TRANSCRANIAL ADAPTIVE BEAMFORMING VIA ULTRASONIC PHASED ARRAYS AND ITS APPLICATION TO 3D IMAGING OF CERTAIN TYPES OF HEAD INJURIES

by

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A new adaptive beamforming method for ultrasonic imaging via small-aperture phased arrays through composite layered structures, such as human skull, is developed. If there is a scattering layer between the phased array and the imaged volume, acoustic phase aberration and wave refraction at undulating interfaces between the barrier and the rest of the propagation media can cause significant distortion of an ultrasonic image pattern produced by conventional beamforming techniques. This distortion takes the form of defocusing the ultrasonic field transmitted through the skull and causes loss of resolution, overall degradation of image quality and generation of non-informative final sonograms. To compensate for the phase aberration and refractive effects, an adaptive beamforming algorithm is developed and examined. After accurately assessing the skull’s local geometry and sound speed, the method calculates a new timing scheme to refocus the distorted beam at its original location. The procedure is in fact a construction of a matched filter that automatically adapts the transmission and reception patterns.
of the phased array to the local geometry and acoustical properties of the skull and cancels its distorting effects. Results of numerical simulations, developed to accommodate and verify the proposed theory, are provided and discussed. The simulation results are verified experimentally by applying the method on realistic human skull phantoms in water immersion setups. The developed adaptive beamforming algorithms were implemented on an open-platform phased array controller and a lab prototype of the imaging system was delivered. Results of 2D and 3D adaptive imaging through skull phantoms via 2MHz linear and matrix phased arrays are presented.
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Chapter 1

Introduction

1.1 Motivation

Tagged as a “silent epidemic” nearly a decade ago, head injuries are a leading cause of death and disability among a predominately young population. Estimates run as high as 10 million cases of head injuries per year worldwide. Annually, within the U.S., there are about 2 million emergency room visits for head injuries, roughly 575,000 admissions for brain trauma, nearly 52,000 deaths and approximately 80,000 cases of severe long-term disability [1]. About 10% of combat injuries sustained during conventional land warfare involve head injuries, including skull and brain traumas [2]. The spectrum of these injuries ranges from minor fragment wounds to mutilating blast injuries. It is critical for optimal treatment to be able to identify the internal head injuries in minutes after occurrence.

In the past few decades, ultrasonic imaging with phased array systems has been widely under study as a state-of-the-art diagnostics and inspection technique for various application areas including Sonar, NDT and biomedical imaging. Applying different beamforming techniques (multiple steering and focusing of ultrasonic waves in front of an array of elements) through phased array systems with a very small physical footprint, large pyramid-shaped volumetric images of underlying material can be generated (sonograms).
A noticeable number of head injuries involve foreign objects of various sizes lodged into the brain tissue, such as bone fragments, pieces of shrapnel, etc. If conventional imaging techniques via phased arrays are used for ultrasonic transcranial imaging of such objects, strong phase aberration and refractional effects of the human skull could result in significant distortion of beam patterns causing defocusing of the ultrasonic field. These effects, along with human skull's natural strong attenuation, lead to significant degradation of the image quality, loss of spatial resolution, and possibly large displacement or total disappearance of reflecting targets in the resulting image.

The main motivation for this study was to enable first responders to diagnose existence and accurate location of small foreign objects in a brain trauma immediately in injured persons, in both conscious and unconscious conditions. The early identification of internal head injuries can play a significant role in facilitating transportation of critically injured people to stationary medical facilities, which can both verify early diagnosis and offer therapeutical and/or surgical intervention.

### 1.2 Thesis objectives

The main objectives of the proposed research are:

**Algorithm development**- To develop and verify (through simulation and laboratory tests) an adaptive ultrasonic imaging algorithm for detection and 3D imaging of high impedance objects through thick human skull bone using matrix arrays. The targets in question have characteristic dimensions of at least 0.5 mm; positioning uncertainty of these targets should not exceed 1mm. During this project phase, skull phantoms that closely mimic geometrical and ultrasonic
properties of a real human skull (i.e. attenuation, longitudinal wave speed and density), need to be designed and developed. These phantoms will be used to evaluate the simulation results.

**Hardware implementation** - Following objective one, the imaging algorithm should be integrated into a deliverable prototype of an ultrasound imaging instrument, including a biomedical matrix array, an imaging screen, and an ultrasonic open platform phased array controller. The open platform phased array controller needs to be programmed to integrate the developed imaging algorithm. During this phase, study and implementation of applicable optimization techniques to the beamforming algorithms, signal processing methods, data storage, data transfer, and software-hardware communication must be considered. A final frame rate of 2 frames/sec is desired on a final adaptive imaging system. However, for the proof-of-concept study covered in this thesis, it is recognized that this frame rate may not be achieved, and will need to await further development and optimization.

These two objectives are to be first accomplished for the case of a linear (one-dimensional) array for 2D image generation. Pending success of this first stage, the process will be extended to matrix (two-dimensional) arrays for 3D image construction.

### 1.3 Thesis organization

This thesis is divided into seven chapters:

Following this introduction, chapter 2 starts with an overview of the main available imaging techniques used in the diagnosis of head injuries and brain disorders. A discussion is provided which explains why Ultrasonography is picked to be the focus of the current study. The chapter
then continues with a brief review of the fundamental concepts involved in this work. It also describes the anatomy and acoustical properties of human skull and brain tissue. This chapter will conclude with a review of previous work relevant to this study in the field of ultrasonic skull-induced phase aberration correction methods.

In chapter 3, the main theory and algorithm of the proposed adaptive beamforming method are described. The theory and algorithms are initially explained for the case of linear phased arrays and then an extension to matrix phased arrays is described. The numerical simulations developed to accommodate and verify the proposed theory in both cases of linear and matrix arrays are described and the simulation results are provided.

To verify the theory and simulation results of the proposed adaptive beamforming method, a series of laboratory experiments were conducted. A description of the laboratory equipment and materials involved in the testing is provided in chapter 4. Development of acoustically and geometrically realistic human skull phantoms is also described in this chapter.

In chapter 5, experimental verification of the developed theory and numerical simulation, employing the custom-designed skull phantoms, is described. Results of the laboratory experiments in cases of both linear and matrix probes are presented. Quantitative assessment and error analysis of the experimental results are presented and discussed.

Chapter 6 explains hardware implementation of the developed adaptive beamforming method for final image construction. Results of 2D and 3D adaptive imaging through realistic skull phantoms are presented.

Conclusion and recommendations for future work are presented in chapter 7.
Chapter 2
Fundamentals and literature review

2.1 Conventional brain imaging techniques

In the past few decades, a set of imaging techniques has been employed to assist in the diagnosis of head injuries, brain disorders and detection of brain tumors and abnormalities. The most famous methods are Computed Tomography (CT), Magnetic Resonance Imaging (MRI), Electroencephalography (EEG), Positron Emission Tomography (PET), and Ultrasonography (US). In this section, a comparison among these methods is provided by briefly describing the physics, along with major advantages and disadvantages of each method. A discussion is also provided at the end which justifies why Ultrasonography is picked to be the focus of the current study.

Computed Tomography (CT)- The use of CT to image the brain has been a well-established technique ever since its clinical application in the 1970s [3]. The medical applications for head CT scans are diverse and include, but are not exclusive to, inspecting tumour size and location, quickly diagnosing head trauma, and finding structural anomalies [4]. Even though CT scanners have evolved through several scanning generations (single source/detector to helical and multi-slice CT), the general principles remain the same. As the x-rays used for CT (energy of keV) travel through the body they are attenuated depending on the tissue encountered (Z-dependence).
Consequently, by changing position and angle, a large number of projections containing transmission information can be obtained and processed using filtered back-projection into a slice [3]. Notable advantages of head CT compared to alternative imaging modalities include relative short scanning times (order of seconds up to a minute) and cheaper scanning costs compared to PET and MRI, and good anatomical contrast and better resolution (order of mm) compared to PET and US [5]. A major disadvantage of CT compared to MRI, EEG and US is that it uses ionizing radiation. It is well known that any dose of radiation contributes to an increase in the overall cumulative stochastic risk of developing cancer.

**Magnetic Resonance Imaging (MRI)**- MRI has become one of the most useful and common imaging techniques for the brain. Modern uses of MRI include diagnosing strokes, and displaying noticeable changes in the brain associated with multiple sclerosis, epilepsy, and dementia [5]. The physical processes involved in MRI are rather complex but can be broken down into several basic steps. When the object of interest, for example, a brain, is placed into the bore of the MRI unit and exposed to a strong magnetic field, ranging from 0.5 to 4.0 Teslas, the magnetic moments of fundamental particles such as protons will become aligned and a net magnetic moment is produced. By tipping the magnetic moments with a certain sequence of radiofrequency pulses (order of MHz) by a certain degree, information, such as proton density, spin-lattice relaxation time, and spin-spin relaxation time, can be obtained from pickup coils. Lastly, with the use of a set of gradient coils, a final image can be obtained [5]. MRI has major advantages such as a higher ability to contrast soft tissues, such as white and grey matter, compared to CT and US; a spatial resolution comparable to CT and better than PET, US or EEG, and absence of ionizing radiation. Some negative aspects include the higher costs compared to US and CT, the lack of portability, not being able to image patients with ferromagnetic implants
such as neurostimulators, and long scan times (order of minutes) required to produce an image compared to US and CT [5].

**Positron Emission Tomography (PET)**- One of the first imaging methods used to overcome the functional limitations of imaging the brain with MRI, CT, and US was PET. Current cranial applications for using PET include the ability to detect tumors prior to structural changes, diagnosis of neurological illnesses such as Alzheimer’s disease, the detection of movement disorders such as Parkinson’s, and the ability to measure blood flow [6]. PET is a nuclear medicine technique that measures human physiology by localizing and measuring the concentration of positron-emitting radiopharmaceuticals that are intravenously administered into the body. Due to the principle of annihilation, the emitted positron quickly interacts with an electron to release two photons, each with energy of 511 keV, in opposite directions. Using coincidence detecting and filtered-back projection, an image is formed [6]. A great strength of PET is that a large number of the positron-emitting radionuclides used for labeling are organic molecules (oxygen-15, carbon-11, nitrogen-13, etc.) that can be easily incorporated in the body. However, PET has some significant drawbacks including: (i) long scan times (order of hours) compared to US, MRI, EEG and CT; (ii) the need for on-site cyclotrons or generators for producing radioisotopes that typically have short half-lives contributing to an overall increase in expense (iii) poor spatial resolution (order of cm) compared to US, MRI and CT; (iv) lack of anatomical information; and (v) the negative consequences of delivering low doses of radiation to the body [5], [6].

**Electroencephalography (EEG)**- EEG is one of the oldest clinical techniques used to measure human brain activity, dating as far back as 1924 when Hans Berger amplified the electrical signals picked up on the human scalp. Despite EEG being a relatively old concept compared to
PET and fMRI, it still is used to study epileptic seizures, and diagnose coma, brain death, and sleep disorders [7]. In general, EEG can be defined as the recording of electrical signals caused by the potential difference fluctuations of ions flowing through neurons in the brain. The electrical signals are measured by multiple electrodes, which are placed in precise locations around the scalp [7]. By interpreting amplitudes and the frequency spectrum of the resulting rhythmic activity of the processed “brain wave”, comparisons can be made between normal signals and those that are a possible result of an abnormality. In addition to being relatively cheap, safe, and portable, a major advantage of EEG is its good temporal resolution which allows brain activity to be measured on a microsecond scale. Disadvantages include long preparation time prior to scanning, the inability to produce three-dimensional images, and extremely poor spatial resolution.

**Ultrasonography (US)**- Compared to the aforementioned imaging techniques, the current clinical applications of ultrasonography to image the brain are limited. The basic idea in ultrasonography is to send an initial pulse into the body, measure the reflected ultrasonic field, process the data, and generate an image (sonogram). The sound wave, usually of frequency between 1 to 5 MHz, is focused using an acoustic lens or more commonly a beamforming phased array; the latter may have to be optimized in order to accurately image the brain. Due to strong distortional effects of skull bone, mainly attenuation and high acoustical impedance mismatch, imaging of the brain using ultrasound is currently mostly limited to thin regions of the skull called insonation windows, located in the temporal region above the zygomatic arch, through the eyes, or back of the head [8], [9]. For instance, Transcranial Doppler (TCD) has the ability to measure the velocity of blood flow of the brain’s vessels and screen for stenosis, emboli, and Sickle Cell Disease, by taking advantage of imaging through these regions. A technique that
could overcome the inability of ultrasound to image through thick sections of skull bone, would yield the advantages of low costs, greater portability than MRI, CT, or PET, fast real-time imaging, and no use of radiation.

2.1.1 Why Ultrasonography?

It was mentioned earlier that signs and symptoms of internal head injuries are difficult to diagnose in civilian emergency settings due to the lack of portable transcranial diagnostic equipment. Some of the existing imaging techniques (CT, MRI, EEG and PET) are not field deployable while the existing portable ultrasonic diagnostic machines are also not able to image through thick skull bones. Ultrasonography has the advantages of low cost and much greater portability compared to other methods, fast real-time imaging, and no use of radiation.

Ultrasonic imaging of soft tissue brain abnormalities through intact skull is not a realistic target with current technology, due to the weak signals generated by such abnormalities and the high attenuation of skull bone. However, the author believes that ultrasonic imaging of small high-density foreign objects lodged inside brain tissue should be possible due to their high reflectivity. This would require development of an ultrasonic imaging system with a robust adaptive technique to compensate for large phase aberrations induced by the skull. The proposed method in this thesis suggests a way to meet that challenge, and would enable first responders to detect/characterize small foreign objects in a brain trauma in conscious and unconscious patients.
2.2 Human skull and brain tissue

2.2.1 Human skull and brain tissue anatomies

*Brain tissue* - The brain is the command center for the human nervous system. It is comprised of over 1 billion neurons. Although it has the same general structure as the brains of other mammals, it is over three times as large as the brain of a typical mammal with an equivalent body size, and much more complex. The adult human brain weighs on average about 1.5 kg (3 lbs). Almost 80% of the brain consists of water (mainly in the cytoplasm of its cells), with a further 10-12% being fatty lipids and 8% protein. It is protected by the thick bones of the skull, suspended in cerebrospinal fluid, and isolated from the bloodstream by the blood-brain barrier; but the delicate nature of the brain nevertheless makes it susceptible to many types of damage and disease [10]–[12].

The human brain is hugely interconnected but three major components can be identified [13]: the cerebrum, the cerebellum and the brain stem (Figure 2.1).
The brainstem which includes the medulla, the pons and the midbrain, controls breathing, digestion, heart rate and other autonomic processes, as well as connecting the brain with the spinal cord and the rest of the body. The cerebellum plays an important role in balance, motor control, but is also involved in some cognitive functions such as attention, language, emotional functions and in the processing of procedural memories. The cerebrum (or forebrain), which makes up 75% of the brain by volume and 85% by weight, is divided by a large groove, known as the longitudinal fissure, into two distinct hemispheres. The left and right hemispheres are linked by a large bundle of nerve fibers called the corpus callosum, and also by other smaller connections called commissures. The two hemispheres look similar, but are slightly different in structure and perform different functions. The right hemisphere generally controls the left side of the body, and vice versa[10]–[12].

The cerebrum is covered by a sheet of neural tissue known as the cerebral cortex, which envelops other brain organs such as the thalamus and the hypothalamus. About 90% of all the
brain’s neurons are located in the cerebral cortex, mainly in the "grey matter", which makes up the surface regions of the cerebral cortex, while the inner "white matter" consists mainly of myelinated axons. The cerebral cortex itself is only 2-4 mm thick, and contains six distinct but interconnected layers.

As the brain has no ability to maintain long term energy reserves, it must be constantly supplied with blood. This occurs through two sets of arteries that serve the anterior and caudal portions of the brain, respectively. They are known as the vertebral arteries and the internal carotid arteries. Although it accounts for just 2% of body weight, it uses fully 20-25% of the body's oxygen supply and nutrients, and as much as 70% of the body's glucose supply as fuel, all of which is supplied by constant blood flow[10]–[13].

