Design of Piezoelectric Micromachined Ultrasonic Transducer (PMUT) Arrays for Intrabody Networking Applications

A Thesis Presented
by

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<td>3.10</td>
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List of Acronyms

ADC  Analog to Digital Converter
ADS  Advanced Design System
AlN  Aluminum Nitride
BAN  Body Area Network
BW   Band Width
CMUT Capacitive Micromachined Ultrasonic Transducer
DC   Direct Current
DRIE Deep Reactive Ion Etching
DSP  Double Side Polished
ECG  Electro Cardio Gram
EEG  Electro Encephalo Gram
EMG  Electro Myo Gram
FEM  Finite Element Modeling
FPGA Field Programmable Gate Array
GSG  Ground Signal Ground
IVUS Intra Vascular Ultra Sound
KOH  Potassium Hydroxide
LTO  Low Temperature Oxide
MAC  Medium Access Control
MEMS Micro Electro Mechanical Systems
MPPT Maximum Power Point Tracking
PMUT  Piezoelectric Micromachined Ultrasonic Transducer

RF  Radio Frequency

SiO2  Silicon Dioxide

SiN  Silicon Nitride

SOI  Silicon On Insulator

SONAR  SOund Navigation and Ranging

UCD  University of California Davis

XeF2  Xenon Difluoride
Acknowledgments

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Abstract of the Thesis

Design of Piezoelectric Micromachined Ultrasonic Transducer (PMUT) Arrays for Intrabody Networking Applications

by

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Master of Science in Electrical and Computer Engineering
Northeastern University, August 2017
Dr. Matteo Rinaldi, Advisor

The present work details the design process for Piezoelectric Micromachined Ultrasonic Transducer (PMUT) arrays for wireless intrabody networking applications. Transducers operating at 0.7 and 2MHz are designed for creating a communication link in the body medium interfacing with Field Programmable Gate Array (FPGA) based circuitry and transmission protocols developed in previous work. After definition of the transducer geometry, design constraints such as dimensions, source power, directivity, transmission distance, and bandwidth are addressed for allowing the transducers to operate within miniaturized implantable devices.

A matrix model is obtained for the transducer and, along with a medium transmission model, a system-wide representation is obtained allowing simulation of the transmitter, medium and receiver arrangement with varying parameters. An equivalent circuit model is also developed for prediction of performance parameters. The necessary quantities for these models are extracted from Finite Element Modeling (FEM) simulation.

A micro fabrication process is also developed for the designed topology and progress in its implementation is shown.
Chapter 1

Introduction

SOund Navigation and Ranging (SONAR) and Ultrasound Imaging are well established technologies that, albeit operating in widely distant scales, have the commonality of being based on ultrasonic wave transmission. Both use timing information between emission and reception of an ultrasonic pulse that is reflected from an object of interest to estimate the distance to it. The same principle is used but the dimensions, operating frequencies and intensities allowed for safe operation are orders of magnitude away.

The operation of SONAR is characterized by low frequency, large transducers designed for long distances. Applications of the technology include bottom topology mapping in oceanography, submarine guidance and communication systems and sonobuoy beacons [1]. Contrasting to this, ultrasound imaging uses high frequency ultrasound at short distances having the advantage of improved resolution.

The point of this digression is to illustrate the versatility of ultrasound to adapt to different scenarios. Even so, it has not reached the popularity of Radio Frequency (RF) wave transmission due to the superiority of electromagnetic wave transmission in air at long distances. When the medium is changed to water, the story is quite different however, as radio waves have a very high attenuation in this medium.

This fact, plus the developments in Micro Electro Mechanical Systems (MEMS), open up new possibilities for utilization of ultrasonic waves. The small form factor, plus better matching to aqueous media, has already been exploited by miniaturized ultrasound imaging devices and Intra Vascular Ultra Sound (IVUS) for imaging or kidney stone ablation [2].

Nevertheless, the implementation of ultrasound for medical applications has been limited to stand-alone devices, leaving the known possibility of communication by ultrasonic waves behind.
CHAPTER 1. INTRODUCTION

It is here that a qualitative leap has been proposed \cite{3, 4} to exploit this capability for developing ultrasound-based Body Area Network (BANs). The present work fits within this framework by exploiting the advantages of MEMS technology to develop transducers optimized for the requirements of intra-body communication.

1.1 Motivation

The main motivation for pursuing wireless communication within the body is to provide a wide platform to enhance the functionality of implantable devices. The objective is not to address a specific illness or condition but to provide a wide ranging platform for several kinds of sensors and actuators that either only function within the body or for which the performance or accuracy is greater if they are implanted.

The most widespread and evident of these applications is pacemakers. Advancements in pacemaker technology currently allow them to be reprogrammed based on the user’s progress and even some dynamically adjust the pace relative to the level of activity measured by accelerometers, metabolic or blood oxygen saturation implantable sensors \cite{5}. A distributed pacemaker, in which the stimulation not only occurs on a particular region of the heart but on several spots with a coordination scheme can also make the heart response smoother and better controlled. For all of this possibilities, a wireless communication network is a requirement.

Pacemakers also operate on batteries that require periodic replacement. This involves a surgical procedure with a higher associated risk of infection on each iteration. The same wireless ultrasonic link developed for communication can be used for energy transmission for battery recharging. Furthermore, the ultrasonic transducer may work as an energy harvester by converting available acoustic energy from ambient noise to electrical energy that can charge the batteries when the communication link is not in use.

Metabolyte (specific chemicals present in the body regulating or aiding a particular function) regulation is also a major possible field of application. The most known of these is insuline, which needs to be artificially provided in patients with diabetes. Implantable chemical dispensers are a possible solution to this which could be networked to an insuline level sensor or a programmable control unit in general \cite{6}. Other examples of chemical regulation include the endocrine system (hormone regulation) or drug dispensers localized on the affected area, like a cancerous tumor.

Telemetry and logging of measurements from several implanted sensors on a centralized unit is another possibility for monitoring patients that are recovering from surgery or who suffer from...
a chronic disease. The centralized unit can then be connected to a mobile device or to the internet to be accessed by a physician. Electro Cardio Gram (ECG) signals, heart rate, blood oxygen level, pressure and position and orientation of the body are measurements that these implanted sensors could provide to obtain an integral snapshot of the current state of the patient [6].

Figure 1.1 shows a summary of sensors and other more complex devices that could be interfaced to a BAN with their data transmission rate and corresponding bandwidth requirements.

