT-5H	438	150	150

Fig 4.4 Finite element model of 2-D array of Hybrid Composite material

Table 4.7 Colour code and dimensions of various materials used in finite element model of 2-D hybrid transducer array

Colour	Material	Dimensions in μm	Dimensions in μm in Y direction	Dimensions in μm in z direction
Purple	Backing material	400	150	150
Pista	Matching material	110	150	150
Pink	Silver electrode	14	1000	1000
Pink-2	PZT-5A	145	150	150
Green	Water	440	1000	1000
Blue 1	PZT-5H	145	150	150
Blue 2	PZT-5H	145	123	150
Red	Epo-Tek 301-2	130	150	27

Fig 4.5 Input function and the corresponding frequency spectrum

4.3 Results and Discussion

In this section the detailed comparison of the PZT-5H transducer, piezoceramic transducer with hybrid array, single crystal transducers using PNN-PT-PZ material has been presented with respect to resonant frequencies, and pulse echo responses. The fractional bandwidth (@ -6dB) has been used to compare the performances of the transducer array models. Another parameter, which was introduced by Selfridge [47] as a figure of merit, called “badness” was also used for evaluating the design criteria. This uses the impulse response of the transducer.

$$\text{"badness"} = \frac{\int_{t_0}^{\infty} |V(t)|(t-t_0) dt}{|V|_{pk}^2}$$

Where $V(t)$ is the pulse-echo response of the transducer, and V_{pk} is the peak amplitude of the pulse. “Badness” is the energy in the ring-down after time t_0 divided by the peak amplitude of the pulse squared.

4.3.1 Comparison between Hybrid array and PZT-5H array

The Fig. 4.6 and fig. 4.7 shows the PZflex simulated impedance plots of magnitude and phase for PZT-5H control and hybrid array elements in water. As shown in the Fig. 4.6, the resonant frequency for both the arrays was at 2 MHz with a magnitude of 1750 Ω for the PZT control element and 256 Ω for the hybrid element.

Fig. 4.8 and Fig. 4.9 compare the simulated pulse-echo results. The PZT control element had an amplitude of 20mV, a -6dB fractional bandwidth of 43.44%, and a normalized badness of 3.5. The hybrid element echo amplitude was 48mV, a -6dB fractional bandwidth of 64.94%, and a normalized badness of 1. Thus, the hybrid array had increased echo amplitude of 10.9dB, increased bandwidth by a factor of 1.42 and a decreased badness of 4.77 as compared to the PZT-5 H transducer. Table 4.4 shows the summary of FE analysis for PZT-5H and Hybrid array. It can be inferred from above values that the hybrid arrays are giving better performance in pulse echo response than PZT-5H array.

Table 4.8 Summary of FEM results

	PZT-5H	Hybrid
$Z_s (\Omega)$	1750	256
BW	43.44%	64.94%
Norm. Badness	3.5	1

Fig 4.6 Magnitude of complex impedance for the PZT-5H control and Hybrid array elements

Fig 4.7 Phase of complex impedance for PZT-5H control and Hybrid array elements

Fig 4.8 Phase of complex impedance for PZT-5H control and Hybrid array elements

Fig 4.9 FEM pulse echo results: Spectra for the PZT-5H control and Hybrid array

4.3.2 Multi-layer Composite Hybrid 2-D Transducer Array

The Fig. 4.10 and Fig. 4.11 show the PZflex simulated impedance plots of magnitude and phase for 2-D hybrid array elements in water. As shown in the Fig. 4.10, the resonant frequency for the transducer array was at 2 MHz with a magnitude of 240Ω for the hybrid array elements. The second mode appears at 3 MHz and width mode appears at 6 MHz. Fig. 4.10 and fig. 4.11 also show the PZflex simulated impedance plots of magnitude and phase for 2-D PZT-5H control array elements in water. As shown in the Fig. 4.10, the resonant frequency for the transducer array was at 2 MHz with a magnitude

of $1250\ \Omega$ for the hybrid array elements. The second mode appears at 3 MHz and the width mode appears at 7 MHz.

Fig. 4.12 and Fig. 4.13 compare the simulated pulse-echo results of 2-D hybrid array and 2-D PZT-5H control array. The PZT control element had an amplitude of 40mV, a -6dB fractional bandwidth of 16.72%, and a normalized badness of 4.88. The hybrid element echo amplitude was 120mV, a -6dB bandwidth of 38.93%, and a normalized badness of 1. The simulated pulse-echo results are summarized in Table 4.5. Thus the hybrid 2-D array increased bandwidth by a factor of 1.33, and decreased badness of 4.88 compared to PZT-5H 2-D transducer array. However, it has reduced the bandwidth by a factor of 0.66 as compared to the 1-D array of the transducer of same material. This is due to the ultrasound energy dissipation in the cross talk between the 2-D array elements as compared to that of 1-D array elements.

Table 4.9 Summary of FEM results for 2-D array

	PZT-5H	Hybrid
$Z_s\ (\Omega)$	1250	246
BW	16.72%	38.93%
Norm. Badness	1.33	1

Fig 4.10 Magnitude of complex impedance for the Hybrid 2-D array elements in water

Fig 4.11 Phase of complex impedance for the Hybrid 2-D array elements in water

Fig 4.12 FEM pulse echo results: Pulse for the 2-D hybrid array elements in water

Fig 4.13 FEM pulse echo results: Spectra for the 2-D hybrid array elements in water

4.3.3 Comparison between Single Crystal (PNN-PT-PZ) and PZT-5H array

The comparison of pulse echo response of 1-D array of PZT-5H control and PNN-PT-PZ material is studied in this section. Table 4.5 shows the details of the optimized design parameters for each transducer. Fig. 4.14 and Fig. 4.15 show series resonant frequency from the PZflex simulation impedance plots for both PZT-5H, and PNN-PT-PZ transducer arrays. The first resonant frequency of PZT-5H and PNN-PT-PZ transducer array has been found to be 1.8 MHz and 2.75 MHz respectively with magnitude of 271 Ω and 535 Ω respectively. The second natural frequency mode in the PZT element appears at 6.43 MHz where as in the case of PNN-PT-PZ arrays, it appears at 4.85 MHz.

Fig. 4.16 and Fig. 4.17 show the simulated pulse echo results. The fractional bandwidths are calculated at -6dB. The PZT-5H control element has an amplitude of 160mV, and bandwidth of 60.76 %. It has normalized badness of 1.41. For the PNN-PT-PZ transducer element the echo amplitude and bandwidth have been found to be 140mV and 79.76 % respectively. It has normalized badness of 1.0. The simulated pulse-echo results are summarized in Table 4.6. Thus the bandwidth for PNN-PT-PZ transducer element has been found to be increased by a factor of 1.31, and the badness factor has been found to be decreased by a factor of 1.41 compared to PZT-5H control arrays.

Table 4.10 Summary of FEM results

	PZT-5H	PNN-PT-PZ
$Z_s (\Omega)$	271	535
BW	60.76 %	79.76 %
Norm. Badness	1.41	1

Fig 4.14 FEM complex impedance magnitude in water for the PZT-5H control and PNN-PT-PZ array elements

Fig 4.15 FEM complex impedance phase in water for the PZT-5H control and PNN-PT-PZ array elements

Fig 4.16 FEM pulse echo results: Pulse for the PNN-PT-PZ array elements in water

Fig 4.17 FEM pulse echo results: Spectra for PNN-PT-PZ array elements in water

4.3.4 Comparison of 2D Single Crystal (PNN-PT-PZ) and PZT-5H array

The Fig. 4.18 and fig. 4.19 show the PZflex simulated impedance plots of magnitude and phase for single crystal material (PNN-PT-PZ) array elements in water. As shown in the Fig. 4.18, the resonant frequency for the transducer array is at 2.5 MHz with a magnitude of 120Ω for the single crystal array elements. The second mode appears at 5 MHz and the width mode appears at 6.8 MHz. The Fig. 4.18 and fig. 4.19 also shows the PZflex simulated impedance plots of magnitude and phase for 2-D PZT-5H control array

elements in water. The resonant frequency for the transducer array is at 2 MHz for the PZT-5H array elements. The second mode appears at 3 MHz and the width mode appears at 7 MHz.