**Skull**- Human skull is composed of 22 bones that are fused together except for the mandible [13], [15]. These 21 bones are separate in children to allow the skull and brain to grow, but fuse to give added strength and protection as an adult. The superior portion of the skull is itself made of eight plate-like bones and is known as the cranium which holds and protects the brain in a large cavity, called the cranial vault [13], [16]. These eight plate-like bones are named as Occipital, two Parietals, Frontal, two Temporals, Sphenoidal and Ethmoidal [16]. They are fitted together at joints called sutures. The cranium, which is the focus of this study, is inhomogeneous and in many regions consists of compact (cortical) versus porous (cancellous or trabecular) bone as shown in Figure 2.2. Note that in this thesis, the term “skull” is used instead of cranium, as is common practice among scientists in this area.
In general, cortical bone is a fairly solid and dense material made up of minerals, organic materials and water [18]. The mineral ingredient is hydroxyapatite (~69%); the organic parts are fibrous protein collagen and non-collagenous materials (~22%); the water portion is about 9% of the total mass. The mineral part is composed of extremely small crystals (4 nm by 50 nm by 50nm) in a variant of hydroxyapatite Ca_{10}(PO_4)_6(OH)_2. The crystals are impure with about 4-6% of carbonate replacing the phosphate groups. The organic component is mostly a collagen Type I, but there are small amounts of Type III and Type VI as well. Trabecular bone, on the other hand, consists primarily of lamellar bones which are arranged in packets that make up an interconnected irregular array of plates and rods, called \textit{trabeculae}. The space between trabeculae is filled with bone marrow. Such a structure makes the trabecular bone a highly heterogeneous material that is not very conducive to ultrasound transmission.
The middle porous layer of the skull, Diploe, is the main contributor to the distortional effects of attenuation and scattering of ultrasonic waves [19], [20]. The inner layer of the skull at many areas is curved and has an undulating surface boundary. The total skull thickness variation in adults ranges from 3 to 15 mm [21]–[23]. At specific skull locations with an area of the order of 1-2 cm² (such as the effective area of a conventional transcranial ultrasonic imaging probe), a skull segment can be effectively represented as a layer with flat top surface and an undulating inner surface. The skull thickness variation could be up to 5mm even across such small areas.

2.2.2 Acoustical properties

In the past few decades, numerous studies have been conducted to measure and tabulate acoustical properties of human skull and brain tissue. Major acoustical properties include density $\rho$, attenuation $\alpha$, longitudinal and shear sound speeds $c_L$ and $c_T$; from such measurements the acoustical impedance $Z$ can then be determined as $Z = \rho c_L$. The reported values for acoustical properties of brain tissue and especially human skull differ from one study to another. This is due to skull’s complex structure, dependence on the individual, maintenance procedure, and even conditions and methods of measurement. Therefore assigning a specific value to each parameter is impossible. Nevertheless, here the author has tried to collect and present an average value or a common range of values for each parameter that are consistent with the published literature, as listed in Table 2.1.
Table 2.1: Acoustical properties of human skull and brain tissue

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Density [Kg/m$^3$]</th>
<th>Shear speed [m/s]</th>
<th>Longitudinal speed [m/s]</th>
<th>Acoustic impedance [MRayl]</th>
<th>Young's modulus [GPa]</th>
<th>Poisson’s ratio</th>
<th>Attenuation [dB/(cm.MHz)]</th>
<th>Thickness [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Brain tissue</td>
<td>1040</td>
<td>N.A.</td>
<td>1560</td>
<td>1.62</td>
<td>9210 Pa</td>
<td>0.46</td>
<td>0.6</td>
<td>N.A.</td>
</tr>
<tr>
<td>Cortical layers</td>
<td>2050</td>
<td>1510</td>
<td>2900 - 3500</td>
<td>5.64- 6.02</td>
<td>14 - 20</td>
<td>0.2 - 0.5</td>
<td>6 – 10</td>
<td>3 - 15</td>
</tr>
<tr>
<td>Diploe</td>
<td>1742</td>
<td>N.A.</td>
<td>2450 - 2530</td>
<td>4.27-4.41</td>
<td>0.05-0.5</td>
<td>0.01-0.35</td>
<td>14 – 30</td>
<td>0 - 12</td>
</tr>
<tr>
<td>Full skull</td>
<td>1830-2050</td>
<td>1510</td>
<td>2800 - 3500</td>
<td>5.12- 6.79</td>
<td>N.A.</td>
<td>N.A.</td>
<td>8 – 22</td>
<td>3 - 15</td>
</tr>
</tbody>
</table>

It should be noted that the range of measured ultrasonic attenuation values for human skull is very broad [24]. The wide range in the results can be attributed to many factors, such as different porosity levels of the skull segments used in each study. For example, Aubry et al. (2003) reported sound absorption by skull bone in the range 2–80 dB/cm at 1.5 MHz within the same ex-vivo skull [25]. Pichardo et al. (2011) reported 27 dB/cm attenuation for cortical bone at 1.4 MHz [26]. Attenuation values ranging from 33 dB to 105 dB/cm were measured at 2 MHz by Ammi et al. [27]. Fry and Barger (1978) give 13 (dB/cm) for cortical bone and 34 (dB/cm) for diploe skull at 2 MHz [21]. Pinton et al. (2012) reported 16.6 dB/cm at 1 MHz, which can be extrapolated to 33 dB/cm at 2 MHz [28]. On another ex-vivo measurement, attenuation of 14–19 dB/cm at 1.7MHz was reported by Baykov et al. in 2003 [29]. During the development of simulation algorithms in this study, as will be explained in chapter 3, an attenuation range of 18–35 dB/cm is considered for human skull at 2MHz which is typical according to previous studies.
It is also important to note that ultrasonic attenuation significantly depends on the frequency of the propagating ultrasound. It has been previously determined that it rapidly increases in skull when frequency increases \([24]\). Therefore, the conventional range of frequencies used in ultrasonic transcranial imaging is limited to the low range of 0.25-2.5MHz. Depending on the application and the imaging targets inside the brain, a compromise between the penetration depth and image resolution is inevitable. As explained in chapter 1, the imaging targets in the presented study are foreign solid objects lodged in brain tissue, which are in fact highly reflective. Therefore a central frequency of 2MHz for the propagating ultrasound signals is used in development of the theory, simulation and experimental verification to maximize the final image quality factors.

2.2.3 Ultrasound propagation in Human skull

As a complex structure, human skull causes a variety of distortional effects on propagating ultrasound \([21]\), \([26]\)--\([28]\), \([30]\), \([31]\). In transcranial imaging applications, major contributions come from attenuation, strong impedance mismatch between the skull and rest of the propagation media, as well as complex surface topography of the skull (large variations of thickness-to-wavelength ratio over the probe's area). Depending on the application and the frequency range, each of these factors could gain different weights in the skull’s total distortional effect on the ultrasound transmission. In addition to skull’s attenuation, the last two factors, i.e. the acoustic mismatch between the skull and the brain tissue as well as the skull's thickness variation, cause strong reflection, wave refraction and re-direction at undulating interfaces between the skull and the brain tissue. This results in significant distortion of an ultrasonic image pattern produced by conventional imaging techniques.
2.3 Ultrasonic phased arrays

Phased arrays enable ultrasonic images to be obtained without the need for mechanical scanning by single element transducers. They can provide an image with a sufficiently high frame rate that avoids significant motion distortion caused by fast moving structures such as those present in the heart [32]. The development of ultrasound imaging arrays was in large measure due to the much earlier invention and development of radio frequency antenna arrays that began at the end of the 1800’s. Some of the first publications concerning the design and use of ultrasonic arrays originated from Tucker’s group in 1950’s, [33], [34], for underwater echo-ranging applications. The first publication of an electronic sequential scanning phased array system for medical applications appears to be the 10 element ultrasonic array described by Bushman in 1965 [35] which was used for imaging the eye.

Ever since, ultrasonic imaging with phased array systems has been widely studied as a diagnostics and inspection technique for application areas including sonar (e.g. [36]), NDT (e.g. [37]) and biomedical imaging (e.g. [38], [39]). Early systems generally relied on analog circuits, such as tapped delay lines. Subsequently, with the availability of high-speed analog to digital (A/D) converters and processors, digital systems have been developed which are referred to as digital beamformers [40], [41].

Generally speaking, a phased array imaging process can be divided into two major steps, Transmission (Tx) and Reception (Rx). A typical ultrasonic phased array image (sonogram) is obtained via multiple steering and focusing of ultrasonic waves in both steps. By applying various beamforming techniques, sonograms representing a relatively large volume of the
underlying material can be generated using phased array systems with a very small physical footprint. The ability to accurately focus an array in both transmission and reception modes is vital for obtaining high-quality images. A brief introduction to each mode is provided in this section, while a comprehensive discussion and further details on ultrasonic phased arrays and beamformers can be found in [32].

**Transmission (Singular Focusing)** - Focusing in transmission mode is achieved by appropriate timing of signals for each array element. In each desired focusing scenario in this mode, a time delay pattern is calculated to cause the ultrasonic fields from all elements to interfere constructively with each other at the intended focal area and destructively elsewhere. This way the focus can be placed at any desired angle and depth within the accessible inspection area or volume of the phased array. Further details on this mode will be explained in section 3.2 where the theory of the proposed adaptive beamforming method is described and also in chapter 6 (section 6.2.1) where the final imaging system is designed.

**Reception (Dynamic Focusing)** - Dynamic focusing in reception, as the transmitted signal is proceeding through the medium, makes it possible to track a reflecting or scattering source encountered by the transmitted signal. The idea of dynamic focusing on reception is to focus at various depths along each transmission line so that each scattered wave that arrives at the receive aperture, is optimally detected. Similar to the transmission mode, the focusing is achieved by appropriate timing of the elements in the receiving aperture. A very simplified block diagram illustrating a typical digital beamforming receiver system is shown in Figure 2.3. As shown in the diagram, for each focal point in the dynamic focusing process, the following steps are executed at the receiving beamformer:
- Pre-amplification of the received signals at each receiving element

- Analog to Digital (A/D) conversion of the received signals at each element

- Applying appropriate signal processing on digitized signals, e.g. high-, low- or band-pass filters and amplitude adjustment via TGC (time gain compensation [32])

- Spatial apodization over the receiving channels

- Applying time delays to bring the received signals in phase

- Summation of the processed signals and appointing a brightness or color level to the final output signal

The above steps will be explained in detail in chapter 7 (section 6.2.2) where the final imaging system is designed.

![Figure 2.3: Simplified block diagram illustrating a digital beamforming receiver system that uses a low-noise preamplifier, a high-speed (40 MHz) A/D converter (≥12 bit), and a digital delay (shift register) controlled by a digital signal processor (DSP) on each of 128 signals received by the array elements. (Reproduced with permission from [32])](image-url)
2.4 Ultrasonic phase aberration correction techniques

2.4.1 Introduction

Ultrasound imaging of brain tissue structures has been the focus of considerable research. Some potential diagnostic applications include blood flow imaging, hydrocephalus monitoring, detecting internal hemorrhage, tumor imaging, and detecting foreign objects in emergency situations [42]. Initial studies by Aaslid et al., [44] and [45], date back to the early 1980’s, in which authors reported Doppler measurement of flow velocity and non-invasive evaluation of vasospasm in cerebral arteries based on these velocities. Bogdahn et al. reported on the application of transcranial sonography to identify intracranial physiologic structures such as the circle of Willis and lateral ventricles [45].

Almost all the initial studies and also many more recent studies in transcranial imaging have concentrated on transmitting ultrasound through the temporal acoustic windows of the skull, using relatively low frequency ultrasonic pulses in the range of 1-2 MHz [46]–[48]. The skull is thin in these windows (approximately 1–3 mm thick), and falls within the near-field of the transducer given its relatively low compression wave velocity of about 2650 m/s. Therefore it introduces minimum distortion compared to other areas of human skull. But the available inspection volume through the temporal windows is only a fraction of the total brain tissue in adults. Transcranial ultrasonic brain imaging through thick human skull is still a major focus of study. Furthermore, strong attenuation and aberrating effects of human skull (see section 2.2) make it an extremely challenging field.

Over the past several years, many ultrasonic imaging algorithms, including algorithms for transcranial imaging, have continued their growth with trends towards the utilization of phased
arrays with a large number of elements. Such systems hold the promise of increased target
definition and sensitivity. However for many clinical applications, this promise is not realized
due to phase aberrations induced by differences in ultrasound propagation speeds between brain
and skull tissues. An acoustic wavefront propagating through regions with different acoustic
velocities will be phase-shifted. This effect is known as phase aberration, or more precisely as
arrival-time errors. As an example, in the case of transcranial imaging, the speed of sound
propagation in human skull varies in the range of 2650-3400 m/s while brain tissue has an
average velocity of 1520m/s [21], [22], [24]. As another example, in the case of abdominal
imaging, abdominal fat has an acoustic velocity of 1460 m/s while liver and kidney have
velocities of approximately 1560 m/s [49].

In order to obtain a good ultrasound image via phased arrays, it is necessary to focus ultrasound
beams in the area under investigation: the narrower the focal spot, the better the image
resolution; the lower the sidelobe levels, the better the contrast. Therefore, the above-mentioned
phase aberrations degrade image quality mainly by reducing resolution and contrast. The phase
aberrations may also introduce geometric distortions by either displacing a target or totally
eliminating it from the image.

Almost all of the commercially available medical ultrasonic phased array imaging systems
assume a single acoustic velocity in tissue while steering and focusing the acoustic beam, e.g.
[50]. In such systems, the inter-element timing differences of transmitted signals, used to steer
and focus beams via phased arrays, are calculated assuming a constant acoustic velocity of
typically 1540 m/s. But non-uniform tissue velocities and undulating boundaries between the
transducer array and the target organ will cause acoustical pulses transmitted from different array
elements to arrive at the nominal focal spot out of phase with respect to each other. Similarly,
reflected signals received across the array will be summed out of phase. In extreme cases, where the there is a large velocity difference in the propagation media such as in the case of human skull and brain tissue, the focal spot may be severely displaced or even never established in the inspection area. This results in a non-informative reflected field and consequently a non-informative and distorted final image. Previous studies showed that this problem cannot be fixed by using a single average velocity for all the different media.

Therefore, it is necessary to use adaptive methods to compensate for phase aberration and accurately focus through an inhomogeneous inspection area. The term “adaptive” in this context refers to the adaptation of excitation signals and their time delay patterns to the velocity inhomogeneity and undulating boundaries of propagation media.

During past years, many adaptive algorithms have been developed to correct phase aberrations for both therapeutic and diagnostic medical ultrasound applications. More specifically, the main adaptive imaging techniques, suggested for phase aberration correction for diagnostic medical ultrasound applications, can be listed as:

- Time reversal mirror (TRM) [51]–[54]
- Cross-correlation peak measurement [55]–[57]
- Speckle brightness [49], [58], [59]
- Least-mean-square estimation [60], [61]
- Direct estimation using a $k$-space approach [62]
- Near-field signal-redundancy (NFSR) [63]–[66]
In the following subsection, the basic principles, the key mathematical concepts, and a summary of some promising results for each technique are explained in brief. A final discussion of advantages and limitations of each method is also provided.

2.4.2 Biomedical phase aberration correction methods

2.4.2.1 Time reversal mirror (TRM)

Originally proposed by M. Fink in 1992, [51]–[53], time reversal of ultrasonic fields represents an adaptive method for focusing ultrasound through inhomogeneous media. This may be accomplished by a time-reversal mirror (TRM) made from an array of transmit-receive transducer elements that each allow an incident acoustic pressure field to be digitized. The source of this field is normally an object in the inspection area which reflects the array’s original emitted field back towards itself. It also could be a transducer, independent from the array, which transmits towards the array. In either case, the digitized pressure field is then time-reversed at each array element and re-emitted to focus at the location of the aforementioned source. Time reversal, in its early stages, was only proposed for lossless media or media with low attenuation, but further studies led to some extensions of the original method in transcranial brain imaging techniques which are the focus of discussion in this section.