<table>
<thead>
<tr>
<th>Application</th>
<th>Data rate</th>
<th>Bandwidth (Hz)</th>
<th>Accuracy (bits)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ECG (12 leads)</td>
<td>288 kbps</td>
<td>100–1000</td>
<td>12</td>
</tr>
<tr>
<td>ECG (6 leads)</td>
<td>71 kbps</td>
<td>100–500</td>
<td>12</td>
</tr>
<tr>
<td>EMG</td>
<td>320 kbps</td>
<td>0–10,000</td>
<td>16</td>
</tr>
<tr>
<td>EEG (12 leads)</td>
<td>43.2 kbps</td>
<td>0–150</td>
<td>12</td>
</tr>
<tr>
<td>Blood saturation</td>
<td>16 bps</td>
<td>0–1</td>
<td>8</td>
</tr>
<tr>
<td>Glucose monitoring</td>
<td>1600 bps</td>
<td>0–50</td>
<td>16</td>
</tr>
<tr>
<td>Temperature</td>
<td>120 bps</td>
<td>0–1</td>
<td>8</td>
</tr>
<tr>
<td>Motion sensor</td>
<td>35 kbps</td>
<td>0–500</td>
<td>12</td>
</tr>
<tr>
<td>Cochlear implant</td>
<td>100 kbps</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Artificial retina</td>
<td>50–700 kbps</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Audio</td>
<td>1 Mbps</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Voice</td>
<td>50–100 kbps</td>
<td>–</td>
<td>–</td>
</tr>
</tbody>
</table>

Figure 1.1: Summary of implantable devices with their corresponding transmission requirements (from [7])

1.2 Previous Work

In [3], ultrasonic waves are proposed as a physical medium for communication in BANs. An overview of the requirements and challenges is also presented. After an initial model for transmission of acoustic waves within the body is developed, the wide range of possible dimensions of the transducers and their operating frequencies is narrowed (Figure 1.2). Considerations such as bandwidth, reflection and scattering issues and health concerns are initially addressed.

A Medium Access Control (MAC) scheme is proposed which is further developed in [4]. An Adaptive Time hopping scheme, based on short pulses of ultrasound, is detailed. The short-duration pulses reduce detrimental heating effects in tissue while limiting reflections and scattering.
CHAPTER 1. INTRODUCTION

![Table of Ultrasonic Transducer Frequency Limits](from [3])

<table>
<thead>
<tr>
<th>Communication range</th>
<th>Distance</th>
<th>Frequency Limit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Short Range</td>
<td>(\mu\text{m} - \text{mm})</td>
<td>&gt; 1GHz</td>
</tr>
<tr>
<td>Medium Range</td>
<td>mm - cm</td>
<td>(\approx 100\text{MHz})</td>
</tr>
<tr>
<td>Long Range</td>
<td>&gt;cm</td>
<td>(\approx 10\text{MHz})</td>
</tr>
</tbody>
</table>

Further work from the same research group has developed FPGA-based circuitry implementing this protocol and interfacing it with commercially available ultrasound transducers operating at a frequency of 700kHz. The setup was also successfully tested on a medium consisting of an artificial gel with organ phantoms, replicating the acoustic behavior of real tissue and organs.

The present work, which is part of a joint effort project with the mentioned research group, aims to design and fabricate MEMS-based transducer arrays, optimized for intra-body transmission to interface with the existing setup. Better acoustic matching, power efficiency, smaller form factor and capability of beamforming and focusing are expected outcomes.

1.3 State of the Art

Work from the University of California Davis (UCD) and Berkeley Sensor and Actuator Center has been the most thorough regarding design of PMUT arrays for intra-body applications. [2] details the design, fabrication and testing of circular diaphragm transducer arrays based on silicon dioxide as a base layer and aluminum nitride as the piezoelectric actuation layer. High fill factor, with a large number of elements (1261 elements in a 1.2 mm diameter circle), a wide bandwidth (4.9MHz) and a highly focused acoustic beam are features of this design. The center frequency is 18.6MHz. The application target for this design is IVUS.

From the same research group, [8] and [9] develop transducer arrays based on Silicon On Insulator (SOI) wafers with cavities previously defined below the device layer (a process commercially available at the Irish company ICEMOS). The application for these is ultrasound imaging. The frequencies of operation range from 10 to 55MHz in air and beamforming was successfully demonstrated.

Regarding innovative topologies for the PMUT construction, [10] designs and fabricates transducers based on a double piezoelectric layer with dual electrodes (concentric separated electrodes) actuated out of phase in such a way that the resulting bending moments add up to four times...
CHAPTER 1. INTRODUCTION

the moment of a single layer, thus producing a four fold greater output pressure. Figure 1.3 shows
the topology for the transducer.

Figure 1.3: Principle of operation illustration for bimorph PMUTs. a) Single layer, b) Single layer
with differentially-driven electrodes, c) Bimorph and differential drive (from [10])

[11] shows a stress engineering approach to create curved diaphragm PMUTs optimizing
the fill factor for the same radius and increasing the output pressure. The construction, based on the
difference of residual stresses after deposition of silicon nitride and Low Temperature Oxide (LTO)
films, can be seen in Figure 1.4.

A ring shape for the diaphragm is used in [12], which allows for an additional degree of
freedom in the geometry of the PMUT as both the radius of the structure as well as its width can
be separately varied. Reportedly, the width sets the main resonant frequency while the radius sets a
second, acoustically matched, frequency in the same mode of vibration when the device operates
in fluid. Selection of an adequate radius that produces a peak on the second resonant frequency
allows both resonant modes to merge resulting in a large bandwidth spectrum. A 620kHz, tunable
bandwidth was reported. Figure 1.5 shows the geometry of the device.
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Figure 1.4: Structure of curved PMUT (from [11])

Figure 1.5: Structure of ring PMUT (from [12])
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An option for achieving a higher bandwidth is shown in [13], where long, rectangular diaphragms are built that have several mode shapes at resonant frequencies in air that are close together. Once the devices operate in water, the peaks become less narrow and merge in a wide region of large response. Figure 1.6 illustrates the geometry, modes of operation and frequency response in air (when in fluid, the peaks of the frequency response merge together).

Figure 1.6: a) Membrane maximum displacement frequency response and b) Mode shapes for rectangular PMUT (from [13])

The performance data obtained in the main state-of-the-art papers is collected in Table 1.1.

<table>
<thead>
<tr>
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<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Structure</td>
<td>Cavity</td>
<td>Cavity</td>
<td>Bimorph</td>
<td>Curved</td>
<td>Ring</td>
<td>Rect.</td>
</tr>
<tr>
<td>Radius (µm)</td>
<td>25</td>
<td>50</td>
<td>170</td>
<td>200</td>
<td>275</td>
<td>1550x250</td>
</tr>
<tr>
<td>Passive layer (um)</td>
<td>0.8</td>
<td>2.5</td>
<td>0.15</td>
<td>4.65</td>
<td>1</td>
<td>11</td>
</tr>
<tr>
<td>Piezoelectric (um)</td>
<td>0.95</td>
<td>0.8</td>
<td>0.95</td>
<td>2</td>
<td>0.8</td>
<td>2</td>
</tr>
<tr>
<td>Array Number</td>
<td>1261</td>
<td>72x9</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Air $f_0$ (MHZ)</td>
<td>18.6</td>
<td>19</td>
<td>0.345</td>
<td>0.647</td>
<td>2</td>
<td>1.12</td>
</tr>
<tr>
<td>Fluid $\Delta f$ (MHZ)</td>
<td>4.9</td>
<td>3.4</td>
<td>-</td>
<td>-</td>
<td>0.8</td>
<td>1</td>
</tr>
<tr>
<td>Air $\Delta y_{max}$ (nm/V)</td>
<td>-</td>
<td>13.7</td>
<td>452</td>
<td>40</td>
<td>100</td>
<td>-</td>
</tr>
<tr>
<td>P (kPa/V)</td>
<td>2</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Quality Factor</td>
<td>45</td>
<td>140</td>
<td>-</td>
<td>-</td>
<td>-</td>
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</tr>
</tbody>
</table>

Table 1.1: State-of-the-art PMUT Parameters
Chapter 2

Background

2.1 PMUTs

There are several methods for electroacoustic actuation, including magnetic fields on a coil producing motion of a magnet, an electrostatic force creating the vibration or piezoelectric strain being generated from applied electric potential. The first has been widely used in speakers and underwater transducers but is not easily subject to miniaturization. The second is the basis for operation of the Capacitive Micromachined Ultrasonic Transducer (CMUT), where the "capacitive" term comes from the geometry of the transducer being an electrode separated by air (dielectric) to another one and is characterized by a varying capacitance. Even though CMUTs do benefit from MEMS techniques to be miniaturized, they present the drawback of requiring a high Direct Current (DC) bias for achieving high performance. The final option, piezoelectric actuation, allows for miniaturization with less geometric constraints, and lower power requirements [14]. The topology of the small form factor transducers can be seen in Figure 2.1.

