Design of a Flexible Ultrasound Phased Array with Adaptive Phasing for Curvature

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Abstract

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Diagnostic ultrasound has become invaluable to healthcare professions for the purpose of imaging soft tissue without the risk of exposure to damaging, ionizing radiation. However, the majority of commercially available transducers have rigid, fixed interfaces that cannot conform to the surface of the human body. Such limitations both introduce a potential air gap (requiring the application of ultrasound gel) and make long-term monitoring impractical. In this work, I propose a novel flexible 2D ultrasound phased array with adaptive phasing that is capable of compensating for the radius of curvature. I describe the phasing algorithm and illustrate the detrimental effect of a lack of phase correction through simulation. I conduct phase detection by using time of arrival (TOA) without additional external hardware. In addition to simulations, I provide details of the fabrication process of a flexible 16 by 16 element array. The manufactured array, with an operating frequency of 1.4MHz and bandwidth of 41.3%, was capable of generating pressures up to 600 kPa. Finally, I conduct an in-vivo human study to demonstrate the functionality of the array on a human humerus. Although visible without phase correction, the location of the bone can easily be tracked in real-time after applying the correction algorithm.
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1 Introduction

Diagnostic ultrasound technology has provided a means to safely and accurately image the inside of the human body. Advances have been made in transducer technology to improve the signal-to-noise ratio (SNR), as well as in imaging modalities, to allow for the collection of new information from medical ultrasound scans. However, human subjects do not have perfectly flat or uniform interfaces on the skin, which makes for imperfect coupling to the measurement device. Ultrasound gel has been used as a means of removing interfacial air pockets, but the ideal solution would be for the probe to conform to the patient. Flexible ultrasound arrays already exist, but the phasing method for arrays is still performed in the classical sense, introducing errors. In this work, I describe the development of a two-dimensional (2D) flexible ultrasound phased array with an adaptive phasing algorithm. I demonstrate the design and fabrication procedures for the array, benchmark its performance on gelatin phantoms, and present in-vivo data from human organs. In developing the fabrication techniques for the array, laser micromachining of the piezoelectric polymer polyvinylidene fluoride (PVDF) is also explored. This material is difficult to mechanically manipulate into defined features once it is already in a poled piezoelectric state. Patterning PVDF with an excimer laser allows for rapid and precise feature development with minimal debris, translating readily to ultrasound transducer fabrication.
1.1 Flexible Arrays

While traditional rigid phased arrays are regularly utilized in clinical settings, flexible arrays have started to emerge in both research and industry. Most commercially available transducers are designed for non-destructive testing (NDT) of inanimate objects [1, 2]. For NDT, these one-dimensional (1D) ultrasound probes are wrapped around the object under test (e.g. a pipe, nozzle, etc.), and used in traditional imaging modes to check for cracks or defects. Current commercial probes typically operate in the 1-10 MHz range, with between 8-128 elements. For example, the Olympus probe [1] is a 5 MHz device with 64 transducer elements with 1 mm element pitch. As will be discussed in Chapter 3, such a wide pitch at 5 MHz in water, or when interfacing biological tissue, will lead to grating lobes, degrading the imaging performance.

In addition to what is available commercially, several researchers have explored the concept of creating flexible or stretchable ultrasound arrays. These generally involve the direct integration of traditional ultrasound transducer elements to a flexible circuit [3]. These arrays use a variety of types of transducers; however, they do not explicitly compensate for the resulting curvature of the array in the phase calculations. The fabrication of these devices differs from source to source depending on the desired size and characteristics. For example, McGough et al. use a custom silicone mold to individually mount and place pre-cut lead zirconate titanate (PZT) transducer elements that are 7 mm by 7 mm by 2.8 mm each, to create a 96-element 2D array [4]. While functioning well at 617 kHz, manually manipulating and placing transducers that are smaller than 1 mm\(^2\) for common MHz ultrasound ranges becomes impractical with this method. More commonly, PZT or piezocomposites are directly integrated onto a polymer backing substrate with electrodes for the signal and ground connections [5, 6, 7]. Zhu et al. recently introduced a stretchable
patch based on 100 separate 1-3 piezocomposite elements, arranged in a 10 by 10 array on a stretchable polydimethylsiloxane (PDMS) substrate [7]. Although this was not a traditional B-mode imaging device using a phased array, they successfully demonstrated a flexible patch using a synthetic aperture focusing (SAF) technique to find defects in an aluminum sample. Pashaei et al. developed a flexible 1D PZT array of 64 elements for imaging nerves in the human body [6]. While the effect of the curved array on the focus was described in simulation, no mention is made to corrections in phasing to compensate for the bending radius.

1.2 Potential Applications

A flexible ultrasound phased arrays offers a plethora of applications, particularly in the field of biomedical imaging. Biological tissue rarely conforms perfectly to rigid surfaces, and as such the air gap that forms in-between the two interfaces can lead to reflections. Potential applications for an array which removes these reflective pockets include common medical needs such as the confirmation of proper endotracheal tube insertion as well as long-term monitoring of intensive care unit (ICU) patients. Diagnostic ultrasound is routinely used in ICUs for electrocardiography, abdominal, and even remote, tele-mentored ultrasound imaging [8]. Tele-mentored ultrasound involves an inexperienced trainee applying the probe to the patient, while a remote technician interprets the data to provide feedback. By having a fixed patch that is able to provide three-dimensional (3D) scans, a few trained medical professionals can monitor multiple patients simultaneously, freeing valuable personnel resources and increasing ICU monitoring capacity.

One of the most commonly performed procedures in an ICU is tracheal intubation [9]. This is a procedure by which a tube is inserted orally, through the trachea, and into the
lungs to provide air to the patient in an emergency. In a study of 3,423 cases by [10], 10.3% reported difficulty during the intubation, with up to 4.2% causing further complications. In another study, 25% of 253 intubation required multiple attempts [11]. The insertion is often performed by junior staff in the ICU, increasing the probability of error [9]. Out of 3,423 intubations that are not in an operating room (OR) 60% were in an ICU, showing the majority occur where longer term care is required [9].

Figure 1.1: Example placement of a flexible ultrasound patch to assist with endotracheal tube insertion.
One possible solution is using a flexible ultrasound phased array patch, as shown in Fig. 1.1. 2D phased arrays possess the additional advantage of steering the generated ultrasound beam in three dimensions at arbitrary focal lengths, which can be used for detailed imaging and focused ultrasound (FUS) therapy to accurately ablate tumours with a narrow beam [12, 13, 14]. Potential applications of these conformable versions of the transducer arrays include common medical needs such as the confirmation of proper endotracheal tube insertion, as well as long-term monitoring of intensive care unit (ICU) patients. During tracheal intubation [9], a tube is inserted through the trachea, and into the lungs to provide air to the patient in an emergency. In a study of 3,423 cases [10], 10.3% of personnel reported difficulty during the intubation, with up to 4.2% cases causing further complications. In another report, it was revealed that 25% of 253 intubations required multiple attempts [11]. Out of 3,423 intubations that were not performed in an operating room, 60% happened in an ICU – showing that the majority occur where long term care is required [10]. Ultrasound is already used as a secondary methodology for confirming that the tube is correctly placed; however, this is performed with traditional rigid probes [15, 16, 17, 18]. Depending on the anatomy of the patient, the individual tube diameter for insertion is typically 7-9 mm for males, and 6.5-8.5 mm for females [19]. As the patch must, at the smallest extent, stretch to conform to a cylinder of this size, this requires the radius of curvature for a potential imaging array to be 1.5 cm. Flexible ultrasound phased arrays, as described in this thesis, may provide a convenient solution for real-time ultrasound imaging-guided tracheostomies, reducing all of these complications.

While X-ray imaging has become the standard for identifying bone fractures, ultrasonic scanning has been examined as offering a rapid alternative. The non-ionizing nature of ultrasound is particularly important when examining fractures in minors, where unnecessary exposure to X-rays is dangerous. In a study of 224 suspected fractures in children, 86.6%
of the cases were correctly identified using ultrasound [20]. Ultrasound provided the highest sensitivity for identifying fractures in long bones, specifically the humerus, with a sensitivity as high as 98% across 233 individuals [21]. The diameter of an adult humerus varies along the length of the humeral shaft from 17mm-23mm for males and 14mm-20mm for females [22, 23], easily detectable by low ultrasonic frequencies. Flexible ultrasound phased arrays, as described in this paper, may provide a convenient solution for safe, rapid, and convenient detection of fractures.
2 Literature Review

Ultrasound has seen wide applications in biomedical instrumentation, material processing, and imaging. In this section, I provide a brief overview of how ultrasound transducers operate. While this is not intended to be a comprehensive overview of ultrasound physics, the basic operating principles and applications are described to provide context to the design decisions used when making the flexible ultrasound phased array.

2.1 Brief Review of Ultrasound Physics

Ultrasonics have a variety of applications due to the physical effects that rapid acoustic vibrations have on materials. These include: localized heating, mechanical mixing, welding components together by causing electrolytic corrosion, forced diffusion, and many more [24, 25]. Depending on the energy levels, these can vary from vaporizing biological tissue, to cleaning devices (rapidly vibrating samples in a solvent to loosen contaminants), to non-destructive imaging. To understand ultrasound physics, it is important to first describe how ultrasound travels through space. Acoustic energy travels as a wave by applying a localized compression of matter that then exerts a force on neighboring particles. The localized pressure at a given point is often given as a function of time, \( p(t, z) \), whereas the velocity of the wave is given by \( u(t, z) \). It is important to note that there are two general methods
of travel for acoustic waves: shear and longitudinal. Shear waves, also known as transverse waves have a wave propagation that is orthogonal to the motion of particles in the host medium. Conversely longitudinal, or congressional waves, travel in the same direction as the particle motion [26, 27]. Imaging modalities primarily focus on longitudinal waves.

The propagation of an ultrasonic field through space and time is described by the wave equations. Full derivations of the wave equations can be found in [26, 27] among numerous other texts. Ignoring the shear waves and focusing on only the \( z \) dimension of travel, one can consider a unit cell of cross-sectional area, \( A \), over which the applied force is \( F = \Delta pA = (p(z) - p(z + \Delta z))A \). Simultaneously, the force can also be described by Newton’s second law:

\[
F = m \left( \frac{\partial u}{\partial t} + u \frac{\partial u}{\partial z} \right) \tag{2.1}
\]

Here, \( m \) is the mass of the medium in question, which is assumed to be homogeneous in this example. Setting these equal to each other, and taking any infinitesimally small unit cell of area, gives us an expression for the movement of pressure through space:

\[
- \frac{\partial p}{\partial z} = \rho \left( \frac{\partial u}{\partial t} + u \frac{\partial u}{\partial z} \right) \tag{2.2}
\]

Assuming the small fluctuations in the density of the propagation medium, \( \rho_1 \), are much smaller than the average density, \( \rho_0 \), we can rewrite this as:

\[
\frac{\partial p}{\partial z} + \rho_0 \left( \frac{\partial u}{\partial t} \right) = 0 \tag{2.3}
\]

We can see that Eq. 2.3 gives us one expression for the pressure with respect to time, distance, space, and wave velocity. However, the velocity is not always known, thus it is beneficial to decouple it with another equation. Defining the compressibility constant as
\[ K = \frac{\rho_0}{p\rho_0} \], we can use the same process to describe the conservation of mass entering and leaving a given space as:

\[
\frac{\partial (\rho u)}{\partial z} + \frac{\partial \rho}{\partial z} = 0 \quad (2.4)
\]

\[
\frac{1}{K} \cdot \frac{\partial u}{\partial z} + \frac{\partial p}{\partial t} = 0 \quad (2.5)
\]

After taking the partial derivative with respect to time, and combining Eqs. 2.3 and 2.5, we arrive at the wave equation describing the pressure in a single dimension with respect to time:

\[
\frac{\partial^2 p}{\partial z^2} - \rho_0 K \left( \frac{\partial^2 p}{\partial t^2} \right) = 0 \quad (2.6)
\]

The wave equation can be expanded to multiple dimensions; however, it generally must be solved numerically in simulation. To generate such solutions, the boundary conditions must be appropriately defined. This includes pressure sources, radiation boundaries at the edge of the domain, and impedance boundaries when the wave propagates through a surface [28]. Acoustic impedance is a material property that is defined as the ratio between the pressure and wave velocity, and can thus also be described by the longitudinal speed of sound in a medium, \( c \) [26]

\[
Z = \frac{p}{u} = \rho_0 c \quad (2.7)
\]

As with electromagnetic waves, Snell’s law applies when a wave strikes a boundary between media of differing refractive indices. When a plane wave crosses the boundary, part of the power is reflected back and not transmitted to the other medium. Here, the ratio of acoustic impedances determines the reflection coefficient, \( R \), as opposed to the ratio of the relative permittivities. If a plane wave intersects normal to a material boundary, the reflection coefficient is given by Eq. 2.8 [26].

\[
R = \frac{Z_2 - Z_1}{Z_2 + Z_1} \quad (2.8)
\]
This reflection coefficient becomes an important design consideration when discussing the power transmitted by and received from a piezoelectric transducer. As explained above, this coefficient is solely dependent on the material properties of the surrounding media. The interactions between an ultrasound transducer and surrounding media are often complex due to varying angles at interfaces, differing boundary conditions, higher order effects, etc. To this end, accurate predictions require full wave solvers to model fluid dynamics and the piezoelectric material interaction in space, using software capable of solving the wave equations in space, time, and frequency. This can be performed using techniques such as the finite element method (FEM) [29], the Tupholme-Stepanishen impulse response method [30, 31], the fast near-field method (FNM) [32], and the angular spectrum algorithm (ASA) [33].

2.2 Piezoelectric Materials

The foundation of an ultrasonic transducer is the ability to generate a pressure wave from a controlled electrical impulse. To accomplish this, the transducer element can be constructed from a piezoelectric crystal that mechanically deforms according to an applied electric field. An example of a piezoelectric crystal generating charge on opposite faces upon compression is shown in Fig. 2.1.

The field quantities in a piezoelectric transducer, in terms of displacement and stress, are often expanded in matrix quantities, as demonstrated in Eq. 2.9 [34, 35], where the subscripts indicate the dimensions. Here, $D_n$ is the electric displacement field, which is a result of the contribution from the existing electric field, $E_n$, combined with the product of the piezoelectric coefficient, $d_{nm}$, and the stress tensor, $\sigma_n$. Traditionally, in a non-piezoelectric, the displacement field is given only as $D = \epsilon_0 \varepsilon_r \vec{E} + \vec{P}_p$, where $\varepsilon_r$ is the relative permittivity.
Figure 2.1: Conceptual illustration of the compression of a piezoelectric crystal in the thickness mode and the corresponding charge generation

of the material, and $\varepsilon_0$ is the permittivity of free space. The $\tilde{P}_p$ term represents the electric displacement field due to residual dipole moments within the material.

$$\begin{bmatrix} D_1 \\ D_2 \\ D_3 \end{bmatrix} = \varepsilon_0 \begin{bmatrix} \varepsilon_{11} & \varepsilon_{12} & \varepsilon_{13} \\ \varepsilon_{21} & \varepsilon_{22} & \varepsilon_{23} \\ \varepsilon_{31} & \varepsilon_{32} & \varepsilon_{33} \end{bmatrix} \begin{bmatrix} E_1 \\ E_2 \\ E_3 \end{bmatrix} + \begin{bmatrix} d_{11} & d_{12} & d_{13} & d_{14} & d_{15} & d_{16} \\ d_{21} & d_{22} & d_{23} & d_{24} & d_{25} & d_{26} \\ d_{31} & d_{32} & d_{33} & d_{34} & d_{35} & d_{36} \end{bmatrix} \begin{bmatrix} \sigma_1 \\ \sigma_2 \\ \sigma_3 \\ \sigma_4 \\ \sigma_5 \\ \sigma_6 \end{bmatrix}$$ (2.9)

While applying stress to a crystal can generate a corresponding field, the converse is also true. One can describe the relationship through the resulting strain, as in Eq. 2.10. Here, the strain, $\delta_n$, in a particular dimension is the sum of effects from the elastic compliance
matrix, \( S \), and the stress tensor, \( \sigma \), combined with the product of the piezoelectric constant, \( d \), and the electric field, \( \vec{E} \) [35], [36].

\[
\begin{bmatrix}
\delta_1 \\
\delta_2 \\
\delta_3 \\
\delta_4 \\
\delta_5 \\
\delta_6 \\
\end{bmatrix}
= 
\begin{bmatrix}
S_{11} & S_{12} & S_{13} & S_{14} & S_{15} & S_{16} \\
S_{21} & S_{22} & S_{23} & S_{24} & S_{25} & S_{26} \\
S_{31} & S_{32} & S_{33} & S_{34} & S_{35} & S_{36} \\
S_{41} & S_{42} & S_{43} & S_{44} & S_{45} & S_{46} \\
S_{51} & S_{52} & S_{53} & S_{54} & S_{55} & S_{56} \\
S_{61} & S_{62} & S_{63} & S_{64} & S_{65} & S_{66} \\
\end{bmatrix}
\begin{bmatrix}
\sigma_1 \\
\sigma_2 \\
\sigma_3 \\
\sigma_4 \\
\sigma_5 \\
\sigma_6 \\
\end{bmatrix}^T
+ 
\begin{bmatrix}
d_{11} & d_{12} & d_{13} & d_{14} & d_{15} & d_{16} \\
d_{21} & d_{22} & d_{23} & d_{24} & d_{25} & d_{26} \\
d_{31} & d_{32} & d_{33} & d_{34} & d_{35} & d_{36} \\
\end{bmatrix}^T
\begin{bmatrix}
E_1 \\
E_2 \\
E_3 \\
\end{bmatrix}
\]

(2.10)

Although piezoelectric crystals can show weak piezoelectric effects in their natural forms such as in the Rochelle salt [37] and quartz [34], the performance is limited by the random orientation of the dipole moments in the crystalline domain structure. Here, the dielectric undergoes electrostriction, where the change in dimension is the same regardless of the applied electric field [24, 38]. Under relaxed circumstances, the gravity centers of the dipoles are randomly distributed to produce a net zero charge. [34]. However, if the material is heated past the Curie temperature, and a sufficient external voltage is applied, the domains can align to form permanent dipole moments, enhancing the piezoelectric effect. This process is referred to as piezoelectric poling. Examples of poling parameters of selected piezoelectric materials are provided in Table 2.1.