Fig. 4.20 and Fig. 4.21 compare the simulated pulse-echo results. The PZT control element has an amplitude of 50mV, a -6dB fractional bandwidth of 43.44%, and a normalized badness of 2.33. The PNN-PT-PZ element has an echo amplitude of about 200mV, a -6dB bandwidth of 64.30%, and a normalized badness of 1. Thus, the PNN-PT-PZ array has an increased bandwidth by a factor of 0.48, and a decreased badness of 2.33 compared to PZT-5H transducer.

Fig 4.18 Magnitude of complex impedance for the PZT-5H control 2-D array elements in water

Fig 4.19 Phase of complex impedance for the PZT-5H control and 2-D array elements in water

Fig 4.20 FEM pulse echo results: Pulse for the 2-D hybrid array elements in water

Fig 4.21 FEM pulse echo results: Spectra for the 2-D hybrid array elements in water

4.4 Conclusion

In this chapter, a study of the conventional transducers used in present ultrasound systems has been done using Finite element analysis. The finite element model has been constructed with required boundary conditions. The different transducer materials have been used in the study to simulate the performance of the each type of transducer. The performance of the transducer has been studied by using the fractional bandwidth @ -6dB as a figure of merit for the study. The performance of the transducer has been also studied by comparing the ‘badness parameter’ of the transducer. The performance of the PZT-5H transducer has been compared with that of the piezoceramic transducer, hybrid

array and single crystal transducer. It is found that the hybrid array transducer provides a broader bandwidth as compared to the PZT-5H transducer. It is also observed that some single crystal transducer materials provide broader bandwidth as compared to hybrid arrays and also PZT-5H transducers. The PNN-PT-PZ transducer material gives a broader bandwidth as compared to the PZT-5H transducer. This material can be used as an alternative to PZT-5H transducers, if the cost of the transducer is not the major parameter.

The finite element model 2-D array of the transducer has been constructed to study the pulse echo response and to evaluate their bandwidth and badness parameter. It can be established from FE analysis that 2-D array of this transducer does not provide a broader bandwidth than that of the linear 1-D array due to the cross-talk between the elements of the transducer array. Moreover, the fabrication of the 2-D array is more difficult and challenging. In addition to these limitations, the cost of the transducer array is much higher. The newly developed capacitive micromachined ultrasonic transducer arrays are a potential and promising alternative to the conventional transducer, as these transducers overcome the fabrication difficulties and also provide broader bandwidth. The cross talk between the transducer elements of the capacitive micromachined ultrasonic transducers can be reduced by providing some suitable structural modifications in the cMUT structure. The next two chapters will discuss in detail, the design of capacitive micromachined ultrasonic transducer and the cross talk between the transducer elements.

Chapter 5 : Finite Element analysis of cMUT cell

5.1 Introduction

As seen in the previous chapters, capacitive micromachined ultrasonic transducers are a promising alternative to conventional piezoelectric transducers. These transducers work based on the principle of electrostatic force generated in two capacitor plates. This is described in detail, in chapter 2. It is critical to design the cMUT cell keeping in view the major parameters such as thickness of the membrane, gap thickness, electrode thickness and electrode size. The various research groups, working on cMUTs have done substantial work on the design of cMUT cells [12-14], [22] using equivalent circuit approach and the finite element analysis approach. These parameters define the resonant frequency, centre frequency, and the transducer efficiency. Hence, these parameters affect the performance of the transducer as well as the image quality of the ultrasound system, which incorporates this transducer. The frequency of the transducer can go up to 50 MHz to 60 MHz or beyond that depending upon all the parameters such as thickness of the membrane, diameter of the membrane and the gap thickness as mentioned above. Omar Oralkan *et. al.* developed high frequency transducer, which has resonant frequency of 60 MHz in and air [72].

The finite element analysis is a generalized tool to design these parameters of the cMUT cell. In this chapter the finite element analysis is done on the cMUT cell. A 2-D finite element model is constructed for single cMUT cell. The impulse response analysis is done in the time domain on this finite element model. The collapse voltage is evaluated for different configurations of cMUT cell. The FE model of the transducer is subjected to

AC voltage to evaluated impulse response of the cMUT cell. The use of finite element analysis is extended further to handle array structures in next chapter. This would allow one to study the cross talk between the elements. The cross talk study involves a thorough understanding of multidisciplinary fields such as electrostatics and acoustics.

5.2 Analysis of cMUT cell

The finite element analysis is carried out on the cMUT membrane to find out the static deflection under the combined effects of atmospheric pressure and the applied DC biased voltage. In the present study, a polysilicon membrane is used. It is mounted on a silicon nitride post over the silicon substrate. The material properties used in the analysis are shown in the Table 5.1. The cMUT membranes studied in this analysis are of circular in shapes. The typical dimensions of the cMUT cell used in the analysis are shown in Table 5.2. The resonant frequencies were obtained from the impedance plot of the transducer, showing both magnitude as well phase. The cell is subjected to different bias voltages and the collapse voltage is obtained for every possible configuration of the cMUT cell. The pulse echo response is studied for the single cMUT cell. The performance of the transducer in the form of a single cell has been thoroughly studied in time domain as well as in the frequency domain. The results are described in detail in the following sections. The fractional bandwidth of -6dB for the transducer has been obtained. This is used as a parameter for comparison between various combinations of the cell parameters.

5.3 Finite Element Model

The finite element model of the cMUT membrane is constructed using the commercial FE PZflex (Weidlinger Assoc.; New York, NY and Los Altos, CA). The two

dimensional axisymmetric FE model is used in the study. Fig. 5.1 shows typical finite element model of the cMUT used in the study. The finite element model consists of a single cell. The components of the cMUT transducer single cell are modeled and the symmetric boundary condition in lateral direction and absorbing boundary condition in axial direction are implemented in FE model. Water is considered as a medium in which ultrasonic waves are transmitted. The time domain transient analysis has been considered here to study the pulse echo response of the transducer. The critical dimensions and material properties used in the FE analysis are shown in Table 5.1 and Table 5.2. To maintain the accuracy; 8 to 20 mesh elements per wavelength at the highest frequency have been used. The cMUT cell parameters and the array parameters were optimized for the maximum bandwidth. While maintaining the accuracy, variation in mesh size and time steps in relation to the longitudinal velocity of sound in each material also have been taken into account. These simulations have been performed on Windows 2000 on a 1400MHz Intel ® Pentium ® III CPU (dual processor).