PMUTs operate by the conversion of electrical potential to mechanical force through the piezoelectric material. A thin layer of piezoelectric material, together with some thin passive layer, form a membrane. When the piezoelectric material tries to contract or expand, as the membrane is clamped on the edges, it produces a deformation that displaces the membrane out of the horizontal plane, displacing the medium around it and creating a pressure wave.
CHAPTER 2. BACKGROUND

Figure 2.1: Ultrasonic electroacoustic transducer topologies: a) $d_{33}$ mode bulk piezoelectric ultrasonic transducer, b) CMUT, c) PMUT

2.2 Piezoelectric Actuation

Piezoelectricity is a phenomenon appearing in several crystalline materials in which a stress applied on a material (piezo derived from the greek piezein meaning squeeze) creates an electric potential. The converse effect also occurs: applying a voltage to a piezoelectric material causes it to expand or contract depending on the polarity of this voltage [1].

As both the mechanical, elastic stress-strain relationship and the piezoelectric effect coexist in the material, a matrix relationship is used to express both simultaneously. The equations relating these variables are as follow, where $S$ and $T$ are 1x6 column matrices containing the longitudinal and shear components of stress and strain respectively, $d$ is a 3x6 matrix of the corresponding piezoelectric coefficients, $\epsilon^T$ is a 3x3 matrix containing permittivity coefficients, and the superscripts denote which variable was held constant when measuring the coefficients in order to represent only the effect of the variable that is being changed [1]:

$$S = s^E T + d^t E$$  \hspace{1cm} (2.1)

$$D = d^T + \epsilon^T E$$  \hspace{1cm} (2.2)

Many of the coefficients involved are null and the relationships from (2.1) and (2.2) reduce to four linear equations for two cases of practical interest: the electric field and the corresponding strain is parallel, related to piezoelectric coefficient $d_{33}$, and the electric field is perpendicular to the resulting strain, relating to $d_{31}$. Talking specifically about the PMUT geometry, $d_{33}$ relates a potential applied on the faces of the membrane to expansion or contraction in a direction perpendicular to this faces. $d_{31}$ relates the same applied potential to a lateral contraction of the membrane, causing it to deflect out of plane as it is clamped on the edges (Figure 2.1).
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2.3 Mechanics of a Vibrating Membrane

Any material, within its elastic regime, is subject to a linear stress-strain relationship when it is compressed or expanded. This is characterized by the bulk material property Young’s modulus \(E\) and the geometry. The contraction or expansion perpendicular to the applied force (as the volume of the element must remain constant) is characterized by the material’s bulk Poisson ratio \(\nu\). When bending is involved, the Young’s modulus does not characterize the problem fully, but the flexural rigidity, which is obtained by use of the elastic modulus and integration over the geometry.

Additional considerations must be taken in our scenario as the membrane of the PMUT does not only statically deflect but vibrates as a specific frequency. The mass of the membrane itself affects this dynamic behaviour, depending on it’s density \(\rho\). The resistance of the medium affects the vibration as a load, characterized by it’s acoustic impedance, which is frequency dependent.

Also, the piezoelectric material relates the stiffness of the membrane to the electrical domain, where frequency dependent loading effects also occur. It is because of this multidomain dependance that the vibrating membrane cannot be characterized in isolation but the equivalent circuit approach will be taken as explained in Chapter 2.5.1.

However, a useful link between the mechanics of the membrane and its resonant frequency in air, where very little medium loading occurs, can be seen in equation (2.3). This relationship only holds if the residual stress (present in the membrane without actuation, due to the fabrication only) is low [8]. On equation (2.4), it is important to note that the Young’s modulus depends on the distance to the neutral axis \(z\) as there are several materials in the stack.

\[
f_0 = \left(\frac{3.2}{r}\right)^2 \sqrt{\frac{D}{\rho}} \quad (2.3)
\]
\[
D = \int_{bot}^{top} \frac{E(z)z^2}{1 - \nu(z)z} dz \quad (2.4)
\]

2.4 Acoustic Radiation

An element vibrating within a medium produces a disturbance in the latter. Regarding a vibrating membrane, when it bulges out into the surrounding medium, it pushes the particles composing it closer together (compression), and when it retracts, it pulls them further away from each other (rarefaction). This effect progressively spreads out through the medium as a pressure wave.
A model for the spatial pattern of this acoustic radiation is desirable to predict the intensity at point in space where a receiving element could be situated, to guarantee safe operating intensity levels and to design transmitters for a specific application. Such a model has been developed in Acoustics theory by posing a wave equation and solving it relating to the geometry and parameters of the setup. This equation is formulated from the application of Newton’s second law on a small fluid element along with continuity equations of the medium [15].

There are complex interactions between the emitted waves close to the surface of the transducer but, from a certain distance, it acts as a concentrated source emitting a single spherical wave. This region is known as the far field, and the solution for the pressure \( p \) at a certain radial distance \( r \) is as follows [8]:

\[
p(r) = j \frac{ka^2}{2r} P_0 e^{j(\omega t - kr)} D(\theta) \tag{2.5}
\]

\[
D(\theta) = \frac{48 J_2(ka \sin \theta)}{(ka \sin \theta)^3} \tag{2.6}
\]

\[
P_0 = \rho c u_0 \tag{2.7}
\]

\[
k = \frac{2\pi}{\lambda} = \frac{\omega}{c} \tag{2.8}
\]

\( k \) is a parameter called the wave number related to the angular frequency of the wave \( \omega \) and the sound speed for the particular medium \( c \) and is also related to the wavelength \( \lambda = c/f \). \( a \) is the radius of the membrane. \( D(\theta) \) is the directivity factor, which increases or reduces the pressure at a certain angle. The directivity is smaller (narrower beam) for higher frequency of the wave. A general guide for the directivity is that if \( 2a < \lambda/2 \), the response is omni directional, while larger dimensions in relation to the wave number have a narrower beam width. Figure 2.2 shows the resulting directivity function for varying dimensions.