Assuming compressibility of \( \kappa(r) \), density of \( \rho(r) \), and introducing the local compression wave velocity as \( c(r) = \left( \rho(r) \kappa(r) \right)^{-1/2} \), the wave equation for an acoustic pressure field \( p(r, t) \) can be written as

\[
\nabla \cdot \left( \frac{\nabla p}{\rho} \right) - \frac{1}{\rho c^2} \frac{\partial^2 p}{\partial t^2} = 0
\] (2.1)
In a typical wave propagation experiment, initial and boundary conditions determine a unique solution $p(r, t)$ to this propagation equation. The equation contains only a second-order time-derivative operator; therefore, it can be concluded that if $p(r, t)$ is a pressure field solution of the propagation equation, then $p(r, -t)$ is another solution of the problem. This property is specific to the invariance under a time-reversal operation for lossless media and was the starting point of the time-reversal principle.

The goal, in time-reversal experiments, was to modify the initial conditions in order to generate the dual solution $p(r, -t)$. However, due to causality requirements, $p(r, -t)$ is not an experimentally valid solution. Therefore, another solution was suggested. It consisted of measuring during a time interval $T$, the pressure $p(r, t)$ on a TRM. Such a mirror can be planar or focused, 1D or 2D. The time interval $T$ should be long enough so that the pressure field vanishes for $t > T$.

As shown in Figure 2.4, TRM focusing through an inhomogeneous media requires three steps. The first step consists of transmitting a wide beam-width wave field through the inhomogeneous medium from the array to the target (Figure 2.4.a). The target generates a backscattered pressure field that propagates through the inhomogeneous medium and is distorted. In the second step, the backscattered pressure field is recorded by the transducer array elements (Figure 2.4.b). The circuitry associated with each element then computes a voltage train that is a time reversal of the received backscattered field. In the last step, each transducer array element generates a pressure field according to the time reversed voltage sequence. This pressure field then propagates through the inhomogeneous medium, and focuses on the target (Figure 2.4.c).
Time reversal by itself could not totally correct for skull aberrations because of ultrasonic absorption in the bone that breaks the time reversal symmetry of the wave equation [67]. To address this challenge, Fink et al. applied an amplitude compensation of time reversed signals before emission [67], [68]. This gain compensation acts as an optimal correction technique for absorption effects as long as the only attenuative part of the propagation media is a very thin absorbing layer in contact with the array. Although this first refinement improved the focusing quality, this assumption of a thin absorbing layer is still invalid in many areas of the skull.

Later, Ebbini et al. tried to enhance the TRM method by defining more realistic inverse filters between the focus and array elements [69]–[72]. In a continuation of their work, a more general approach was suggested by Tanter et al. [73], which enabled a practical adaptive focusing by building a complete inverse filter of the transcranial propagation both in the space and time domains. This spatiotemporal inverse filter, known as the STIF, could be computed from the experimental measurement of the impulse responses on the main array and a second array placed in the focal area. The STIF technique was applied to transcranial focusing by Aubry et al., who demonstrated that a good correction of skull aberrations could be achieved [74]. Later, a faster and more robust extension of this technique based on the concept of *iterative time reversal* between two arrays was introduced by Montaldo et al. [75]. However, these methods based on the STIF concept, were totally unrealistic in practical configurations as they required the
presence of a second array of transducers in the focal area (inside the brain!). They have been only tested theoretically and for focusing through ex-vivo skull segments in water tanks or similar setups in a laboratory environment. Eventually another strategy overcoming this major drawback was proposed by Tanter et al. [76]. The second array was no longer located inside the brain but was moved outside the head. The major theoretical aspects, experimental verification, and feasibility of the delivered prototype are explained in [76].

In addition to the above methods, another technique based on the matched filtering process and analogous to TRM was suggested by Svet et al. in 2003 [77]. In this method, the inverse wave front procedure for wideband probe signals is used, and information about the skull bone structure is extracted from the ultrasound field. Before the matched filtering process, the phased array is used in reflection mode to estimate the local skull bone thickness under each array element. In this step, the complex amplitude was measured and, as a result, the skull bone transfer function was formed. The complex transfer coefficients contain information about both local thickness and local absorption. In the next step the matched filter process was applied, via consideration of local thickness and absorption, to focus ultrasound through the skull. This method was suggested for both therapeutic and imaging applications. Theoretical and model studies of this method are presented in [77], [78]. This method was later studied by Maev et al. for transcranial imaging of hi-contrast brain tissue traumas and the result were presented in [79].
2.4.2.2 Cross-correlation peak measurement

Among the various methods suggested to correct for phase aberration in ultrasonic imaging systems, the simplest method is reception mode cross-correlation to focus on a known point source, first described by O’Donnell and Flax in 1988 [55].

The basic principles of the theory can be described with help from Figure 2.5. Figure 2.5.a shows the principle of focusing in a homogeneous medium when there is no aberrating layer between the source and the array. The pressure signals coming from an acoustic source are sensed by each transducer element and digitized. The reception mode time delay pattern (thick bars) can then be found through simple geometry of array elements and the reflecting source and the expected acoustic velocity of the tissue, 1540m/s. In this simple case, the thick bars exactly compensate for the geometrical time delays the signals sensed on the array and bring them into phase before their summation.

The influence of an aberrating layer is shown in Figure 2.5.b where the received signals are no longer aligned by the simple geometrical concept of Figure 2.5.a, leading to a net phase aberration of the element signals. To remove this aberration, the final alignment step before summation is to apply secondary time delays, determined by time shifting according to the peak of the cross-correlation between signals from neighboring transducers [55], [56].

![Figure 2.5: Configurational steps of cross-correlation phase aberration correction [51].](image-url)
O’Donnell et al. later enhanced the above method such that an ideal point source is no longer needed. They demonstrated an iterative multi-bit cross correlation phase correction procedure that could be used to obtain estimates of arrival time differences using signals from diffuse scatterers [55], [56].

It should be noted that the cross-correlation method described in this subsection is capable of only partially correcting for phase aberration in ultrasonic imaging, as it only applies a correction in the reception mode of an adaptive beamforming process. Transmission mode corrections are not possible with this technique. No pre-determination of acoustical or geometric properties of the inspection area is required with this method.

2.4.2.3 Speckle brightness

In biomedical imaging, ultrasonic speckle arises from the coherent interference of reflections from many randomly-positioned sub-resolution scatterers in the inspection area. Most of these scatterers are part of regular tissue structures; this is why ultrasonic images often have a random grainy appearance which is called speckle. Originally proposed in 1986, Smith et al. suggested to use speckle brightness as a quality factor to correct for unknown phase aberration [49], [59]. They found and described a relationship between the magnitude of phase aberrations and the regional brightness of speckle and point-like targets. The proposed correction technique was in fact analogous to its optical counterpart, used to adaptively focus incoherent optical telescopes [80].

The technique involved an iterative adjustment of the transmit and receive phasing of individual or groups of elements, utilizing the area-wise average brightness of region-of-interest (ROI) speckle as the quality factor. The ROI position within the imaging plane and its axial and lateral
dimensions are operator-selectable, as is the number of A-scans to be collected from the ROI [49]. According to the proposed method, an initial phasing of the system is calculated through a conventional phasing scenario as if the entire inspection area were homogenous with a tissue velocity of 1540 m/s. A single transmit focus and a few dynamically updated receive foci are first selected to start the process. Using a simple iterative procedure, the element phasing that maximizes the average ROI speckle brightness is found. In their original work, element phasing was adjusted in steps of 80 nanoseconds and for operator-selectable element groups (i.e. groups of 1, 2 or 3) from one side of the array to the other. For a given element group, the phase delay is increased from its initial value by 80 ns, and the ROI brightness for the initial and modified phase values are compared. If the change increased the ROI brightness, then successively larger delay values are tried until a maximum brightness is found. If the initial phase delay increment decreases the brightness, then the delay values are decremented until a maximum brightness is found. The ROI brightness is then calculated for a new element(s) group phasing until the desired phasing is found for all elements. Once it is completed, the modifications to the initial normal phasing are applied to the transmit and receive phasing of lines throughout the full size sector (entire sonogram), and real-time imaging with this “corrected” machine phasing continues.

The original approach described above has been used ever since for biomedical ultrasound imaging applications. For example, in 2004, Trahey et al. described the development of a 3D transcranial imaging system including real-time 3D color Doppler images of cerebral vessels, as they claimed in [81], based on the above method. Later developments were reported in 2012 with a combination of the above phase aberration correction and the dual apodization with cross-correlation technique as explained in [82].
2.4.2.4  Least-mean-square estimation

As another phase aberration correction approach, the least-mean-square error-fitting method was suggested by T. Sato et al. [60], [61]. The proposed method forms an image of targets on a plane parallel to the transducer array surface, when there is an inhomogeneous layer close to the array. First, a set of data is acquired by collecting A-scans corresponding to all possible combinations of pairs of transducers on the array operating in transmission and reception mode. This builds an over-determined equation group which has sufficient equations to estimate the spatial frequency components of the target plane across the array, by means of a least-mean-square error fitting routine. Basic principles of this method can be understood through the configuration shown in Figure 2.6. In this system, if ultrasonic waves of amplitude 1 and wavenumber $k$ were transmitted from the transducer at $x_T$ to the target object at depth $z$ with reflection coefficient distribution $\rho(v)$, then the signal received at $x_R$ is given by

$$ R(x_T, x_R) = \int \xi_T(x_T)\xi_R(x_R)\rho(v)\exp\{jk[z^2 + (x_T - v)^2]^{1/2} + jk[z^2 + (x_R - v)^2]^{1/2} \} \, dv $$

(2.2)

where $\xi_T(x_T)$ is the effect of the inhomogeneous layer at $x_T$ for transmission and $\xi_R(x_R)$ is the effect for reception. The attenuation factor due to the propagation distance is omitted.
Now assuming that the object is located in the far field and the area of the object is so small that the following approximation can be applied

\[ [z^2 + (x - v)^2]^{1/2} \approx z - xv/z + x^2/2z , \quad (2.3) \]

then Eq. 2.2 is reduced to

\[
R(x_T, x_R) = \\
\int \xi_T(x_T)\xi_R(x_R)\rho(v)\exp(2jkz)\exp[-jk(x_T + x_R)v/z]\exp[jk(x_T^2 + x_R^2)/2z] \, dv.
\]

Now knowing that the spatial frequency representation of the reflection \( \rho(v) \) is

\[
\rho(v) = \int \lambda(\mu)\exp\left(\frac{jk\mu v}{z}\right) \, d\mu \quad (2.5)
\]

or

\[
\lambda(\mu) = \int \rho(v)\exp\left(-\frac{jk\mu v}{z}\right) \, dv , \quad (2.6)
\]
the datum \(d_{m,n}\) obtained for the transmission at \(m\Delta x\) and reception at \(n\Delta x\) (\(\Delta x\) being the array’s pitch) can be written as

\[
d_{m,n} = ca_mb_n\lambda_{m+n},
\]

(2.7)

where

\[
a_m = \xi_T(m\Delta x)\exp(jkm^2\Delta x^2/2z),
\]

\[
b_m = \xi_R(n\Delta x)\exp(jkn^2\Delta x^2/2z),
\]

(2.8)

\[
\lambda_{m+n} = \lambda[(m + n)\Delta x],
\]

and \(c\) is a constant. The proposed method suggests to write Eq. 2.7 for all possible transmission-reception combinations (\(N\)). When \(N\) is sufficiently large there is redundancy in the generated set of equations for the required parameters. Thus these parameters can be estimated from the data by means of least-mean-square error fitting. Then the image is reconstructed from the estimated spatial frequency components (\(\lambda\)) by using Eq. 2.5. Further details of the method can be found in [60], [61].

### 2.4.2.5 Common-midpoint signal redundancy using a k-space approach

**Farfield approach**- Another phase aberration measurement and correction method, based on the phase-closure method in radioastronomy [83], was introduced by Rachlin [62]. The method uses the signal-redundancy principle, [62], to measure the phase aberration profile directly. First, common midpoint signals are cross-correlated to find the relative time-shift between them, then an over-determined matrix is used to derive the phase-aberration value on the location of each element.
To demonstrate the basic principles of the method, a four-element subset of a linear array is depicted in Figure 2.7. In this figure, the two elements indicated by primed subscripts \((m'\) and \(n'\)) are receivers and the other two elements \((m\) and \(n\)) are their corresponding transmitters. Both transmitting/receiving pairs \((n-n'\) and \(m-m'\)) share a common midpoint. The transmitters separately emit identical signals. Let \(g_{n,n'}(t)\) represent the analytic signal acquired by the simple system consisting of the \(n^{th}\) transmitter and the \(n^{th}\) receiver elements, and define \(s(y, t)\leftrightarrow S(f_y, f)\) as the target scatterer distribution. The transmitter and receiver element midpoints are located at \(y = y_n\) and \(y = y_{n'}\), respectively.

![Diagram](image)

**Figure 2.7:** four-element subset of a linear array used to explain the signal redundancy method [62].

The \(k\)-space Fourier transform of the signal from the \(n-n'\) pair yields a single line of spatial frequency domain data

\[
G(f; n, n') = H(f_y, f; n, n')S(f_y, f)
\]  

(2.9)

where \(H(f_y, f; n, n')\) represents the system frequency response. The complete set of discrete Fourier transformed data can be then represented as
\[ G(f_j; n, n') = |G(f_j; n, n')|e^{j \varphi G(f_j; n, n')} = |H(f_j; n, n')||S(f_j; n + n')|e^{j2\pi f_j(\tau_n + \tau_{n'})e^{\varphi S(f_j; n + n')}} \]

(2.10)

where \( \tau_n \equiv \tau(y_n) \), \( \tau(y) \) being the one-way aberration delay. In this method, it was suggested to choose any two random pairs of elements with common midpoints and develop a set of equations from Eq. 2.10. The set of possible equations available using transmitter-receiver element pairs in an array contains a high degree of redundancy, enabling many equations to be excluded. The approach is then to determine the time delay between each two pairs in a way that it maximizes the magnitude of the cross-correlation function estimate

\[ \hat{R}_{n,n',m,m'}(\Delta t) \equiv \frac{1}{f} \sum_j g(t; n, n')g^*(t + \Delta t; m, m'). \]

(2.11)

Now by calculation of such time delays for a combination of these two pairs, and by applying a few mathematical manipulations, [62], a time delay estimate matrix of all desired array elements can be calculated. The reverse of such delay estimates is the compensating time delay pattern that is applied to the array’s aperture to remove focusing aberrations. Details of this method can be found in [62]. The scope of this method is confined to situations in which nearfield propagation inhomogeneities can be modeled as a fixed delay screen superimposed upon the array aperture. Therefore its efficiency is restricted to imaging targets in an array’s farfield zone.

**Nearfield Approach** - The above-mentioned signal redundancy method was further developed and experimentally tested by Y. Li [63]–[65], to also account for nearfield targets. Mathematical details of the approach can be found in [63], and here only the major difference between the two methods is explained: In the previous method, when there is no phase aberration, the relative time-shift between common midpoint signals is zero according to Eq.9 and Eq.10 in [62], which
is true only in the far-field. The analysis proposed by Y. Li, shows that for targets in the near-field, there is a near-field term in the relative time-shift between common midpoint signals. Therefore he proposed a dynamic near-field-correction process to reduce its effect on the measurement [63], [64].

2.5 Discussion

A discussion of the major advantages and disadvantages of all above methods for transcranial imaging application, and justification for the author’s proposed approach is presented in this section.

The time reversal mirror (TRM) method, section 2.4.2.1, is a viable process for adaptive focusing through inhomogeneous but not highly attenuative media. No pre-determination of acoustical or geometric properties of the inspection area is required with this method. It also compensates for any geometrical distortions of the array structure. But, in consideration of its application to transcranial imaging, this method has major drawbacks: Firstly, time reversal by itself cannot totally correct for skull aberrations because of ultrasonic absorption in the bone that breaks the time reversal symmetry of the wave equation, Eq. 2.1. Even the gain compensation method which was suggested as a correction technique for absorption effects was not an optimal solution. This correction is only valid as long as the only attenuative part of the propagation media is a very thin absorbing layer in contact with the array; this restriction is invalid in many areas of the skull. In the time reversal process the presence of a highly reflective source (or a small catheter-injected source) was necessary. This contradicts the non-invasive nature of the ultrasonic imaging processes. Further developments based on the STIF concept tried to address
this difficulty but they require a dual array imaging system to be held still during the imaging process.