Table 5.1 Material properties of Si_3N_4 and Polysilicon used in analysis

	Si_3N_4	Polysilicon
Young's Modulus (N/m^2)	2.25E+11	1.69E+11
Density (Kg/m^3)	3270	2332
Poisson's ratio	0.263	0.22
Permittivity	5.7	11.8

Table 5.2 Cell dimensions used in the analysis

Cell diameter (d_{cell}), μm	50
Membrane thickness (t_m), μm	2.0
Gap thickness (t_g), μm	0.16

Fig 5.1 Finite element model for single cMUT cell

5.4 Results and Discussion

The typical dimensions of the cMUT cells are shown in Table 5.2. First, the cMUT cell is subjected to a DC bias voltage to get the static deflection of the cMUT membrane. The DC bias voltage is increased gradually to get the collapse voltage. The collapse voltage for the cMUT cell as shown in Table 5.2 is 55 V. It is necessary to apply the DC bias voltage, which is close to the collapse voltage of the cMUT to get the maximum efficiency of the transducer. Hence, a DC bias of 50 V is applied as a good choice of an operating voltage in further studies. The AC voltage of 5V peak to peak at 15MHz driving frequency is applied on the top of the DC biased voltage after the membrane has achieved the static equilibrium condition. The input voltage in time domain is shown in Fig. 5.2. The Table 5.3 compares the collapse voltage for the different combinations of the parameters of the cMUT cell. The pulse echo response of the cMUT cell is studied by keeping the reflector at 10mm away from the transducer. The transmitted and received pressure has been calculated at Kirchoff plane in the near field of the transducer. The Fig. 5.2 shows the material plot of the transducer and pressure field in the water. It also illustrates the deflection at the centre of the cMUT membrane in the time domain. The static deflection due to DC bias voltage is about 0.06 μm . The impedance plot is shown in Fig 5.3 in magnitude as well as in phase. The impedance plot shows that the first resonant frequency of the transducer is 6.5 MHz. The pressure of the transmitted ultrasound waves is shown in Fig. 5.4. Fig. 5.5 shows the pressure field in the water due to the reflected ultrasound waves. The peak pressure is 1.5Kpa for this configuration of the cMUT cell. Fig. 5.5 also shows the deflection of the centre of the cMUT membrane due to the reflected wave forms. It also illustrates the material plot and the input voltage with only DC bias voltage. Fig. 5.6 and 5.7 show the pressure of the received ultrasound

waves in time domain as well as in frequency domain. It can be deduced from the Fig. 5.7 that the fractional bandwidth @ -6dB is 101.10%. The comparative study has been done on the centre frequency and the bandwidth of the transducer for different combinations of the cMUT cell parameters and is shown in Table 5.3.

Table 5.3 Comparison of centre frequency, Fractional bandwidth and collapse voltage of different cMUT cells

Cell Parameters			Collapse Voltage in V	Fractional bandwidth @ -6dB	Centre Frequency in MHz
Diameter of Membrane in μm	Thickness of Membrane in μm	Gap Thickness in μm			
50	2.0	0.16	52	101.10%	3.316
54	2.0	0.16	42	132.78%	4.875
44	2.0	0.16	62	115.28%	6.441
36	1.5	0.22	90	104.44%	7.887
32	1.0	0.22	60	130.84%	8.915
36	1.8	0.16	75	93.22%	8.967

Fig 5.2 Material plot of the cMUT and pressure field in water, Plots of input voltage, deflection of cMUT membrane at the centre in time domain

Fig 5.3 Impedance plots of single cMUT cell in magnitude and phase

Fig 5.4 Pressure of the transmitted ultrasound waveform at the centre of the cMUT cell in time domain

Fig 5.5 Material plot, pressure field of reflected waveforms, and deflection at the centre of the cMUT membrane due to reflected waveforms

Fig 5.6 Pressure of the reflected waveforms near the cMUT membrane in time domain

Fig 5.7 Response of the transducer in time domain and frequency domain

5.5 Conclusion

In this chapter, a detailed study has been performed on the single cell of the capacitive micromachined transducers (cMUT). The 2-D axisymmetric FEM model is generated for a single cMUT cell. Impedance plots from FEM simulation were studied. Pulse echo results from cMUT cell were studied. The various characteristics such as collapse voltage, fractional bandwidth, and centre frequency were studied. The purpose of the study is to find the collapse voltages for the different configurations of the cMUT cell. Table 5.4 shows the comparison of the collapse voltages for the various configurations of the cMUT cells. The fractional bandwidth @ -6dB for the above configuration of the transducer is found to be 101.10%. In the next chapter the cross coupling between the transducer elements will be discussed in detail. The effect of the cross coupling on the bandwidth of the transducer is presented in the next chapter using Finite Element analysis.

Chapter 6 : The effect Cross talk on bandwidth of the transducer

6.1 Introduction

The capacitive micromachined ultrasonic transducer is becoming a popular alternative to the piezoelectric transducers in many applications. They offer the advantage of increased bandwidth with comparable sensitivity to piezoelectric transducers as well as ease of fabrication and electronic integration [14]. There has been an appreciable increase in the research effort on cMUTs in the last few decades. Research groups are now investigating various aspects of the cMUT design, performance and manufacture [12-25]. The performance of the cMUT has been studied using a simple Mason type equivalent circuit model to predict behaviour [12, 15, and 21]. However, the equivalent circuits are unable to consider the important effect such as the coupling into the substrate and cross talk [46] between the elements of an array of transducers. As the cross talk between the elements of an array of transducer affect the bandwidth of the transducer, it is necessary to study this phenomenon and to provide a method to reduce the cross talk between the elements of an array transducer. The cross talk between the elements is due to propagation of various spurious waves such as Lamb waves, which propagate through wafer; and Stoneley waves, which propagate through the interface between solid and fluid. Very recently other research groups have also studied this phenomenon through finite element analysis [18, 19], [22-24] and have suggested some structural modifications to reduce cross talk.

In the present study, finite element analysis has been used as a tool to investigate the cross talk phenomenon. The study has been done on the cross talk between the elements of the transducer and between the cells with in the element of the transducer. The major

focus of the study is on the effect of Lamb wave propagation on cross talk, which further affects the bandwidth of the transducer. A novel structure is proposed in the present study to reduce the cross talk between the transducer elements. The proposed structure consists of an acoustic cavity in the silicon substrate below the cMUT cell. The acoustic cavity is filled with an acoustically attenuating material such as polymers, elastomers. In the present research different attenuating materials such as PDMS, tungsten loaded epoxies are explored. A number of shapes for the acoustic cavity were considered and studied. The 2-D FE models of the transducer show that the acoustic cavity is of rectangular and trapezoidal shape. The 2-D FE models, which are shown in Fig. 6.1 to Fig. 6.4, give better understanding of an acoustic cavity below the cMUT cells. The lateral and axial dimensions of the acoustic cavity are varied and compared in order to study their effect of the cross coupling and on the bandwidth of the transducer. Similarly the different acoustically attenuating materials have been used to see their effect on the acoustic coupling between the cMUT cells and the bandwidth of the transducer.

6.2 Finite Element Model of Transducer Array

The finite element analysis is used as a tool to study the cross talk effect. The commercial FE code PZflex (Weidlinger Assoc.; New York, NY and Los Altos, CA) has been used in the present study.

The study is based on various two dimensional finite element models. All finite element models consist of the array of three cells of cMUT, in which the first model is without acoustic cavity and other models have an acoustic cavity, which is filled with the acoustically attenuating material below the center cMUT cell. Fig 6.1 and Fig 6.2 show

the different views of the part of the transducer element. Fig. 6.3 shows the typical finite element model of the cMUT without acoustic cavity used in the study. The dimensions along the lateral direction are modeled along x-axis and the dimensions in the thickness direction are modeled as along y-axis. The finite element models of the cMUT with acoustic cavity of different shapes are shown in Fig. 6.4, Fig. 6.5, and Fig. 6.6. These Finite element models consist of three cMUT cells. The part of the cMUT transducer is modeled and the symmetric boundary condition in the lateral direction and absorbing boundary condition in the axial direction are implemented in FE model. Water is considered as the medium in which ultrasonic waves are transmitted. The time domain transient analysis has been considered here to study the pulse echo response of the transducer. The reflector is kept at a distance of 5cm from the top of the cMUT membrane and the membrane material used in the study is silicon nitride. Critical dimensions and material properties used in the FE analysis are shown in Table 6.1. The material filled in the acoustic cavity is tungsten loaded epoxy with 10% volume fraction (vf) having an acoustic impedance of 5.8 MRayl. To maintain the accuracy, 8 to 20 mesh elements per wavelength at the highest frequency have been used. While maintaining the accuracy, variation in mesh size and time steps in relation to the longitudinal velocity of sound in each material also have been taken into account. These simulations have been performed on Windows 2000 on a 1400MHz Intel ® Pentium ® III CPU (dual processor). The pulse echo response study has been done on all the models. The depth of an acoustic cavity is varied in each model to study its effect on acoustic coupling. Moreover, the effect of different filler materials in the acoustic cavity, on the cross coupling is critically observed.