Equation (2.7) relates the membrane surface pressure \( P_0 \) to the velocity at the center of the membrane \( u_0 \). This velocity is also the product of the displacement amplitude and the angular frequency. For the case of a vibrating membrane, the average displacement value must be used, which is a third of the maximum value (at the center of the membrane). In a more general case, the relation between the pressure and velocity is called acoustic impedance \( Z_{ac} \):

\[
P = uZ_{ac} \tag{2.9}
\]

Equation (2.5) assumes an ideal, lossless medium. In reality however, there is attenuation besides the simple spherical \( (1/r) \) radiation predicted by this equation. Attenuation is higher with
Figure 2.2: Single element directivity for membrane diameter: a) $0.1\lambda$, b) $0.5\lambda$, c) $\lambda$ and d) $3\lambda$

increasing frequency. Equation (2.10) empirically models this effect as a factor that should be multiplied by the value previously obtained for pressure. The equation shown for the $\alpha$ factor is for water.

\[ p_{\text{attenuation}}(r) = p(r)e^{-\alpha r} \]  
\[ \alpha [\text{dB/cm}] = 2.17 \cdot 10^{-15} f^2 \]  

2.5 PMUT Models

2.5.1 Equivalent Circuits

The equivalent circuit model for the transducer aims to treat every element of the electrical, mechanical and acoustical domains as a single, lumped element of an electrical equivalent circuit. For this, every parameter must be converted to an equivalent resistor, inductor, capacitor or transformer \[1\]. Table 2.1 shows the analogies between elements and how they are converted to equivalent circuit components. Electrical elements do not need to be converted. The radiation resistance and reactance (mass) are the real and imaginary parts of the radiation impedance. Compliance is the reciprocal of the stiffness (bending equivalent to the spring constant).
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<table>
<thead>
<tr>
<th>Equivalent Circuit</th>
<th>Acoustic</th>
<th>Mechanical</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistance</td>
<td>Radiation Resistance ($R_r$)</td>
<td>Damping ($R_m$)</td>
</tr>
<tr>
<td>Inductance</td>
<td>Radiation Mass ($M_r$)</td>
<td>Mass ($M$)</td>
</tr>
<tr>
<td>Capacitance</td>
<td>-</td>
<td>Compliance ($C_m = 1/K_m$)</td>
</tr>
<tr>
<td>Voltage</td>
<td>Force (on an area) ($P$)</td>
<td>Force ($F$)</td>
</tr>
<tr>
<td>Current</td>
<td>Velocity ($U$)</td>
<td>Velocity ($U$)</td>
</tr>
</tbody>
</table>

Table 2.1: Equivalent circuit analog variables

By this definitions, it is possible to represent all the interactions in the transducer with the equivalent circuit of Figure 2.3. The turns ratio $N$ directly relates the voltage on the PMUT to the force exerted on the membrane and is related to the electromechanical transduction. If the mechanical damping and radiation resistance are added, the same is done for the mechanical and radiation reactance and the mechno-acoustical element values are multiplied by $N$, the simplified Van Dyke circuit of Figure 2.4 is obtained.

Solving the equivalent circuit is the analogue of finding the solution to the linear differential equations characterizing the behaviour of the electrical, acoustical and mechanical elements all together. By solving for the electrical impedance, equation (2.12) is obtained [1]:

\[
Y = j\omega C_0 + \frac{N^2}{(R + R_r) + j[\omega(M + M_r) - 1/\omega C_m^E]} \quad (2.12)
\]

The resonant frequency $\omega_r$ is a particular frequency value that maximizes the admittance, and thus the transducer response, and occurs when $\omega(M + M_r) - 1/\omega C_m^E = 0$:

\[
\omega_r = 1/\sqrt{(M + M_r)C_m^E} \quad (2.13)
\]

The antiresonant frequency is a minimum in the admittance and its value is:

\[
\omega_a = 1/\sqrt{(M + M_r)C_m^D} \quad (2.14)
\]

\[
C_m^D = C_m^E/(1 + N^2C_m^E/C_0) \quad (2.15)
\]

As can be seen, resonant and antiresonant frequencies have an explicit relation to the equivalent circuit parameters. Thus, measuring the resonant and antiresonant frequencies from experimental admittance frequency response curves can be used to determine the circuit component values to represent a transducer.
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Figure 2.3: PMUT equivalent circuit (from [1])

Figure 2.4: Van Dyke equivalent circuit (from [1])
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2.5.2 Matrix Models

As the transducer can be thought of as a linear system, at least for relatively small vibration amplitudes, another option to model its behaviour is to obtain linear coefficients relating the variables involved and collecting all the relevant equations in simple matrix identities. This reduces the complexity of design.

Equation (2.16) shows the relationship between the variables and their corresponding coefficients. Each pair of coefficients is obtained by fixing one of the input variables to zero while measuring the output with the other input variable being non zero.

\[
\begin{bmatrix}
I \\
U
\end{bmatrix} =
\begin{bmatrix}
Y_{11} & Y_{12} \\
Y_{21} & Y_{22}
\end{bmatrix}
\begin{bmatrix}
V \\
P
\end{bmatrix}
\]

\[Y_{11} = \left. \frac{I}{V} \right|_{P=0} \quad Y_{21} = \left. \frac{U}{V} \right|_{P=0} \quad Y_{12} = \left. \frac{I}{P} \right|_{V=0} \quad Y_{22} = \left. \frac{U}{P} \right|_{V=0}
\] (2.17)

2.6 Transducer Arrays

As PMUTs can be miniaturized, their arrangement in large arrays whose dimensions are still small is possible. Each transducer can contribute to a larger added acoustic output. Also, as each element can be individually actuated, control of timing and sequencing allows for beam forming and steering.

Similarly to the individual case, the dimensions of the array in relation to the wavelength of operation determine the directivity of the beam and the possibility of side lobes appearing. For the array case, the geometric limit in order to have an omni directional pattern is for the pitch (distance between elements) to be less than half a wavelength \[15\].

Equation (2.18) relates the number of elements in the array (M), the single element pressure and the array directivity function \(D_a(\theta)\):

\[p_a(r) = MD_a(\theta)p(r)\] (2.18)

\[D_a(\theta) = \frac{\sin[(Mks \sin \theta)/2]}{M \sin[(ks \sin \theta)/2]}\] (2.19)

Beam forming is a way of focusing the beam at a certain distance from the array by making the peaks of the emitted acoustic waves from each single element to coincide at a point and interfere constructively. This is done by delaying the electrical pulses to elements as they are further away.

\(^1\)Zero input pressure corresponds to emitter operation and zero input voltage corresponds to receiver operation
CHAPTER 2. BACKGROUND

Figure 2.5: Schematic illustration of beam forming and steering (from [15])

from the center of the array. Beam steering can also be done by activating consequent elements in cascade. The delay within activations determines the steering angle. These techniques can also be applied at the receiver by activating reception of individual PMUTs at specific times. Figure 2.5 illustrates these techniques.

2.7 Performance Parameters

2.7.1 Electromechanical Coupling Factor

The electromechanical coupling factor, as it name suggests, is a measure of conversion of energy from the electrical to the mechanical domain. It can also be interpreted as a factor representing the change in capacitance and stiffness of the transducer due to the piezoelectric actuation relating these mechanical and electrical variables. It can be defined from the difference among the clamped (no motion) capacitance \( C_0 \) and the free (transducer in operation) capacitance \( C_f \):

\[
k^2 = \frac{C_f - C_0}{C_f} \tag{2.20}
\]
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The coupling factor can also be determined from the resonant and antiresonant frequencies:

\[ k^2 = 1 - \left( \frac{\omega_r}{\omega_a} \right)^2 \]  
(2.21)

2.7.2 Quality Factor

The quality factor Q is related to the bandwidth of the transducer, that is how sharp the resonant frequency response is. It can be defined from the frequencies \( f_1 \) and \( f_2 \) where the response falls 3dB from the resonant value:

\[ Q = \frac{f_r}{f_2 - f_1} \]  
(2.22)

Another definition comes from equivalent circuit parameters:

\[ Q = \omega_r \frac{M + M_r}{R + R_r} \]  
(2.23)
Chapter 3

PMUT Array Design

The multi-domain nature of the working principles of the PMUTs require their design to tackle several, usually interrelated, perspectives. To limit the large amount of possibilities resulting from this, clear design requirements are first determined from both the available interface circuits and intra-body focus described in Chapter 1. Next, geometry and materials are defined from the fabrication process best suited to the needs and available resources. Comsol Multiphysics simulation is then performed to determine dimensions that achieve the desired center frequency for a single transducer.