The cross-correlation technique, section 2.4.2.2, has been realized as a practical technique for enhancing ultrasonic image quality factors. Among all the described methods, the cross correlation technique is the simplest and it can be employed in many ultrasonic imaging scenarios with any array geometry. The major disadvantage of this method though, is that it is capable of only partially correcting for phase aberration, as it only applies a correction to the reception mode of an adaptive beamforming process. Therefore, it does not resolve the phase aberration in transmission mode. If only this method is used for transcranial imaging, poor focusing will be achieved in transmission mode due to high impedance mismatch between the skull and brain tissue and the consequent refraction.

In the author’s opinion, the Speckle Brightness method is a more viable method for phase aberration correction in transcranial imaging than the other methods. It is applicable on small aperture phased array imaging systems and it does not require any predetermination of acoustical and geometrical properties of the skull or brain tissue. But since the method is based on evaluation of speckle brightness of a ROI as a quality factor, it follows a time consuming trial-and-error procedure for each and every focusing scenario in transmission and reception modes. This is a drawback when high resolution (both lateral and depth) image generation is desired. Besides, this approach corrects for phase aberration based only on a small ROI in the inspection area. This means that the adaptive phasing of the array elements is only optimal for imaging of the corresponding ROI and not optimally valid for the entire inspection area.
The least-mean-square and midpoint signal redundancy methods are based on their counterparts in optics. They both do not require any predetermination of acoustical and geometrical properties of the aberrating media. But their scope is confined to situations in which nearfield propagation inhomogeneities can be modeled as a fixed delay screen superimposed upon the array aperture. Therefore their efficiency is restricted to imaging targets in an array’s farfield zone. This is not a realistically valid assumption for transcranial imaging, since there are situations in which the targets are located in the nearfield zone of the imaging probe. When the system is too close to the array (i.e., in the Fresnel zone), only an approximated image can be obtained. This limitation was addressed in the nearfield approach presented in section 2.4.2.5. But still the common-midpoint nearfield signal redundancy technique is not a reliable solution to image large inspection areas such as entire brain tissue. Aside from heavy processing load associated with this technique, the correction term introduced in this method is only valid when the inspection area is small [63].

A new method for adaptive imaging through highly aberrating layers such as human skull is presented in this thesis. The objective is to overcome the major disadvantages of existing phase aberration correction methods when their application for ultrasonic transcranial imaging of high impedance targets is considered. The presented method is based on a custom-designed predetermination of acoustical and geometrical properties of the segment of the aberrating layer in contact with the array as will be explain in the following chapters.
Chapter 3
Single-point adaptive focusing – Theory and simulation

3.1 Introduction

The background theory and a new algorithm for single-point adaptive focusing in transmission mode are presented in this chapter. The proposed algorithm will later be implemented in a full adaptive beamforming process, including complete transmission and reception modes (chapter 6). The method suggests a new phase aberration compensation technique for adaptive focusing of ultrasound through randomly shaped scattering layers via small aperture phased arrays. In this study, the proposed method is optimized for the case of human skull as the scattering layer and its application for transcranial imaging is discussed. The theory and algorithms are initially explained for the case of linear phased arrays for 2D adaptive beamforming. The proposed theory is then extended for the case of matrix (or planar) phased arrays for 3D adaptive beamforming. Numerical simulations with wave propagation modeling were developed to accommodate and verify the proposed theory in each case. Although the physical theory of the proposed method remains the same for both cases of linear and matrix probes, simple extension of the proposed algorithm from linear to matrix phased arrays was not possible. This mainly stems from the presence of more complicated boundary conditions in the case of matrix arrays and also the high processing time of the original approach. Therefore, a more accurate, efficient,
and robust approach was designed by the author in transition to matrix arrays. The chapter contains simulation results for both cases. Discussion of the simulation results is provided.

The simulation algorithms, designed and coded by the author in MATLAB are capable of conducting all necessary computations of the proposed theory. The author believes that to date no commercial simulators are capable of these particular computations. This in fact was the main motivation to develop the independent simulation software used in this study.

In order to explain different aspects of the proposed method, it is important to emphasize all of the major physical effects of a typical human skull contributing to the reduction of sonogram quality in an ultrasonic beamforming process. In the transcranial imaging application discussed in this thesis, image degradation originates from ultrasound attenuation, strong impedance mismatch between the skull and rest of the propagation media, as well as the complex surface topography of the skull. The distortional effects of scattering and absorption will be discussed in more details later in this section. The main focus of the proposed phase aberration correction technique, however, is to compensate for the two remaining factors, i.e. for the acoustic mismatch between the skull and the brain tissue as well as for the skull's thickness variation. These last two factors cause unpredictable bending of the wave trajectories leading to angle-dependent displacement and blurring of the focal spot. The proposed method automatically adapts the time delay transmission patterns of the phased array to the local geometry and acoustical properties of the skull and cancels its refractional distorting effects.

It should be noted that, due to their minimal distortional effects on the application in hand, the following simplifications were made during this study:
Consideration of nonlinearity effects of the barrier (skull) and propagation media (skin and brain tissue) is outside the scope of this study. This, in fact, is a worthwhile sacrifice for our final imaging purposes since accounting for nonlinearity could substantially increase the algorithm processing time. Accounting for nonlinearity delivers relatively minor sonogram corrections compared to the refraction-related algorithm presented here.

Dispersion across the spectral bandwidth of the transmitted pulse is very minor. Based on previous studies on the human skull bone [26], we estimate a maximum phase velocity variation of ±30m/s over the implemented bandwidth of this study (55%@-6dB, 2MHz-CF). In our simulation, this variation is neglected, and an average sound speed of 2980m/s is used for full skull segments.

In the frequency range of our study, the radii of pores in the trabecular layer in many areas of a human skull are much smaller than the average wavelength in cortical layers of skull bone - of the order of 1.5mm at 2MHz. In other words, we are in the Rayleigh wave scattering regime in many areas of human skull. Under this condition, scattering patterns of pores lead to very low mis-directional effects, as explained in [84], and only their attenuation effect needs to be considered.

Also, shear wave transmission inside the skull and its mode conversion at the boundaries are not considered since generation of this mode is only considerable at incident angles above ~20 degrees [85].

The targets to be imaged via application of the method presented in this work are highly reflective small objects lodged in the brain tissue. Therefore the above-mentioned simplifications
are allowable, as opposed to the case of normal brain tissue imaging via collection of reflected pulses from inhomogeneities.

Moreover, the application of our beamforming model remains limited to those skull areas where the porous layer is not extremely thick or the pores are not extremely large. In other areas, the attenuation and scattering effects are predicted to become sufficiently large that the transmitted signals will be buried in noise at the observation point. An approximate map of applicable and non-applicable areas on the center-line of the sample skull in Figure 3.1 is marked as green versus red regions respectively.

Figure 3.1: Example of applicable and non-applicable regions of human skull for the reported beamforming method. The red zones refer to the skull areas possessing extreme porosity and/or great thickness [17].
Such applicable and non-applicable areas can be identified by looking at the skull profile image obtained with a phased array probe placed in contact with the head skin. If the reflection from the back wall (inner boundary of the skull interfacing the brain tissue) is clearly visible in the scan, then the segment in contact with the probe falls in the green zone and vice versa. Examples of such in-vivo B-Scans for green zones, taken in our lab via a linear phased array are shown in Figure 3.2. The two examples show different behavior of skull thickness variation at different regions.

Figure 3.2: In-vivo B-scans of human forehead profile imaged with ULA-OP in our lab. In these images the effect of the porous layer is not too extreme; the reflection from the inner boundary is detectable and therefore the inspection area is in the green zone.
3.2 Single-point adaptive focusing in transmission mode - linear arrays

Phased array imaging is achieved via multiple steering and focusing of ultrasonic waves in both transmission and reception modes. As shown in Figure 3.3.a, in a linear array of transducers, single point focusing is achieved by appropriate timing of signals for each array element. The delayed ultrasonic fields interfere constructively with each other at the intended focal area and destructively elsewhere. This way the focus can be placed at any desired angle and depth, within the eligible inspection area of the probe. In the transmission mode of an ultrasonic beamforming process, single focusing is repeated at different steering angles and focal depths until the whole area of interest in front of the array is scanned.

Figure 3.3: (a) Beam steering and focusing in transmission mode. $\tau_i$ are delays that vary as a function of each element's position. (b) Geometry used for calculating the required delays. (Reproduced, with permission, from [32])
In a homogeneous medium, the path of a beam from each array element to any observation point in front of the array can be assumed to be a straight line. Focusing can be achieved by introducing the time delays, $\tau_n$, as indicated in Figure 3.3.a. To focus at desired coordinates, $(R_F, \phi_s)$, the delays to each element are

$$
\tau_n = \frac{1}{c_0} \left[ \sqrt{R_F^2 + \frac{(N-1)^2d^2}{4}} + (N - 1)R_Fd \sin|\phi_s| - \sqrt{R_F^2 + (nd)^2 - 2nR_Fd \sin \phi_s} \right]
$$

(3.1)

where the index $n$ denotes the element number for $-(N-1)/2 \le n \le (N-1)/2$, $N$ is the total number of elements in the array, and $c_0$ is the compression wave sound speed in the propagation medium. $R_F$ and $\phi_s$ are the focal distance and the steering angle measured from the center of the array, respectively, and $d$ is the array’s pitch (Figure 3.3.b).

The next step is the calculation of the far-field response of the array of point sources under pulse excitation and generation of the array’s corresponding beam profile. It must be noted that pulse excitation is preferred over continuous wave (CW) excitation in this study. The primary effect of pulse excitation is to remove the sidelobe scalloping without affecting the main lobe dimensions. As an example, Figure 3.4 compares the far-field pulse and CW responses for a 65 element linear array [32]. The presence of a wide distribution of frequencies in the excitation waveform reduces the interference that arises from the periodic nature of the array in combination with the periodicity of the excitation.
Another advantage of using pulse excitation over CW is that it avoids overheating of the skull-brain boundary. Although CW excitation helps with delivering more power to the inspection area and therefore acquiring better contrast on the final sonogram, it also causes higher temperature, particularly at reflective boundaries and the focal point. Multi-cycled tone-burst signals are also not appropriate choices for the problem in hand, due to the fact that for phased array ultrasonic imaging systems, depth resolution is directly dependent on the duration of the excitation signal.

Therefore, a Gaussian modulated sinusoidal waveform was used in our simulation studies as an appropriate choice of excitation signal. The use of a temporal window in the excitation signal reduces the spectral leakage in the transmitted pulse. Figure 3.5 shows a time-domain plot of such a waveform and its frequency spectrum. The signal is digitized with a sampling frequency of 50MHz, which is the fixed-value sampling frequency of the probe’s controller used in the experiments (see section 4.3). The simulation is capable of accommodating any other desired waveform, temporal window and acceptable (above Nyquist) sampling frequency.
The complex pressure \( p \) at a distance \( r \) from a single point source excited by a Gaussian modulated sinusoidal waveform is given by

\[
p(r; t) = \frac{A}{r} e^{i\omega_c (t - r/c_0)} e^{-\sigma_\omega^2 (t - r/c_0)^2 / 2},
\]

where \( \omega_c \) is the angular center frequency. The -6 dB bandwidth is equal to \( 2.36 \sigma_\omega / \omega_c \) where \( \sigma_\omega \) defines the width of the temporal Gaussian window. The waveform at any observation point \((R, \theta)\) can be found by summing over all \( N \) elements of the array, i.e.

\[
p(R, \theta; t) \approx \frac{A}{R} e^{i\omega_c t} \sum_{n=-(N-1)/2}^{(N-1)/2} \left[ e^{-i\omega_c r_n/c_0} e^{-\sigma_\omega^2 (t - r_n/c_0)^2 / 2} \right] \omega_n
\]

where

\[
r_n = \sqrt{(R \cos \theta)^2 + (n d + R \sin \theta)^2}
\]

\( R \) represents the distance from the middle point (origin) of the array to the observation point; \( \omega_n \) is the spatial apodization coefficient on the \( n^{th} \) element.
Various spatial apodization functions were programmed in the simulation and are available for applying to phased array transmission. Major implemented windowing functions include “Rectangular”, “Gaussian”, “Cosine”, “Sinc”, “Hamming”, “Hanning” and “Four-term Blackman-Harris” [86]. The use of spatial apodization over active elements of an ultrasonic array will reduce the magnitude of sidelobes relative to the main lobe. However, since some increase in the angular width of the main lobes (and therefore a loss in resolution) accompanies this reduction, it is important that the trade-off be carefully examined. As an example, in Figure 3.6 the effects of three different apodization functions are illustrated for a 128 element linear array and are compared to a simple rectangular window whose first sidelobe is just 13 dB below the peak [32]. For example, with the cosine-squared window the first sidelobe is at $–32$ dB, though the width of the main lobe at the $–3$ dB level is increased by a factor of $\sim 1.6$.

An investigation on the influence of nine different apodization schemes for wideband excitation of a linear ultrasound array was reported in [86]. For each scheme, the relation between the

![Figure 3.6: Illustrating the effect of apodization on the farfield response of a 128 element linear array. (i) Rectangular window, (ii) Gaussian window, (iii) cos window, (iv) $\cos^2$ window. For all four windows the main lobe angular full width at half maximum (FWHM) and the first side lobe amplitude (S. Lobe) are given in the table. The window shapes are also shown. Reproduced with permission from [32].](image)
FWHM (which governs the lateral resolution) and the beam width at –25 dB (which is a measure of the sensitivity to artifacts) was examined. As a conclusion, the “Cosine”, “Hamming”, “Sinc” and “10% truncated Gaussian” were reported to be optimal in regard to image quality. Therefore, “10% truncated Gaussian” was used as the default apodization function ($\omega_n$) in the simulation. Simulation results of section 3.3 are also based on this choice of apodization.

To proceed with the transmission mode beamforming algorithm, calculation of the far-field response of the linear array is accomplished in the following way. The excitation signal of Eq. 3.2, (Figure 3.5), can be digitized and addressed as $S_{r_{ef}}(m)$, with $m$ being the sample number in the excitation signal. A sampling frequency of 50MHz was used in the simulation as mentioned previously. The total intensity at each observation point (pixel) in front of the array could be then calculated by a first summation over the $m$ samples of the spatially apodized excitation signal for each element, $S_n$, and a second summation over all $n$ active elements on the array as

$$P(R, \theta) = \sum_m \left| \sum_n S_n(R_F, \phi_S, m + \xi_n) \right|^2$$  \hspace{1cm} (3.4)

where

$$S_n(R_F, \phi_S, m) = EXP(-\alpha r)S_{r_{ef}}(R_F, \phi_S, m + \tau_n)\omega_n$$  \hspace{1cm} (3.5)

In the above equations, $\xi_n$ is the difference between the paths from the observation point to the $n^{th}$ element and from the observation point to the middle of the array in signal sample units. The parameter $\alpha$ is the attenuation coefficient ($Np/m$), and $r$ is the beam path length (meters) from the observation point to the $n^{th}$ element of the array.
In the development of Eqs. 3.1-3.5, straight paths are assumed from the array elements to desired focal coordinates or observation points. This is a key assumption in most existing conventional beamforming techniques. But in the presence of a barrier of non-uniform thickness, e.g. human skull, ultrasonic fields are phase aberrated and ultrasonic beams can be refracted away from their intended targets. Under such circumstances, the use of original timing patterns (calculated based on the straight path assumption) results in distortion and displacement of the beam from its intended focal spot, and compromises the final image. An adaptive beamforming algorithm is required to avoid such image degradation.

In the proposed method, geometrical and acoustical properties of the simulated skull layer and brain tissue are used to find physically possible refracted acoustic ray paths between the active elements of the array and the desired focal point. In simulation, the acoustical and geometrical properties of the skull layer are input by the user. In practice, as will be explained in section 5.2, these properties will be measured by the imaging probe, via the method recently disclosed by the author and his team in [24].

In use of state-of-the-art phased arrays with small footprint, the probe surface can be practically held parallel to the skull’s outer boundary in most areas. For this reason the simulated skull layers were designed with flat outer boundaries and undulating inner boundaries. In this thesis outer boundary refers to the skull-skin boundary and inner boundary refers to the skull-brain tissue boundary. Also to mimic the real situation (see Table 2.1), the thickness of the simulated skull layers varies from 3 to 15mm.