Fig 6.1 3-D view of the part of the transducer element

Fig 6.2 Semitransparent view of the part of the transducer element

Fig 6.3 Finite element model of cMUT without acoustic cavity

Fig 6.4 Finite element model having trapezoidal cavity having base of the trapezoid on the top

Fig 6.5 Finite element model having rectangular acoustic cavity filled with acoustically attenuating material

Fig 6.6 Finite element model having trapezoidal cavity having base of the trapezoid at the bottom

Table 6.1 Critical dimensions and material properties

	Parameters
Silicon Membrane diameter	18 μm
Vacuum gap diameter	18 μm
Silicon Nitride membrane thickness	0.8 μm
Vacuum gap thickness	0.2 μm
Silicon Nitride density	3270 kgm^{-3}
Silicon Nitride longitudinal Velocity	11000 ms^{-1}
Silicon Nitride shear velocity	6250 ms^{-1}
Silicon Nitride relative permittivity	7.5

6.3 Results and Discussion

In this section, the results obtained from the pulse echo response analyses are studied in detail. The pulse echo response from the all FE models have been obtained and compared with each other. The effect of the depth and width of an acoustic cavity and the width of on the bandwidth of the transducer are discussed. The effect of different materials used in an acoustic cavity is also evaluated.

6.3.1 Model without acoustic cavity

The pulse echo response of the transducer model has been studied in detail. To analyze the pulse echo response, a reflector is placed in the form of plane surface placed at a distance of 5 cm from the transducer model. Fig. 6.7 through Fig. 6.9 show the results for the model of the transducer without the acoustic cavity. The transducer is subjected to 300V DC bias voltage and -10V sine pulse. A driving frequency of 26MHz is applied to generate the ultrasound signals. The input function and its corresponding frequency spectrum are shown in Fig. 6.17. Fig. 6.7 shows the driving voltage in time domain. The displacement of the top of the centre of the cMUT membrane is also shown in time domain in Fig. 6.7. It can be inferred from the displacement plot in time domain that the membrane does achieve its static equilibrium before it is subjected to the pulse of AC voltage. The pressure field of the transmitted ultrasound waveforms in water is also shown in the Fig. 6.7. The typical received response of the cMUT in time domain for this transducer model is shown in Fig. 6.8. Fig 6.8 illustrates the pressure field of the reflected waveforms in water and the pressure of the reflected waveforms in the time domain. Fig. 6.9 illustrates the pressure of the reflected ultrasound waveforms in time

domain as well as in the frequency domain. The high frequency components present in the figure are due to ringing of the transducer membrane. The membrane of the transducer is not damped. The sufficient damping can reduce the high frequency components in pulse echo response. The fractional bandwidth @ -6dB of the transducers obtained from Fig. 6.9 is 47.32%. The maximum amplitude of the transmitted pressure is about 1.50KPa.

Fig 6.7 The input voltage in time domain, 2-pressure in water, 3-displacement of membrane in time domain

Fig 6.8 The pressure in water and received signal in time domain for second model

Fig 6.9 Pressure of reflected waveforms in time domain and frequency domain

6.3.2 Model with trapezoidal cavity with base at the top

The second finite element model studied here consists of a trapezoidal cavity with the base of the cavity close to the cMUT membrane as shown in Fig. 6.4. As in the case of the previous model, the transducer is subjected to 300V DC bias voltage and -10V sine pulse with driving frequency of 26MHz is applied to generate the ultrasound signals. Fig. 6.10 shows the driving voltage in time domain. The displacement at the top of the centre of the cMUT membrane is shown in time domain in Fig. 6.10. The membrane is allowed to reach its equilibrium after DC bias voltage is applied before it is subjected to AC voltage. The pressure field of the transmitted ultrasound waveforms in the water is also shown in the Fig. 6.10. The typical received response of the cMUT in time domain for this transducer model is shown in Fig. 6.11. Fig 6.11 illustrates the pressure field of the reflected waveforms in water and the pressure of the reflected waveforms in time domain. Fig. 6.12 illustrates the pressure of the reflected ultrasound waveforms in time domain as well as in the frequency domain. The fractional bandwidth @ -6dB of the transducers, which is calculated from Fig. 6.12 is 65.029%. The maximum amplitude of the transmitted pressure is about 1.50KPa. The fractional bandwidth of the transducer is increased from 47.32% to 65.029%. The increase in the fractional bandwidth is due to the acoustic cavity filled with an acoustically attenuating material, which absorbs the backward propagating acoustic waves. The effect of the depth and the width of the acoustic cavity on the bandwidth of the transducer are presented in detail in the next section.

Fig 6.10 The input voltage, pressure in water and transmitted signal in time domain

Fig 6.11 Pressure in water and received signal in time domain

Fig 6.12 Pressure of reflected waves in time domain and frequency domain

6.3.3 Rectangular acoustic cavity

The third finite element model of the transducer studied in the research work is with a rectangular acoustic cavity below the cMUT cell as shown in Fig 6.5. The transducer

model is subjected to same DC bias and AC voltage as in the previous cases to generate the ultrasound signals. Fig. 6.13 shows the driving voltage in time domain. The displacement at the top of the centre of the cMUT membrane is shown in time domain in Fig. 6.13. The membrane is allowed to reach its equilibrium after DC bias voltage is applied just before it is subjected to AC voltage. The typical received response of the cMUT in time domain for this transducer model is shown in Fig. 6.14. Fig 6.14 illustrates the pressure field of the reflected waveforms in water and the pressure of the reflected waveforms in time domain. Fig. 6.15 illustrates the pressure of the reflected ultrasound waveforms in time domain as well as in the frequency domain. It is shown in the Fig. 6.15 that the fractional bandwidth @ -6dB of the transducers is 61.048%. The maximum amplitude of the transmitted pressure is about 0.60 KPa. Here the fractional bandwidth @ -6dB of the transducer is increased to 61.048% from 47.32%, which is lesser than that obtained in the previous case. It is observed from these two cases that an acoustic cavity of trapezoidal shape used in previous case is better than the rectangular acoustic cavity used in the present case. There is increased gain of 5.01% of fractional bandwidth @ -6dB in the trapezoidal acoustic cavity as compared to the rectangular acoustic cavity.

Fig 6.13 The input voltage, pressure in water and transmitted signal in time domain

Fig 6.14 Pressure in water and received signal in time domain

Fig 6.15 Pressure of reflected waves in time domain and frequency domain

6.3.4 Trapezoidal Cavity with base at the bottom

Fig. 6.16 shows the results for the model of the transducer with out the acoustic cavity. As in the previous cases studied, here too the transducer is subjected to 300V DC bias voltage and -10V sine pulse with driving frequency of 26MHz is applied to generate the ultrasound signals. Fig. 6.16 shows the driving voltage in time domain. The displacement at the top of the centre of the cMUT membrane is shown in time domain in Fig. 6.16. The membrane is allowed to reach its equilibrium after DC bias voltage is applied before it is subjected to AC voltage. The pressure field of the transmitted ultrasound waveforms in the water is also shown in the Fig. 6.16. The typical received response of the cMUT in time domain for this transducer model is shown in Fig. 6.17. Fig 6.17 illustrates the pressure field of the reflected waveforms in water and the pressure of the reflected waveforms in time domain. Fig. 6.18 illustrates the pressure of the reflected ultrasound waveforms in time domain as well as in the frequency domain. It is shown in the Fig. 6.18 that the fractional bandwidth @ -6dB of the transducers is 48.38%. The maximum amplitude of the transmitted pressure is about 1.50KPa. It is observed that there is not much increase in the fractional bandwidth @ -6dB in this case. The cases studied with trapezoidal acoustic cavity with base close to the cMUT and the rectangular acoustic cavity has considerable effect on the bandwidth of the transducer as compared to this case.