There exists a variety of different types of piezoelectric materials with different properties. Table 2.2 provides a short summary of a small selection of commonly used materials in piezoelectric devices. Polymer based materials, such as PVDF, offer better acoustic impedance matching to water (1.48 MRayl) in comparison to ceramics such as PZT. However, the
# 2.3 Ultrasonic Transducers

Ultrasonic transducers are generally categorized as bulk piezoelectric crystals, capacitive micromachined ultrasonic transducers (CMUTs), or piezoelectric micromachined ultrasonic transducers (PMUTs). Each of these offers unique advantages and disadvantages, particular to a given application. An overview of the stack-ups of typical bulk piezoelectric transducers, PMUTs, and CMUTs are shown in Fig. 2.2.

Bulk piezoelectric transducers are so called because each transducer uses a single bulk pre-poled piezoelectric crystal to vibrate and couple to the vibration medium. When in thickness mode, these transducers are operated at an anti-resonance of $\lambda/2$ [26, 44], determined by the distance that the acoustic wave must travel from one side to another. Traditionally, these transducers consist of a resonating piezoelectric layer, metal electrodes on either side,
Figure 2.2: Example stack-up of a) a bulk piezoelectric transducer, b) a CMUT, and c) a PMUT.

A matching layer on the side of propagation, and a backing layer on the reverse. An acoustic lens may also be included after the backing layer. An example of a typical cross-section of an ultrasonic transducer using a bulk element is provided in Fig. 2.3.
Figure 2.3: Cross-sectional example of a traditional diagnostic ultrasound probe.

In a traditional bulk piezoelectric transducer probe, the row of transducer elements is surrounded on either side by single sheet layers on the front and back. To prevent ultrasonic reflections created by the acoustic impedance mismatch between the piezoelectric material and the propagation medium, a $\lambda/4$ matching layer is included in front of the elements. To minimize losses, this layer ideally has an acoustic impedance that is $Z_{ML} = \sqrt[3]{Z_1 Z_2}$, based on Chebychev’s equations [26, 45]. Although this is the most common matching criterion, others can be used to enhance particular qualities. For example, the DeSilets’ criterion weights the transmission medium, $Z_{medium}$, greater than the piezoelectric, $Z_{piezo}$ to minimize distortion due to the pulse shape: $Z_{ML} = \sqrt[3]{Z_{piezo} Z_{medium}^2}$ [46, 47]. For maximum peak amplitude, the Souquet criterion can be used: $Z_{ML} = \sqrt[3]{2Z_{piezo} Z_{medium}^2}$ [45, 48]. As these methods involve using the wavelength of sound in the material, they are inherently narrow-band matching methods. This bandwidth can be widened by the stacking of multiple $\lambda/4$ matching layers,
each with an acoustic impedance governed by a set of the above criteria with respect to the adjacent layers [45]. The improved bandwidth comes at the cost of increased design and fabrication complexity.

The majority of commonly used piezoelectric materials found in transducers are either single crystals or ceramics with high densities and longitudinal speeds of sound, resulting in acoustic impedances on the order of $\approx 30$ MRayl, as per Table 2.2. Given that the impedance of water (similar to biological tissue) is $1.48$ MRayls [42], the required acoustic impedance of a desirable matching layer, according to the Chebychev method is $Z_{ML} \approx 6.66$ MRayl. A multitude of ceramic and epoxy materials have been used as matching layers, which generally involves creating a mixture or composite to achieve the desired acoustic impedance. For example, Wang et al. [49] performed acoustic property testing on an epoxy matrix of EPO-TEK 301 (Epoxy Technology, Inc, Bellerica, MA) with varying volumes of alumina and tungsten particles. This technique is commonly used in PZT transducers [50, 51]. Similar composites can be formed from alumina-Hysol [52]. Other methods include those with mixed conductive particles such as gold with parylene [47], silver-loaded polymers [53], and commercially available conductive films such as Ablefilm 5025E (Ablestik, Rancho Domingo, CA) [54]. Conductive films allow for easier electrical connection to the top of the transducer, at the cost of additional attenuation. Piezoelectric polymers have a substantially lower acoustic impedance (2-4 MRayl) and as such, polymers such as polycarbonate, polyester, and parylene C are frequently used for matching [55].

The backing layer can be found on the opposite side of the transducer, and can fill two possible roles: maximizing the forward power transfer or improving bandwidth. If an extremely high acoustic mismatch exists at the rear face (e.g. if the transducer is air-backed), the ultrasound energy is reflected forward towards the target, ideally doubling the transmitted pressure. As the piezoelectric element effectively rings longer, the axial resolution is
degraded [26]. Conversely, an alumina or tungsten loaded epoxy backing layer is capable of reducing reflections at the rear interface, and attenuating the signal sufficiently by the time the echo returns to the transducer. This results in a lower quality factor, and thus, improved bandwidth [54, 51]. However, the power radiated in this undesired direction is effectively lost. Moreover, a sufficiently rigid backing can cause a transducer to operate in a $\lambda/4$ resonance mode as opposed to the expected $\lambda/2$ resonance [55]. Therefore, a backing material allows for a compromise between resolution and power (sensitivity), while potentially de-tuning the designed resonant frequency of the transducer.

Fundamentally, a CMUT is a thin capacitive membrane with electrodes on either side, vibrating above a cavity. An example of such a device is shown in Fig. 2.2 (b). A constant DC voltage is applied to the membrane to allow electrostatic forces to pull a thin oxide film towards the cavity. The stiffness of the membrane provides an opposite force to restore it to the original flat state. When an AC voltage is applied in addition to the static DC voltage, the membrane will oscillate and create vibrations in the medium [44, 56]. For flexible ultrasound transducers, CMUTs can be undesirable because the silicon substrate can act as a secondary resonator and generate additional longitudinal waves at other frequencies [7]; although there are examples of semi-rigid CMUT arrays being used on flexible PDMS substrates [57].

PMUTs also operate by creating a deflection of a membrane, but this 0.01-3 $\mu$m film is now a piezoelectric material such as PZT, PVDF, aluminum nitride (AlN) or zinc oxide (ZnO) [44, 58]. Decreased mechanical stiffness (leading to a lower quality factor and a higher bandwidth) and up to a 60% increase in the piezoelectric coefficient has been reported by using a sputtered ScAlN film [59]. Although multiple methods of PMUT fabrication and integration have been explored, the most common is the sacrificial layer technique. The process starts with a thin layer of silicon nitride deposited on a sacrificial silicon carrier wafer. Bottom electrodes are then defined on the nitride layer before small holes are plasma-
etched that will later allow the diaphragm to be released. At this stage, a thin film of a piezoelectric material is deposited and patterned into the desired resonator shape (e.g. a ring). The top layer electrode layer is then deposited before a wet etch is used to release the sacrificial layer [44, 60].

A comparison of the advantages and disadvantages of bulk piezoelectric, CMUT, and PMUT transducers with regards to making a phased array transducer is provided in Fig. 2.4 [44, 61]. Specific features that are considered to be beneficial, such as a higher bandwidth, are displayed in green, whereas drawbacks, such as a DC bias requirement, are shown in red. A major reason for selecting bulk piezoelectric materials for this work was the availability of appropriate fabrication tools. Although the lack of CMOS compatibility is often cited as a deterrent for using bulk piezoelectric elements, the process described here is, in fact, compatible with CMOS [61].

<table>
<thead>
<tr>
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<th>Bulk Piezoelectric</th>
<th>CMUT</th>
<th>PMUT</th>
</tr>
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<tbody>
<tr>
<td>TX Power</td>
<td>Highest</td>
<td>Moderate</td>
<td>Moderate</td>
</tr>
<tr>
<td>RX Sensitivity</td>
<td>Moderate</td>
<td>Highest</td>
<td>Moderate</td>
</tr>
<tr>
<td>Bandwidth</td>
<td>Low</td>
<td>High</td>
<td>High</td>
</tr>
<tr>
<td>Matching Layer</td>
<td>Yes</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>DC Bias Required</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Process Variation</td>
<td>High</td>
<td>Low</td>
<td>Low</td>
</tr>
<tr>
<td>Processing</td>
<td>Dicing/RIE</td>
<td>Wafer bonding, Micromachining</td>
<td>Micromachining, Wafer transfer diaphragm</td>
</tr>
<tr>
<td>Notes</td>
<td>Generally not CMOS Compatible</td>
<td>Additional Longitudinal Modes</td>
<td>Difficult to tune resonance Residual Stress in membrane</td>
</tr>
</tbody>
</table>

Figure 2.4: Comparison of advantages of disadvantages of different transducer element types.
2.4 Phased Arrays

The concept of a phased array is to control the constructive and destructive interference of waves by precisely timing the excited/received phase of signals from a set of independently-controlled elements. Although experiments with electromagnetic antenna arrays for radar applications had been conducted for decades, the first steerable ultrasound array, the Electroscan I, was only developed in 1967 by Jan Somer [62, 63]. The Electroscan I had 21 transmit, and a single receive element, but was difficult to tune because of the fixed delay circuitry [62]. This limitation was alleviated in 1979, when Samuel Maslak heterodyned the pulses to an intermediate oscillator frequency, similar to common radio systems [64]. Through the 1990s, 2D arrays were introduced to allow for 3D imaging in real-time used today for cardio-vascular imaging [65]. The first arrays by Olaf von Ramm utilized a sparse Mill Cross layout for spacing 32 transmit and receive elements [66], whereas George Stetten arranged his 256 transducers in an annular ring [67]. Today, ultrasound phased array transducers are regularly used for detailed imaging and focused ultrasound (FUS) therapy to accurately ablate tumors with a narrow beam [12, 13, 14]. Further research is being conducted in sparse arrays in the interest of reducing connectivity complexity, while maintaining resolution performance [68].

Arrays can be modelled as a series of point sources, that is, a group of individual, infinitesimal pressure radiators. The Rayleigh-Sommerfeld integral can describe the pressure amplitude, \( p \), radiating in space from source \( S \) as a function of angular frequency, \( \omega \) [69]:

\[
p(\vec{r}, \omega) = \frac{j \rho_0 c k v_0}{2\pi} \int_S \frac{e^{j[\omega t - k(\vec{r} - \vec{r}_0)]} A(\vec{r}_0) dS}{|\vec{r} - \vec{r}_0|} \quad (2.11)
\]

Here, the term \( A(\vec{r}_0) \) indicates the local source function amplitude. Also, \( |\vec{r} - \vec{r}_0| \) represents the difference of the distance vectors from the origin to the target point, and the transducer

19
to the target point.

The time required for pressure to arrive from a given element to a target point, \( P(r, \theta) \), is dependent on the distance between the point and the element, as well as the wave velocity in the medium. Consider the geometry of a simple array of radiating elements arranged in a line, as shown in Fig. 2.5. In this simple 1D case, we can describe the distance of the \( n^{th} \) element to a point by Eq. 2.12.

![Figure 2.5: Distances to a target point for phased array elements in a flat 1D array.](image-url)
To maximize constructive interference between the waves, the peak pressure for all elements should arrive simultaneously. Elements that are physically closer to the target must be delayed to compensate for the extra distance the other pulses must travel. The delay will be equal to the propagation time between elements, thus the difference in distance, $\Delta r_n$, can be defined by Eq. 2.13.

$$\Delta r_n = r - r_n = r - \sqrt{r^2 + x_n^2 - 2rx_n \sin(\theta)}$$  (2.13)

The Fresnel approximation can be used to simplify Eq. 2.13 by using the geometrical assumption that $r \gg x_n$ [70, 71]. Such a claim implies that the focus is further away than the distance from the center of the array to a given element, which is true for distant foci, or short arrays. This assumption allows us to rewrite Eq. 2.13 as Eq. 2.14.

$$\Delta r_n \approx x_n \sin(\theta) - \frac{x_n^2}{2r}$$  (2.14)

Using the spatial difference between each element and the target, the delay equation for the given $n^{th}$ element is given by Eq. 2.15. In this form, the first term determines the delay due to the steering angle, whereas the second term is the contribution from the focus. It is therefore possible to only apply a steering angle to generate a plane wave in a desired direction.

$$\Delta t_n = \frac{\Delta r_n}{c} = \frac{x_n \sin(\theta)}{c} - \frac{x_n^2}{2cr}$$  (2.15)

A simulation can be used to visualize the pressure field generated by an array for various
Figure 2.6: Phase delays and the corresponding simulated pressure field for an 8-element array steered at $-30^\circ$ with no focusing.

excitation phase distributions. Using the MATLAB FOCUS package [32], a simple 8-element array can be simulated with custom excitation functions. By staggering the delays between elements as shown in Fig. 2.6 and Fig. 2.7, the pressure beam can be steered with or without a focus. Such steering will be useful in the reconstruction of sectors of a final B-mode image.
Figure 2.7: Phase delays and the corresponding simulated pressure field for an 8-element array steered at $+20^\circ$, and focused at 5 mm from the center of the array.
3 Flexible Ultrasound Phased Array

In this section I discuss the design, fabrication, and testing of the flexible ultrasound phased array. I explore the need for flexible phased arrays in diagnostic ultrasound and suggest relevant applications. After discussing compromises between system parameters, I explain my design decisions as the design evolved over iterations. I first confirm the baseline functionality of the phased array in the traditional flat orientation before demonstrating the ability to adapt to curvature on both gelatin phantoms and a human.

3.1 Need for Flexible Ultrasound

Diagnostic ultrasound has become an invaluable imaging modality to healthcare professionals for the purpose of non-invasive examination of soft tissue, as it carries with it no associated risks of exposure to cell-damaging ionizing radiation. This method of non-destructive imaging is usually achieved with linear arrays of piezoelectric ultrasound transducers, which convert mechanical sound energy to electrical signals and vice versa. The majority of commercially available transducers are housed in a rigid probe and have fixed interfaces that cannot conform to the surface of the human body in three dimensions. These geometric limitations can serve to introduce air pockets between the imaging probe and the patient’s body, creating localised strong ultrasound reflectors that disturb image formation. Clinical
applications thus require the application of an intermediate ultrasound gel layer to match the acoustic impedance of the probe and the imaged body part, rendering long-term monitoring impractical and cumbersome on uneven surfaces. Thus, many incentives exist to create flexible, body-conformable and reliable ultrasound probes that can match the frequency and resolution performance of contemporary rigid-interface arrays for biomedical applications.