A system-wide model consisting of emitter, medium and receiver elements is desirable for guaranteeing an acceptable output signal level. Thus, matrix and equivalent circuit models were obtained for the single transducer. Transmission in the medium for varying distance was also modeled. These components were linked together in a system-wide circuit, allowing to simulate changes in distances between transmitter and receiver, medium, source values and number of array elements.

Optimization was also performed for the source and load (impedance matching). Directivity of the array is further analyzed. The specific sequence followed is summarized below:

1. Design constraints
2. Single element frequency response
   (a) Set resonant frequency
3. Source impedance matching
4. Modeling
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(a) Single element matrix modeling
(b) Single element equivalent circuit modeling
(c) Medium transmission modeling

5. Load impedance matching

6. Directivity

7. System-wide simulation
   (a) Array factor
   (b) Overall system input and output check

3.1 Design Constraints

As was detailed in Chapter 1, the design goal is to achieve intra-body acoustic communication. From the previous work, interface circuitry for ultrasound transducers has already been designed working at 700kHz and 2MHz. Therefore, the first design constraint is to operate at both these frequencies.

A diagram of the connections to the interface circuit is shown in Figure 3.1. This network can be switched for transmitter or receiver operation using the same PMUT element. As a transmitter, an output pin of the FPGA acts as a voltage source for the transducer. A square wave is generated at the resonant frequency of the PMUT, but due to its electromechanical characteristics, it acts as a low pass filter in such a way that the source wave can be approximated as a sinusoid. Then, the source can be simulated as an AC source with a 50 ohm resistance. This is the second design constraint.

As will be further explained in Chapter 4, the PMUT structure is basically a layer of aluminum nitride in between patterned top and bottom electrodes. This stack lays on top of a silicon dioxide diaphragm. A cavity in the underlying silicon wafer is opened in order to release this membrane. The values chosen for this structure are shown in Table 3.1. This defined structure is the third design constraint. However, a choice remains on whether the cavity is left open to the medium (vented) or if another silicon wafer is vacuum bonded on the bottom to seal it. Both possibilities are discussed in Chapter 3.2.1.

The propagation of acoustic waves in the body medium also needs to be taken into account. Even though the organs, bones and fluids within the body have different densities and some of the
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Figure 3.1: Connections to FPGA Diagram

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Oxide Thickness (um)</td>
<td>0.8</td>
</tr>
<tr>
<td>Aluminum Nitride Thickness (um)</td>
<td>0.75</td>
</tr>
<tr>
<td>Electrode Thickness (um)</td>
<td>0.1</td>
</tr>
<tr>
<td>Top Electrode to cavity radius ratio</td>
<td>0.8</td>
</tr>
<tr>
<td>Top Electrode Material</td>
<td>Gold</td>
</tr>
<tr>
<td>Bottom Electrode Material</td>
<td>Platinum</td>
</tr>
</tbody>
</table>

Table 3.1: PMUT Parameters
structures are flexible while others are not, a first approximation is to design the transducers to operate in water, as the body is mostly composed of it. However, as was mentioned in Chapter 2.7, some of the performance parameters must be measured in air, so behavior in air must be considered as well. This, along with a reasonable average distance between transmitters and receivers within the body, constitutes a fourth design constraint.

Single transducers whose dimensions are smaller than the wavelength of the acoustic wave they are emitting can be considered to have an omni directional emission pattern. However, this changes when an array of such elements is composed. As the devices can displace or rotate along with motions of the body, if the directivity is too narrow, the transducers can slide out of the field of view and lose connection even though they were originally facing each other. Therefore, a minimum directivity must be set.

Depending on the necessary amount of data to be transmitted, a minimum bandwidth must be set. This constitutes the fifth design constraint.

Table 3.2 summarizes the requirements.

<table>
<thead>
<tr>
<th>Constraint</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Operating Frequency (MHz)</td>
<td>0.7, 2</td>
</tr>
<tr>
<td>Source Voltage (V)</td>
<td>3</td>
</tr>
<tr>
<td>Maximum Source Current (mA)</td>
<td>8</td>
</tr>
<tr>
<td>Source Resistance (Ω)</td>
<td>50</td>
</tr>
<tr>
<td>Load Resistance (GΩ)</td>
<td>1.82</td>
</tr>
<tr>
<td>Medium</td>
<td>Water, Air</td>
</tr>
<tr>
<td>Transmitter-Receiver Distance (m)</td>
<td>0.2</td>
</tr>
<tr>
<td>Directivity (degrees)</td>
<td>30</td>
</tr>
<tr>
<td>Chip Size</td>
<td>1 cm x 1 cm</td>
</tr>
<tr>
<td>Bandwidth (MHz)</td>
<td>1</td>
</tr>
</tbody>
</table>

Table 3.2: Design Constraints

3.2 700 kHz Resonant Frequency

3.2.1 Resonant Frequency

Even though equation (2.3) defines the relationship between the rigidity and geometry of a deflecting membrane with it’s resonant frequency, finite element simulation in Comsol Multiphysics
CHAPTER 3. PMUT ARRAY DESIGN

was used to verify the resonant frequency for the vibrating membrane. Figure 3.2 shows the results for both vented and sealed cavities.

Setting the corresponding radii for each of the cases, as is shown in Table 3.3, Comsol simulation was performed. The sealed back cavity advantages are clear in Figure 3.2, where the vented design under performs its sealed counterpart in both of the chosen metrics of performance (electric and acoustic). This is because in the vented case, the PMUT transfers acoustic energy to the medium on its backside while for the sealed case all of the energy is transferred to the front. Wafer bonding, however, adds to the complexity of the fabrication process. Therefore, for the first implementation of this design, a vented architecture will be chosen for the further design steps, leaving the sealing of the back cavity for further work.

Regarding the performance in air, the increased medium loading in water compared to air accounts for the lower frequency and larger bandwidth in fluid, as well as the higher far-field pressure (due to a lower attenuation coefficient). However, as was discussed in Chapter 2.4, behaviour in air is useful for determining transducer parameters. The narrower peaks for resonant and antiresonant frequencies are clearly seen in Figure 3.3.

![Figure 3.2: 700 kHz transducer vented and sealed cavity comparison](image)

![Table 3.3: Cavity Radii for 700kHz resonant frequency](table)

<table>
<thead>
<tr>
<th>Cavity Radius (um)</th>
<th>Vented</th>
<th>Sealed</th>
</tr>
</thead>
<tbody>
<tr>
<td>46</td>
<td>52.1</td>
<td></td>
</tr>
</tbody>
</table>

In the described FEM simulation, all the necessary parameters are accounted for in the material definitions from the software libraries but a mechanical damping in the structure was set to 1/100 as a reasonable value for MEMS transducers of this scale. This makes the simulations mostly accurate for resonant frequency and order of magnitude of output pressure, but precise prediction of quality factor and bandwidth requires experimental validation.
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3.2.2 Source Impedance Matching

Due to the PMUT basically being a capacitive structure, its resistance is relatively high. In comparison, the 50 ohm source impedance is almost negligible. No resistive matching can be done. However, the case is different for the reactive portion as the capacitance of the structure can well be compensated by adding a matching inductor.