Fig 6.16 The input voltage, pressure in water, and transmitted signal in time domain

Fig 6.17 Input function and the corresponding frequency spectrum

Fig 6.18 Pressure in water and received signal in time domain

6.3.5 Effect of acoustic cavity parameters on the bandwidth

The parameters of an acoustic cavity studied here are the depth of the cavity, the width and the material filled in the acoustic cavity. These parameters affect the bandwidth of the transducer and need to be optimized for the maximum fractional bandwidth (@ -6dB). The depth of an acoustic cavity formed below the cMUT cells is one of the critical parameters which affect the bandwidth of the transducer. The depth is varied in all the three types of acoustic cavities considered. Table 6.2 shows the comparison between the fractional bandwidth (@ -6dB) of the transducer of different acoustic cavity shapes and for different depths. It can be deduced from Table 6.2 that for the given configuration of the cMUT transducer shown in Table 6.1 for the trapezoidal acoustic cavity the maximum bandwidth is obtained at about 15 μm of the depth of the cavity. The fractional bandwidth decreases as the depth of cavity is increased further. It is observed that the fractional bandwidth decreases if the depth of the cavity is below 15 μm . A similar relation is obtained in the case of a rectangular acoustic cavity. In this case, the maximum bandwidth is obtained at 19 μm of the depth. The fractional bandwidth is reduced, if the depth of acoustic cavity is above and below the optimum value of 19 μm . It is also observed that in the case of the trapezoidal acoustic cavity in which the base is at the bottom of the substrate, the fractional bandwidth is affected little, due to the variation in the depth.

Table 6.2 Bandwidth of the transducer of different shapes of acoustic cavity as compared to 47.32% bandwidth without acoustic cavity

Depth of an acoustic cavity in μm	Fractional Bandwidth of the transducer @ -6dB		
	Trapezoid acoustic cavity	Rectangular acoustic cavity	Reverse trapezoid acoustic cavity
10	65.029	48.41	47.51
15	78.14	47.60	48.01

19	52.08	62.03	48.21
20	40.95	61.048	48.38
25	39.07	28.86	48.55

Fig 6.19 Bandwidth of the transducer of different shapes of acoustic cavity as compared (wrt depth) to 47.32% bandwidth without acoustic cavity

The width of an acoustic cavity is also one of the critical parameters which affect the bandwidth of the transducer. Table 6.2 shows the comparison between the fractional bandwidth @ -6dB of the transducer of different acoustic cavity shapes and for different depth of an acoustic cavity. It can be concluded from Table 6.3 that the maximum bandwidth is obtained at 30 μm of width of an acoustic cavity of trapezoidal shape with base close to the cMUT cell. It is also found that the fractional bandwidth is reduced further if the width of an acoustic cavity is decreased from a normal width of 30 μm as well as increased further. Similarly in the case of the rectangular acoustic cavity, the maximum bandwidth is obtained at 20 μm of width of an acoustic cavity for the given

configuration of cMUT. The fractional bandwidth decreases further, as the width of an acoustic cavity is increased and decreased. It is observed for the trapezoidal acoustic cavity, with base at the bottom of the substrate there is little change in the fractional bandwidth of the transducer due to variations in the width of an acoustic cavity.

Table 6.3 Bandwidth of the transducer of different shapes of acoustic cavity as compared to 47.32% bandwidth without acoustic cavity

Width of an acoustic cavity in μm	Fractional Bandwidth of the transducer @ -6dB		
	Trapezoid acoustic cavity at the depth of 13 μm	Rectangular acoustic cavity	Reverse trapezoid acoustic cavity
15	54.94	56.72	48.04
20	59.37	70.04	48.20
25	65.59	63.98	48.23
30	79.94	63.30	48.38
35	66.17	61.048	48.46
40	32.84	32.73	48.45

Fig 6.20 Bandwidth of the transducer of different shapes of acoustic cavity as compared (wrt width) to 47.32% bandwidth without acoustic cavity

The filler material in the acoustic cavity affects the bandwidth of the transducer. In the above analysis the material used in the acoustic cavity is tungsten loaded epoxy, 10%vf (volumetric fraction) with acoustic impedance of 5.8 MRayl. The effect on the bandwidth of the transducer is studied by changing the material to tungsten loaded epoxy, 5%vf with acoustic impedance of 4.6 MRayl and polydimethylsiloxane (PDMS). Table 6.4 shows the comparison of the bandwidth of the transducer, for different acoustic attenuating material and for different shapes of the acoustic cavity. The depth of the acoustic cavity is kept at 20 μm and the width of an acoustic cavity is kept at 40 μm . It is found from Table 6.4 that for a given configuration of an acoustic cavity the tungsten loaded epoxy with 5% vf provides maximum bandwidth amongst the three materials studied. In the case of rectangular acoustic cavity, the tungsten loaded epoxy with 5% vf gives maximum bandwidth amongst all the three materials. While, in the case of the trapezoidal cavity, with base at the bottom of the substrate, PDMS gives maximum bandwidth. However, the difference in the fractional bandwidths obtained for these materials is very little as compared to the other two configurations of an acoustic cavity. It is known from literature that the fabrication of the trapezoidal acoustic cavity with base near the cMUT cell is more difficult process as compared to the other two configurations. The rectangular acoustic cavity can be created by using the Deep RIE process and the trapezoidal cavity, with base at the bottom of the substrate, can be created by using wet etching with KOH as an etchant. By considering the fabrication challenges and the fractional bandwidth comparison, it can be concluded that rectangular acoustic cavity is more suitable for the cMUT transducer. It can further be optimised to reduce the cross coupling.

Table 6.4 Bandwidth of the transducer for different materials filled in acoustic cavity

Acoustically attenuating material	Fractional Bandwidth of the transducer @ -6dB		
	Trapezoid acoustic cavity at the depth of 20 μm and 40 μm width	Rectangular acoustic cavity at the depth of 20 μm and width of 40 μm	Reverse trapezoid acoustic cavity at the depth of 20 μm and width of 40 μm
Tungsten loaded Epoxy 10%vf	59.37	61.04	48.38
Tungsten loaded Epoxy 5% vf	68.15	52.11	48.59
PDMS	58.97	55.14	49.63

6.4 Conclusion

In this chapter, we investigated the cross-talk between the cells and the elements of the transducer by using FE analysis as a tool. We have investigated only acoustical coupling, which is in the form of Lamb waves between the elements. The electrical cross talk is beyond the scope of the present study. We studied the pulse echo response of the transducer and carried out a detailed examination on the effects of the acoustic cross talk between the elements and on the bandwidth of the transducer. The FE simulation results show that the bandwidth of the transducer is affected considerably by the cross talk between the elements of the transducer. It is also found during the study that the acoustically attenuating material in the cavity below the cMUT cell absorbs Lamb waves propagating through the wafer and hence, it further reduces the cross-talk between the elements and cells of the transducer. The effect of this phenomenon increases the bandwidth of the transducer, which can further improve the image quality in Ultrasound Imaging. It is also shown in this chapter that the parameters of the acoustic cavity, below the cMUT cell, such as the depth, the width and the material filled in an acoustic cavity affect the bandwidth of the transducer. It is found that for a given configuration of the

transducer there is an optimum depth of the acoustic cavity, at which maximum bandwidth is obtained. The bandwidth of the transducer decreases, as the depth of an acoustic cavity is increased or decreased further. Similar relation is also obtained for the width of an acoustic cavity, which is another parameter of an acoustic cavity. The material filled in acoustic cavity does affect the bandwidth of the transducer. The effect of acoustically attenuating material on the bandwidth of the transducer depends on the shape of an acoustic cavity. There is negligible effect on the bandwidth of the transducer, if the shape of an acoustic cavity is trapezoidal. The change in acoustically attenuating material affects the bandwidth of the transducer considerably, in case of rectangular cavity and trapezoidal cavity. The steps involved in the fabrication process for the proposed structure of the transducer are presented in the next chapter.