2D phased arrays possess the additional advantage of steering the generated ultrasound beam in three dimensions at arbitrary focal lengths, which can be used for detailed imaging and focused ultrasound (FUS) therapy to accurately ablate tumours with a narrow beam [12, 13, 14]. Potential applications of these conformable versions of the transducer arrays include emergency medicine needs, such as the confirmation of proper endotracheal tube insertion. During tracheal intubation, a tube is inserted through the trachea and into the lungs to provide air to the patient in an emergency [9]. However, 10-25% of cases involve multiple intubation attempts and other complications [10, 11], while most happen in an intensive care unit where long term monitoring is also required [9]. Ultrasound is already used as a secondary methodology for confirming that the tube is correctly placed; however, this is performed with traditional rigid probes [15, 16, 17, 18], and depending on the anatomy of the patient, the individual tube diameter for insertion may be as small as 6.5 mm [19]. Flexible probes for these and other similar applications must therefore, at the smallest extent, stretch to conform to a cylinder with a radius of curvature of $\approx 1.5$ cm to ensure accurate image reconstruction.

Ultrasound imaging is also being explored as a viable high-throughput standard for identifying bone fractures, offering a rapid alternative to X-rays. The non-ionising nature of ultrasound is particularly important when examining fractures in minors, where unnecessary exposure to X-rays can be dangerous. In a study of 224 suspected fractures in children, 86.6% of the cases were correctly identified using ultrasound [20]. Ultrasound also provided
the highest sensitivity for identifying fractures in long bones, specifically the humerus, with a sensitivity as high as 98% across 233 individuals [21]. The diameter of an adult humerus in the midshaft position varies statistically from 17 to 23 mm for males, and 14-20 mm for females [22], comfortably resolvable by low ultrasonic frequencies. Flexible ultrasound phased arrays, as described in this work, may thus provide a convenient solution for safe, rapid, and accurate detection of fractures in the upper limbs, as well as other musculoskeletal abnormalities.

In general, the concept of creating flexible or stretchable ultrasound arrays involves the direct integration of ultrasound transducer elements onto a flexible interconnect substrate [3]. Prior efforts in this area have generally integrated bulk samples of the 5H phase of lead zirconate titanate (PZT-5H) or piezo-composites onto a polymer backing substrate with electrodes for the signal and ground connections [4, 5, 6]. A notable advance in the field of flexible ultrasound imagers was the recent introduction of a stretchable patch containing 100 separate 1-3 piezo-composite elements, arranged in a two-dimensional (2D) array on a stretchable polydimethylsiloxane (PDMS) substrate [72]. One-dimensional (1D) wrap-around arrays have also been demonstrated, but they typically operate in the 1-10 MHz range, with between 8-128 transducer elements [73]. However, many of these systems have technical shortcomings, such as too wide a pitch between elements, leading to the formation of grating lobes at the designated frequencies of operation, and no consideration for array curvature and its effect on phase interference; ultimately degrading the performance of the imaging system by introducing errors and artefacts in the final image.

Here, we demonstrate the FlexArray – a flexible 2D ultrasound phased array containing 256 piezoelectric transducers, which utilises adaptive phasing to compensate for the changing radius of curvature when imaging curved objects. We describe the fabrication process and functionality of the flexible array and discuss the phasing algorithm employed for image
correction. We compare and contrast the imaging performance of the conformable array on phantom soft matter of various shapes with and without phase correction. The FlexArray demonstrates controllable beam steering functionality and operational pressures of > 500 kPa at the focal point, at resonant driving frequencies of ~ 1.4 MHz. The piezoelectric transducer elements have a wide bandwidth (41.3%), making them ideal for interfacing and interrogating biological tissue. The cross-talk between neighbouring elements is suitably low ( ~ -52 dB to the first neighbour, ~ -74 dB to the second neighbour), allowing for high-contrast imaging with minimal interference. Finally, we use the device to image the cross-section of the human upper arm, proving its capability for a non-invasive, non-ionising biomedical imaging apparatus that can examine bone topology.

3.2 Results

3.2.1 Device Fabrication and Benchmarking

Our flexible ultrasound array utilises bulk piezoelectric transducers; each fabricated from a single pre-polled piezoelectric PZT crystal coupling directly to the vibrating medium to be imaged. When operated in thickness mode, these transducers should have their height optimised to $\lambda/2$ (18,19), where $\lambda$ is the wavelength of the used ultrasound wave. The resonating piezoelectric layer also has contact electrodes on the top and bottom to transduce the electrical signal generated by the piezoelectric effect. We chose to fabricate a bulk transducer array, rather than opting for capacitive micromachined ultrasonic transducers (CMUTs), or piezoelectric micromachined ultrasonic transducers (PMUTs) for several reasons. In the case of the former, the longitudinal modes can cause significant cross-talk between elements [72], reducing the resolution. The required DC bias for receive mode also presents a complication that would require additional circuitry and a transmit/receive mode detection method local
to the board housing the array. A PMUT design, in turn, eliminates the need for the DC bias, but a delicate transfer process would need to be devised to protect the flexible printed circuit board (PCB) substrate from the backside etching used to define the diaphragm [44]. Fig. 3.1a presents a 3D diagram of the piezoelectric PZT transducer array mounted on a flexible PCB. Briefly; the FlexArray is fabricated by lithographic patterning and dicing of a single block of PZT into four square quadrants each containing $8 \times 8$ pillars, with individual pillars measuring $825 \, \mu\text{m} \times 825 \, \mu\text{m}$ in area, and $1.006 \, \text{mm}$ in height. The detailed descriptions of the fabrication process for the transducer array are included in Methods. The scanning electron micrograph in Fig. 3.1b shows a corner of the metallised array after dicing, while Fig. 3.1c presents the typical surface morphology of the PZT pillar after coating with the top electrode metal layer of Cr/Au. Energy-dispersive X-ray spectroscopy was used to confirm the elemental content of PZT in the fabricated pillars (Fig. 3.1d), while readings of the piezoelectric coefficient ($d_{33}$) indicate average values of $534 \, \text{pC/m}$ – sufficient to obtain ample voltages at typical propagation pressures generated by sound waves in soft matter. The table summarises the ($d_{33}$) values after each fabrication step, demonstrating that the piezoelectric properties of the material are not affected by the processing. We note here that as the bulk transducers are operated in longitudinal mode, the array is fabricated such that the PZT-5H poling direction is out of the plane of the device [74].

The design, manufacturing and composition of the flexible PCB are described in Methods. The final thickness of the array is approximately $0.26 \, \text{mm}$ in bending regions and $1.26 \, \text{mm}$ at the piezoelectric pillar locations. Figs. 3.1e-h demonstrate several bending modalities for the as-fabricated substrate. A diagonal routing design allows for conformable wrap-around patching of the imaging target, with the exposed electrical contacts for the piezoelectric array located in the center of the board. The minimum radius of curvature (R) supported by the array before the adjacent pillars on the mounted array touch is $R > 1.4 \, \text{cm}$, meet-
Figure 3.1: a 3D rendering of the flexible array with mounted piezoelectric transducers. b Scanning electron micrograph of diced PZT pillars on the edge of an array quadrant. The pillars are contacted with a Cr/Au top metal layer. Scale bar, 200 µm. c Scanning electron micrograph presenting the surface morphology of the gold-covered PZT crystals. Scale bar, 2 µm. d Energy dispersive X-ray spectrum of the pillar, demonstrating the elemental content of the fabricated transducer. e The as-manufactured flexible PCB used to house the piezoelectric array in the center, seen held in a neutral position. f The diagonal connector design allows for the easy bending of the array by hand. The board is flexible enough to also support g convex bending, and h shear bending.

...ing the geometrical requirements to image most human body parts. The bonding process that combines the PCB substrate and the diced PZT array is carried out using standard semiconductor industry fabrication techniques (see Methods). Fig. 3.2a shows the diced PZT blocks after the bonding step is carried out to mount them on the electrical pads in the center of the flexible PCB. A cross-sectional diagram of two neighbouring transducer...
pillars is shown in Fig. 3.2b. The transducers are electrically separated from each other, and from the conductive traces on the PCB, with an insulating layer of parylene. In order to minimise potential electrical cross-talk, the signal pads are made taller with an intermediate electroplating step before bonding, such that the pillars are propped up above the height of the board traces. Once the FlexArray is fabricated, it retains all of the flexibility and ease of handling of the bare PCB, while providing a curved imaging modality with the piezoelectric transducer array, as seen in the photographs in Figs. 3.2c,d.

We benchmark the ultrasound transduction performance of the finished device by first considering the impulse response of the array in a neutral (unflexed) position. Fig. 3.2e presents four experimental pulse-echo traces, each from a randomly-chosen transducer in a different quarter of the 16 × 16 pillar array, and their associated frequency spectra. As evident from the nearly-identical plots, the element-to-element variation in the performance across the array is negligible, confirming the reliability of the fabrication process and ensuring consistent device functionality regardless of bending direction. The ultrasonic pressure generated by the array, when tested in water, is 1.4 kPa V⁻¹ (∼ 15.2 kPa V⁻¹ at focus), which is comparable with other flexible ultrasound platforms [72, 75, 76, 77]. These response parameters meet the general requirements for high-frequency ultrasound biomedical imaging [78], even before phase correction is implemented to improve image quality. A frequency sweep (Fig. 3.2f) for the array, both when unfocused and focused, reveals the resonant frequency of the device to be ∼ 1.4 MHz, with a bandwidth of 41.3%, with a small anti-resonance peak emerging at ∼ 2.7 MHz. In the inset, we plot the generated ultrasonic pressure, at resonance, as a function of the voltage applied across the array transducers, demonstrating the expected linear scaling at both $f = \infty$ and $f = 2$ cm.
Figure 3.2: a Photograph of the bonded PZT array quarters after back-side dicing to separate out the individual transducer pillars. b Cross-sectional illustration (not to scale) of the finalised pillar mount. Cu electroplating is used to raise the metal connection before bonding with the bulk PZT transducer on the ACF substrate. Parylene isolates the signal pads from the electrical traces on the board. c Photograph of the completed FlexArray, relative to a human hand. d Close-up photograph of a bent piezoelectric array in the center of the device after metallization and encapsulation. e Examples of pulse-echo responses, and their associated frequency spectra as insets, from arbitrary pillars in each quarter of the array. The temporal envelope for three pulses is $\sim 4000 \mu s$ with a wide bandwidth of $\sim 41\%$. f Ultrasonic pressure generated by the FlexArray in water when focused at $f = \infty$ (blue) and at $f = 2\text{cm}$ (red), across a range of ultrasound frequencies. The pressure value peaks at over 600 kPa. The inset shows the linear scaling of the generated ultrasonic pressure with applied peak-to-peak voltage across the transducer elements.

### 3.2.2 Ultrasound Beam Steering

Each transducer on the device can be biased with a voltage of an arbitrary phase. The relative offset of these individual phase delays creates patterns of constructive/destructive...
interference. This allows the FlexArray to steer the ultrasound beam in specific directions in three dimensions at an arbitrary focal point. Such phase steering implementations can have multiple applications, including highly concentrating a pressure beam at a specific location for increased contrast, or to stimulate electrical responses in neurons at pressures of $\sim 100$ kPa [79, 80]. To demonstrate ultrasound beam steering functionality with the FlexArray, we used a custom configuration script to excite the array elements in order to produce a deflected ultrasonic pressure beam in water, in both the x-z and y-z planes, and we measured the pressure at each location in the z-direction, using a hydrophone on a motorised stage. The FlexArray was mounted to a custom 3D-printed part, designed to hold the device within a 50 mm $\times$ 70 mm $\times$ 60 mm water tank, as shown in Fig. 3.3a. Simulated beam patterns were produced using the FOCUS II MATLAB package [4], to predict the behaviour of the device when focused at 2 cm and steered at $-20^\circ$, $0^\circ$, and $+20^\circ$ of phase difference along the x-z plane. The resulting simulated beam patterns for those respective phase shifts are shown in Figs. 3.3b-d, demonstrating a clear concentration of ultrasound pressure to narrow areas of the array’s field of view. The measured experimental results from the FlexArray for the same respective phase shifts and focus are shown in Figs. 3.3e-g. The experimentally-realised ultrasound beam steering matches well with the patterns predicted by simulations, demonstrating spatial ultrasound beam steering capability as well as axial pressure hot spot localization. The maximum value of focal pressure achieved with the FlexArray is $\approx 615$ kPa, with a power supply of $\pm 20$ V at a focal length of 2 cm. An embedded wire phantom was also used to determine the axial and lateral resolutions to be approximately 3 mm and 2 mm, respectively, at a depth of 2 cm. Improvements to ultrasound coupling to biological media with the potential inclusion of a matching layer on the surface of the PZT pillars may enable these types of 2D arrays to be used for efficient neural stimulation in the future [79, 80].
Figure 3.3: a Photograph of the experimental setup used to record the distribution of pressure in a water tank at a focal length of 2 cm. b-d Simulated pressure maps at this focal length when the array elements are excited at -20°, 0° and 20° of phase shift, respectively. e-g Experimental pressure maps collected with the FlexArray at the same respective phase shift values. Note that all colormaps are self-normalised to the colorbar on the right of the figure.

To demonstrate the robustness of the device, we performed additional FIELD II simulations to predict the functionality of the transducer array at various radii of curvature; with and without phase correction. As seen in 3.4a-d, the phase correction for the B-mode images has an expectedly less-pronounced effect for wider radii of curvature. In Figs. 3.4e-g, we also present simulation results of beam steering with an unfocused flat array (i.e. focused at $f = \infty$), which we compare to experimental results in Figs. 3.5e-g.

The experimental results from the FlexArray in Fig. 3.5 should be compared directly with the simulated images presented in Fig. 3.4. We produced additional gelatin phantoms with larger radii of curvature to demonstrate the effect of phase correction as R increases,
Figure 3.4: a Simulated B-mode image without phase correction when R = 2 cm. b Simulated B-mode image with phase correction when R = 2 cm. c Simulated B-mode image without phase correction when R = 3 cm. d Simulated B-mode image with phase correction when R = 3 cm. Note that a-d are all self-normalised to the colorbar on the right. e Simulated beam steering pattern with focus at infinity, when the phase shift is -20°. f Simulated beam steering pattern with focus at infinity, when the phase shift is 0°. g Simulated beam steering pattern with focus at infinity, when the phase shift is +20°. Note that e-g are all self-normalised to the colorbar on the right.

shown in Fig. 3.5a-d. As expected, as the radius increases the array becomes flatter, and the correction has less of a demonstrable effect on the resulting reconstructed B-mode image. Furthermore, experimental pressure maps for the unfocused (plane wave) beamforming at different steering angles are demonstrated in Figs. 3.5e-g.
3.2.3 Transducer Cross-Talk Characterization with Increasing Curvature

To ensure that the PZT pillars are electromechanically isolated at all instances of bending, we examined the inter-pillar electrical cross-talk on the array as a function of the radius of curvature, $R$. Fig. 3.6a shows a photograph of a detachable 3D-printed model which
allows for bending the FlexArray to a specific radius of curvature between 1–5 cm. We characterised the electrical cross-talk between an individual powered element and its nearest two neighbours across this radius of curvature range based on their power spectral densities (extreme cases of $R = 1.5$ cm and $R = \infty$ are presented in Fig. 3.6b). In Fig. 3.6c, we plot the relative cross-talk power between the powered-on element and its first and second neighbours, as a function of $R$. Each radius was tested 10 times on the same set of elements, and the error bars correspond to one standard deviation from the mean. The average recorded cross-talk magnitudes are $\sim -50$ dB for the nearest neighbour element and drop to under -70 dB when measured two elements away. Importantly, the cross-talk levels remain unchanged as the array is flexed away from the neutral position all the way to extreme radii of curvature below 2 cm. This establishes the fabrication methodology for the array as reliable and suitable for these curved imaging applications. In addition, the array was mechanically cycled hundreds of times throughout this testing and did not incur any noticeable change in performance. Although piezoelectric composites may offer higher cross-talk suppression than PZT due to their anisotropic phonon dispersions [72, 49, 81, 82], they are considerably more expensive and labour-intensive to fabricate. We argue that the resultant trade-off for a small SNR increase and additional cross-talk suppression of $\sim 10$ dB to the nearest element [72] is not as crucial for improved imaging functionality as phase correction; as we go on to show in the following sections.