After defining the skull layer in the simulation, the proposed method first discretizes the inner boundary of the skull profile. The step size between the boundary points is a user-defined value.
Use of finer steps results in calculation of more accurate refracted paths but with cost of higher processing time. The process continues by first solving the Helmholtz linear wave propagation equation in each medium:

$$\nabla^2 \Phi + k^2 \Phi = 0$$  \hspace{1cm} (3.6)

In Eq. 3.6, $k = k / \sqrt{1 + j \omega \kappa \mu_o}$ is a complex wave number in which $k = \omega / c_o$ and $c_o = 1 / \sqrt{\kappa \rho_o}$. $\kappa$ and $\mu_B$ are the compressibility coefficient and bulk viscosity of the propagation media respectively. The algorithm then searches through the boundary points and picks the one at which all boundary conditions are satisfied. Boundary conditions include Snell’s law, Fermat's principle, and consideration of reflection/transmission coefficients. The identified boundary point defines the physically possible refracted path between an active element on the array and a desired focal point. This process is then repeated for all the active elements on the array which participate in the beamforming process. The calculated refracted paths are then used in Eqs. 3.1-3.5, replacing their original straight counterparts $R_F$, $R$, $r$ and $r_n$, to derive a new timing pattern and generate a corrected beam profile. Accommodating refracted paths into Eq. 3.4, the far-field response of the array is then calculated at each pixel and the corrected beam profile is finally reconstructed.

### 3.3 Single-point adaptive focusing via linear arrays – Simulation results

The mathematical approach of the proposed adaptive beamforming technique was verified through computer simulation. The program receives ultrasonic and geometrical properties of the skull and brain tissue (e.g. sound velocities, densities, geometrical profile of skull layer, etc.) as
inputs. The simulation results presented in this section were obtained with the following input parameters:

a) Propagation media - sound velocities of 2980m/s and 1540m/s and compressional wave attenuation of 24dB/cm@2MHz and 1.8dB/cm@2MHz for the skull and the propagation media respectively. These values were picked based on the average values for most areas of the human skull and brain tissue respectively (see section 2.2.2).

b) Transducer array - The simulation results are presented for an array of 64 point source transducers pitched at 0.17 mm making a total aperture size of 10.71mm. Spatial apodization of Gaussian type was used to weight the element outputs to minimize the appearance of side lobes in the beam profile pattern.

c) Excitation signal - A 6μs-long Gaussian-modulated sinusoidal waveform (Figure 3.5) with central frequency of 2MHz and 40% bandwidth, digitized at the sampling frequency of 50MHz and quantized with 16-bit resolution was introduced to the simulated beamformer.

After calculating the corrected time delay pattern for a desired focal point, the simulation is capable of generating an intensity profile of the array’s ultrasonic field. As an example, Figure 3.7 shows (a) simulated original transmitted beam intensity profile without accounting for the skull layer, (b) simulated distorted intensity profile with the skull present, but with no adaptive beamforming, and (c) beam from Figure 3.7.b corrected by the new beamforming technique. Each image covers an area of 6cm x 10cm with a pixel resolution of 0.5mm. The white cross shows the location of the intended focus. It can be seen that the misdirected beam of Figure 3.7.b is redirected to a path through the white cross in Figure 3.7.c after the compensation algorithm is applied.
Figure 3.7: (a) Simulated original transmitted beam profile in the absence of a simulated skull, (b) distorted beam in the presence of a simulated skull with no adaptive beamforming, and (c) beam from Fig. 2 (b) corrected by the new beamforming technique. Results for focal coordinates of (5cm, 0°) are shown. The intended target is designated by a white cross.

In terms of timing the simulated array elements, Figure 3.8 represents time delay patterns over 64 transmitting elements used in Figure 3.7. The curve labeled as “original” shows the excitation timing sequence of the array elements when beamforming at focal coordinates of (R=5cm, 0°) is intended in absence of a simulated skull layer (Figure 3.7.a). The curve labeled as “corrected”, on the other hand, represents excitation timing sequence of array elements in the same scenario, but this time in the presence of the simulated skull layer used to generate Figure 3.7.c. The timing sequence is now adapted to the local geometry and acoustical properties of the simulated skull layer in contact with the probe. Through application of this modified timing pattern, the out-of-focus beam of Figure 3.7.b was brought back to its desired direction as was shown in Figure 3.7.c.
Figure 3.8: Original and corrected time delay patterns responsible for beamforming at focal coordinates of (5cm, 0°), corresponding to Figure 3.7.a and Figure 3.7.c respectively.

At first glance, it is notable that the original timing sequence of the array elements in Figure 3.8 is distributed symmetrically around the center of the transmitting aperture for the example focal coordinates of (5cm, 0°). This is due to the symmetry of the propagation medium in the absence of a simulated skull layer. The corrected timing sequence, on the other hand, is not symmetric as it was adapted to the asymmetric geometry of the simulated skull layer.

As a simple verification of the two plots, comparison between the original and corrected timing of single elements of the array can be made. In the specific case of the above simulated system, when the simulated skull is present, the 40th element of the array is excited with a delay of 4.86x10^{-7} seconds compared to its excitation in the absence of the layer. A ray-tracing
calculation of the straight and refracted paths from this element to the focal coordinates verifies the result.

3.4 Single-point adaptive focusing in transmission mode via matrix arrays - Theory

Upon completion of the theory and simulation for the case of linear probes and 2D adaptive beamforming, the developed method was extended to the case of matrix probes and 3D adaptive beamforming. The general advantage of matrix probes compared to linear arrays is in generating volumetric images and therefore covering a larger inspection area in a single positioning of the probe.

Transition from linear arrays to matrix probes was a necessity for another reason as well: In linear ultrasonic arrays, array elements are long strips of piezoelectric material and therefore they are point-like only along the width of the probe (x-axis in Figure 3.3). In contrast, the random thickness variation of human skull has a two dimensional nature, i.e. along both width and length of an ultrasonic probe. Since the proposed algorithm depends on accurate knowledge of the thickness profile of the skull segment immediately below the array (a 2D phenomenon), linear arrays are not able to handle the task. Therefore development of the proposed algorithm in case of linear probes was only accomplished as a proof-of-concept for the original theory. Although it maintains its potential for being implemented on other biomedical or industrial ultrasonic applications, it is not directly applicable to beamforming through scattering layers which do not possess a linear symmetry (thickness variation limited to one axis).
In conventional 3D beamforming via matrix arrays, the principal physical theory of focusing and steering remains the same as for linear arrays. In conventional 3D beamforming systems, the propagation media is assumed homogeneous and symmetric. Governing equations in the absence of skull bone expressed in three dimensions can be derived with the addition of a third dimension into Eq. 3.1 and Eqs. 3.3 - 3.5. With this consideration, to focus at desired coordinates \((R_F, \phi, \psi)\) in Figure 3.9,

\[
\tau_{nm} = \frac{1}{c_0} \left[ \sqrt{R_F^2 + \frac{d^2}{4} [(N-1)^2 + (M-1)^2]} + [(N-1) \cos \phi + (M-1) \sin \phi] dR_F \cos \Psi \right] - \sqrt{R_F^2 + (n^2 + m^2)d^2 - 2(n \cos \phi + m \sin \phi)dR_F \cos \Psi} \tag{3.7}
\]
where the index \( n \) denotes the element number along x-axis for \( -(N-1)/2 \leq n \leq (N-1)/2 \), \( N \) is the total number of elements in width of the array. The index \( m \) denotes the element number along the y-axis for \( -(M-1)/2 \leq m \leq (M-1)/2 \), \( M \) is the total number of elements in length of the array. \( c_0 \) is the compression wave sound speed in propagation medium and \( d \) is the array’s pitch.

Using the same Gaussian modulated sinusoidal waveform of Eq. 3.2, the waveform at any observation point \((R, \theta, \gamma)\) can be found by summing over all \( N \times M \) elements of the array, i.e.

\[
p(R, \theta, \gamma; t) \approx \frac{A}{R} e^{j\omega t} \sum_{m=-(M-1)/2}^{(M-1)/2} \sum_{n=-(N-1)/2}^{(N-1)/2} \left[ e^{-j\omega c r_{nm}/c_0} e^{-\sigma_0^2 (t-r_{nm}/c_0)^2/2} \right] \omega_{nm} \quad (3.8)
\]

where

\[
r_{nm} = \sqrt{(n^2 + m^2)d^2 + R^2 + 2dR \cos(n \cos \theta + m \sin \theta)}
\]

\( R \) represents the distance from the centre of the array to the observation point; \( \omega_{nm} \) is the spatial apodization coefficient on the \((n,m)^{th}\) element. The excitation signal of Eq. 3.2, (Figure 3.5), can be digitized and addressed as \( S_{\text{ref}}(k) \), with \( k \) being the sample number in the excitation signal.

The total intensity at each observation point (pixel) in front of the array could be then calculated via a three dimensional version of Eq. 3.4:

\[
P(R, \theta, \gamma) = \sum_k \left| \sum_{n,m} S_{nm}(R_F, \psi, \phi, k + \xi_{nm}) \right|^2 \quad (3.9)
\]

where
\[ S_{nm}(R_F, \psi, \phi, k) = EXP(-\alpha r)S_{ref}(R_F, \psi, \phi, k + \tau_{nm})\omega_{nm} \quad (3.10) \]

In the above equations, \( \xi_{nm} \) is the difference between the paths from the observation point to the \( (n,m)^{th} \) element and from the observation point to the center of the array in signal sample unit. \( r \) is the beam path length, in meters, from the observation point to the \( (n,m)^{th} \) element on the array.

It must be noted again that in development of Eqs. 3.7-3.10, straight paths are assumed from the array elements to desired focal coordinates or observation points. But, in the presence of a barrier of non-uniform thickness, e.g. human skull, the propagation media are not symmetric anymore. A 3D adaptive beamforming algorithm is required which is described in the next section.

### 3.5 Single-point adaptive focusing via matrix arrays – Algorithm and Simulation

The simulation of the 3D adaptive beamforming was developed in polar coordinates, as opposed to the Cartesian coordinates used in the previous case of linear arrays. To be able to focus the transmitted ultrasound field on an intended focal point, a new transmission time delay pattern, adapted to local geometry and acoustical properties of the skull, is required for the array. Similar to the linear array’s case, to calculate this delay pattern and compensate for the phase aberration of skull, the refracted path between each element of the array and the intended focal point must be found. These refracted paths will then replace the conventional straight paths assumed in the development of Eqs. 3.7-3.10; these refracted paths will be used to calculate the adapted time delay pattern for the individual array elements and generate a corrected beam profile. As a first step in this task, simulation of two major components, i.e. the matrix array and the propagation media, are explained.
3.5.1 Matrix array simulation

A MATLAB code was developed by the author to geometrically design and simulate matrix arrays with square or rectangular elements. As an example, Figure 3.10 shows the MATLAB configuration of the simulated probe used in the transmission mode adaptive beamforming calculations of this section. Dark elements refer to the active elements on the probe and light gray elements refer to inactive elements. The small red crosses show the center of each element. Any combination of array elements can be picked as the active aperture of the array. As shown in Figure 3.10.b, each array element can itself be simulated by as many sub-elements (point sources) as desired for optimum accuracy of the simulation results. The choice of the number of sub-elements for each element could be made based on the frequency of the probe and the size of array elements. Normally when the element size is less than the wavelength, which is the case of this study, each element can be simulated as a single point source, although this approximation is only valid as long as maximum steering in a beamforming scenario does not exceed the -6dB main-lobe limit of each element’s directivity function.
Table 3.1 lists the geometrical and acoustical characteristics of the simulated probe of Figure 3.10 based on which simulation results are provided later in this section.

**Table 3.1: Geometrical and Acoustical Characteristics of the simulated probe**

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of channels (NxM)</td>
<td>16x16</td>
</tr>
<tr>
<td>Total number of active elements</td>
<td>192</td>
</tr>
<tr>
<td>Pitch along x-axis</td>
<td>0.55 mm</td>
</tr>
<tr>
<td>Pitch along y-axis</td>
<td>0.55 mm</td>
</tr>
<tr>
<td>Kerf along x-axis</td>
<td>0.1 mm</td>
</tr>
<tr>
<td>Kerf along y-axis</td>
<td>0.1 mm</td>
</tr>
<tr>
<td>Central frequency</td>
<td>2 MHz</td>
</tr>
<tr>
<td>Bandwidth</td>
<td>40 %</td>
</tr>
</tbody>
</table>
3.5.2 Propagation environment simulation

*Skull layer* – Another MATLAB code was developed for design of a skull layer. The code and its graphical interface provide the flexibility to design a skull layer with any desired acoustical and geometrical properties. The thickness profile of the skull can be input using boundary functions, or by means of a manually defined thickness matrix. A simulated skull layer in our case possesses a flat outer boundary and an undulating inner boundary. A longitudinal sound speed of 2980m/s, density of 1950kg/m³ and attenuation of 24dB/cm were used for the simulated skull layers. The simulated skull profile is then discretized with a mesh-grid of boundary points. Figure 3.11 shows an example of such simulated skull layer. The grid size is a user defined value which must be chosen based on the thickness gradient of the skull layer.

![Example of a simulated skull layer with an undulating inner boundary](image)

*Figure 3.11: Example of a simulated skull layer with an undulating inner boundary*
In the next step, normal and gradient vectors are calculated over the inner boundary of the skull layer at each grid point, as shown in Figure 3.12. The small blue arrows at the grid points show the field of normal vectors of the layer while the contoured green arrows depict the field of gradient vectors of the layer. The two green and red curves on the figure represent projections of the x and y axes on the inner boundary of simulated skull layer. The normal and gradient vectors fields are then output to the main body of the simulation code. Such vectors fields will be later used to apply boundary conditions and find corrected refracted paths from each array element to a desired focal point.
Figure 3.12: Normal and gradient vectors fields of the simulated skull layer
**Brain tissue** – In the same manner as the previous case (linear arrays), the brain tissue was simulated as a homogeneous propagation medium with compressional wave speed of 1540 m/s, density of 1040 kg/m$^3$ and attenuation of 1.8 dB/cm corresponding to a frequency of 2 MHz.

### 3.5.3 Adapted time delay calculation algorithm

After the simulated planar array and the skull layer are input to the main simulation engine, the next step is to find the corrected time delay pattern for the array elements. The pattern should be adapted to the designed propagation media for a desired focusing point in the brain. First, physically possible refracted paths from each active array element (dark elements in Figure 3.10) to a desired focal point are to be found. For this purpose, the simulation first solves for the Helmholtz wave propagation equation in polar coordinates:

\[
\frac{1}{r^2} \frac{\partial}{\partial r} \left( r^2 \frac{\partial \Phi}{\partial x} \right) + \frac{1}{r \sin \theta} \frac{\partial}{\partial \theta} \left( \sin \theta \frac{\partial \Phi}{\partial \theta} \right) + \frac{1}{r^2 \sin^2 \theta} \frac{\partial^2 \Phi}{\partial \varphi^2} + k^2 \Phi(r, \theta, \varphi; \omega) = 0 \tag{3.11}
\]

During the next step, the algorithm searches through the boundary points and picks the one at which all boundary conditions are satisfied. For optimum results, more points are interpolated in between the original grid points on highly undulating areas of the inner boundary; on the other hand, when the thickness profile has a smooth variation, some boundary points are skipped.

In the proposed algorithm, there are three major boundary conditions to be satisfied:

1. **The normal vector at each grid point should be perpendicular to the norm of the communal plane of incident and refracted beams**; It must be noted and realized that this boundary condition was naturally satisfied in the previous case of 2D adaptive beamforming with linear arrays, section 3.2. In the current 3D beamforming scenario, first the points on the inner boundary which
satisfy this condition are found. To find these points, the angle between the two decisive vectors (i.e. the normal vector and the norm of the communal plane) is calculated at all the inner boundary points. An example of such angle distribution is shown in Figure 3.13. This angle distribution corresponds to the simulated skull profile of Figure 3.12 when seeking for the refracted path between first active element on the simulated array (top-left corner element in Figure 3.10) and an intended focal coordinates of \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\).