Chapter 7 : Transducer Fabrication and Characterisation

7.1 Introduction

The capacitive micromachined ultrasonic transducer cMUT, designed in the present study is specifically to reduce the cross coupling between the transducer elements. This can further increase the bandwidth of the transducer. The transducer design in the present research work has the following unique modules.

- An acoustic cavity of any shape is formed in the substrate below the cMUT by using Deep RIE process.
- Each acoustic cavity encapsulates at least four cMUT cells.
- The acoustic cavity is filled with acoustically attenuating material such as, polymers, elastomers, or various epoxy matrices that have dispersed solid metallic, ceramic, polymeric filler particles of selected density to absorb the backward transmitted energy. This prevents it from propagating in the form of lamb waves. In the present research work the acoustically attenuating material used to fill the acoustic cavity is tungsten epoxy with 10% volume fraction.

In this chapter, the process to fabricate the transducer is described in detail. The fabrication process involves eutectic wafer bonding method. This is done by using titanium gold is used as an intermediate layer to bond silicon nitride to silicon.

Fig 7.1 Cross sectional area of the single element of the cMUT

Fig 7.2 Top View of the part of the element of the cMUT structure

Fig. 7.1 shows the cross sectional view of the cMUT transducer array with rectangular acoustic cavity below the transducer cells. The cMUT transducer array consists of the various elements formed by cMUT cells. The cells may be of circular, square or hexagonal in shape. In the present design the cells are defined to be of circular in shape as shown in Fig. 7.2. The array transmits the ultrasound energy in the human body and is also capable of receiving ultrasound waveforms and generating output electric signals.

The input and out signals are further transferred to the ultrasound system, through interconnects and front-end electronics of the system.

7.2 Fabrication Process

The cMUT is formed on the surface of the low resistive 4 inch silicon wafer through the application of a series of well established silicon micromachining processes such as patterning a surface of the substrate using a photolithographic process, and successively deposition of material layers. The various elements of the structure are further formed by removing selected portion of the deposited material or substrate itself through the application of RIE and DRIE etching processes. The following figures show the cross sectional views that illustrate the steps in a method for fabrication a cMUT array in the present research work. The process uses eutectic wafer bonding technology to form the cMUT array.

The various steps are involved in the fabrication of the transducer array. The process starts with the photolithography, which is one of the important process steps and is used frequently. In this step the wafer is patterned according to the requirements of user. In this case the wafer is patterned by using photolithography process to form the square openings of the acoustic cavities on the wafer. The mask aligner (Karl Suss MA6) used in this process is shown in Fig. 7.3. The Karl Suss MA6 is a double-side contact and proximity mask aligner system for pattern down to 1 μ m. The UV source is broadband, emitting between 350nm and 450nm. The process uses low resistivity silicon wafers (0 – 0.4 $\mu\Omega$) and of 500 micron thickness. The wafers are cleaned in wet bench for about 4

minutes and are dried in spin drier. The positive photoresist (AZ7220) is deposited on the wafer by using spin coater shown in Fig. 7.4. The r.p.m and other recipes of the spin coater are selected so as to keep uniform layer of $5\text{ }\mu\text{m}$ of photoresist on the top of the wafer. The wafer is then baked for 4 minutes. Meanwhile the desired mask is fixed on the mask aligner. Subsequently wafer is loaded on the mask aligner and then exposed to the UV light for 14 seconds. The exposed wafer is then developed, cleaned and spin dried for further processes.

The square cavities are etched in the patterned wafer by using Deep RIE process as shown in Fig 7.5. The equipment used for Deep RIE is STS Deep RIE is shown in Fig. 7.6. The STS machine uses a mechanical clamp to hold the wafer to a chuck. This clamp is made up of eight ceramic 'fingers' which bear down near the edge of the wafer to hold it onto the chuck. The chuck provides Helium (He) backside cooling to the wafers. This keeps the wafer cool enough during the plasma process so that the resist does not etch very quickly (selectivity to Si is about 75:1) and also minimizes resist burning. The square cavities are about 350 to 450 μm deep. Helium is delivered to the bottom of the wafer at about 10 torr of pressure. The temperature is maintained at 45°C . Coupled with the downward pressure of the clamp fingers, it is easy to see that the wafer is subjected to a large amount of force. This force must be kept in mind when etching, for as the etch goes deeper, the wafer gets weaker. In many cases wafers have broken in the chamber. This problem is solved by bonding two wafers by photoresist before etching. The bonding aids in thermal transfer enough to keep the resist from burning. The depth of the

cavity is measured after subsequent runs of the Deep RIE process, by using Wyko Optical profile meter.

Fig 7.3 Karl Suss MA6 Mask aligner

Fig 7.4 Spin coater used for coating photoresist

Fig 7.5 Deep RIE etched cavities in the substrate of the transducer

Fig 7.6 Deep RIE STS

The second wafer is processed in parallel to form cMUT cell in it. Silicon oxide is grown is about $1\ \mu\text{m}$ over the silicon substrate. The Tystar furnace used for oxidation is shown in Fig. 7.7. The oxidation is performed at high temperature of about $1100\ ^\circ\text{C}$ in pure oxygen or in wet oxygen ambient. Subsequently, a layer of polysilicon is deposited on the oxide layer by using Low Pressure Chemical Vapor Deposition (LPCVD) process. Fig 7.8 shows the second wafer after deposition of polysilicon layer on it. The insulating layer of silicon nitride is deposited over polysilicon. The Nitride furnace is used to deposit silicon nitride thin-film. Depending on the ratio of gas used, the thin-film can be stoichiometric Si_3N_4 or silicon rich nitride, which will then have a low stress. This silicon nitride layer defines the thickness of the vacuum gap in the cMUT cell and it also provides the support to the polysilicon membrane. Moreover, it serves as an insulating

layer between the top electrode and the bottom electrode, which is silicon substrate. The silicon nitride is deposited on the substrate is shown in Fig. 7.9. Further, the silicon nitride is patterned by using standard photolithography process described earlier and etched using Reactive Ion Etching (RIE) process to form the cMUT cells as shown in Fig. 7.10.

Fig 7.7 Furnace Tystar

Fig 7.8 Second wafer with oxide grown on it and polysilicon deposited by LPCVD

Fig 7.9 Insulating layer of silicon nitride is deposited by PECVD

Fig 7.10 Silicon nitride layer etched by RIE to form cMUT cells

The critical step in the fabrication process is wafer bonding. In the present research work silicon nitride layer one wafer is to be bonded with silicon on another wafer. Hence, it is convenient to use eutectic bonding technique. To use eutectic wafer bonding technique, it is required to deposit a thin layer of chromium-gold on the silicon nitride layer. The silicon nitride deposited wafer is first patterned to fill up all the RIE etched cavities, which are future cMUT cells as described previously. It is then patterned using photolithography to cover all the etched pattern of the silicon nitride layer. The thin layer of 2000 Å Au/Cr is sputtered on it using the sputtering process. The cMUT cells are then subsequently released by using lift off process. Fig. 7.11 shows the second wafer after the deposition of the chromium-gold layer on it. It is found that the chromium-gold layer is not sufficient to bond the wafer, as the bonded wafers got separated during Deep RIE etching process, which follows wafer bonding. Hence, the polymer layer is also deposited to make the bonding between the two wafers stronger. The wafer is further cleaned and kept ready for the wafer bonding process.