3.2.4 B-mode Imaging of Test Phantoms

After confirming the electrical functionality of the FlexArray, we imaged the medical-grade 84-317 Multi-Purpose Tissue/Cyst Ultrasound Phantom (Fluke Biomedical) in a flat position ($R = \infty$). The composition of the 84-317 phantom is designed to mimic the acoustic properties of the human liver, with additional buried ultrasound reflectors consisting of
0.24-mm diameter monofilament nylon rods. To demonstrate the resolution of the array, we imaged a set of coplanar parallel rods located 1 cm apart in the axial direction (see photograph in Fig. 3.6d). A composite image was created by weighted averaging of scans focused at depths of 4 cm, 5 cm, 6 cm, and 7 cm. This resultant B-mode image (Fig. 3.6e) matches well with the known locations of the reflectors in the phantom. Additional scatterers from an adjacent set of rod targets are also seen in the photograph; however, the reflected intensity in the B-mode from the low scatter cyst (Lo-S) region remains unresolved, as the core is designed to be more transparent to ultrasound. We extracted the axial and lateral resolutions (see Methods) as a function of imaging depth from the phantom images (circular data points in Fig. 3.6f). We performed a series of simulations in FIELD II utilising the same array geometry at 1.4 MHz, with a single point reflector located at arbitrary points along the z axis, and plotted the resulting resolutions (dashed lines) along with the theoretical expected resolutions (solid lines). The experimental data points show a good trend correlation with the measured point targets in the phantom, expectedly trailing off at larger depths due to increased scattering in the medium. We note that the discontinuity in the simulated lateral resolution is caused when a reflection expands to the point of being on an additional ray line. In both simulations and experiments, we limited the number of x-z raylines to 96.

Having established a well-performing electrical functionality away from the neutral position, we proceed to discuss the imaging performance of the FlexArray when significant curvature is introduced, and make comparisons with simulations. To accurately compute the phase delays, the precise relative positions of the individual transducer elements must be known. We developed a custom adaptive phasing algorithm to compute these positions based on a known radius of curvature in the x and y axes. More details of this corrective algorithm are presented in Methods. For simple cylindrical bending around an axis, the new coordinates, \( x_0 \), \( y_0 \), and \( z_0 \) (along with the azimuth and elevation), can be computed from
Figure 3.6: a Photograph of the detachable 3D-printed apparatus for the testing of electrical cross-talk across different radii of curvature. b Power spectra of the voltage response of a single powered transducer (green), relative to those of its nearest neighbor element (blue) and its second nearest neighbor (pink). Extreme cases of a highly-curved array ($R = 1.5 \text{ cm}$) and a flat array ($R = \infty$) are compared. c Cross-talk between neighboring elements, extracted at 1.4 MHz, as a function of radius of curvature of the FlexArray. The device shows excellent inter-channel shielding characteristics at high operational frequencies, regardless of shape. d Photograph of the cross-section of the tissue phantom with embedded nylon wires (white dots). The yellow dot marks the vertical location where the FlexArray was placed during imaging. e B-mode scan of the phantom taken with the FlexArray, demonstrating both its axial and lateral resolution. The locations of the nylon rods can be resolved axially at $\sim 1 \text{ cm}$ apart. f Plot of the theoretical, simulated, and measured axial/lateral resolutions as a function of scatterer depth. Note that error bars are smaller than the marker size of the experimental data points.

their respective original coordinates, $x$, $y$, and $z$, using the transformations in Table 3.1. In the FIELD II simulations used here [30], arbitrary radii of curvature can also be applied to the transducer array to reconstruct the image of a rectangular $10 \text{ mm} \times 5 \text{ mm}$ air pocket suspended in gelatin at a distance of 2 cm from the imaging aperture [31]. For the case of R
= 1.5 cm, which is most relevant for complex biomedical applications, we compare simulations with the experimental performance of the FlexArray – with and without the adaptive phasing correction. In order to accurately bend the array to a known R for experimental imaging of the air pocket, custom moulds were 3D-printed and blocks of gelatin were cast therein. These exchangeable shapes could then easily be used to cap the air pocket with a curved piece of a known R for each measurement. Figs. 3.7a,b present labeled photographs of the experimental set-up from the side and top views, respectively, highlighting the location of the air pocket. Fig. 3.7c shows the simulated B-mode scan without phase correction. The two main high-intensity features seen in the plot are regions of high reflectivity from the top and bottom of the air block, i.e. material interfaces where the acoustic impedance mismatch is at its highest. Due to the introduced curvature, the constructive interference of ultrasound waves results in spurious reflections around the air pocket target, artificially increasing its observed lateral width. When using our phase delay correction in simulation (Fig. 3.7e), the edges of the air pocket become clearly defined in the B-mode image, confirming the functionality of the correction algorithm. Figs. 3.7d,f present the same B-mode comparison, but for experimental data obtained with the FlexArray, without and with phase correction, respectively. As seen readily from the plots, spurious reflections are strong around the air-gelatin interfaces when no phase correction is applied, while the lateral width of the air pocket is artificially increased. When the phasing algorithm is implemented, the air-gelatin interface is correctly resolved in the highly-curved case of R = 1.5 cm, with the spurious reflections suppressed and a noticeable decrease in the lateral extension of the target. The extent of the strong reflection is also narrowed in the axial direction for the two interfaces. We note that the experimental sonograms are compound images of B-modes taken at three different focal lengths (f = 2, 3, 4 cm), appropriately weighted with windowing functions to remove the high-intensity speckle region from the near field. In addition to assisting in routing on
### Table 3.1: Translated locations of elements after bending

<table>
<thead>
<tr>
<th></th>
<th>Curve XZ</th>
<th>Curve YZ</th>
<th>Curve Both</th>
</tr>
</thead>
<tbody>
<tr>
<td>(x')</td>
<td>(R_x \sin \left( \frac{x}{R_x} \right))</td>
<td>(x)</td>
<td>(R_x \sin \left( \frac{x}{R_x} \right))</td>
</tr>
<tr>
<td>(y')</td>
<td>(y)</td>
<td>(R_x \sin \left( \frac{y}{R_y} \right))</td>
<td>(R_x \sin \left( \frac{y}{R_y} \right))</td>
</tr>
<tr>
<td>(z')</td>
<td>(R_x \left( 1 - \cos \left( \frac{x}{R_x} \right) \right))</td>
<td>(R_y \left( 1 - \cos \left( \frac{y}{R_y} \right) \right))</td>
<td>(R_x \left( 1 - \cos \left( \frac{x}{R_x} \right) \right) + R_y \left( 1 - \cos \left( \frac{y}{R_y} \right) \right))</td>
</tr>
<tr>
<td>Azimuth</td>
<td>(\tan \left( \frac{x}{R_x} \right))</td>
<td>0</td>
<td>(\tan \left( \frac{x}{R_x} \right))</td>
</tr>
<tr>
<td>Elevation</td>
<td>0</td>
<td>(\tan \left( \frac{y}{R_y} \right))</td>
<td>(\tan \left( \frac{y}{R_y} \right))</td>
</tr>
</tbody>
</table>

The board, the 1 mm element pitch prevents grating lobes at low frequencies. The 2D image of the x-z slice generated by the FlexArray is displayed on the computer in real time and its intensity, contrast and colormap can be adjusted easily with user-friendly toggles in the controlling software by the operator.

#### 3.2.5 In-vivo Imaging of the Human Humerus

After validating the curved functionality of the FlexArray on the gelatin phantoms, the system was tested by wrapping the device directly around the arm of a healthy human male. To demonstrate the effectiveness of the system on a curved body part, we imaged the humerus bone at three different levels along the arm, with the scan plane orthogonal to the length of the arm (Figs. 3.8a,b). Details of the experimental procedure can be found in Methods. The B-mode images in Figs. 3.8c,e,g were taken without any phase correction at each level, whereas those in Figs. 3.8d,f,h were taken at the same corresponding levels, but after applying a correction for a radius of curvature of 4 cm in both the x-z and y-z dimensions. The phase correction algorithm allows the center feature to be more clearly defined, and removes spurious signals seen at the peripheral scan angles. In both uncorrected and corrected images, the predominant feature is the surface of the humerus where a fraction of the top part of the bone, approximately 2 cm in diameter, can be seen. When phase
Figure 3.7: a Photograph of the cast gelatin phantom used to collect B-mode images at arbitrary curvatures. The FlexArray is seen conforming to the surface. b Plan-view photograph of the air pocket-containing gelatin mould. c Simulated B-mode image when the radius of curvature is 1.5 cm and phase correction is not implemented. d The same simulation with phase correction implemented. e Experimental B-mode image of an air pocket in gelatin captured with the FlexArray when the radius of curvature is 1.5 cm; without phase correction. f The same B-mode image with implemented phase correction. Both images in e and f are compound averaged focal series images between $f = 2, 3$ and 4 cm.

correction is applied, this reflection narrows laterally by more than 30% in all three locations. Assuming a circular cross-section for the bone, the imaged bone surface corresponds to a
bone diameter falling within the expected range for an adult human male [21, 22]. Similarly, Figs. 3.9a,c,e and Figs. 3.9b,d,f show 3D representations of 16 y-z slices, without and with the corrective algorithm.

![Figure 3.8](image)

Figure 3.8: a Close-up photograph of the array area on the arm during operation, also showing the 2D plane along which the scan was taken. b Photograph of the subject’s arm with the device in the Level 2 position. Three levels were chosen to image the bone along the length of the arm. c-d Level 1 B-mode images of the humerus without and with correction for the radius of curvature of 4 cm, respectively. Analogous images were taken at e-f Level 2 and g-h Level 3.

### 3.3 Discussion

In this work, we have demonstrated a passive flexible ultrasound phased piezoelectric array with implemented curvature correction. The adaptive phasing algorithm allows for signifi-
Figure 3.9: Combined slices of 16 y-z scan angles to produce 3D images of the humerus at the level 1 (a,b) level 2, (c,d), and level 3 measurement locations. (a,c,e) The resulting image using traditional phasing and reconstruction, regardless of the radius. (b,d,f) The same position on the arm, now imaged using the corrected radius of curvature.
cant improvements to image quality when visualising the internal structure of soft matter with curved surfaces, including those of the human body. Further improvements to device performance can be achieved in the future by optimising the frame rate by reducing the processing time, and by fabricating a matching layer on top of the transducers to increase the acoustic pressure coupling, bettering the sensitivity range of the device. To push the technology into fully realised and minimally-invasive ultrasound diagnostic bio-electronic interfaces, future iterations ought to also reduce the transducer pitch, while integrating a larger amount of them on the flexible board. Low-power wireless signal actuation and transmission are also future goals which would allow for distributed and real-time subsurface health monitoring of the human body.

3.4 Methods

3.4.1 Device Design, Fabrication, and Packaging

The overall design flow aims to electrically contact the piezoelectric transducer sheet by patterning pads on either side. The flexible printed circuit board (PCB) is prepared for integration with the transducer by building up the pad height through electroplating, and creating isolating parylene pockets. Subsequently, PZT pillars are then diced with a saw and bonded to the substrate. A metallization step by sputtering is used to provide the top ground connection. Finally, the device is encapsulated with parylene to prevent electrical shorts from contact with the environment. We discuss the fabrication flow in detail below and present a sketch of the cross-section of the array at each key step in Fig. 3.10:

A disk of pre-poled PZT-5H (diameter approx. 75 mm, thickness approx. 1016 µm, Piezo.com) was used as the starting material. Utilizing a pre-poled piezoceramic sheet is amenable to migrating this fabrication flow to complementary metal-oxide-semiconductor
Figure 3.10: a The as-produced flexible PCB with marked board traces and contact pads. b Parylene coating isolating the pads from the traces. c PZT block with lithographically defined pads, before bonding to the board. d The PZT block is bonded to the pads on the board with an intermediary anisotropic conductive film (ACF) layer. e Final parylene coat after dicing, electrically isolating all the individual pillars from one another.

(CMOS) processes, as electrical poling procedures involve large electric fields that will destroy integrated circuits during poling. The processing ought to remain, however, under half of the Curie temperature of PZT ($T_C = 340^\circ C$) in order to preserve a meaningful piezoelectric effect. Commercial PZT disks are pre-coated with nickel, which is susceptible to oxidation in air, thus the disks were placed in ferric chloride solution for approximately 1 minute to strip the nickel film from the surface of the PZT. Standard photolithographic techniques were then used to pattern a grid of contact pads on either side. These pads were designed to have the same dimensions as the pads on the board, i.e. $16 \times 16$ symmetric element arrays, with 425 $\mu m$ pads, with 1 mm pitch. To optimise the amount of usable material per piezoelectric disk, a $4 \times 4$ grid of these arrays was defined, although the size of the piezoelectric disk usu-
ally limited the yield to only the center-most 12 arrays being fully usable. There is a spacing of 1575 µm from edge to edge, between the pads of each array. Cross-shaped markers were used both for alignment during lithography and as dicing lanes. The used mask is shown in Fig. 3.11a.

To pattern the pads, the PZT block was first treated in a plasma asher (Anatech) with O₂ plasma at 100 W for 5 minutes to remove adventitious organic residue, and chemically prime the surface for adhesion. S1818 photoresist was spun at 500 rpm for 10 seconds to spread the resist initially, and then at 3000 rpm to produce a layer of thickness approx. 1.5 µm. The resist was then soft-baked at 110°C for 60 seconds. The UV light exposure was performed using a Suss MA6 mask aligner with the aforementioned mask at 150 mJ/cm². The pattern was then developed in AZ 300 MIF developer for 60 seconds. To ensure no resist scum remained in the patterned area, the same plasma asher was used to expose the sample to O₂ plasma at 100 W for 2 minutes. Subsequent metallization was carried out in an electron beam evaporator (Angstrom Engineering EvoVac Multi-Process), depositing a 5 nm adhesion layer of chromium followed by 100 nm of gold – forming the top pads. An example of a PZT sheet immediately after metal deposition is shown in Fig. 3.11b. Rather than immediately performing lift-off, the backside of the PZT was patterned first. The PZT disk is flipped to the yet-unexposed back-side, and the same photolithographic process is repeated. During the alignment stage, however, backside alignment was performed on the Suss MA6 mask aligner, i.e. the microscope lens was situated below the stage. Backside development was performed in AZ 300 MIF again using the same parameters. We note here that over-development of the already-exposed and metallized side is not a concern as the developer does not react with gold. The same plasma ashing and evaporation steps were repeated on the newly exposed backside. The entire disk was then placed in acetone for lift-off and sonicated for 5 minutes to completely remove the resist. This resulted in a PZT
disk containing grids of gold pads on either side, which are aligned with each other across
the piezoceramic, yielding approximately 12 grids which can be used to make 12 separate
arrays.

Figure 3.11: a Computer-aided design pattern of the piezoelectric array. b PZT disk after lithographic
pattern exposure and metal deposition. c 3D rendering of the dicing progression for an
individual array. d Array quarters after dicing. e Camera shot of the array cross-section
during the bonding procedure to the flexible PCB. Note the metal toolhead pressing down
from the top. f Isolated PZT pillars directly mounted to dicing tape to demonstrate flexibility.
g Photograph of the diced array on the PCB after bonding. h Polyimide shadow mask on
the board, preparing the array for top metal deposition.