![Angle distribution](image)

Figure 3.13: Angle distribution for the simulated skull profile of Figure 3.12 when seeking for the refracted path between first active element on the simulated array (top-left corner element in Figure 3.10) and an intended focal coordinates of \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\).

In general, on the angle distributions, only boundary points possessing a value of 90° (contour level of 90°), satisfy the above boundary condition.

2. On a fluid-solid boundary (brain-skull in our case), the fluid pressure must be equal to the normal component of the stress just within the solid; the tangential stress in the solid must
vanish, and the normal component of the velocity must be continuous; Mathematical formulation and full description of this boundary condition can be found in [87]. Snell’s law of refraction is a direct consequence of this boundary condition. In simulation, to identify the points on the skull-brain interface which satisfy this boundary condition, a second distribution is calculated for all boundary points. This distribution will be referred to as “Snell’s variance” distribution in this context. The Snell’s variance distribution for the simulated skull profile of Figure 3.12, when seeking for the refracted path between the first active element on the simulated array (top-left corner element in Figure 3.10) and an intended focal coordinates of \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\), is presented in Figure 3.14. In general, on Snell’s variance distributions, only boundary points possessing value of 0 (contour level of 0) satisfy the above boundary condition.

![Snell’s variance distribution](image)

**Figure 3.14:** Snell’s variance distribution for the simulated skull profile of Figure 3.12 when seeking for the refracted path between first active element on the simulated array (top-left corner element in Figure 3.10) and an intended focal coordinates of \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\).
3. *Fermat’s principle*; Contour plots of the skull-brain boundary points which satisfy the above two boundary conditions are plotted in Figure 3.15. The green curves are the 90°-level contour plot of the angle distribution (Figure 3.13) and the dotted-blue lines are the 0-level contour plot of the Snell’s variance distribution (Figure 3.14). The intersection points of the two plots satisfy both boundary conditions. These points are identified by small black circles in Figure 3.15. Among these selected points, the one that will correspond to the shortest refracted path between the selected array element and the intended focal point is selected based on the Fermat’s principle. This point is shown by a small red cross in Figure 3.15. Therefore the ultrasound transmitted from the array element under consideration to the desired focal point in the brain will pass through this point on the skull-brain interface.

![Figure 3.15](image-url)

*Figure 3.15*: Contour plots of the boundary points which satisfy the first two boundary conditions when seeking for the refracted path between first active element on the simulated array (top-left corner element in Figure 3.10) and an intended focal coordinates of \((R, \varphi, \psi) = (5\text{ cm}, 0^\circ, 0^\circ)\).
Finally, the boundary point that corresponds to the \((x,y)\) coordinates of the red-cross in Figure 3.15, must be found. This point does not necessarily relate to an original skull boundary mesh grid point. Therefore a skull thickness value at this location is interpolated from values at the boundary grid points. This interpolation method results in fairly accurate calculation of the final boundary point even in case of a coarse mesh grid and could also dramatically improve the overall processing time of the algorithm.

To calculate a final corrected time delay pattern adapted to the simulated skull layer, the above process now needs to be repeated for all the active elements on the array. Figure 3.16.a shows the original straight paths from the array element to the intended focal coordinates of \((R, \varphi, \psi) = (5cm, 0^\circ, 0^\circ)\) when the skull layer is absent in the simulation engine. Figure 3.16.b, on the other hand, shows the physically possible refracted paths calculated via the algorithm explained above when the skull layer is present in the simulation.
Figure 3.16: Comparison of the original straight and physically possible refracted paths between the simulated array elements and the intended focal point.

With the knowledge of these refracted paths for each single focusing scenario, the transmission mode corrected time delay pattern for the array elements can be now calculated through Eq. 3.7. The results of such calculation are shown in Figure 3.17 in which the original and corrected time delay patterns are compared. Figure 3.17.a shows the timing sequence of the array elements when beamforming at the same focal coordinates of \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\) is intended in absence of a simulated skull layer. Figure 3.17.b, on the other hand, represents the excitation timing sequence of array elements in the same scenario, but this time in the presence of the simulated skull layer. The timing sequence is now adapted to the local geometry and acoustical properties of the simulated skull layer in contact with the probe.
Figure 3.17: (a) Original and (b) corrected time delay patterns

At first glance, it is notable that the original timing sequence of the array elements, Figure 3.17.a, is distributed symmetrically around the center of the transmitting aperture. This is due to the symmetry of the propagation medium in the absence of a simulated skull layer. The corrected timing sequence, on the other hand, is not symmetric as it was adapted to the asymmetric geometry of the simulated skull layer.

After calculating the corrected time delay pattern for a desired focal point, the simulation is capable of generating an intensity profile of the ultrasonic field transmitted by the array. Figure 3.18 shows three cross-sectional beam profiles numerically generated through the simulation. Each image represents a 4cmx5cm area parallel to the plane of the 2D array and 5 cm away from it. The intended focus (white cross) has the coordinates of \((R, \varphi, \psi) = (5 \text{ cm}, 0^\circ, 0^\circ)\).
Figure 3.18: Simulated cross-section beam profiles; (a) original, with no simulated skull layer (b) distorted with no correction, in presence of a simulated skull layer and (c) corrected in presence of the simulated skull layer. White cross shows the intended focus.

Figure 3.18.a shows the cross-sectional beam profile without the phantom. Figure 3.18.b shows the beam pattern through the simulated skull layer of Figure 3.12 when the conventional delay scheme (without correction) of Figure 3.17.a is applied. It can be seen that the beam is strongly deflected from its intended focal point. Figure 3.18.c shows the simulated beam profile after applying the corrected delay pattern of Figure 3.17.b. It can be seen that the misdirected beam of Figure 3.18.b is now brought back to the desired orientation and passes through the intended focus.

3.6 Discussion

The numerical simulation results in both cases of linear and matrix arrays validated the proposed theory of adaptive beamforming. It must be noted that in the simulation results presented in this chapter, no major sources of error can be identified. All the input parameters, e.g. sound speeds in the propagation media, are user-defined and single-valued. Some very minor sources of error, related to numerical solutions of systems of equations, can be identified, although their effect can be kept small. Examples include errors associated with sampling of the excitation signal or
discretization of the simulated skull’s inner boundary. By use of a very high sampling frequency (e.g. high order multiples of the Nyquist frequency) or very fine mesh-grid on the skull’s inner boundary (e.g. small fraction of the wavelength) the corresponding contribution to simulation errors can be kept very small. Comprehensive quantitative assessment and error analysis of the developed adaptive beamforming method is therefore delayed to chapter 5 where the new beamforming strategy is implemented in experimental setups.

3.7 Chapter conclusion

The main theory and a new algorithm for single-point adaptive focusing in transmission mode were presented in this chapter. The theory and algorithms were initially described for the case of linear phased arrays used for 2D adaptive beamforming. The proposed theory was then extended to the case of matrix phased arrays for 3D adaptive beamforming, employing some simplifications to reduce processing time. These simplifications were addressed and discussed in this chapter.

During this study, numerical simulations were developed in MATLAB to implement and verify the proposed theory in the cases of both linear and matrix arrays. Simulation results for linear and matrix arrays were presented in section 3.3 and section 3.5 respectively. The numerical simulation results validated the proposed theory for single-point adaptive focusing in transmission mode. The single-point adaptive focusing simulation results of this chapter are an indicator of the potential of the approach to correct for phase aberration effects of skull layers in experimental setups. Experimental verification of the simulation results of the current chapter will be conducted in chapter 5.
Chapter 4

Equipment and materials

4.1 Introduction

To verify the theory and simulation results of the proposed adaptive beamforming method, the author conducted a series of laboratory experiments. Before discussing the details and results of these experiments, it is necessary to describe the laboratory equipment and materials involved in the testing.

This chapter starts with description of a novel composite material specifically developed for human skull phantoms. The final generation of our phantoms has its main geometrical and acoustical properties, i.e. sound speed, attenuation and density, well within the ranges reported in the literature for the human skull. The acoustical and geometrical characteristics of the developed phantoms are described in this chapter. The specific skull phantoms used in the experimental setups of the presented study are also described.

The chapter then continues with describing the main electrical and mechanical hardware components used in the experimental setups. The Ultrasound Advanced Open Platform (ULA-OP) phased array controller (Università di Firenze, Florence, Italy), the linear phased array and the custom designed planar phased array are major elements to be described in this chapter.
4.2 Skull phantoms

Obtaining and maintaining *ex-vivo* human skulls for a research-based study would be faced with major difficulties. Such difficulties include extensive and time-consuming permission requirements from ethical institutions, fast and continuous degradation of acoustical properties of real skulls, storage and maintenance difficulties, and chance of contamination of the ex-vivo samples. Therefore, from early stages of the presented study, development of ultrasonically realistic human skull phantoms has been a major goal of the project. In contrast to ex-vivo tissues, the bone phantom material can maintain its custom designed physical and acoustical properties unchanged for long time.

As explained in Section 2.2, human skull bone consists of two cortical bone layers (inner and outer layers) and a middle trabecular bone, which in this particular case is called *diplo’e*. The acoustical properties can be considered as a weighted average of the properties of cortical and trabecular bones. The main acoustical parameters to be mimicked in skull phantoms include attenuation, speed of sound and density. It was previously mentioned (section 2.2.2) that according to published data, the ranges of attenuation and sound speed values in actual skull bone are very broad. The properties of the composite material presented in this section fall within this broad range. It is also important to point out that our procedure allows us to form skull bone phantoms with custom-designed properties.

To date, three generations of skull phantoms have been developed by our team, with the development of the first generation phantoms by the author himself. It included the design, fabrication of the composite material, and measurement of the acoustical parameters of the fabricated phantoms. Since the first generation, the development of more realistic skull phantoms has been pursued by a M.Sc. student in our research team, under general supervision of the
author. This collaboration resulted in development of second and third generations of our skull phantoms. The description of the composite material and the fabrication process, along with explanation and discussion of the techniques used in calibration of major acoustical properties of the developed phantoms are provided in [18], [88]. Only a summary of the major milestones and acoustical properties of the skull phantoms is presented in this section.

**First and Second generations** – The first generation of skull phantoms was developed with ultrasonic properties roughly approaching those of the human skull. A custom combination of ‘Abocast epoxy’ and titanium powder (grit size 50μm) was used as the composite material for the bone phantoms of this generation. It was possible to change acoustical properties by changing the mass ratio of the components. Different calibration and test samples with different compositions were prepared and investigated to correlate between physical (e.g. powder grit size and curing process) and major acoustical properties of the phantoms. Figure 4.1.a shows a picture of some calibration samples and Figure 4.1.b shows a picture of one of the final phantoms of the first generation.

![First generation skull phantoms](image)

**Figure 4.1: First generation skull phantoms**

As a result of our continuing research, the second generation of skull phantoms was prepared from a new and specially designed composite material, with more realistic acoustical properties.
Compared to the previous composite material, some unique properties of the new material such as viscosity, curing time and curing temperature allowed us to have a better control over the major acoustical properties. This property alongside a custom designed fabrication process developed by A. Wydra, [18], [88], resulted in creating various bone phantoms with desired shapes. These phantoms mimic all the required acoustic properties of the *cortical* human bones (see Table 2.1). In order to find a correlation between the ultrasonic properties and the density of the material, a group of calibration sample phantoms were developed and studied. A picture of such sample phantoms is shown in Figure 4.2.a. Figure 4.2.b shows a picture of a final phantom from this generation.

![Second generation skull phantoms](image)

**Figure 4.2: Second generation skull phantoms**

Geometrical and acoustical properties associated with final composite materials of the first two generations are listed in Table 4.1. In phantoms of both first and second generations, the skull was approximated as a single layer of cortical bone with thickness variation limited to one axis (linear symmetry), in the range of 3-15 mm.
Table 4.1: Properties of first and second generations of skull phantoms

<table>
<thead>
<tr>
<th>Property</th>
<th>Skull phantom (1&lt;sup&gt;st&lt;/sup&gt; gen.)</th>
<th>Skull phantom (2&lt;sup&gt;nd&lt;/sup&gt; gen.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density [g/cm&lt;sup&gt;3&lt;/sup&gt;]</td>
<td>1.85 ± 0.03</td>
<td>2.1 ± 0.03</td>
</tr>
<tr>
<td>Speed of sound [m/s]</td>
<td>2361 ± 10</td>
<td>3011 ± 30</td>
</tr>
<tr>
<td>Acoustic impedance [MRayl]</td>
<td>4.36 ± 0.3</td>
<td>6.3 ± 0.3</td>
</tr>
<tr>
<td>Attenuation @ 2.25MHz [dB/cm]</td>
<td>11.1 ± 0.5</td>
<td>14.6 ± 0.7</td>
</tr>
<tr>
<td>Thickness [mm]</td>
<td>3 – 15</td>
<td>3 – 15</td>
</tr>
</tbody>
</table>

**Third and final generation**- Presence of an intermediate trabecular layer in between the inner and outer cortical layers of the skull, was not considered in the previous generations. Development of realistic trabecular bone phantoms was the major achievement for this third generation. Representative examples of some sample trabecular phantoms developed for calibration studies are shown in Figure 4.3. A list of measured acoustical properties of these samples is also provided in Table 4.2.

![Figure 4.3: Trabecular bone calibration phantoms](image_url)
Table 4.2: Acoustical properties of the sample trabecular bone phantoms of Figure 4.3

<table>
<thead>
<tr>
<th>Sample</th>
<th>Freq.</th>
<th>Density (g cm$^{-3}$)</th>
<th>Porosity (%)</th>
<th>Speed of sound (ms$^{-1}$)</th>
<th>$\alpha$ (dB cm$^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>1 MHz</td>
<td>1.90</td>
<td>~19</td>
<td>2862 ± 10</td>
<td>18.9 ± 0.7</td>
</tr>
<tr>
<td></td>
<td>2.25 MHz</td>
<td></td>
<td></td>
<td>2923 ± 10</td>
<td>33.5 ± 1.0</td>
</tr>
<tr>
<td>B</td>
<td>1 MHz</td>
<td>1.85</td>
<td>~24</td>
<td>2828 ± 10</td>
<td>21.8 ± 1.0</td>
</tr>
<tr>
<td></td>
<td>2.25 MHz</td>
<td></td>
<td></td>
<td>2885 ± 10</td>
<td>39.0 ± 0.7</td>
</tr>
<tr>
<td>C</td>
<td>1 MHz</td>
<td>1.83</td>
<td>~26</td>
<td>2817 ± 10</td>
<td>22.2 ± 0.4</td>
</tr>
<tr>
<td></td>
<td>2.25 MHz</td>
<td></td>
<td></td>
<td>2832 ± 10</td>
<td>39.3 ± 0.7</td>
</tr>
<tr>
<td>D</td>
<td>1 MHz</td>
<td>1.76</td>
<td>~33</td>
<td>2720 ± 10</td>
<td>24.9 ± 0.5</td>
</tr>
<tr>
<td></td>
<td>2.25 MHz</td>
<td></td>
<td></td>
<td>2760 ± 10</td>
<td>39.4 ± 0.3</td>
</tr>
<tr>
<td>E</td>
<td>1 MHz</td>
<td>1.63</td>
<td>~46</td>
<td>2514 ± 10</td>
<td>43.6 ± 0.5</td>
</tr>
<tr>
<td></td>
<td>2.25 MHz</td>
<td></td>
<td></td>
<td>2570 ± 10</td>
<td>47.6 ± 0.7</td>
</tr>
<tr>
<td>F</td>
<td>1 MHz</td>
<td>1.57</td>
<td>~51</td>
<td>2393 ± 10</td>
<td>54.1 ± 0.6</td>
</tr>
<tr>
<td></td>
<td>2.25 MHz</td>
<td></td>
<td></td>
<td>2431 ± 10</td>
<td>58.0 ± 0.6</td>
</tr>
</tbody>
</table>

One can notice the trend that higher porosity and lower density result in reduced observed sound velocity. Another attenuation-related trend previously reported in [89]–[91] is that the attenuation is proportional to the porosity up to a certain porosity level (60% or 70%) after which the trend is inverted. An explanation for this trend has not been reported.