Fig 7.11 Gold-Chromium layer deposited on the silicon nitride layer

Before undergoing the wafer bonding process, both the wafer are cleaned and aligned properly by using predefined alignment marks on the respective patterns of the wafers. The same mask aligner (Karl Suss MA6) is used for wafer alignment by using different wafer holding chuck. This is done in order to keep the acoustic cavities exactly below the cMUT cells. Each acoustic cavity encapsulates four cMUT cells. The aligned wafers are subjected to eutectic wafer bonding process at a temperature of 400°C and a pressure of about 2 bar. The equipment used for wafer bonding process is shown in Fig 7.12.. Fig 7.13 shows the two wafers bonded together to form cMUT structure. The top wafer is removed by subsequent RIE and Deep RIE process to release the polysilicon membrane to form the cMUT cell over the substrate as shown in Fig 7.14.

Fig 7.12 The equipment used for wafer bonding

Fig 7.13 Two wafers are bonded together using eutectic wafer bonding process

Fig 7.14 Top layers are removed by using RIE and Deep RIE process

The polysilicon layer is further patterned using photolithography to deposit the top electrodes and interconnects. Further gold electrodes and interconnects of 200 nm thick are formed by sputtering gold on the polysilicon membrane as shown in Fig. 7.15. Finally, the acoustic cavity is filled with the acoustic attenuating material that has a relatively high acoustic attenuation to provide an acoustically damped region below the cMUT as shown in Fig. 7.16. The material to be filled in the acoustic cavity should be sufficiently lossy to dissipate the acoustic energy present in the waves. Hence, the material chosen here is an epoxy called tungsten epoxy with a volume fraction of 10%. The material is deposited in acoustic cavity by direct injection or by spin coating method, which is used for depositing photoresist on the silicon wafer. The cMUT array is

positioned on an acoustic backing member to support the array and to provide further acoustic attenuation. The pictures of the transducer under microscope are shown in Fig. 7.17. The pictures of the transducer taken under Scanning Electron Microscope (SEM) are shown in Fig. 7.18 to Fig. 7.20. The fabrication yield of 4 inch wafer is 48 1-D transducer arrays of 32 elements. All transducer arrays are not found in good condition. The Deep RIE process is very critical for successful fabrication. The etching is not uniform all over the wafer. Hence, the final thickness of the cMUT membrane is uniform and even at some places the membrane is found to be broken because of low tensile strength. In addition, as the cMUT membrane thickness is different, the characteristics (resonant frequency) of the transducer elements will be different

Fig 7.15 cMUT structure with gold electrodes deposited on it

Fig 7.16 cMUT array with acoustic cavity filled with PDMS and supported by acoustic backing material

Fig 7.17 Picture of fabricated transducer under microscope

Fig 7.18 SEM pictures of the cMUT transducer

Fig 7.19 SEM pictures of the transducer array closer views

Fig 7.20 SEM pictures of the transducer array angled view

7.3 Transducer characterization

The 1-D linear transducer array of 32 elements is fabricated as described previously. To summarize, firstly an acoustic cavity of rectangular shape is formed below the cMUT cell as shown in Fig. 7.1 by using DRIE process. The acoustic cavity of the transducer is the filled with tungsten loaded epoxy with 10% volume fraction. This fabricated prototype of the transducer is used in the characterization process to study the pulse echo response of the transducer and to validate one of the finite element results obtained for the same configuration of the transducer in chapter 6. The pulse echo response of the transducer has been studied by using FE analysis. The pressure of the reflected waveforms is plotted in time domain as well as in frequency domain as shown in Fig. 6.14.

The pulse echo response of the transducer is studied in the transducer characterization. The test set up used in the transducer characterization is shown in Fig. 7.21. The schematic diagram of the transducer characterization set up is shown in Fig. 7.22, which describes all the equipments used in the characterization process. The transducer is placed on the top of the water surface for transmitting ultrasound waves in downward direction. The reflector is place in the water at about 5 cm away from the transducer. The DC bias voltage of 300 V is applied from high voltage supply (P059066 from Precision Acoustics) through DC bias network analyzer. The AC part of the input voltage i.e. 10V pulse is supplied from the AC function generator (5077PR from Panametrics). The pressure of the reflected ultrasound waves from the reflector is measured by a calibrated hydrophone placed near the transducer. The hydrophone frequency range is 1-10MHz, the aperture size is less than 500 μm . The hydrophone

response signal is amplified by using preamplifier with a 32 dB mid-band gain and a 3-dB bandwidth of 3.5 MHz centered at 2.5 MHz and plotted in oscilloscope shown in Fig. 7.22 and taken to the computer for analysis. The measured impulse response and corresponding frequency spectrum is shown in Fig. 7.23 and Fig. 7.24 respectively. The maximum amplitude of the pressure of the reflected waves is around 1.5KPa. It is shown in Fig. 7.24 that the fractional bandwidth @ -6dB is about 56.15%. The fractional bandwidth @ -6dB of the transducers obtained from finite element analysis is 61.048%, which is shown in Fig. 6.15.

The fractional bandwidths obtained from the impulse response experiment and from the pulse echo response in Finite element analysis are fairly matching with each other. The difference between the analysis and testing results is due to the dissimilarities in finite element model and the fabricated prototype. The fabricated prototype is not exactly similar to the finite element model because of the limitations in fabrication processes. Firstly, due to the limitations in the Deep RIE process, the shape of the acoustic cavity is not exactly rectangular. Hence, the depth of the acoustic cavity is not uniform. Secondly, again due to the limitations in the Deep RIE process the membrane thickness is not uniform. As Deep RIE process is not uniform over the membrane, the silicon wafer is not completely etched at some places on the silicon membrane. These defects can be removed by fabricating transducer prototype in professional silicon microfabrication companies, which use sophisticated and automated tools to fabricate the devices.

Fig 7.21 Experimental set up for the transducer characterisation

Fig 7.22 Schematic diagram of the experimental set up

Fig 7.23 Pressure obtained from hydrophone in Time domain

Fig 7.24 Frequency spectrum of the impulse response obtained from the hydrophone

7.4 Commercial Applications

The present invention develops capacitive micromachined ultrasonic transducer, which reduces the cross coupling in the transducer elements and it improves the image quality of the ultrasound system. The use of this transducer in diagnostic ultrasound systems will provide appreciable assistance to doctors and clinicians to make the correct diagnosis. This design helps in producing more dense 2D arrays of the ultrasonic transducer, which further improves the image quality. The design can be further used in portable ultrasound systems and in the future can be implemented in hand held ultrasound systems, where transducer arrays can be integrated to front-end electronics of ultrasound systems.

Chapter 8 Conclusion and Future plan of work

8.1 Conclusion

A detailed study has been done on piezoceramic transducers and capacitive micromachined transducers under the scope of this thesis. To propose a cost efficient transducer, various materials and geometries were evaluated with the help of Finite Element Modeling (FEM). FEM models were generated for 1-D and 2-D PZT-5H control arrays and Hybrid arrays. Impedance plots and Pulse echo results from FEM analysis were studied. It is inferred from the research that hybrid arrays give a better performance as compared to the PZT-5H control arrays. To optimize the FEM model, a new material, PNN-PT-PZ was considered. Transducer arrays of new material (PNN-PT-PZ) give better results as far as bandwidth and badness factor is concerned. However, this material is more expensive than PZT-5H control elements.