The size of the pads on which the PZT pillars are mounted is limited by the routing of the
board traces. These traces between pads also received the same ENIG (two layer metallic
coating of 2-8 µin Au over 120-240 µin Ni) surface finish during manufacturing – meaning
that on as-produced boards, the pads and the traces both have the same height. This would result in short-circuiting between elements after pillar bonding, thus the pad areas needed to be selectively built-up in order to comfortably protrude above the height of the board traces; if their area is large enough to result in this overlap. This was accomplished by electroplating the gold pads and isolating these with parylene to prevent shorts. First, a masking layer was used to only expose the pads as targets for plating. Thick (> 10 µm) AZ P4620 photoresist was spun on at 500 rpm for 5 seconds, followed by 2500 rpm for 45 seconds, and baked at 110°C for 90 seconds to remove any remaining solvent. Exposure was done with the Suss MA6 mask aligner, at 300 mJ/cm², followed by development in AZ 400K:DI water (1:4) for 3 minutes. O₂ plasma was used again to de-scum the sample for 5 minutes at 100 W. A wafer electroplating set-up (Yamamoto-MS) utilising cupric sulfate solution was used to build up a copper layer on top of the gold pads. The pads were temporarily shorted with edge card connector pins using acetone-soluble silver paint solution (TedPella) so that they were at the same electrical potential and attracted an equal deposition rate for the copper. The whole board was masked in Kapton tape to prevent plating in unwanted areas. The desired exposed area for plating was 0.7164 cm². Given a target current density of 20 mA/cm² to ensure a constant deposition rate, the current was limited to 14.33 mA, which required a 140 mV bias. The throw distance between the copper anode and the target PCB was fixed to 4.5 cm. The deposition rate was approximately 6.9 nm/sec for a target height of 12.45 µm after 30 minutes of plating. After plating, the entire board was placed in acetone for 10 minutes to remove both the resist mask, as well as the silver paint. To ensure the board was clean, an additional acetone clean, followed by an isopropanol wash, was performed. The entire PCB was then coated with approximately 3.3 µm of parylene using a SCS Labcoater 2 parylene deposition system. A 193 nm excimer laser (IPG Photonics IX-255) was then used to open holes in the parylene. The stabilization energy was selected to be 7 mJ, with
a pulse repetition rate of 100 Hz, with a 70 µm by 70 µm spot size, and variable attenuator angle (VAT) of 40 degrees (measured to give a fluence of 2.9 J/cm²). The laser was then set to raster over each pad, with an overlap of only 7.1% (65 µm per pulse), for a total of 8 passes. The result is a flexible PCB with pads that are now higher than the surrounding traces, and those traces are now insulated by Parylene C.

We note here that some boards were fabricated with the electroplating step bypassed entirely. In those cases, the pillar area was reduced to 425 µm by 425 µm to avoid shorting to board traces, which gives a lower fill factor of only about 18.06% and ultimately a lower ultrasound pressure.

Piezoelectric crystals are known to return to a non-piezoelectric state after exposure to thermal energy above their respective Curie temperatures. This loss in piezoelectricity, due to a permanent realignment of the electric dipoles in the material, is manifested in a drop in the piezoelectric coefficient \( d_{33} \). As such, maintaining a low temperature throughout the array fabrication process is necessary to maintain the electro-mechanical performance, obtaining optimal sensitivity and pressure generation for imaging. In addition, exposing additional surfaces of the PZT crystal by dicing, and treating the top and bottom surfaces chemically throughout lithography and metal deposition, may have unwanted effects on the piezoelectric coefficient. We used a Piezotest PM300 \( d_{33} \) meter to confirm the piezoelectric coefficient of our utilised material after each major fabrication step for the FlexArray. The samples were tested with a static force of 10.3 N and a dynamic force of 0.25 N, with a frequency of 110 Hz. As testing individual 825 µm by 825 µm pillars is impractical, a larger 1.6 cm by 1.6 cm piece of the same PZT-5H was used for the entire process. Three samples underwent processing independently, and each was measured three times after each stage, starting from a baseline measurement before any processing, i.e. fresh after purchasing from the manufacturer. The ACF tacking stage was emulated by heating to 90°C and applying
100 N on the sub-micron bonder for 20 seconds, undergoing the same thermal and pressure gradients experienced by the final array. The backside was diced down 500 µm (half of the height) using the same parameters as for making the actual array. Note that the topside was not diced, as the pillars would have come loose without a supporting substrate. The sample was then heated to 160°C under 375 N of force from the bonder to simulate the bonding procedure, without the substrate. Finally, the sample was sputtered with 10 nm of chromium and 1 µm of copper to test the effects of the top metal deposition on the $d_{33}$.

As seen in Fig. 3.12, the overall piezoelectric properties do not degrade throughout processing. The final array has a $d_{33}$ that is, on average, 5.25% higher than the initial value before fabrication. Note that the largest increase in the piezoelectric coefficient occurs immediately following the dicing procedure (17.97%). This is unlikely to be caused by thermal effects (which are expected to degrade the performance, as opposed to enhance it), but rather by the structural changes undergone by the material, i.e. the change of the aspect ratio and exposure of additional interfaces. After partial dicing, the resulting PZT may be analogous to a 1-3 piezocomposite, with air serving as the filing medium, rather than the standard epoxy.

**Board Design and System Interconnectivity**

The edge card adapter used here, MEC5-080-01-L-RA-W1-TR, has two rows of 500 µm pitch pads, totaling 160 connections. Of these, 128 per connector are used for carrying signal lines for the elements, leaving 32 for the return ground path. We can calculate the approximate current flowing through the ground network. If we assume we are using PZT-5H and the input impedance is approximately dominated by the capacitance, the input impedance of a single 825 µm × 825 µm × 1 mm element at 1.7 MHz is given as:
The piezoelectric coefficient increases throughout the fabrication process for the FlexArray, relative to the as-purchased large PZT crystals.

\[
C \approx \varepsilon_r \varepsilon_0 A \frac{3400(825\mu m \times 825\mu m)}{1mm} \approx 20.5pF \quad (3.1)
\]

\[
Z_{in} \approx \frac{1}{j\omega C} = 4.57k\Omega \quad (3.2)
\]

To reduce the technological development time, the Verasonics Vantage 256 system was used as the analog/digital front-end. While providing the design flexibility of being able to control most electronic parameters through MATLAB, it does impose certain restrictions on the design. Most important is the limitation of a maximum of 256 elements. While it is possible to multiplex multiple transducers to each signal pin to increase this number, such an approach is made impractical because it would require active multiplexing circuitry directly
Figure 3.13: a Custom-made board for interfacing the FlexArray with the Verasonics Vantage ultrasound system. b The edge-card adapter at the opposite end of the cable allows for direct plug-in to the FlexArray, facilitating rapid exchange of test boards and securing the electrical connections to the device.

on the board, and the Vantage system does not support an early trigger signal to allow safe separation of transmit and receive signal paths. An overview of the system integration with the Vantage 256 tool is shown in Fig. 3.15. The host controller is a PC running MATLAB 2020a, and has the ability of configuring a number of internal parameters, allowing for the customisation of the transmitted and received electronic signals. Moreover, the so-called "event" based system in the independent hardware and software sequences allows for asynchronous collection of data and arbitrary parameters to be set for each rayline. The image reconstruction algorithm is then performed locally, in real-time, on the host controller.
Figure 3.14:  

a Computer-aided design of the routing of the top layer of the flexible PCB.  
b FlexArray before processing.  
c Detailed cross-sectional stack-up of the FlexArray prior to processing.
Figure 3.15: **a** Block diagram for system and diagram for the *Verasonics Vantage* interface. **b** Connectivity diagram and number of channels for the FlexArray system.
Scanning Electron Micrographs

The scanning electron micrographs of the whole pillar array were taken before parylene encapsulation at a beam energy of 20 keV using the secondary electron detector, while the high-magnification images of the PZT surface were taken at 4 keV using the same detector. For energy-dispersive X-ray spectroscopy spectra collection, the stage was tilted at 15° towards the detector, using a 30 µm aperture at 20 keV.

3.4.2 Ultrasound and Electromechanical Testing of the Array

The array was first tested on a flat surface, mounted within a 3D-printed (Stratasys Connex3 Objet260), water-tight testing apparatus. These initial tests confirmed the operation of the array by producing beam patterns. Beam steering and focusing were independently controlled by means of setting the delays on the graphical user interface of the MATLAB software controlling the Verasonics Vantage system. An Onda HGL-0200 hydrophone was used in conjunction with a motorised stage to sample the pressure in a grid along the x-z plane, with a grid resolution of 1 mm. For the B-mode imaging of the curved gelatin phantoms, 85 g of uncoloured gelatin (Knox) were dissolved in 500 mL of boiling water before pouring this mixture into 3D-printed moulds of a pre-defined radius of curvature (in a range of $R = 1-3$ cm). After curing the moulds at 4°C for 3 hours, the gelatin phantoms were removed for testing. One set of moulds was always used as the bottom part – it was cast as a flat 5 cm by 5 cm by 2 cm block, with an air cavity located at the top center, with dimensions of 2 cm by 1 cm by 0.5 cm. In this way, the top half of the gelatin phantom could be easily changed out between measurements and offered an exact radius of curvature, while always measuring the same air pocket target. All tests were conducted on an optics table, where the FlexArray was manually wrapped around the curvature of the gelatin phantom.
to record each B-mode image.

### 3.4.3 Defining the Resolution of the FlexArray on a Medical-Grade Phantom

We defined the resolution in the lateral and axial dimensions by finding the half-power beamwidth (-3 dB) of the corresponding reflected echo along the horizontal and vertical axes, respectively. The theoretical axial resolution should be constant with depth, and depends on the quality factor, $Q$, wavelength of ultrasound, $\lambda$, and the number of pulses, $n$:

$$AR_{Theo} = \frac{Q\lambda n}{4}$$  \hspace{1cm} (3.3)

Similarly, given a depth, $z$, and array aperture, $L$, the theoretical lateral resolution as a function of depth can be given by:

$$LR_{Theo} = \frac{2\pi\lambda}{L}$$  \hspace{1cm} (3.4)

Experimentally, the B-mode was taken on the top surface of the 84-317 Multi-Purpose Tissue/Cyst Ultrasound Phantom (Fluke Biomedical), with the array laying flat ($R = \infty$), centered above the column of buried nylon strings, each separated axially by 1 cm. The transducer excitation voltage was ±10 V, with 2 pulses used to form the image.

### 3.4.4 In-vivo Imaging of the Human Humerus

Approval was obtained from the Columbia University Institutional Review Board to carry out ultrasound imaging experiments on volunteer human subjects (IRB no. AAAS6542). During imaging, the subject was conscious for the entirety of the procedure and was able
to provide feedback in real time. The FlexArray was placed directly on the skin at various locations on the arm, and standard ultrasound coupling gel was used (Aquasonic 100). The subject’s arm remained stationary as the scans were taken. The radius of curvature of the arm was measured at three different points, by matching to a 3D-printed mould of a fitting size. This R was then used for image correction during the scan. Images were taken with an applied voltage of ± 10 V to the transducers, which corresponds to a peak pressure of less than 200 kPa at the focal point.

3.5 Derivation of the Theoretical Minimum Radius of Curvature

It is difficult to define a practical metric for the flexibility of the final array in terms of strain, as different parts of the board and the array itself may experience varying strain depending on the topology of the imaged surface and materials used in each section of the finished device. Instead, quantifying a minimal radius of curvature before mechanical failure can serve as a more relatable guide for performance in the highly-curved limit. If rigid piezoelectric transducer elements are used, a major limiting factor in the flexibility of the overall array is the inevitable physical contact between the tops of the elements during extreme bending.

The most simple and practical case may be studied by considering cylindrical bending along a single axis, as demonstrated in Fig. 3.16a, with two elements of width \( a \), and fixed pitch \( d \). Assuming that \( d' \approx d \), that is, there is no buckling or deformation when the array is bent, the arc distance between elements across the bend ought to be preserved. If there were deformations, this likely would damage or destroy connecting circuitry, rendering the array nonfunctional. The height, \( h \), of the elements can be assumed to be the total height of
the pillar, including any matching layer on the top in more advanced designs. This can help
determine the compromise on flexibility that additional layers could incur. Furthermore, it
is assumed that the rectangular elements are rigid, thus width and height do not change
with curvature.

In Fig. 3.16a, the primed variables indicate the new flexed distances, which are dependent
on the radius of curvature, \( R \). For the pillars to physically not be in contact, the arc between
elements must be \( x' > 0 \). To find this arc length, the newly-formed segment, \( R' \), can be used. \( R' \) is given by the hypotenuse of the right triangle formed by the pillar and the radius of
curvature:

\[
R' = \sqrt{\frac{a^2}{4} + (R - h)^2} \tag{3.5}
\]

The inner arc length, \( x' \), can be defined as:

\[
x' = \theta_2 R' \tag{3.6}
\]

while the outer arc length, \( d' \), which is described by \( \theta_2 \) can be found by subtracting away
the arc segments labeled \( y' \) in Fig. 3.16a:

\[
y' = \theta_1 R \tag{3.7}
\]

Whilst the angles \( \theta_1 \) and \( \theta_2 \) are not explicitly known, they trigonometrically relate the arc
segment given by \( d' - 2y' \) to the radius of curvature:

\[
\theta_1 = \frac{0.5a}{R - h} \tag{3.8}
\]
Figure 3.16: a Diagram presenting the key variables which change when flexing in a single axis. b Graph of the theoretical minimum radius of curvature for various fixed element pitch values.

\[
\theta_2 = \frac{d' - 2y'}{R} \tag{3.9}
\]
where \( a \) is the width of the transducer elements. An expression for the arc lengths can then be found:

\[
x' = \frac{R'(d' - 2y')}{R}
\]

(3.10)

\[
y' = \frac{R(0.5a)}{R - h}
\]

(3.11)

Substituting the expressions for \( y' \) and \( R' \) into the definition of \( x' \) provides the resulting arc length and flexibility condition in terms of the known design variables (\( a, d, h \) and \( R \)):

\[
x' = \frac{1}{R} \sqrt{\frac{a^2}{4} + (R - h)^2} \left( d - 2R\tan^{-1}\left( \frac{0.5a}{R - h} \right) \right) > 0
\]

(3.12)

This transcendental inequality cannot be solved analytically to provide explicit expressions for each of the aforementioned design variables. However, most of these values are fixed because of physical constraints in real devices. For example, the element pitch is limited by the compromise between grating lobes and resolution while the element height is restricted by the resonant frequency of operation. Element width, in turn, has an effect on output pressure, and thus the SNR, so it needs to be optimised to a single value. As such, \( x' \) can be solved numerically for realistic device cases. In Fig. 3.16b, we compute the theoretical minimum radius of curvature that is achievable with an element pitch of 1 mm. Note that these are geometrical limits, and may be further limited by the rigidity of the chosen substrate.

For all cases tested in this work, \( a < 825 \mu m \) and \( h > 1 \) mm, so the theoretical minimum radius of curvature achievable with the FlexArray is \( R < 1 \) cm.
4 Laser Micromachining of PVDF

4.1 Introduction

Piezoelectric devices are used in a variety of applications from simple microphones, to actuators, to more complex ultrasonic transducers. Traditionally, piezoceramics such as lead zirconium titanate (PZT) have filled these roles because of their high electromechanical coupling factor. However, piezoelectric polymers have recently gained attention by bringing unique material properties that provide advantages for many applications.

In comparison to ceramics, PVDF has the ability to flex and conform to a shape. This is useful in biological application to either make good contact with the curvature of the skin, or to minimize foreign-body responses when implanting a PVDF device. Moreover, PVDF is biocompatible, chemically inert, and lead-free.

PVDF has an acoustic impedance of 2.5 MRayls in comparison to 33.7 MRayls for PZT-5H. [42] This implies that if a transducer made of each of these materials is transmitting or receiving directly into water, 6.6% and 85.1% of the pressure from PVDF and PZT-5H are reflected, respectively. The piezoelectric coupling coefficient, $k_{33}$ of PVDF is only 0.27 compared to 0.69 for PZT. Thus, most of the PZT signal is lost if a matching layer is not added (requiring additional fabrication stages). [36]
4.1.1 State of the Art PVDF Machining Methodologies

As a fluoropolymer, PVDF is inherently difficult to micromachine. It is chemically inert, making it resistant to most wet etching processes. In order to maintain the piezoelectric properties of $\beta$-phase PVDF, processing temperatures must be kept below 80$^\circ$C. This requires the development of non-traditional photolithography techniques in order to create photoresist masks for selective etching.

Han et al reported high etch rates of PVDF using N,N-dimethylacetamide (DMA). At 40$^\circ$C, an etch rate of $\approx 10000$ nm/min was achieved. [83] This increases to 40000 nm/min at 50$^\circ$C. Unfortunately, this etch is isotropic, making high aspect ratio features impossible. Moreover, etch mask selectivity can be a problem, as strong organic solvents attack both photoresists and bonding materials easily.