Upon successful development of trabecular bone phantoms, the third and final generation of layered skull phantoms has also been developed. The skull phantoms of this generation were designed and manufactured as a combination of two cortical layers (inner and outer layers) and a middle trabecular layer (diplo’e). The phantoms have their main geometrical and acoustical properties, i.e. sound speed, attenuation and density, well within the ranges reported in the literature for the human skull. By controlling the thickness of each layer as well as the density
and the distribution of pores in the phantoms, it is possible to control the total density, attenuation and sound speed of each phantom. Figure 4.4 shows some sample pieces with different porosity layers.

![Sample skull phantoms of the third generation with different porosity layers](image)

**Figure 4.4: Sample skull phantoms of the third generation with different porosity layers**

Figure 4.5.a and Figure 4.5.b show two final skull phantoms from this generation. These phantoms were used in the experimental evaluation of the author’s proposed adaptive beamforming technique, as will be addressed in future chapters. The phantom of Figure 4.5.a was prepared with a linear symmetry which is suitable for testing with linear arrays. It is flat on one side and has an undulating structure on the other side. The phantom of Figure 4.5.b, on the other hand, is suitable for testing with planar arrays and similar to real human skull, it has curves along both surface axes.
Table 4.3 lists the main acoustical properties of the skull phantoms shown in Figure 4.5 as well as their corresponding average reported ranges for non-extreme regions of human skull based on the literature data.
### Table 4.3: Properties of human skull and the third generation of skull phantoms

<table>
<thead>
<tr>
<th>Property</th>
<th>Human Skull</th>
<th>Skull phantom (3\textsuperscript{rd} gen.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density [g/cm(^3)]</td>
<td>1.8 – 2.1</td>
<td>2.04 ± 0.03</td>
</tr>
<tr>
<td>Speed of sound [m/s]</td>
<td>2800 – 3500</td>
<td>2989 ± 10</td>
</tr>
<tr>
<td>Acoustic impedance [MRayl]</td>
<td>5.12 – 6.79</td>
<td>6.1 ± 0.3</td>
</tr>
<tr>
<td>Attenuation @ 2.25MHz [dB/cm]</td>
<td>17 – 38</td>
<td>24.1 ± 0.7</td>
</tr>
<tr>
<td>Thickness [mm]</td>
<td>3 – 15</td>
<td>3 – 15</td>
</tr>
</tbody>
</table>

Consideration of the acoustic nonlinearity of the human skull is outside the scope of this study and therefore it was not considered while developing the skull phantoms. It should also be clarified that our phantoms at the current stage of development cannot mimic the anatomical microstructure and sub-microstructure of bones.

### 4.3 Open source controlling system for ultrasound probes

An open source controlling system for ultrasound probes, Figure 4.6, was purchased for lab experiments and final implementation of the author’s proposed adaptive beamforming method. A detailed description and capabilities of the Ultrasound Advanced Open Platform, ULA-OP (Università di Firenze, Florence, Italy), can be found in [50]. A brief description and summary of major capabilities are provided in this section.
ULA-OP consists of a metal rack of dimensions 33x23x18 cm, connected to a PC where a dedicated software runs. The backplane in the rack has the probe connector and routes the signals among the power unit and two main boards: an analog board (AB) and a digital board (DB). The AB includes the RF front-end while the DB hosts the devices in charge of numerical signal processing. More detailed features of ULA-OP are described in Appendix A.

4.4 Linear probe

A standard 128-element linear ultrasonic array (ESAOTE PA230) was used for experimental verification of theory and simulation results of the 2D adaptive beamforming method in transmission mode (Section 3.2). It was also used as the imaging probe when the complete 2D adaptive beamforming method (transmission and reception modes) was implemented on ULA-OP to generate corrected 2D sonograms through skull phantoms. Figure 4.7 shows the probe, while its technical characteristics are listed in Table 4.4.
Figure 4.7: A picture of the linear probe used in 2D adaptive beamforming lab experiments

Table 4.4: Technical properties of the linear probe

<table>
<thead>
<tr>
<th><strong>Esaote – PA230 technical properties</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements</td>
</tr>
<tr>
<td>Frequency</td>
</tr>
<tr>
<td>Single element dimensions</td>
</tr>
<tr>
<td>Pitch</td>
</tr>
<tr>
<td>Kerf</td>
</tr>
<tr>
<td>Total active area</td>
</tr>
<tr>
<td>Piezo material</td>
</tr>
<tr>
<td>Off-plane mechanical focusing</td>
</tr>
<tr>
<td>mechanical lens thickness</td>
</tr>
<tr>
<td>Acoustical Impedance matching</td>
</tr>
<tr>
<td>Connector</td>
</tr>
</tbody>
</table>

Frequency response of a representative element of the probe under pulse excitation is shown in Figure 4.8.
4.5 Planar (Matrix) probe

A square matrix ultrasonic transducer array with 16x16 elements was designed by the author and manufactured by Imasonic, France. The probe was initially used for experimental verification of the author’s 3D adaptive beamforming method in transmission mode (section 3.4). It was also used as the imaging probe when the complete 3D adaptive beamforming method (transmission and reception modes) was implemented on ULA-OP to generate corrected 3D sonograms through skull phantoms.

In order to accurately design a customized planar probe suitable for beamforming purposes of this study, a software simulation was developed by the author. The simulation is capable of estimating the field profile produced by various planar array geometries. It accepts the geometrical parameters related to design of a matrix array as inputs (e.g. number of elements on the array, elements' shape, size and orientation, array elements' pitch). It then uses such inputs...
along with a desired focal point, to calculate delay times for each array element and generate the beam profile of the probe. Studying these profiles for different array geometries and comparing key factors, provided input for the final design of the matrix phased array. The key factors studied in each profile were the level of side-lobes, possible grating lobes, and focal spot size. The probe was designed to be capable of generating 3D sonograms with up to 60° angle of view (-30° to 30° along each surface axes with respect to the normal of the probe). Sketches of the original design of the probe, prepared by the author in SketchUp Pro software (version 8.0.15158), are shown in Figure 4.9. Table 4.5 also lists the technical properties proposed by the author associated with this original design. The author and his team then started to negotiate with a number of manufacturers and finally Imasonic Corp., France, was chosen to place the order. During the fabrication process, some compromises were required on the specifications proposed in the original design due to fabrication limitations of the manufacturer. For instance, for a probe with central frequency response of 2MHz, a minimum element size of 4.5mm was the limitation of the manufacturer at the time of placing the order. As another example, a 6dB bandwidth of 75% or higher specified on the original design which was adjusted to 40% by the manufacturer.
Table 4.5: Desired technical properties of the matrix probe proposed on the original design

<table>
<thead>
<tr>
<th>Proposed technical characteristics</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements (Nₓ x Nᵧ)</td>
<td>16x16</td>
</tr>
<tr>
<td>Frequency</td>
<td>2 MHz</td>
</tr>
<tr>
<td>Bandwidth</td>
<td>&gt;75% @ -6dB</td>
</tr>
<tr>
<td>Element size</td>
<td>0.4mm x 0.4mm</td>
</tr>
<tr>
<td>Pitch</td>
<td>0.45 mm</td>
</tr>
<tr>
<td>Kerf</td>
<td>0.05 mm</td>
</tr>
<tr>
<td>Total active area</td>
<td>7.15mm x 7.15mm</td>
</tr>
<tr>
<td>Piezo material</td>
<td>1-3 piezocomposite</td>
</tr>
<tr>
<td>Mechanical focusing</td>
<td>None</td>
</tr>
<tr>
<td>Acoustical Impedance matching</td>
<td>Water (1.5 MRayl)</td>
</tr>
<tr>
<td>Connector</td>
<td>ITT CANNON DLM5-260P</td>
</tr>
</tbody>
</table>
A picture of the delivered probe and its technical specifications are shown in Figure 4.10.

Figure 4.10: Pictures of the delivered matrix probe and its technical specifications

Representative frequency response of an arbitrary element of the probe under pulse excitation is shown in Figure 4.11.
After receiving the probe from the manufacturer, the author programmed ULA-OP to work in a single-element-inspection mode for preliminary verification of the probe’s quality. This mode allows to transmit and receive with any selected single element and to examine its sensitivity and bandwidth individually. It is also useful for excitation of a single element and collecting its beam profile to inspect for any possible leakage or cross-talk between the elements. The author then used this mode to examine forty arbitrarily chosen array elements. No sign of leakage or cross-
talk was observed. The observed sensitivities and bandwidths of the examined elements were also consistent with the data provided by the manufacturer.

4.6 Ultrasound hydrophone

An ultrasonic hydrophone (ONDA, HGL-0400) was used to collect intensity profiles at any desired cross-section or volume in front of the linear and planar probes. The submersible hydrophone has a small effective aperture (400 μm) and desirable sensitivity (~160 nV/Pa) at the 2MHz frequency of this study. It was supplied with a cordless pre-amplifier (ONDA, AH-2010) which helps to lower the noise level in received signals. The hydrophone and its pre-amplifier are shown in Figure 4.12.a. The hydrophone’s directivity plot, supplied by the manufacturer, is shown in Figure 4.12.b (orange curve).
Figure 4.12: (a) Picture of the hydrophone and its pre-amplifier. (b) The hydrophone’s directivity plot (orange curve).

Technical specifications of the hydrophone and its pre-amplifier are listed in Table 4.6.

Table 4.6: Technical specifications of the hydrophone and its pre-amplifier

<table>
<thead>
<tr>
<th>HGL-0400</th>
<th>AH-2010</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frequency range (±3dB)</td>
<td>0.25 - 20 MHz</td>
</tr>
<tr>
<td>Electrode aperture (μm)</td>
<td>400</td>
</tr>
<tr>
<td>Nominal Sensitivity [nV/Pa]</td>
<td>160</td>
</tr>
<tr>
<td>Acceptance angle (-3dB at 2 MHz)</td>
<td>70°</td>
</tr>
<tr>
<td>Capacitance</td>
<td>30 pF</td>
</tr>
<tr>
<td>Max. Operating Temperature</td>
<td>50 °C</td>
</tr>
</tbody>
</table>
4.7 **Four-dimensional scanner and other peripherals**

A 4-axis ultrasonic immersion scanning system with a positioning and scanning accuracy of 1µm, was used in the experiments. The system included:

- Mechanical scanner (IAI, Japan) with the IAI XSEL-4 motion controller;
- Acrylic water tank with dimensions of 70 cm x 40 cm x 45 cm;
- Low-noise preamplifier (5660C, Olympus NDT, USA),
- A 14-bit, 65 Ms/s PCI data acquisition board (Instrumental Systems, Russia);

The motion control and data acquisition were realized via a set of custom LabView (National Instruments, USA) and C/C++ programs running on a desktop computer.

The immersion scanning system was used throughout the project for a variety of proof-of-concept testing tasks, for studying the propagation of acoustic waves through skull phantoms, and to test two- and three-dimensional adaptive beamforming algorithms. It was also used for ultrasonic field mapping and calibration purposes via the HGL-0400 hydrophone. A photograph of the immersion scanning system is shown in Figure 4.13.
Figure 4.13: Photograph of the laboratory tabletop scanning system
Chapter 5

Single point adaptive focusing –
Experimental verification

5.1 Introduction

To check the validity of the developed theory and numerical simulation, the author designed and conducted a series of laboratory experiments employing the custom-designed skull phantoms. The first set of experiments was designed to verify the simulation results of adaptive beamforming in transmission mode. The theoretical basis and simulation results for a single-point adaptive focusing in transmission mode were discussed in chapter 3. This chapter presents experimental verification of those results and practical realization of the method.

The chapter starts with the description of a custom-designed time-of-flight measurement technique specifically developed to obtain information about the geometry of the skull phantom prior to the beamforming process. This experimental mode was designed and programmed on ULA-OP to accurately measure the skull profile in front of the array. The extracted skull profile is then used as an input to the numerical simulation preceding the beamforming process. The chapter then continues with a description of the experimental setup. Results of the laboratory experiments in cases of both linear and matrix probes are presented. Quantitative assessment and error analysis of the experimental results are presented and discussed.
5.2 **Skull profile reading module**

To obtain information about the geometry of the skull phantom prior to the beamforming process, a customized time-of-flight measurement technique was developed via collaboration of the author with ULA-OP’s product support department. The algorithm was programmed in MATLAB and then was linked to ULA-OP. Knowledge of sound propagation speed in the skull phantom is a necessity for profile extraction. The sound speed value can be input to the simulation either from previously measured and tabulated values or using an in-line measurement technique, developed and reported in [24] by our team. In this technique, thickness variation and sound speed can be simultaneously measured just before the beamforming process. The reader is encouraged to see the reference for details.

Two separate control modules were programmed on ULA-OP for skull profile extraction via linear and matrix probes.

5.2.1 **Linear array control module**

In this module, a custom-designed B-scan of the skull phantom profile is recorded using all 128 elements of the array. To get a better SNR on receiving RF signals of the B-scan, the author arranged five elements at a time to focus at a 1cm distance from the probe's surface. Starting from one end of the array, this small aperture was moved in single element steps to the other end. At each step, multiple acquisitions of the RF signals were recorded, averaged, and subjected to standard five-element beamforming and summation processes. Figure 5.1.a shows a typical result of such a scan. IQ signals (In-phase/Quadrature demodulated signals) of such scans are then transferred to the MATLAB code, and a custom-designed cross-correlation algorithm is applied to accurately define the two boundaries of the skull phantom. Yellow lines in Figure 5.1.b show...
the result of the above process. (The cross-correlation algorithm is explained in Appendix B). The skull phantom profile, extracted from the B-scan, is then fed into the simulation engine to calculate a new corrected delay pattern adapted to the acoustical and geometrical properties of the skull phantom.

Figure 5.1: Example of original and extracted profiles of a section of skull phantom using the linear probe

5.2.2 Matrix array control module

In this experimental module, a two-dimensional scan of the skull phantom is recorded using all 192 available elements of the matrix array (section 4.5). (That is, the procedure of Section 5.2.1 is now applied to the matrix array). To get a better SNR on receiving RF signals of the scan, nine elements at a time were arranged to focus at a 1cm distance from the probe's surface. Starting from one corner of the array, Figure 5.2, this small aperture is moved in single element steps to the opposite corner of the matrix probe in a raster scanning manner. The red line in Figure 5.2 shows the scanning path of the centre element of the nine-element aperture.
Figure 5.2: Arrangement of the elements and their scanning path used for the profile extraction via the matrix probe

At each step, multiple acquisitions of the RF signals were recorded, averaged and fed through standard nine-element beamforming and summation processes. Figure 5.3.a shows a typical result for such a scan. It must be noted that although the current version of ULA-OP is capable of acquiring volumetric data, its software does not have volumetric screening capabilities. Figure 5.3.a can be interpreted as horizontal scan lines of Figure 5.2 placed adjacent to each other. IQ data of such scans (14x14 complex signals) is then transferred to the custom-designed cross-correlation (Appendix B) MATLAB code to accurately define the two boundaries of the skull phantom. Figure 5.3.b shows a typical result of such process, plotted as a grid of 14x14 inner boundary points with 0.55 mm grid size.
Figure 5.3: A snapshot of the skull profile extraction software which works at a frame rate of 2 frames/sec; (a) raw data from ULA-OP and (b) processed skull profile

The current version of this in-line profile extraction software works at a frame rate of 2 frames/sec. While moving the probe over the skull phantom’s surface, the operator can on demand save the profile at a desired location. The software then generates the final skull phantom profile at the location of the probe. An example of such a final profile comprised of the inner and outer boundaries is shown in Figure 5.4. As opposed to the arbitrary axes units of the in-line software, Figure 5.3.b, the saved version of the profile includes its true dimensions.
Figure 5.4: Example of a final extracted skull phantom profile via the matrix probe

To examine the accuracy of the explained profile extraction method, a 2D calibration skull phantom was fabricated as shown in Figure 5.5. The thickness of the calibration phantom was first measured with a digital caliper (Mastercraft, Canada; resolution: 0.01mm, error: ±0.02mm) at 16 points. The above time of flight measurement method was then applied at those 16 points. A maximum error of 70 µm with a standard deviation of 31 µm was observed. The overall order of this error would lead to a final focusing error of less than 0.2mm, when focusing up to 5cm from the surface of the probe. This is an acceptable error for the adaptive focusing purposes of this study.
It was previously explained in section 4.5 that the matrix probe is capable of steering a beam at up to ~30° from the normal to its surface. When steering and focusing at wide angles, it is necessary to know the skull profile to a certain extent beyond the area immediately under the probe. The above profile extraction method works only in the area immediately underneath the array elements; this area might not be wide enough to include all refracted paths between each array element and the desired focal coordinates. Therefore an extrapolation code was developed by the author to estimate the skull profile beyond the originally calculated boundaries.