The finite element analysis is done on the 2-D array of the transducer for each material. The 3-D finite element models were constructed for the 2-D array of the transducer. It is found that the bandwidth of the 2-D array is narrower than that of 1-D array. The bandwidth of the 2-D array is affected due to cross talk between the elements and as compared to the 1-D array. Studies show that it is very expensive to fabricate the piezoceramic arrays. Moreover, fabrication of a 2-D transducer array is difficult because the size of the element is smaller. Due to small size of the elements, the acoustic impedance of the transducer is much higher than that of the tissue. This impedance mismatch affects the signal to noise ratio and the quality of the image. Another

drawback, in case of a 2-D array is that the number of elements involved is large. This implies more number of electrical connections, which in turn affect the quality of the image.

The focus of research shifted to the upcoming Capacitive Micromachined Ultrasonic Transducers (cMUTs) as alternative transducers to piezoceramic transducers to overcome some of the aforementioned difficulties. cMUTs satisfy most of the requirements of the ultrasound system that primarily include good sensitivity and image quality. Hence, further study has been done on cMUTs. Analysis and design of membrane of cMUTs is a critical part, as far as performance of the transducer is concerned. To accomplish this, FEM analysis of the membrane is done at the cell level. The collapse voltage is calculated for the designed membrane. A new design is proposed for the membrane. The cross coupling between the transducer elements of the capacitive micromachined ultrasonic transducer is the major factor, which affects the bandwidth of the transducer, and it in turn affects the image quality of the ultrasound system. This problem is studied and analysed in detail using finite element analysis. A novel structural design has been proposed, which consists of an acoustical cavity filled with acoustically attenuating material to reduce the cross coupling between the transducer elements. The FEM modeling results show the bandwidth of the transducer is reduced considerably due to the cross coupling. The FEM results further illustrate that the proposed structure reduces the cross coupling between the transducer elements and increases the bandwidth of the transducer eventually.

The FEM analysis of the transducer has been done by using commercial software PZflex. The software is basically used for the conventional piezoelectric transducer. To complete the FE analysis of the cMUT, an electrostatic analysis has to be performed. The PZflex software needs to be carefully configured to effectively carry out electrostatic analysis of the cMUT. The ringing in the cMUT transducer is one of the critical issues found during the analysis. The ringing has to be reduced to observe clearly the effect of cross talk on the bandwidth of the transducer. The ringing of the transducer has been reduced by a careful selection of damping frequency.

The fabrication process is designed for the novel proposed structure of the transducer. The 1-D array of 32 transducer elements is fabricated using the convening MEMS fabrication processes. The wafer bonding is one of the critical steps involved in the transducer fabrication. It is difficult to bond silicon oxide with the silicon wafer. Here the technique used is eutectic bonding with gold layer deposited between silicon oxide and silicon. The pulse-echo experiment has been carried out on the fabricated prototype of the cMUT to get the bandwidth of the transducer. The characterization results match fairly with the FEM results.

In summary the following conclusions are drawn from the above study

1. The cMUT transducer has been designed for centre frequency of 3.3 MHz and it can be used for abdominal medical imaging

2. The finite element analysis has been used as a tool to study the cross-talk between the transducer elements and it has been found that the cross-talk reduces the fractional bandwidth @ -6dB of the transducer from 101.01 % to 47.32 %.
3. The novel structure has been designed to reduce the cross-talk. The structure has an acoustic cavity below the cMUT cells and it is filled with the acoustically attenuating material. From finite element analysis it is found that the bandwidth of the transducer has been increased in the novel structure. For rectangular acoustic cavity the fractional bandwidth @ -6 dB has been improved from 47.32 % to 61.048 %. This bandwidth is obtained from the transducer which is without backing and matching layers. If these layers are added to the transducer on front and backside the transducer frequency will increase further.
4. The novel transducer prototype has been fabricated using conventional MEMS processes.
5. The transducer has been characterised by using pitch-catch experiment to obtain the bandwidth of the transducer. The fractional bandwidth @ -6dB obtained from pitch-catch experiment is 56.15%. This is fairly matching with the finite element analysis results (61.048 %). The difference between the analysis and testing results is due to the dissimilarities in finite element model and the fabricated prototype. The fabricated prototype is not exactly similar to the finite element model because of the limitations in fabrication process as mentioned earlier.

8.2 Scope for Future Research Work

The capacitive micromachined ultrasonic transducers are a promising alternative to the conventional ultrasonic transducers. There is a lot of scope for the future research work. The research in the area of cross coupling problem in the cMUT transducer is still at its early stage. The performance of the transducer can be still improved with some innovative solution to this problem apart from the one discussed in this thesis.

The membrane formation and the fabrication of the cMUT device is also an area where a lot of research can be done to make the fabrication more efficient and simple. The integration of the transducer with the front end electronics of the ultrasound is one of the most challenging areas in near future. The integrated transducer allows the transducer fabrication along with the front end electronics in the same wafer. The fabrication of this type of transducer is another major area for future research. Many research groups all over the world are involved in this research. There are various methods that can come up for integrating the transducer with front end electronics. To reduce parasitic capacitance, while maintaining a low leakage current and high break down voltage is a major engineering challenge. As a part of the characterization the measurement of directivity and the spatial function of the cMUT transducer useful future work. The measurement cross-talk between the transducer elements due to lamb waves is also a challenging work. The cMUT technology is advantageous as it uses the conventional MEMS fabrication processes. This technology and the research in this area can be further implemented in other device fabrications such as silicon microphone, pressure sensors, accelerometers, and other MEMS devices.

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List of Publications

- [1] Vinay S. Sawant, M. K. Lim, “Design and Fabrication of Capacitive Micromachined Ultrasonic Transducer for reduced Cross-coupling between transducer elements”, Invention disclosure submitted.
- [2] Vinay S. Sawant, M. K. Lim, “Comparison of a new material PNN-PT-PZ with PZT arrays for medical ultrasound transducers by finite element analysis,” presented at 2nd IEEE-EMBS International Summer School and Symposium on Medical Devices and Biosensors (ISSS-MDBS), July 2004.
- [3] Vinay S. Sawant, M. K. Lim, S. M. Krishnan, “Comparison of a new material PNN-PT-PZ with PZT arrays for medical ultrasound transducers by finite element analysis, “ Presented at MIDAS symposium of the Joint 3rd International Conference on Materials Processing for Properties and Performance (MP³) and International Conference on Nanostructures and Nanotechnology (N²), November 2004.
- [4] Vinay S. Sawant, M. K. Lim, “A New Design and Fabrication Process for Capacitive Micromachined Ultrasonic Transducer to reduce the cross-talk between the transducer elements,” Symposium F: Nano-Optics and Microsystems of the 3rd International Conference on Materials for Advanced Technologies (ICMAT 2005) and the 9th International Conference on Advanced Materials (ICAM 2005), July 2005
- [5] Vinay S. Sawant, M. K. Lim, “A New Design and Fabrication Process for Capacitive Micromachined Ultrasonic Transducer to reduce the cross-talk between the transducer elements,” submitted to *Sensors and Actuators A: Physical*.
- [6] Vinay S. Sawant, M. K. Lim, “Finite element analysis of cross coupling between the transducer elements of cMUT and its effect on bandwidth”, submitted to *Applied Acoustics*.

[7] Vinay S. Sawant, M. K. Lim, “Comparative study of piezoelectric, hybrid piezoceramic and single crystal (PNN-PT-PZ) transducer arrays using Finite Element analysis”, submitted to *Applied Acoustics*.

[8] Vinay S. Sawant, M. K. Lim, “Design, Fabrication and Characterisation of cMUT based transducer array for reduced cross-talk”, submitted to *Ultrasonics*.