Dry etching using reactive ion etching (RIE) techniques allows for a degree of anisotropicity, which is necessary for tall, high aspect ratio pillars commonly found in many applications of piezoelectric materials. Miki et al achieved an etch rate of 167 nm/min with an etch chemistry of O$_2$. [84] Later experiments involving increased CF$_4$ concentrations allowed for etch rates as high as 667 nm/min. However, this higher vertical etch came at the cost of a horizontal etch rate of 1333 nm/min, making high aspect ratio features impossible. [85] Lower rates were obtained with a similar O$_2$ etch chemistry at a lower power.[83] Substantially higher etch rates of near 770 nm/min were reported when using a combination of CF$_4$ and O$_2$ chemistries. [86] While the features show sharp sidewalls, the comparatively slow etch rate prohibits the creation of tall features greater than 100 $\mu$m in height. In addition to the gas species concentration, the masking material has a significant effect on sidewall angle. Shi et al used a photoresist mask to obtain a gentle sloped sidewall for easier metal deposition for the top metal contact of fabricated features. [87]

Attempts have been made to micro-pattern features into PVDF using laser irradiation.
Lee et al used a 775 nm Ti:Sapphire femtosecond laser to etch metalized PVDF. At this long wavelength, gas bubbles formed from released $H_2$ during polymer decomposition, and significant thermal effects were noted. [88] Liu et al used a 248 nm laser to selectively increase resistivity through surface modification. However the fluence of the particular laser was not high enough for machining and separation of the PVDF into freestanding pillars. [89]

Success has been shown using extreme ultra-violet (EUV) sources to etch PVDF in a clean and controlled manner. Bartnik et al used a 10 Hz, 11 nm laser-plasma source to focus a 60 mJ/cm$^2$ beam on PVDF, producing very clean sidewalls for depths of 50 µm. Unfortunately, the etch rate is very slow at 70 nm/pulse, or 42000 nm/min for a 60 µm by 60 µm spot. [90] Such a method provides excellent sidewall quality and resolution, but EUV sources are rare, and the etch rate may be prohibitively slow for tall features.

### 4.1.2 Excimer Laser Ablation

Excimer lasers operate in the UV spectrum, providing high photon energy to break chemical bonds. Unlike traditional lasers that use thermal ablation mechanisms to vaporize a material, excimer lasers use a photoablative effect to break chemical bonds, allowing for the removal of the resulting smaller particles. [91]

ArF lasers emitting at 193 nm are regularly used to cleanly ablate glass, ceramics, and organic tissue with superior quality in comparison to high-power machining lasers at longer wavelengths. Moreover, PVDF has a higher absorption coefficient at shorter wavelengths, as shown in Fig. 4.1, allowing for efficient use of the beam energy.

The two major mechanisms used to describe the laser removal of material are photothermal ablation (often referred to as thermal ablation), and photochemical ablation (often referred to as photoablation). In thermal ablation the absorbed energy penetrates to a given depth determined by the material’s absorption coefficient, causing the material to heat up, ul-
Figure 4.1: Measured UV-Vis absorption spectrum of β-phase piezoelectric PVDF.
timately vaporizing. Provided that the photon energy is greater than the bond strength, photoablation may occur. This process uses the photon energy directly to break chemical bonds, vaporizing the shorter polymer molecules.[91] As some heat is still transferred to the surface, most ablation processes contain some thermal effects.[92] An advantage of ArF lasers is the high photon energy of about 6.42 eV, in comparison to 1.6 eV for a femtosecond Ti-sapphire femtosecond laser. [88, 91] This high photon energy allows for the breaking of C–C (3.90 eV), C–H (3.51 eV), and C–F (4.99 eV) in PVDF without resorting to melting the material, as one would find with lower energies. [93, 94]

Although surface modification and limited etching can occur at low energy fluences, polymers begin to ablate quickly above a fluence value known as the threshold fluence. This value depends on the material as well as the applied wavelength, but tends to fall in the low mJ/cm² range. For example, the ablation threshold of polyimide at 193 nm is 15 mJ/cm².[92] As the fluence increases, a plasma plume is generated over the surface. This plume can partially shield the material underneath it, preventing higher fluences from linearly increasing the etch rate. If there is not enough energy to vaporize a material, some may melt, and the pressure generated by the plasma plume above may result in the resettling of this molten material in the newly generated crater. This is generally undesirable, as it results in a rough surface, where clean material removal is desired. Below this, is the so-called heat affected zone (HAZ). The depth of this zone is $\sqrt{D \tau}$, which, for PVDF, given a thermal diffusivity of $D \approx 10^{-3}$ cm²/s, should only be $\approx 44$ nm. [95] In reality, this is much larger due to the addition of thermal ablative effects from the residual energy not being used to break bonds. A conceptual illustration of this combined material removal process and the shielding plasma plume is shown in Fig. 4.2.

As excimer lasers rely on the excitation of a pair of rare gas halides, a limiting factor of excimer lasers is the $\approx 10$ ns excitation duration before the gas species return to a relaxed
Figure 4.2: Demonstration of laser ablation and the resulting plasma plume on the surface of a material. Figure is not to scale.
This fundamentally limits the pulse duration to orders of magnitude longer than femtosecond lasers. A longer pulse duration allows for greater heating of the machined material, possibly causing unwanted melting and surface chemical changes. This effect is mitigated by greater coupling to the material at lower wavelengths, resulting in more material decomposition and less melting.

4.2 Experimental Etching Characterization

A 2 cm by 2 cm piece of 110 µm, pre-poled, β-phase PVDF was cut from a commercially purchased sheet (Kureha). This piece of PVDF was measured to have a $d_{33}$ piezoelectric coefficient of 21 pC/N using a PiezoTest PM300 $d_{33}$ meter prior to any processing. We can use the relation $k_{33} = d_{33}/\sqrt{s_{33}\epsilon_{33}}$, where $s_{33}$ is the elastic compliance and $\epsilon_{33}$ is the dielectric constant of the PVDF, to find the $k_{33}$ to be 0.11.

To secure the sample to a flat working surface, a polished 300 µm-thick Si wafer was used as a substrate. The side of the PVDF to be bonded was pre-treated in an Anatech Plasma asher flowing 100 sccm of O$_2$ at 100 W of forward power for three minutes. This chemically activates the surface of the fluoropolymer making it amenable to permanent bonding. Microchem SU-8 3005 was directly applied to the Si wafer and spin coated at 4000 RPM for 45 seconds to provide an approximately 5 µm-thick layer. This was then exposed in a Karl Suss MA6 mask aligner at 100.8 mJ/cm$^2$. The plasma-treated side of the PVDF is placed face-down onto the SU-8 coated Si wafer. A Finetech Lamda Fineplacer is then used to apply heat and pressure to finish the cross-linking process. The Fineplacer applied 68.75 N/cm$^2$ of pressure while simultaneously heating the stack to 70°C for 30 minutes. This fully cures the SU-8, creating a permanent bond to the wafer and a flat PVDF surface for laser ablation.
An IPG IX-255 excimer laser system was used to etch the PVDF. The system uses an ArF 193 nm laser capable of providing fluences in excess of 25 J/cm² for pulse durations of approximately 10 ns. To characterize the material, test grids were exposed, where a constant pulse repetition rate and spot size were used for each grid. Two sets of apertures were used for this experiment, resulting in a 50 µm by 50 µm spot size, and a 70 µm by 70 µm spot size. Both are formed by masking the beam through a rectangular variable aperture (RVA). The smaller spot size is more uniform across the entire beam area, but the larger spot size can provide faster etching when cutting lines or rastering areas. This experiment was repeated for pulse repetition rates of 50 Hz, 100 Hz, and 500 Hz.

Each of the test grids had 6 rows and 31 columns. The rows correspond to the number of test shots: 5, 10, 25, 50, 75, and 100 shots respectively. In this context, a shot is a single pulse of approximately 10 ns that is fired by the laser. When multiple shots are fired, they are striking the same area and effectively drilling down into the material, leaving a square hole. At high pulse repetition rates, the higher shot counts will exhibit greater thermal effects. Each column represents a different energy fluence value, modified by changing the variable attenuator (VAT). The VAT is a coated optical blade that is partially inserted into the beam path to attenuate the fluence based on the amount inserted. The VAT angle was varied from 30° to 45° (corresponding to 21.2 J/cm² down to 2.2 J/cm²), respectively, in steps of 0.5°. The corresponding power is then measured using an Ophir power meter to calculate the energy fluence of each point.

Many etching applications require the machining of lines or traces as opposed to individual points. To help characterize the machinability of piezoelectric PVDF using a 193 nm laser, additional test grids were made of 500 µm-long lines, using the 50 µm by 50 µm spot size. For the case of linear exposures, the concept of shot overlap is introduced to control edge quality. Here, overlap is defined as the percentage of the area a previous shot shares with
Figure 4.3: a) The normalized energy of each energy pulse. The spacing between pulses is determined by the repetition rate, $f$. b) Top view of a single and two consecutive shots. If the beam width is given by $x$ and the overlap is given by $\Delta x$, then the overlap percentage is $OV\% = 100\left(\frac{\Delta x}{x}\right)$
the area of the next pulse. Drilling holes has an overlap of 100% because each shot is exactly placed over the area of the last one without moving in the lateral dimensions. The temporal pulse duration and the overlap ratio is illustrated in Fig. 4.3.

For this test grid, each row corresponds to the amount of overlap between shots as opposed to a number of shots. All lines were scanned with an energy density of 8.32 J/cm². The overlap varied from 0.5 µm/shot (99% overlap) to 10 µm/shot (80% overlap).

Direct measurement of the etch depth of the holes in the PVDF is complicated by the aspect ratio and rough surface quality at the bottom of these etched areas. Surface morphology is an inexact determinant. Depending on the fluence and number of pulses fired, the surface could be scarred by incubation heating, i.e. no perceivable etched depth, but the surface is roughened by 1-5 µm. In the other extreme, the etched hole could be deeper than 110 µm, implying the laser removed all of the PVDF and some of the underlying cured SU-8 and silicon. The bottom of these holes can have a surface roughness comparable to the under-etch case, depending on the repetition rate. Measuring these 50 µm by 50 µm cavities with a comparably sized profilometer tip leads to the probe bottoming out and an underestimate of the actual depth. Optical profilometry fails to measure this correctly as the intensity of light that returns from each hole is over 1000 times weaker than that reflected by the surface because of scattering of light at the bottom of each hole.

To circumvent these issues, the etched pillars were measured indirectly by creating an inverse Sylgard 184 polydimethylsiloxane (PDMS) mold of the etched structures. A 3D-printed 2 cm by 2 cm hollow square enclosure was used to create a dam for the PDMS. It was mounted directly on the PVDF sample, centered around the test area, and the bottom was sealed using silicone to prevent PDMS from leaking. The fenced-in area was then filled with approximately 4 mL of PDMS, desiccated for 30 minutes and allowed to cure at 70°C for 2.5 hours. The PDMS was then manually removed from the mold to produce a freestanding
2 cm by 2 cm by 1 cm PDMS block with pillars imprinted in the same shape and size of the holes on the PVDF. The heights of each of these pillars were then measured using a Bruker Dektak-XT stylus profiler.

4.3 Results and Discussion

For a pure photochemical ablation process, the expected etch rate should follow the Beer-Lambert law:

\[
h_e(F) = \begin{cases} 
0 & F < F_{TH} \\
\frac{1}{\alpha} \ln\left(\frac{F}{F_{TH}}\right) & F \geq F_{TH}
\end{cases}
\] (4.1)

where \(h_e\) is the etch rate in microns per pulse, \(\alpha\) in cm\(^{-1}\) is the absorption coefficient of the material, \(F\) is the applied fluence in J/cm\(^2\), and \(F_{TH}\) is the threshold fluence.

The optics path of the beam delivery system for the laser workstation was not configured to apply low fluence pulses. This precludes testing close to the ablation threshold of PVDF, which is expected to be on the order of hundreds of mJ/cm\(^2\). For example, at 775 nm, Lee et al[88] found the so-called “damage threshold” of PVDF to be 156 mJ/cm\(^2\). The poor coupling of the beam to the PVDF is even seen at 193 nm. For smaller 50 µm by 50 µm spot sizes, the surface morphology is rough at lower repetition rates where thermal effects are less of a factor. The bottom surface of these rough trenches translates into rough pillars after making the PDMS mold. These data points have the greatest variance.

4.3.1 Etching Single Spots

As seen in Fig. 4.4, the measured etch rate of PVDF follows the Beer-Lambert law, and starts to become more linear far from the threshold fluence. One can create a logarithmic
Figure 4.4: Etch rate of PVDF for a 50 µm by 50 µm spot size for various pulse repetition rates.
fit based on Eq. (4.1), and extract $\alpha$ and $F_{TH}$. If one extrapolates the threshold fluence from the fit, the average threshold fluence when using a 50\,\mu m by 50\,\mu m spot size is 467 mJ/cm$^2$, which is considerably higher than reported for 775 nm. The absorption coefficient was measured to be 1.69 cm$^{-1}$, but this simplified model does not take into account the change in absorption under irradiation.

As a smaller spot size is produced by masking the main beam, it should provide a more uniform beam profile. However, the resulting smaller pillars had a consistently lower etch rate than for a 70\,\mu m by 70\,\mu m spot size, as shown in Fig. 4.5. Here, the average threshold fluence was 219 mJ/cm$^2$, with an absorption coefficient of 1.73 cm$^{-1}$. The larger spot size increased the etch rate by approximately 25%. This suggests that smaller spot sizes with higher aspect ratio holes are rate-limited by the ability to remove material from the narrow trench. It is also important to note the large increase in etch rate between the 100 Hz and 500 Hz repetition rates, indicating a transition to a more thermal ablative mechanism as there is less time to cool between pulses.

### 4.3.2 Etching Lines

Figure 4.6 illustrates the increase in etch rate with increasing levels of shot overlap. As expected, greater overlap will expose a given area to a greater number of shots, thus increasing the total amount of material removed per pass.

A higher pulse repetition rate is often desirable to improve throughput in patterning large areas. In Fig. 4.7, one can maintain ablation rates at 100 Hz that are comparable to those achieved at 50 Hz. One important outlier is the 99\% overlap scenario, implying that the close pulse spacing of 0.5\,\mu m is dense enough to start to cause thermal ablation or melting. A closer examination of the surface topology is presented later in this section.

Increasing the pulse repetition rate even further to 500 Hz, as shown in Fig. 4.8, demon-
Figure 4.5: Etch rate of PVDF for a 70 µm by 70 µm spot size for various pulse repetition rates.
Figure 4.6: Etch rate vs energy density with a 50 µm by 50 µm beam at 50 Hz for various levels of shot overlap.
Figure 4.7: Etch rate vs energy density with a 50 µm by 50 µm beam at 100 Hz for various levels of shot overlap.
strates that the etch rate rapidly increases. For high overlap ratios, the entire 110 µm-thick PVDF sheet was completely etched through, even at moderate fluence values of 14 J/cm². However, at lower overlap ratios, a more linear increase in etch rate with fluence is observed. This may also be a result of the now-moving plasma plume generated across the surface of the etched material. With close overlap, the force exerted by the plume directly above the irradiated area may prevent material from escaping the ablation zone. This results in the area retaining heat, causing additional melting in the region.

One of the main advantages of using an excimer laser for PVDF ablation is the trade-off achievable between etch rate and edge quality. As previously mentioned, dry etching through RIE can provide vertical sidewalls, but the throughput is severely limited by the poor etch rate. Faster etch rates can be obtained with wet etching, but patterning fine features is difficult. Lasers at longer wavelengths burn the PVDF, depolarizing it. As seen in Fig. 4.9, a compromise can be made with an excimer laser between the overlap amount and the quality. For the purpose of comparison, a fixed fluence of 8.32 J/cm² was chosen as a point well above the threshold fluence. Similar results are seen for various fluences, corresponding to the previously demonstrated etch rates.

High repetition rates allow for faster processing, resulting in more removed material per pass. However, as the material is removed in discrete steps this may lead to overshooting on a specific target depth. This can be mitigated by doing several passes at higher repetition rates, to remove most of the material, and slowing to a lower etch rate as finer depth control is needed. While an increased repetition rate and overlap percentage provide higher edge quality, it allows less time for cooling in the localized area, potentially causing melting on the sidewalls. This reflow of material is likely the cause of the smooth wall profile at 99% overlap.

For low levels of shot overlap, a coarser line texture is observed. This can be partially
Figure 4.8: Etch rate vs energy density with a 50 µm by 50 µm beam at 500 Hz for various levels of shot overlap.
Figure 4.9: SEM images of single passes of a 50 µm by 50 µm beam at 8.32 J/cm². The approximate processing time for the given conditions is shown in each panel. The scale bar applies for all panels.
overcome with multiple passes, which effectively mimics closer pulse spacing. Closer pulse spacing provides an advantage of a deeper trench, and more vertical sidewalls. Moreover, increasing the repetition rate can lead to localized melting near the edges of the PVDF. This happens on the micron scale, which does not affect the PVDF tens of microns from the ablation area, but does provide a smooth surface where the fibrous texture of the PVDF is less noticeable. Moreover, higher repetition rates generally translate to greater throughput.