An inductor placed in series at the same reactance value as the one corresponding to the PMUT electrical capacitance at the resonant frequency can cancel out the complex impedance (create an LC resonant circuit). To calculate the corresponding value, a parallel plate capacitor approximation is used on the section of the piezoelectric material that is covered by electrode (parasitic capacitances from interconnects are ignored):

\[
C = \epsilon_0 \epsilon_r \frac{A}{d} = \epsilon_0 \epsilon_r \frac{\pi r^2}{d}
\]  

The matching inductance would then be:

\[
\frac{1}{\omega C} = \omega L
\]
CHAPTER 3. PMUT ARRAY DESIGN

Figure 3.4: 700 kHz transducer source impedance matching

Figure 3.5: 700 kHz transducer frequency response with matching inductor

\[ L = \frac{1}{\omega^2 C} \]  \hspace{2cm} (3.2)

A parameter sweep performed in Comsol for the inductance in series with the PMUT verifies the admittance and far-field pressure boost at the calculated value as can be seen in Figure 3.4. A frequency sweep, shown in Figure 3.5, makes it clear that the resonant circuit amplifies both the admittance and far-field pressure. The cancellation of the electrical capacitance is evident in the admittance, where the anti resonant frequency effect is masked.

The agreement between the calculated and measured values for the matching inductor is evident in Table 3.4.

<table>
<thead>
<tr>
<th></th>
<th>Calculated</th>
<th>Simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Inductor</td>
<td>114.41 (mH)</td>
<td>103.03 (mH)</td>
</tr>
</tbody>
</table>

Table 3.4: 700 kHz transducer matching inductor value
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3.2.3 Modeling

3.2.3.1 Single Element Matrix Modeling

Following the procedure introduced in Chapter 2.5, the matrix model for a single transducer shown in Table 3.5 was obtained by these steps in Comsol simulation:

1. Apply 1 volt to the PMUT
   (a) Measure current ($Y_{11}$)
   (b) Measure normal displacement amplitude at the center of the membrane ($Y_{21}$)

2. Apply 1 Pa acoustic pressure to the membrane surface (with the source shorted out)
   (a) Measure current ($Y_{12}$)
   (b) Measure normal displacement amplitude at the center of the membrane ($Y_{22}$)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Y_{11}$</td>
<td>$2.26\mu S$</td>
</tr>
<tr>
<td>$Y_{12}$</td>
<td>$0.1nA/Pa$</td>
</tr>
<tr>
<td>$Y_{21}$</td>
<td>$109nm/V$</td>
</tr>
<tr>
<td>$Y_{22}$</td>
<td>$23.7pm/Pa$</td>
</tr>
</tbody>
</table>

Table 3.5: 700 kHz transducer matrix model parameters

3.2.3.2 Medium Transmission Modeling

By implementing equations (2.5), (2.7) and (2.10) with the specific parameters shown in Table 3.6, Figure 3.6 was obtained, which shows both the effects of geometric spreading and attenuation in the medium. $P_0$, the pressure on the membrane face, is obtained from the output of the emitter simulation.

3.2.4 Load Impedance Matching

For load impedance matching, due to the maximum power transfer theorem, a load resistance value equaling the one of the source allows for optimal power consumption at the load. A

\footnote{It is important to note that, even though the matrix parameters in Table 3.5 show the magnitude of the obtained values, when these values are later used for the system-wide simulation their real and imaginary components are actually used, to be able to simulate effects such as reactive impedance matching.}
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<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\rho$</td>
<td>1000 kg/m$^3$</td>
</tr>
<tr>
<td>$c$</td>
<td>1484 m/s</td>
</tr>
<tr>
<td>$r$</td>
<td>46 $\mu$m</td>
</tr>
<tr>
<td>$\alpha$</td>
<td>1.1283 $\cdot$ 10$^{-4}$ m$^{-1}$</td>
</tr>
</tbody>
</table>

Table 3.6: 700 kHz transducer medium parameters

Comsol simulation of the receiver PMUT, with the predicted acoustic pressure at a 20 cm distance between transmitter and receiver and a varying load resistance, verifies this behaviour (Figure 3.7). The 450 $\Omega$ optimal value corresponds to the sum of $R$ and $R_r$ in the equivalent circuit of Figure 2.3, as the source is acoustic in this case.

However, the load impedance is fixed by the input resistance of the amplifier circuit, so this scheme is not feasible. A transformer designed to match the impedance difference could be an option but, as we are only interested in a detectable voltage for the Analog to Digital Converter (ADC) and not so much on the actual power transferred, this is not necessary.

A sweep of the value of an inductor placed in parallel to the PMUT to cancel out its electrical capacitance causes a negligible increase in the output due to the very high load impedance. This is why a matching inductor is not required at the receiver.

3.2.5 Bandwidth

Figure 3.8 shows the output pressure frequency response where a -6dB bandwidth lower than 100 kHz is measured. This can be attributed to the low operating frequency of the device. A
possible solution for this issue is to have elements of varying radius within the array, in order to produce an output with a wider combined spectrum.

### 3.2.6 Directivity

Once the single element response has been optimized, the effect of placing an array of these elements needs to be considered. The first evident effect is that the output pressure of each single PMUT adds up to the total array output pressure. However, as there can be interference effects and a minimum directivity constraint has been placed, the specific pitch of the array elements needs to be considered.

Similarly to the half-wavelength requirement for element size, the same condition needs to be met on the array pitch (distance between elements) to avoid sidelobes of the beam and to be able to perform beam steering (Chapter 2.4). By use of the equation $\lambda = \frac{c}{f}$, the resonant frequency and speed of sound in water, the wavelength is determined to be $2.12\text{mm}$. The actual pitch that can be
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Figure 3.9: Array directivity

achieved depends on fabrication but a conservative 100\(\mu m\) pitch would correspond to approximately 0.1\(\lambda\).

Matlab’s Sensor Array System Toolbox v3.3 can predict the directivity pattern of an array with specified geometry, number of elements, operating frequency, speed of sound and pitch. Figure 3.9 shows the 2D directivity pattern for 5x5 and 10x10 arrays with 0.1\(\lambda\) and the maximum 0.5\(\lambda\) pitch for comparison purposes. Using a -6dBi cutoff, it is clear that both array sizes comfortably achieve the beam-width requirement for the smaller pitch value with no sidelobes. It is therefore desirable to place more array elements as close as fabrication limits permit, while not exceeding the maximum supply current and chip size requirements.

Figure 3.10 shows the same patterns in a more intuitive, three-dimensional distribution. The effect of a wider pitch in creating side lobes becomes evident here.
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The PMUT matrix model, medium model, and array factor are integrated in the system-wide circuit summarized in Figure 3.11. The specific implementation was done in Agilent’s Advanced Design System (ADS) software and can be seen in Figure 3.12.

The array effect is integrated here as a simple gain factor that accounts for both the emitter and receiver contributions. Thus \( N \) is basically the product of the number of emitter and receiver elements. This can be done without further consideration as more transducers on the emitter side add up to the total acoustic output while on the receiver side, each adds up to the produced current. It must also be emphasized that the circuit predicts the response on the center axis of the transducer (assumes transmitter and receiver are perfectly aligned), the directivity functions described in the previous section predict the reduction due to misalignment.

Figure 3.13 shows a linear increment of the output short-circuit current as the total number of elements increases. The limit to placing as many elements as possible is their current consumption. As Table 3.7 summarizes, the 5x5 array is the largest the FPGA can source. By having a 5x5 array at the receiver as well, an easily detectable potential of 1.4 V can be ideally produced on the receiver circuit.

3.2.7 System-wide Model

Figure 3.10: Three-dimensional directivity plots for PMUT arrays

5x5 Rectangular Array  10x10 Rectangular Array

0.1\( \lambda \) pitch

0.5\( \lambda \) pitch

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Figure 3.11: System-wide components and variables

Figure 3.12: System-wide ADS simulation circuit

Figure 3.13: 700kHz transducer receiver short-circuit current vs number of array elements

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Array Size</td>
<td>5x5</td>
</tr>
<tr>
<td>Maximum Source Current</td>
<td>8mA</td>
</tr>
<tr>
<td>Single PMUT Current Consumption</td>
<td>295µA</td>
</tr>
<tr>
<td>Total Current Consumption</td>
<td>7.38mA</td>
</tr>
<tr>
<td>Single PMUT Voltage Output</td>
<td>2.34mV</td>
</tr>
<tr>
<td>Total Voltage Output</td>
<td>1.46V</td>
</tr>
</tbody>
</table>

Table 3.7: 700 kHz transducer system-wide performance summary
CHAPTER 3. PMUT ARRAY DESIGN

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Array Size</td>
<td>5x5</td>
</tr>
<tr>
<td>Maximum Source Current</td>
<td>8mA</td>
</tr>
<tr>
<td>Single PMUT Current Consumption</td>
<td>6.63μA</td>
</tr>
<tr>
<td>Total Current Consumption</td>
<td>0.166mA</td>
</tr>
<tr>
<td>Single PMUT Voltage Output</td>
<td>46.8μV</td>
</tr>
<tr>
<td>Total Voltage Output</td>
<td>29.3mV</td>
</tr>
</tbody>
</table>

Table 3.8: 700 kHz transducer system-wide performance summary. No source impedance matching

3.3 2 MHz Resonant Frequency

The results of applying the same procedure for a 2MHz resonant frequency are shown.

3.3.1 Resonant Frequency

Table 3.9 shows the radius obtained for a 2MHz resonant frequency while Figure 3.14 shows the response of the transducer in water.

<table>
<thead>
<tr>
<th>Radius (μm)</th>
<th>29.9</th>
</tr>
</thead>
</table>

Table 3.9: Cavity radius for 2MHz resonant frequency

![Figure 3.14: 2MHz transducer frequency response in water](image)

3.3.2 Source Impedance Matching

Figure 3.15 illustrates a frequency sweep of the matching inductor, Figure 3.16 shows the element frequency response with the optimum value for the inductor implemented and Table 3.10 shows a reasonable match between analytic and simulated obtained values for the necessary inductance.
CHAPTER 3. PMUT ARRAY DESIGN

3.3.3 Modeling

3.3.3.1 Single Element Matrix Modeling

Table 3.11 summarizes the matrix model parameters for the transducer.

3.3.3.2 Medium Transmission Modeling

The pressure attenuation and spreading in the medium at a certain distance from the emitter is shown in Figure 3.17.

3.3.4 Load Impedance Matching

Similarly to the 700 kHz case, no benefit is obtained from placing an inductor with the aim to cancel the capacitance of the transducer.
CHAPTER 3. PMUT ARRAY DESIGN

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calculated</td>
<td>21.22 (mH)</td>
</tr>
<tr>
<td>Simulation</td>
<td>30.3 (mH)</td>
</tr>
</tbody>
</table>

Table 3.10: 2MHz transducer matching inductance value

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Y_{11}$</td>
<td>2.69 µS</td>
</tr>
<tr>
<td>$Y_{12}$</td>
<td>0.0325 nA/Pa</td>
</tr>
<tr>
<td>$Y_{21}$</td>
<td>2.67 nm/V</td>
</tr>
<tr>
<td>$Y_{22}$</td>
<td>2.494 pm/Pa</td>
</tr>
</tbody>
</table>

Table 3.11: 2MHz transducer matrix model parameters

3.3.5 Bandwidth

A higher bandwidth (around 250kHz) was achieved in comparison to the 700 kHz transducer due to the higher operating frequency as can be seen in Figure 3.18.

3.3.6 Directivity

By use of the equation $\lambda = c/f$, the resonant frequency and speed of sound in water, the wavelength is determined to be 0.742 mm. In this case, the $0.1\lambda$ pitch would be 74.2 µm, which is still possible to manufacture. The directivity plots are the same as in Figure 3.9 and Figure 3.10.

3.3.7 System-wide Model

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Array Size</td>
<td>5x5</td>
</tr>
<tr>
<td>Maximum Source Current</td>
<td>8 mA</td>
</tr>
<tr>
<td>Single PMUT Current Consumption</td>
<td>305 µA</td>
</tr>
<tr>
<td>Total Current Consumption</td>
<td>7.625 mA</td>
</tr>
<tr>
<td>Single PMUT Voltage Output</td>
<td>0.413 mV</td>
</tr>
<tr>
<td>Total Voltage Output</td>
<td>258 mV</td>
</tr>
</tbody>
</table>

Table 3.12: 2MHz transducer system-wide performance summary
CHAPTER 3. PMUT ARRAY DESIGN

Figure 3.17: 2MHz transducer single element output acoustic pressure vs distance

Figure 3.18: 2 MHz transducer frequency response

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Array Size</td>
<td>5x5</td>
</tr>
<tr>
<td>Maximum Source Current</td>
<td>8mA</td>
</tr>
<tr>
<td>Single PMUT Current Consumption</td>
<td>8.06µA</td>
</tr>
<tr>
<td>Total Current Consumption</td>
<td>0.202mA</td>
</tr>
<tr>
<td>Single PMUT Voltage Output</td>
<td>10.8µV</td>
</tr>
<tr>
<td>Total Voltage Output</td>
<td>6.75mV</td>
</tr>
</tbody>
</table>

Table 3.13: 2MHz transducer system-wide performance summary. No source impedance matching.
Chapter 4

Fabrication

From the literature review, the piezoelectric-electrode stack is either built by metal deposition and sol-gel or sputtering for the Aluminum Nitride (AlN) layer. A variety of dimensions and thicknesses have been reported for this. The passive layer is made out of silicon, Silicon Dioxide (SiO2) or Silicon Nitride (SiN) in some cases [14], [2], [11].

The release of the device (cavity formation) also presents several options such as anisotropic wet Potassium Hydroxide (KOH) silicon etching [14], front side Xenon Difluoride (XeF2) etching [14], sacrificial polysilicon etching and parylene cavity sealing [2] or use of cavity SOI wafers previously manufactured [8].

The manufacturing options more adequate for the current possibilities of our research facilities are the cavity SOI wafers, front side XeF2 etching and backside Deep Reactive Ion Etching (DRIE). Figure 4.1 shows the corresponding topologies while Table 4.1 summarizes their relative advantages and drawbacks.