4.3.3 Creating Square Features

In the previous section, we have shown that it is possible to both drill holes and cut trenches in the PVDF. This can further be expanded to create features of arbitrary shapes. Given the small beam size compared to a working PVDF surface that could potentially measure several centimeters, it is more efficient to simply trace the outline of proposed features as opposed to rastering away the entire unwanted area.

An approximately 1 µm-thick layer of Perminex 2001 adhesive was spin coated at 3000 RPM for 30 seconds on a 1 mm-thick glass slide. Perminex was used to improve bonding, and thus improve pillar yield. [97] The solvent was evaporated by a bake at 70°C for 5 minutes. This substrate was then exposed using the MA6 mask aligner with a photomask that exposed only 425 µm by 425 µm squares arranged in a 16-by-16 grid. The PVDF was exposed to the same O₂ plasma treatment as the previous experiments to improve adhesion. It was then bonded to the Perminex sample using the Fineplacer with 100 N/cm², while baking the stack at 70°C for 2 minutes. The excimer laser was then used to trace around the perimeter of the exposed squares, as the Perminex should only form bonds in these areas. In all cases, multiple passes were required to cut completely through the PVDF. After each pass with given parameters, the beam was manually focused down onto the newly exposed surface in the trench. The process was repeated until the PVDF was entirely etched through. After
etching, the remaining bulk PVDF was manually peeled off with tweezers, and the resulting pillars were cleaned with acetone.

Selected features are shown in Fig. 4.10. As expected, both higher repetition rates and greater overlap ratios resulted in greater melting of the material. Lower repetition rates resulted in voids opening along the length of the PVDF, as seen in Fig. 4.10 (a) and (d). Conversely, at a repetition rate of 500 Hz, substantial thermal damage and redeposition of material is apparent, illustrated by Fig. 4.10 (g) and (j). Lower overlap ratios were not included as strands of PVDF would remain, requiring over 12 passes per pillar. Even under these conditions, those pillars would not always survive the peeling process as fibers of PVDF still would hold the feature to the excess PVDF.

As with etching lines, increasing pulse repetition rate and decreasing the overlap allows for an individual pass to be completed in less time. However, this does not always equate to greater throughput, as less material is removed per pass. The etch rates determined in Fig. 4.6 and Fig. 4.8 can be used to accurately approximate the amount of etched material per pass. For example, using a 96% overlap at 100 Hz as in Fig. 4.10 (a) requires 5 passes to go through all 110 µm of PVDF, but using 99% overlap at 500 Hz as in Fig. 4.10 (j), only requires 2 passes. In all cases, there was some level of overshoot that was intentionally done to ensure that the final pass went entirely through the PVDF. To avoid damaging the substrate, it may be advantageous to etch the final steps at a lower power, but this would require tuning to the specific application. In terms of having the most vertical sidewalls with the least debris, a repetition rate of 100 Hz with a 99% shot overlap ratio is preferred. However, a 7.5-fold increase in throughput can be achieved with minimal decrease in quality by using a 250 Hz repetition rate with a a 96% shot overlap ratio. Higher repetition rates cause excessive melting and damage to the PVDF.
Figure 4.10: SEM images of 425 µm by 425 µm PVDF features etched using a 50 µm by 50 µm beam at 8.32 J/cm². The approximate processing time for the given conditions is shown in the bottom-left corner of each panel. The percentage of shot overlap for each row is shown to the left, the repetition rate used for each column is shown at the top. The scale bar applies for all panels.
4.4 PVDF Laser Etching Conclusions

A 193 nm ArF excimer laser was used to ablate piezoelectric PVDF. The potential for the use of such a laser for the patterning of micromachined piezoelectric transducer structures was investigated. This was accomplished by the characterization of the etch rate and fitting to a Beer-Lambert model, allowing for the extraction of an approximate absorption coefficient and threshold fluence for the material.

Both individual shots and etched lines were separately investigated. Individual shots allow for the more direct characterization of the laser interaction with the material. Practically, this allows for the drilling of high-aspect ratio holes, potentially for electrical vias or other connections. Etched lines are shown to provide a guide to the expected etch rate for various overlap ratios, energy densities, and pulse repetition rates. This information was then applied to the patterning of square PVDF features. These could then potentially be used to form individual transducers or cantilevers. A trade-off between edge quality and fabrication throughput was noted to depend on pulse repetition rate.
5 Future Work

In this work, I have shown the feasibility of producing a flexible ultrasound phased array with curvature correction as a proof of concept. However, there remain many areas that can be expanded upon to enhance usability, which may hopefully lead to a practical commercial product. In addition to general code improvements, such as improving frame rate by reducing processing time, the system as a whole can be expanded upon to improve versatility. For example, while the electronics in this project were located entirely on the Vantage system to reduce development time, these could be moved to a separate PCB. Moreover, designing this PCB will allow for further miniaturization by developing application specific integrated circuits (ASICs) dedicated to specific phasing operations.

5.1 Utilization of Discrete Electronics

As of the time of writing, this flexible ultrasound phased array utilizes the Verasonics Vantage 256 system. While this unit provides all of the front-end hardware (described in Fig. 3.15) and is highly configurable, it is extremely large at 48.9cm × 28.0cm × 47.6cm and weight of 19.1kg [98]. Fig. 5.1 shows a potential expansion of this project, where the front end electronics and phasing are moved to COTS components mounted on an external PCB.

A Field Programmable Gate Array (FPGA) can be used to generate clock signals, as
well as perform the majority of the associated digital logic. By operating at a multiple of the driving frequency, it is possible to utilize counters to dynamically control the delays to individual elements. Such a method is a convenient programmable technique for controlling phase delays for ultrasound signals [99, 100]. These signals can be fed to driving circuits to allow for high-voltage signals to appear on each element.

For the receive mode, commercial amplifiers, anti-aliasing filter, and ADC components can be used for groups of channels. Depending on the number of elements and the operational frequency, multiplexing may be required. Each channel can be placed in a shift-register. The phase delays have already been locally calculated and produced by the FPGA during the transmit mode, so the number of shifts for each channel is precisely known. The registers
can then be summed to produce a single data stream (representing one ray line) that can be streamed serially for display on a host controller PC.

As the electronic components require a large footprint, it would be impractical to directly incorporate them on the flexible array. Instead, these could be soldered to a separate board that communicates with the array by means of an edge card adapter. Such a method would allow for the array to become an interchangeable part, such that it can be easily replaced while using the same electronics board.

5.2 Application Specific Integrated Circuit Scalability

After migrating the front-end electronics to the a purpose built PCB, the next logical advancement is the replacement of COTS components with ASICs. Building the system with currently available discrete components allows for function verification before further miniaturization is attempted. The majority of the driving, phasing, filtering, and sampling circuits can be integrated directly on ICs mounted to the flexible substrate, as illustrated in Fig. 5.2.

To maximize flexibility, the ICs will need to be thinned, a process that is seeing significant attention in the field of ultrasound integration [101, 102]. If the elements are to be directly integrated onto the thinned ICs (as opposed to the package), this may require a series of through silicon VIAs (TSVs) before direct flip-chip bonding [103]. Traditional wirebonding is undesirable because the rigidity of the potting epoxy that encapsulates the wires. An viable alternative to consider would be the use of ACF to directly bond the chip to the flexible package.

The major advantage of direct ASIC integration close to the individual elements is scalability. By mounting transducer elements close to the piezoelectric pillars, the parasitic capacitance of the routing traces can be reduced, thus improving signal quality. Moreover,
Figure 5.2: Rendered image demonstrating a flexing PCB with several mounted, thinned IC controllers.

The array can be subdivided into sub-arrays, each controlled by a local IC. These local ASICs directly control the phase of each member of the corresponding sub-array. Note that the phasing as well the delay-and-sum algorithms used for traditional phasing are completely independent from channel to channel. This independence allows these ASICs to combine the received information into a signal serialized data stream. The resulting data stream from each local sub-array can then be summed to produce the final scanline. Instead of having $N^2$ traces escaping each sub array, the routing can be localized and only control lines and a data line need to escape the full array. This drastically decreases the total routing and allows tiling of several smaller sub-arrays, as shown in Fig. 5.3.

These sub-array ICs are all identical, but are slaved to a signal master IC that provides a
Figure 5.3: Rendered image where slave ICs can control a local region of elements and communicate to a single master IC.

synchronized clock as well as the phase information for each sub-array. This phase information can be communicated by a phase code for each element that is pre-loaded into registers of the corresponding slave IC prior to the start of each scan-line. The master is responsible for producing the appropriate phase codes for each sub-array, combining the data received from each slave, and passing this information onto a host controller for display.
6 Conclusion

In this work, I have designed, developed, and tested a flexible ultrasound phased array. As sensors become more compact and biocompatible, the evolution of fixed, rigid transducers to conformal patch arrays is a logical progression. In exploring the fabrication methods to produce the array, I characterized a novel method of processing PVDF. Ultimately, I selected to use PZT to achieve higher pressures, and produced several arrays on a custom flex-PCB. I designed, simulated, and tested a phase correction algorithm to eliminate phase aberration caused by bending.

6.1 PVDF Microfabrication

My initial designs for the flexible phased array called for a flexible piezoelectric. Ultimately, I found that spacing rigid, PZT transducers that were sufficiently apart were compatible with a conformable patch. As discussed in Chapter 4, the micromachining of PVDF is difficult, making it an underutilised bulk transducer material. In exploring methods to create individual pillars, I found plasma etching to be prohibitively slow, and mechanical cutting to be difficult control without damaging the PVDF or substrate. In this work, I illustrated how PVDF could efficiently be machined by a 193nm excimer laser, allowing for arbitrary array shapes. By varying the fluence dosage, pulse repetition rate, and overlap ratio, I
characterized the etch rate such that future researchers may have another tool available to produce their own PVDF transducers. This work on characterizing the etch rate of PVDF with an excimer laser is published in the Journal of Vacuum Science [104].

6.2 State of the Flexible Array

As demonstrated in Chapter 3, the flexible array is capable of conforming to a human arm, and producing a B-mode image with phasing correction for the curvature. At this stage of development, only cylindrical curvature in the XY, YZ, or both planes are currently considered, but future iterations could expand the algorithm to arbitrary shapes. The current iteration utilizes the Verasonics Vantage 256 system, limiting portability; however, I include a GUI so that the user may change parameters in real-time such that the type of curvature, the expected radius of curvature in each dimension, the focus, and the current YZ-slice. The fabricated arrays function without a matching layer, still being sensitive enough to detect reflectors such as the humerus. While there remains significant areas for improvement, I have successfully built and demonstrated the functionality of a flexible phased array ultrasound transducer that can correct for curvature in both transmit and receive modes. At the time of writing, this flexible array with adaptive phasing has been submitted for publication.

6.3 Additional Publications

In the process of completing this research, I also participated in several other projects that led to publications. These primarily focus around the fabrication of PZT transducers and the investigation of PVDF material properties.

I worked with Chen Shi to explore the possibility of creating an implantable ultrasonic
tag with integrated complementary metal-oxide semiconductor (CMOS) circuitry. Such a
device is to be able to take advantage of 0.18 \mu m 1.8V/5V CMOS sensors, yet be wirelessly
powered by harvesting ultrasonic energy. Initial experiments examined the feasibility of
directly integrating bulk PVDF or PZT transducers. I contributed by helping develop both
fabrication flows. We used 28 \mu m PVDF, which was patterned using reactive ion etching.
The PZT fabrication flow used for this project was very similar to that used for the flex
array, except limited to a single element [105]. This work would be extended to create an
integrated 380 \mu m by 300 \mu m by 570 \mu m monolithic temperature center, powered by an 8.1
MHz ultrasound probe. The implanted wireless device would use the single PZT transducer
to harvest energy from the commercial imaging probe. The sensor integrated into the device
would then modulate the load impedance of the resonator, transmitting the data back to
the imager by modulating the backscattered ultrasound [106].

While investigating PVDF as a potential transducer for the flex array, I would measure the
\( d_{33} \) after processing steps to ensure the heating did not degrade the performance. However,
I noticed short term improvements to the piezoelectric coefficient after apply short one-
minute anneals, commonly found in most photolithographic processes. I worked with Jeffrey
Sherman and Jakub Jadwiszczak to explore the relationship between the exposure of PVDF
to short temperature anneals and the resultant change in \( d_{33} \). We show that temporary
improvements in \( d_{33} \) of up to 40\% for anneals in excess of 70°C decay to a settling point
within hours [107].
References


Appendix A: Design Parameter Details

The design of the phased array refers primarily to the rationale of the performance compromises and the physical fabrication of the elements. In the interest of rapid prototype development, the electronics from the Verasonics Vantage system are used to interface with the developed flexible array. This provides the ability to adjust programming for the phasing paradigm described in a later section.

In this section, we examine the overall board system and the choice of materials. We then explore design requirements and trade-offs between resolution and the appearance of grating lobes. These are, in turn, related to the operating frequency and maximum operating depth. Flexibility is explored from a geometric standpoint, providing theoretical limits to the tightest possible radius of curvature that can be achieved. In addition to the center frequency of operation, \( f_c \), we must choose an appropriate element size, \( a \), pitch, \( d \), and number of elements, \( N \), to achieve the desired performance.

A.1 Resolution

The resolution of a transducer is defined as the minimum distance between two point targets that can be distinguished. A low resolution lead would lead to a blurry image. Resolution can be defined in the axial and lateral directions separately. An example of the axial and lateral resolution in a single plane of a conventional B-mode image is illustrated in Fig. A.1.

Axial resolution is defined as the minimum distance two objects can be resolved in the direction of beam propagation. As such, this resolution is dominated by the ability to
Figure A.1: Illustration of axial vs lateral resolution

distinguish a pulse from additional ringing in the echo. The limit on the axial resolution is half of the pulse duration, as given in Eq. 1. This, in turn, is given by the system quality factor, $Q$, and the wavelength, $\lambda$. Many transducers use 2 to 3 pulses for imaging to improve the transmitted power as a means of counteracting the effects of attenuation [108]. The resolution is then worsened by the number of pulses, $n_{\text{pulses}}$. It follows that the only effective means of lowering the axial resolution of a transducer array is to lower $Q$ or $n_{\text{pulses}}$, as the operating frequency (that controls $\lambda$) is tied to other parameters such as the appearance of
grating lobes. Reducing $Q$ requires damping the ringing of the transducer by the use of a backing layer or using a piezoelectric with greater bandwidth, such as PVDF.

$$AR = \frac{\text{PulseLength} \times n_{\text{pulses}}}{2} = \frac{Q \times n_{\text{pulses}}}{4\lambda}$$ \hspace{1cm} (1)

Axial resolution is defined as the minimum distance two objects can be resolved perpendicular to the direction of beam propagation. For 1D arrays, the resolution in the plane orthogonal to the defined axial resolution and lateral resolution is referred to as the elevational resolution and is determined by the element length in that direction. However, for a 2D array, the resolution in this dimension is equivalent to the lateral resolution. The lateral resolution of an array is dependent on the focal length, $l_f$, and the total array length, $L$, as defined by Eq. 2. Geometrically, the length of the array is given by $L = N(d - 1) + a$. This implies that the ability to focus in the lateral dimension is diminished as the distance from the transducer increases.

$$LR = \frac{2l_f\lambda}{L}$$ \hspace{1cm} (2)

### A.2 Grating Lobes

An array radiating into space can be described by the contributions made by the main lobe, side lobes, and grating lobes. The width of the main lobe defines the above lateral resolution of the array. Increasing the length of the array allows for a tighter focus, narrowing the width of the beam. Side lobes are spurious beams in both the transmit and received mode that occur away from the target angle. These side beams are always weaker than the desired main lobe. Grating lobes are a phenomenon unique to arrays and create repeat images of the main lobe. An example of effect of the number of array elements on beam pattern is
shown in Fig. A.2.

Figure A.2: Example radiation pattern for a linear array of isotropic radiators with 1mm pitch.