Due to the good balance between cost, flexibility in manufacturing at our own facilities and precise cavity definition (to which the resonant frequency is very sensitive), the chosen process is the backside DRIE. The main drawback is the vented cavity, which was shown in the design to be detrimental for the output pressure. However, it avoids the parylene sealing which introduces additional loading to the membrane in the XeF2 process. Vacuum bonding of the manufactured dice to a bare silicon handle wafer is a possibility for future work in order to obtain the sealed cavity benefit from the cavity SOI process while maintaining the freedom to quickly test different configurations locally.
CHAPTER 4. FABRICATION

Figure 4.1: PMUT main fabrication options. Top: cavity SOI, center: backside DRIE, bottom XeF2 etch.

<table>
<thead>
<tr>
<th>Cavity SOI</th>
<th>DRIE</th>
<th>XeF2 etching</th>
</tr>
</thead>
<tbody>
<tr>
<td>Commercially available</td>
<td>Requires manufacturing</td>
<td>Requires manufacturing</td>
</tr>
<tr>
<td>Anisotropic etch</td>
<td>Anisotropic etch</td>
<td>Isotropic, rough etch</td>
</tr>
<tr>
<td>Precise radius</td>
<td>Precise radius</td>
<td>Variable radius</td>
</tr>
<tr>
<td>Back-side alignment needed</td>
<td>Back-side alignment needed</td>
<td>Front side process</td>
</tr>
<tr>
<td>Higher cost</td>
<td>Intermediate cost</td>
<td>Low cost</td>
</tr>
<tr>
<td>Vacuum sealed back</td>
<td>Vented back</td>
<td>Needs Parylene sealing</td>
</tr>
</tbody>
</table>

Table 4.1: Main fabrication options
CHAPTER 4. FABRICATION

4.1 Process

Table 4.2 enumerates the fabrication steps required for the chosen process.

<table>
<thead>
<tr>
<th>No.</th>
<th>Step</th>
<th>Mask</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Grow 800 nm of thermal SiO2 on DSP wafer</td>
<td>1</td>
</tr>
<tr>
<td>2</td>
<td>Bottom electrode photo lithography</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>5nm titanium adhesion layer e-beam evaporation</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>95nm platinum bottom electrode e-beam evaporation</td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>Metal Liftoff</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>750nm AlN sputtering</td>
<td>2</td>
</tr>
<tr>
<td>7</td>
<td>Via photolithography</td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>Bottom electrode phosphoric acid via etching</td>
<td>3</td>
</tr>
<tr>
<td>9</td>
<td>Top electrode photolithography</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>100nm gold top electrode e-beam evaporation</td>
<td></td>
</tr>
<tr>
<td>11</td>
<td>Metal liftoff</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>Wafer Dicing</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>Backside alignment and photo lithography</td>
<td>4</td>
</tr>
<tr>
<td>14</td>
<td>Backside cavity DRIE</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>Wire bonding</td>
<td></td>
</tr>
</tbody>
</table>

Table 4.2: Fabrication process flow

Figure 4.2 details some of the fabrication steps for the piezoelectric electrode stack.

4.2 Layout

The main considerations for layout design were:

1. Cavity Radii
   
   (a) Place arrays of several cavity radii around the designed value to account for fabrication tolerance
   
   (b) Half-wavelength diameter requirement

2. Array Pitch
   
   (a) Half-wavelength pitch requirement

3. Connector and via width allowing low resistance

4. Parasitic capacitance
CHAPTER 4. FABRICATION

Figure 4.2: Optical images of fabrication steps. Left: Deposited bottom electrode and AlN, photolithography for via etch done. Right: Etched vias on AlN

(a) Pattern top and bottom electrodes to avoid overlap creating parasitic capacitance

(b) Set horizontal connectors on bottom electrode and vertical on top to avoid overlap

5. Elements individually addressable for beamforming

6. GSG probe connectors

(a) Individual PMUTs with Ground Signal Ground (GSG) probe pads of different radii available (performance measurement in air)

(b) Arrays connected to GSG probe pads for simultaneous actuation of whole array without wire bonding

7. Arrays of different element numbers available for variability in output pressure prediction

Figure 4.3: Layout components. Left: addressable 5x5 array. Right: single element with GSG probe
Chapter 5

Conclusion

The first step completed in the present work was a clear definition of the requirements of PMUT arrays for intrabody networking applications. Their interface to existing circuitry from previous work and the characteristics of the medium in which they will be performing were addressed. Through FEM simulations, analytic models for the medium, and equivalent models for the transducer, an integral and interrelated design was performed for both the 700 kHz and 2MHz applications. The result is a first design iteration of PMUT arrays for relatively low operating frequencies in comparison to state-of-the-art devices. Additionally, a design framework allowing for adapting the design to changing requirements or correction for differences between the predicted behaviour and the experimental measurements was built.

As FEM simulations formed most of the foundations for prediction of transducer behaviour, their accuracy must be assessed. Resonant frequency, and the order of magnitude of the output pressure obtained can be assumed to be reliable and are also reasonable compared against similar transducers such as [8] and [2]. However, predictions of bandwidth, and of the related quality factor $Q$, are very much dependent on fabrication and actual obtained material properties. This is why, in this regard, experimental measurements are essential for precise assessment of performance of the transducers.

The ADS system-wide model is a valuable output of this work. It is an integral tool for assessing the performance of the system as a whole. Currently, its parameters are closely related to FEM simulation outputs, but parameters from experimental measurements can be easily extracted to increase its accuracy. It allows for impedance matching, variations in number of array elements, and changes in medium and distance between emitter and receiver arrays.

Directivity requirements were easily achieved due to the small element sizes and small
CHAPTER 5. CONCLUSION

array pitch chosen. The wide beam pattern provides a safe margin for possible misalignments due to motion within the body. Bandwidth requirements, in contrast, pose a more difficult challenge due to the relatively low frequencies of operation. Integration of elements of different radii within the array is a possible solution and, once more, experimental validation is required for the predictions.

Due to the possibilities of discrepancy between predicted values and implementation, variations of geometry were implemented in the layout including cavity radius, number of array elements and pitch. The designed fabrication process has been currently successful up to the AlN via etch and is in progress.

Impedance matching was shown to have a considerable impact on the output pressure of an element and thus on the voltage output on the receiver. However, the required inductance for this matching poses a limit due to the size of the corresponding element. For initial bench testing, a sizable inductor could be used, but for the intrabody application this might not be feasible.

A worst case scenario was included in the development within this work where no matching is performed. Conserving the array dimensions of 5x5, a detectable signal was still obtained at the receiver for both the 700kHz and 2MHz cases. Even though the detected potential is of a much lower magnitude, making it more sensitive to noise corruption, the current consumption of each transmitter element is much lower. Larger arrays, that have been included in the current layout can be used to compensate for this fact creating a more robust signal at the receiver.

The equivalent circuit representation of the PMUTs was used as a method of explaining the behaviour of the transducers in terms of frequency response and relation between the involved domains. However, extraction of electromechanical coupling and quality factors, maximum admittance and low frequency reactance from experimental response data can be used to explicitly determine the equivalent circuit variables. Once these are known, variations in geometery (thickness of the membrane stack and radius) can be performed to optimize acoustic matching to the medium, thus maximizing output and increasing the bandwidth.
Bibliography


BIBLIOGRAPHY