As seen in Fig. A.2, increasing the number of elements (effectively increasing the total array aperture $L$) narrows the main beam and reduces the intensity of side lobes. However, grating lobes appear in the same relative position. Further compromises can be made to improve routing between elements with so-called “sparse” arrays. Such arrays have missing rows or columns of elements in a periodic grid at the cost of additional lobes [109, 110]. Side lobes and grating lobes can create a superimposed image over the central beam, resulting in a form of spatial aliasing. Grating lobes occur in locations where the spacing between
elements creates constructive interference. It is then possible to calculate the location of grating lobes by Eq. 3, where \( n \) is an integer [26]. If the angle of the first grating lobe is \( \geq 90^\circ \), then it is outside the field of view and will not affect the image.

\[
GL = \sin^{-1}\left(\frac{n\lambda}{d}\right)
\]

The shape of the individual elements can change the radiation pattern, compromising the main lobe width for side-lobe suppression [111, 112]. However, the use of mechanical dicing limits us to rectangular transducers. A laser would allow for arbitrary shapes, but preliminary experiments showed slow etch rates at 193nm. This would require additional time to optimize, but could prove to be a method of obtaining complex patterned PZT transducers.

As previously seen, design factors such as the element pitch and target frequency affect the resolution and grating lobe location. A simple sweep of Eq. 2 and Eq. 3 can be performed for a fixed focus at 20mm, as shown in Fig. A.3. Note that selecting a resonant frequency of 2MHz with a 1mm pitch, limits the optimal lateral resolution to 1.95mm, with grating lobes limiting the scan angle to \( \leq \pm 50.35^\circ \). Using these same conditions with Eq.1 and a \( Q \approx 8.07 \), the axial resolution is at most 1.6mm. This assumes a single transmit pulse and an array that is 16 elements long.

The quality factor of a transducer can be difficult based on the geometry and surrounding media. A good approximation can be found through an FEM solver capable of solving the full wave equations and piezoelectric coupling the bordering fluid and solid materials. To this end, a simple COMSOL simulation was created to model an \( 825\mu m \) wide by 1mm tall transducer in progressively more complex environments to emulate the proposed environment of the array. The first scenario was the simplest with the PZT pillar suspended in water.
surrounded by plane wave radiation boundaries (absorbing boundary conditions). As seen in Fig. A.4, a resonance occurs approximately at 2MHz as expected. Next, a polyimide backing is considered, the slight mismatch reflecting slightly more pressure forward. When placing air behind the polyimide, we see the expected factor of 2 improvement in pressure. Finally, a 446µm thick matching layer of 38% alumina loaded Epotek 301-2 epoxy is added in front of the transducer. This layer is modeled to have $c = 3140$ (m/s), $\rho = 2.22$ g/cm², and $Z = 6.97$ MRayls. The resulting bandwidth and quality factors from these scenarios are summarized in Table A.1.

As can be seen in Table A.1, matching the transducer to the transmission medium (i.e. water) results in increased forward power transmission at the cost of a reduced bandwidth. This can be thought of as the product of two transfer functions: one of the transducer, and another of matching layer. As with a traditional filter in electronic circuits, the resulting system has increase roll-off outside the pass-band. If manufacturing imperfections result in a thickness that is not $\lambda/4$, it is entirely possible to create filters with orthogonal pass-bands, worsening the result at all frequencies. This can be verified in COMSOL by creating matching layers that are $\lambda/4 \pm 10\%$, as shown in Fig. A.5.
Figure A.3: Analytical calculations for the theoretical resolution and grating lobe range for a given element pitch.
Figure A.4: Simulated surface pressure and input impedance of a single transducer under various background conditions.
Figure A.5: COMSOL simulation results showing the effect of a +/- 10% height mismatch for the matching layer.
Appendix B: Software Integration and Imaging

B.1 Vantage System Integration

As described in Chapter 3, to reduce development time, the Verasonics Vantage 256 system was used as the analog/digital front-end as opposed to developing custom electronics. While providing the design flexibility of being able to control most electronic parameters through MATLAB, it does impose certain restrictions on the design. Most importantly is the limitation to a maximum of 256 elements. While possible to multiplex multiple transducers to each signal pin to increase this number, such an approach is made impractical because it would require active multiplexing circuitry directly on the board, and the Vantage system does not support an early trigger signal to allow safe separation of transmit and receive signal paths.

The host controller is a PC running MATLAB 2020a, and has the ability of configuring a number of internal parameters, allowing for customization of the transmit and received electronic signals. Moreover, the so-called ”event” based system in the independent hardware and software sequences allow for asynchronous collection of data and arbitrary parameters to be set for each rayline. The image reconstruction algorithm is then performed locally, in real-time, on the host controller, which will be described in the imaging paradigm section.
The set-up script that controls the system communicates with 7 primary structures that define the operational parameters.

The resource structure contains general information used for both simulation and image reconstruction. Here, we define the total number of transmit and receive channels (256), the connector configuration, and the computed speed of sound for distance calculations. Moreover, the Receive object determines the buffer size in memory. As we do not use the built-in reconstruction algorithm, only the RcvBuffer is relevant. We configure this buffer to consist of 512 rows of 16 bit integers, fed directly from the 14-bit ADC. Each column of the buffer represents the data from individual transducers. The total number of samples stored in memory can be computed to be:

\[
\text{\#Samples} = \left(\frac{\text{Samples}}{\text{Acquisition}}\right) \times (\#RX\text{Elements}) \times (\#\text{Frames})
\]  

(1)

The transceiver parameters are encoded in the trans object. To accurately compute phase delays, the precise relative positions of the individual transducer elements must be known. A custom function was developed to compute these positions based on a known radius of curvature in the X and Y axes. For simple cylindrical bending, the new coordinates, \(x', y',\) and \(z'\) (along with the azimuth and elevations of the transducer facing), can be computed from Table. 3.1. Note that more complicated spheroidal, spherical, or other non-euclidean cases are not considered. An example of the resultant geometry after the application of a 1cm radius of curvature is applied is shown in Fig. B.1.

The transmit or ”TX” object is a structure that contains all of the information used to direct a single type of pulsed event. The most important two vectors being the phase delays and the transmit apodization values for each element. Phase delays are calculated for the desired focus and steering angle using delay equations, and, if included, compensating for
the radius of curvature. When using multiple raylines to form an image, each transmit pulse will have a different scan angle, and thus require a separate TX object.

Each TX object has an associated waveform, "TW." This study only used only the parametric waveform type, which has 4 controlling parameters: A, B, C, and D. A is the transmit frequency \( f_0 \), B is pulse bandwidth (set to 67%), C is \( N_{\text{pulses}}/2 \), and D is a boolean value indicating whether or not to include an equalization pulse (preventing a DC offset being formed across the transducer).

The receive object defines how data is collected and controls the amplifiers, ADC, filters, and data storage. Each rayline requires a corresponding receive object to sequentially store each acquisition for each frame. In addition the receive apodization vector (which is independent from the TX apodization vector), the receive structure includes the sample mode which dictates the sample rate of the data collected. Samples are stored in 128 sample blocks, with 2 samples per wave for BS100Bw mode (100% BW sampling). From this, we can define a start depth (at which to start recording), \( z_{\text{start}} \), and end depth, \( z_{\text{end}} \), to determine the last sample in an acquisition. This number must be less than or equal to the length of the row defined in Resource. Using the law of cosines for a transducer with an array aperture of \( L \), and the cross-sectional scan angle, \( \theta_{xz} \), the maximum acquisition length is given by Eq. 2.

\[
\text{maxAcqLength} = \text{ceil} \left[ \sqrt{L^2 + z_{\text{end}}^2 - 2Lz_{\text{end}}\cos(\theta_{xz} - \pi/2)} - z_{\text{start}} \right]
\]  

(2)

A script is capable of running multiple process functions, each being a custom MATLAB function operating in VSX (the Verasonics wrapper function to call the GUI) environment. They do, however, have the ability access the base workspace and the received data found in Rcvdata. A custom process function is used to reconstruct the B-mode image in real-time,
and is discussed in detail in the imaging paradigm section.

The event structure contains a sequential list of each action performed by the hardware and software sequencer. This array of structures dictates how the system acts, and which order to perform operations. Most events include a transmit, receive, process, and sequence control code. The transmit and receive code indicate which TX and Receive objects to use. Process, if non-zero, tells the software to run the corresponding MATLAB process function (e.g. a reconstruction and plotting function). The sequence control is a set of codes to unique to the specific command. For example, when defining the event to collect the first rayline, Event(1).tx is set to 1, Event(1).rcv is set to 1, we are not ready to process so Event(1).process=0, and the control sequence should be used to delay until the next acquisition. As we must wait for the pulse to travel to the max depth and return, the transmit time is set to $t_1$ as defined in Eq. 3. After acquiring all rays, the "transferToHost" command is given to initiate the direct memory access (DMA), and the subsequent event will call the process function so the reconstruction can begin on this data as new data is being collected asynchronously.

$$t_1 = \text{ceil} \left[ 2 f_0 \left( \frac{z_{\text{end}}}{c} \right) \right]$$

To summarize, using the incorporation of the Verasonics provides design compromises. The main advantage to using the Vantage 256 system is the ability to rapidly prototype and develop code without designing and testing custom ultrasound PCB front-end circuits controlled by and FPGA. Alternative COTS ultrasound systems are not designed for modification, and offer limited support to introduce custom transducers. The largest limitation is restriction on the number of transducers. However, even if we were capable of incorporating more transducers, the space required by the routing for these elements demands unreasonable
fabrication specifications for commercial board manufacturers.

**B.2 Imaging Paradigm**

A plethora of imaging modalities exist for ultrasound phased arrays. Each have particular use cases, with several techniques being used in concert to provide a more complete diagnostic of a patient. Simple A-mode scans using a simple pulse-echo technique were conducted in 1949 by George Ludwig [113]. Multiple A-lines were combined in 1952 by John Read and later Douglass Howry to create 2-dimensional B-mode images [114, 115]. In 1985, the Doppler effect to determine flow rate by the relative frequency shift of the echo, color Doppler images were superimposed onto B-mode images to indicate motion [116]. Advancements continue in the field of ultrasound imaging to measure relative stiffness using elastography [117, 118, 119], measure fine feature size with photoacoustics [120, 121], and produce 3-dimensional images [122, 123, 124]. This work focuses only on the formation of 3-dimensional B-mode imaging and explores two methods of reconstructing this image: flash and traditional phasing.

A traditional phasing technique for a pulse-echo B-mode image begins with the division phased array region into multiple sectors. A sector is a single slice of the final 2D region. In this work, multiple a 3D scan will be conducted by showing the selected 2D slice in the desired axis, as superimposing multiple images from different XZ slices is difficult for the user to interpret.

An initial rectangular wave pulse is transmitted from each transducer. All transmit elements receive the same waveform, but the delay is calculated according to Eq. 2.15. For traditional phasing, a fixed number of raylines is selected for a particular slice, and each rayline is assigned a unique TX object. This TX object contain the relevant delay information in order to focus the beam in the given direction. To minimize the amount of data being
processed in real-time, only the desired XZ slice is scanned and plotted. This slice can be changed in real-time to scroll functionally scroll through a 3D image. Although a slice is only on a single plane, the entire array is still used in both transmit and receive, as this effectively increases the transmitted power and SNR.

The frequency, bandwidth, number of pulses, and initial polarization can be controlled through the MATLAB interface to optimize the desired properties. For example, to maximize power transfer, the transmit frequency should be as close to resonance as possible, and increasing the number pulses increases the forward pressure (thus the SNR) at the cost of axial resolution.

A unique Receive object is created for each rayline, containing the information describing the data storage location. A simplified example of the reconstruction of a B-mode image from raw received data is provided in Fig. B.2. The data from the 256 elements are combined into a single rayline by delaying the received signal from the corresponding element by the amount specified by the TX object for the corresponding rayline and summed together. The resulting received signal should contain pulses for each echo, representing the presence of an object at the given location. A Hilbert transform is then used to demodulate the signal, extracting the envelope. The process is repeated for each rayline and mapped to the predetermined scan angle of the rayline. A linear 2D interpolation algorithm is then applied to smooth the area where raylines meet. The result is then plotted, representing a single frame.

The frame rate when using traditional imaging relies on the round-trip travel time through the transmission medium and the total number of scanlines. A new pulse cannot be transmitted until the receive signal from the previous pulse has been fully acquired. Transmitting before receiving the final echo can result in spatial aliasing, as it is unknown if an echo is from the current transmit direction or the previous one. Given a maximum scan depth of
max(z), the maximum theoretical frame rate (excluding processing time) is given by Eq. 4.

\[
max(\text{FrameRate}_{\text{theo}}) = \#ScanLines \times \frac{2\text{max}(z)}{c} \tag{4}
\]

The simplest imaging technique used in this study offers no transmit beamforming, as is often called flash echo imaging [125, 126]. The primary difference between Flash and traditional imaging is that each frame requires only a single transmit pulse. This pulse can be a plane wave or focused at a point (improving resolution locally at the expense of worse resolution far from the focal zone). However, the reconstruction from the receive standpoint is nearly identical. Multiple scanlines are still defined, but now the same received response (the only response per frame) is copied and processed using the delays that would have resulted in focusing in the corresponding scanline. This is effectively performing beamforming exclusively on the receive side. In my implementation, I calculate the delays traditionally, but only apply them for the receive mode when performing flash.

The advantage of flash imaging in this context is a substantial increase in frame rate. Instead of waiting for the received echo from multiple scan lines in a single frame, now only a single transmit event is required. Provided the computation for each reconstruction can be completed fast enough, this provides a theoretical improvement in frame rate by \#ScanLines. The drawback of this technique is decrease in lateral resolution. As the initial transmit pulse was not constrained to a particular steering angle, a noticeable smearing of point reflections can occur. This demonstrated by a FIELD II simulated comparison in Fig. B.3.

From a practical standpoint, these frame rates were not reproducible on hardware. Even with most of the phasing code converted into C-coded and executed as a pre-compiled binary from MATLAB, frames rates of 1-7Hz were more common.
B.3 Radius of Curvature Detection

If the flexible phased array is to be conformal to a curved surface, it is highly beneficial to be able to detect the radius of curvature in order to correct the phasing. Two paradigms for curvature detection were considered. The first was the use of an external sensor such as a strain gauge mounted along the major axes of the array. Such a method would require valuable board real estate to be consumed by a large sensor, complicating routing. Moreover, additional hardware such as instrumentation amplifier would be needed to produce a viable signal, necessitating a regulated power source and additional I/O connections on an otherwise passive array. The second method was to use the time-of-arrival (TOA) of signals of the preexisting transducers.

Apodization vectors can used to only transmit or receive from individual elements or groups of elements. By emitting a pulse from a single pillar, all other transducers can operating in the receive mode. Assuming a uniform medium in the near field (a constant phase velocity, \(c\)), the distance between the transmitter and a given receiver is determined by \(d = c \times \text{TOA}\).

Not all transducers need to be pulsed, provided the radius of curvature is cylindrical in nature. By pulsing at least 2 transducers along an axis, it is possible to create a triangle where the points represent the centers of the current location of the pillars. The circle circumscribed by this triangle is the bending radius, \(R\), as shown in Fig. B.4.

Given line segments A, B, C from the calculated distances, it is possible to use determine the radius of the circumscribed triangle by first finding the area of the triangle using Eqs. 5 and 6.

\[
k = \sqrt{p(p-A)(p-B)(p-C)}
\]  

(5)
\[ p = \frac{A + B + C}{2} \]  \hspace{1cm} (6)

The radius of curvature can then be determined by Eq. 7, and by rewriting in terms of the determined lengths, becomes Eq. 8.

\[ R = \frac{ABC}{4k} \] \hspace{1cm} (7)

\[ R = \frac{ABC}{4\sqrt{\frac{(B-A+C)(A-B+C)(A+B-C)(A+B+C)}{2}}} \] \hspace{1cm} (8)

While only 3 segments are needed to define the triangle to determine the radius, more points can be used to form unique triangles to confirm the radius. Points that physically located further apart (producing a higher aspect ratio triangle) provide a more accurate measurement.
Figure B.1: New array geometry of array (a) after a: (b) 1cm radius of bending in the XZ dimension (c) 1cm radius of bending in the YZ dimension. (d) 1cm radius of bending on both axes.
Figure B.2: Simplified processing of a simulated B-mode image in FIELD II using traditional pulse-echo phasing. a) Raw received data from each element b) delayed and summed raylines for each sector c) demodulated raylines d) final B-Mode image
Figure B.3: Simulated B-mode image in FIELD II produced by a) traditional pulse-echo imaging and b) flash imaging for the same phantom. The red crosses indicate the true point positions.

Figure B.4: Illustration of the time-of-arrival calculated distances to determine the radius of curvature.