In the surface extrapolation algorithm, first a two-dimensional polynomial, in the form of

$$P_n(x, y) = \sum_{k=0}^{n} a_k x^i y^j \quad i + j \leq k,$$

is fitted to the measured distribution of inner boundary points. The order of the polynomial is a user-defined value. As a representative example, Figure 5.6 shows the results of fourth-order polynomial fit to the original distribution of Figure 5.4. In this figure, the difference between the measured and fitted points is shown in the distribution plot on the right side. The standard deviation divided by the mean thickness value of the fit is also provided as an indicator of the
accuracy of the polynomial approximation. Similar values when polynomials of various orders are fit to the original points are listed in Table 5.1.

![Figure 5.6: (a) Example of Fourth-order polynomial fit to an original extracted skull profile. (b) Deviation distribution between original and fitted values.](image)

Table 5.1: Calculated standard deviation for the deviation distribution between the original and fitted points when different order polynomials are fitted to the original profile

<table>
<thead>
<tr>
<th>Polynomial order</th>
<th>$\frac{STD}{\bar{Z}}$</th>
<th>Polynomial order</th>
<th>$\frac{STD}{\bar{Z}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>3.5469e-004</td>
<td>6</td>
<td>1.6398e-004</td>
</tr>
<tr>
<td>2</td>
<td>3.2228e-004</td>
<td>7</td>
<td>1.4953e-004</td>
</tr>
<tr>
<td>3</td>
<td>2.0162e-004</td>
<td>8</td>
<td>1.3434e-004</td>
</tr>
<tr>
<td>4</td>
<td>1.8223e-004</td>
<td>9</td>
<td>1.2631e-004</td>
</tr>
<tr>
<td>5</td>
<td>1.7518e-004</td>
<td>10</td>
<td>1.1335e-004</td>
</tr>
</tbody>
</table>

Use of higher order polynomials results in a closer match to the original distribution (lower STD). When focusing at small angles from the normal to the probe’s surface (e.g. less than 10
degrees), the extracted profile is normally wide enough. The polynomial fit in these situations is only used to smoothen the extracted profile and therefore higher orders can be employed. But when extrapolation is desired, a compromise between the amount of surface extension and the polynomial order must be made. A second-order polynomial fit and 5 rings of extrapolated points were set as default settings in the developed code by the author. The skull phantom profile, extracted with the above method, is then fed into the simulation code to calculate a time delay pattern for the array elements, which is adapted to the acoustical and geometrical properties of the skull phantom.

It is noted that the skull phantom profile extrapolation was explained for the case of the matrix array, which is a more comprehensive case than that of the linear array. The same approach is applicable to the linear array with a straightforward dimension reduction.

5.2.3 Discussion

It was explained in chapter 3 that accurate knowledge of the skull profile where it contacts the probe is a necessity for the adaptive beamforming method presented in this thesis. The profile extraction algorithm as described in the previous section will be adversely affected if the ultrasound reflection from the inner boundary of the skull (or skull phantom) is not properly identified by the software. This could occur if the diplo’e layer is too thick, or the reflected pulse from the inner boundary is buried in multiple reflections and reverberations within the diplo’e. These skull areas were indicated in red in Figure 3.1 and are labeled as “unavailable”. An example in which the profiling software was unable to reliably detect the reflection from the skull inner boundary is shown in Figure 5.7.
An immediate and easy solution when the operator confronts such a profile is to move the probe to a new location on the skull or skull phantom, which might be as close as a few millimetres away.

Moreover, the author understands that surface extrapolation cannot be a reliable final solution when an extended profile is desired. The shape of the skull outside the measured area is independent from the shape within that area. The developed surface extrapolation technique is considered as a temporary solution for the available matrix probe which was designed and used for proof of concept in this study. The reliable solution on the other hand, is to use an array which has enough elements to be able to measure the skull over a wider area and then use a sub-aperture in the transmission mode beamforming process. This would be similar to the previous linear array’s case in which the profile was extracted via a 128-elements-wide aperture and yet the transmission mode beamforming is accomplished via the middle 64 elements.
5.3  Transmission mode adaptive beamforming – linear probe

An experimental setup was designed and prepared by the author to verify the simulation results in transmission mode for linear arrays. Figure 5.8 shows a schematic configuration of this setup. In all experiments, water was used to approximate brain tissue, as the two materials have close values of attenuation, density, and compression wave speed.

The adaptive beamforming algorithm in transmission mode was implemented on ULA-OP (section 4.3). ULA-OP simultaneously controls 64 active elements selected out of the 128-element linear phased array (PA230-Esaote, section 4.4) via its 64 independent channels. The middle 64 elements on the array, i.e. elements 33 to 96, were chosen as the active aperture on the probe for this set of experiments. The probe was submerged in water and held in contact with the skull phantom prepared with a linear symmetry for this purpose (Fig.4.3.a). The HGL-0400 ultrasonic hydrophone, section 4.6, was attached to our XSel-IAI 4D scanner, section 4.7, to measure the acoustic field intensity in the plane of the linear probe (x-z plane in Fig.3.1.b) with a 0.5mm step size.
Prior to the beamforming process, the skull phantom profile is extracted using the method described in Section 5.2. The extracted profile is then fed into the simulation code to calculate an adapted delay pattern. Excitation signals with amplitude of 100V_{pp} were applied to the array elements via ULA-OP according to the calculated delay time profile. After applying this pattern to the active array elements via ULA-OP, the hydrophone starts scanning to measure the resulting beam profile. At each scanning point the hydrophone signal was amplified, digitized, averaged, and then processed to calculate the acoustic pressure.

Figure 5.9 shows a sample representative set of beam patterns measured using the above apparatus. Each image in Figure 5.9 is a result of scanning the hydrophone over an area of 5\times5 cm^2 in front of the array and around the intended focus (white cross). In this case, the intended focal coordinates were at R_F = 5 cm and \phi_s = 0^\circ, i.e. on the normal to the probe and 5cm away from its center. (Results are similar for non-zero values of \phi_s).
Figure 5.9: Experimental results of (a) original (b) distorted and (c) corrected beam profiles in absence and presence of the skull phantom. White cross shows the intended focus.

Figure 5.9.a shows the measured beam profile in water without the phantom. Figure 5.9.b shows the measured beam pattern through the skull phantom when the conventional delay scheme (without correction) is applied. Similar to the simulation results, it can be seen how dramatically a beam can be deflected from its intended focal point. Figure 5.9.c shows the measured beam profile through the skull phantom after applying the corrected delay pattern via ULA-OP. It can be seen that the developed adaptive beamforming algorithm in the linear transmission mode is capable of bringing the misdirected beam of Figure 5.9.b back to the desired orientation, so that the beam passes through the intended focus (Figure 5.9.c). The experimental result is consistent with the simulation results of Figure 3.7.

For a better assessment, the pressure plots of the three profiles at the intended focus and along the x-axis are shown in Figure 5.10 on a log scale. On this representative example, a displacement of 7.04mm was observed between the intended focus and the peak of the distorted
profile, as illustrated in Figure 5.10.b. The peak of the corrected profile (Figure 5.10.c) is only 0.013mm away from the intended focus.

![Graph showing pressure plots of original, distorted, and corrected beam profiles.](image)

**Figure 5.10:** Pressure plots of original, distorted and corrected beam profiles of Figure 5.9 along x-axis at the intended focus.

### 5.3.1 Discussion

Compared to the beam of Figure 5.9.a, lower intensity levels are observable in both Figure 5.9.b and Figure 5.9.c; this loss of signal strength is due to the presence of the skull phantom which leads to attenuation and interfacial transmission losses. Based on the HGL-0400 hydrophone’s calibrated chart provided by the manufacturer, a pressure amplitude of 3.577MPa was measured at the intended focus in Figure 5.9.a. On the other hand, a pressure amplitude of only 0.233MPa was measured for the beam profile of Figure 5.9.b at the same intended focus.
The effect of spatial apodization in the farfield response of a phased array was thoroughly discussed in section 3.2. The spatial apodization of the array elements (Gaussian in the presented case) resulted in the almost complete elimination of side lobes and hence generation of clean main lobes in both simulation (Figure 3.7) and experiment (Figure 5.9). However, as explained in section 3.2, some increase in the angular width of the main lobes (and therefore a loss in resolution) accompanies this reduction.

5.3.2 Quantitative assessment and error analysis

The experimental results shown in Figure 5.9 are consistent with the simulation results of the new adaptive beamforming method in transmission mode for linear arrays. As a more stringent test, the author designed and conducted another set of experiments to quantitatively evaluate the developed adaptive focusing method. The evaluation was based on measuring the deviation between calculated time delay patterns via developed simulation and their experimentally measured counterparts in temporal focusing scenarios. A schematic representation of the setup used in this set of experiments is shown in Figure 5.11.
For such assessment, a “single-element-excitation” mode was designed and programmed on ULA-OP. In this operating mode, any single element of the linear array can be individually excited. The linear probe array was submerged in the water tank and the HGL-0400 hydrophone was attached to the 4D scanner for accurate positioning of the hydrophone at any desired point in front of the probe. The experiment was controlled by ULA-OP, which triggered the excitation pulse and the oscilloscope (Tektronix TBS-1000, United States) attached to the hydrophone. A typical received signal on the oscilloscope, collected by the hydrophone when one element of the linear array is excited, is shown in Figure 5.12.

Prior to the experiments, the sound speed in the water tank was measured as 1489 m/s at our laboratory conditions. It was done through a simple time-of-flight measurement between two positions of the hydrophone.
Figure 5.12: Received signal on the oscilloscope when a sample single element on the array is excited.

The first test was conducted to measure the setup’s baseline error in a simple non-adaptive focusing scenario without the skull phantom for future references. For this experiment, the hydrophone was placed in the array’s imaging plane (x-z plane in Figure 3.3.b) in front of the centre of the 64-element aperture and 5cm away from the probe’s surface. This corresponds to focal coordinates of \((R_{F}, \varphi_{s}) = (5\text{cm}, 0^\circ)\) in Figure 3.3.b. For accurate placement of the tip of the hydrophone at these coordinates, its position was calibrated both in depth and laterally. For depth calibration, the hydrophone was moved along the probes centre-line until time-of-flight corresponding to 5 cm was observed on the oscilloscope. For lateral calibration, first and last elements on the probe were excited separately. The probe was moved along the x-axis (Figure 3.3.b) until equal time-of-flights were measured for the two received signals.

After the hydrophone was accurately positioned at the intended focal point, the 64 elements of the array were excited, one by one. The signal received by the hydrophone corresponding to each of these 64 events was digitized and saved. The signals were then transferred to another MATLAB script, developed by the author, for display as shown in Figure 5.13. Only 32 signals out of the 64 (i.e., every second element) are shown for better clarity. To aid in visual understanding of the various time-of-flights for each of the 32 signals, the peaks of the signals
are indicated by red circles. However, for accurate analysis of the data displayed in Figure 5.13, the MATLAB script determined the relative time delays among the signals by calculating the peak of the envelope function of each signal. The envelope function was calculated via the Hilbert transform method.

![Figure 5.13: Representative 32 captured signals from the 64 elements of the linear probe in absence of the skull phantom. The hydrophone is positioned at the intended focal coordinates of \((R_f, \phi_s) = (5\text{cm}, 0^\circ)\)](image)

The physically measured time delay pattern was processed to yield the time delay of each element relative to the central element. To generate a corresponding delay pattern with the simulation code, the appropriate focusing geometry was input to the code, together with the measured speed of 1489 m/s for water; the resulting time delay pattern was then calculated. The resulting measured (black line) and simulated (red line) time delay patterns, both expressed
relative to the central element, are compared in Figure 5.14.a. The deviation between the two curves is shown in Figure 5.14.b.

Figure 5.14: (a) Comparison of measured and calculated delay patterns in absence of the skull phantom. (b) Plot of deviation between the two patterns
A standard deviation of $4 \times 10^{-9}$ sec was observed for this temporal deviation distribution. This standard deviation is about 0.8% of pulse period at its central frequency ($T_{2MHz}$). This small baseline deviation can be attributed to various sources of errors during the measurements. Hydrophone positioning error is recognized as the primary source of error. For example a positioning error of $\sim 8 \mu m$ could cause error of the same order as the mentioned standard deviation ($\sim 1\%$ of the signal period at 2MHz). Other sources include water sound speed measurement error, and the measurement errors associated with the sampling and envelope peak measurements of the received signals. Such baseline error must be recognized and considered in the next step, where temporal deviation is measured in the presence of the skull phantom. In this step, the skull phantom of Figure 5.11 is added to the setup and is held in contact with the linear probe. The above procedure is then repeated to measure the time-of-flight between each array element and the tip of the hydrophone. The relative time delay pattern of the received signals in this setup represents the time delay pattern adapted to the geometry and acoustical properties of the skull phantom segment in contact with the probe. The result of this measurement is shown in Figure 5.15. Again, for visual clarity of the delay pattern, the peaks of the signals are labeled with red circles. The final time delay pattern was obtained by calculating the peak of the envelope function of each signal.
To calculate the counterpart of this measured delay pattern through the simulation algorithm, first the profile of the skull phantom segment in contact with the probe was determined as described in Section 5.2.1. The extracted profile, along with other initial parameters for the same focusing scenario, was then input to the simulation code and the adapted delay pattern was calculated. The measured and calculated patterns are compared in Figure 5.16.a and the deviation plot is shown in Figure 5.16.b.
A standard deviation of $8 \times 10^{-9}$ sec (1.6% of the pulse period at 2MHz) was obtained in this particular example. An overall larger deviation values were observed compared to the previous case where the skull phantom was not introduced. This is mainly due to the errors associated with the skull phantom profile measurement and also to the random phase shifts induced by the
diplo’e in the skull phantom. The standard deviation is still a small fraction of the pulse period and the simulated delay pattern remained in agreement with the experimental pattern.

The above temporal deviation measurement method was then applied to a variety of skull profiles and focusing scenarios. All skull profiles were extracted from the skull phantom with linear symmetry (Figure 5.11) via the linear probe. The results for two representative skull phantom profiles and six focal points through each profile are reported here. These profiles are shown in Figure 5.17.

![Image of two extracted skull profiles used in the temporal deviation measurement experiments via the linear probe](image)

**Figure 5.17:** The two extracted skull profiles used in the temporal deviation measurement experiments via the linear probe

The set of six focal coordinates used for analysis through the skull phantoms shown in Figure 5.17 are listed and graphically shown in Figure 5.18. These points were chosen within the accessible inspection area of the linear probe, shown schematically with dotted lines.
Figure 5.18: The focal coordinates used in the temporal deviation measurement experiments via the linear probe.

Final aberration plots for the two skull phantom profiles of Figure 5.17.a and Figure 5.17.b are presented in Figure 5.19 and Figure 5.20 respectively.

<table>
<thead>
<tr>
<th>Focal point No.</th>
<th>$(R, \varphi)$ [m, deg]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>(0.04, 0)</td>
</tr>
<tr>
<td>2</td>
<td>(0.06, 0)</td>
</tr>
<tr>
<td>3</td>
<td>(0.08, 0)</td>
</tr>
<tr>
<td>4</td>
<td>(0.04, 20)</td>
</tr>
<tr>
<td>5</td>
<td>(0.06, 20)</td>
</tr>
<tr>
<td>6</td>
<td>(0.08, 20)</td>
</tr>
</tbody>
</table>
Figure 5.19: Temporal deviation plots when focusing on representative focal points of Figure 5.18 through skull profile "I" in Figure 5.17. The signal period at its central frequency (2MHz) is $0.5\mu\text{sec}$. 
Figure 5.20: Temporal deviation plots when focusing on representative focal points of Figure 5.18 through skull profile “II” in Figure 5.17. The signal period at its central frequency (2MHz) is $0.5\mu sec$.

### 5.3.3 Discussion

Judging by the observed range of temporal deviation values, the results of Figure 5.19 and Figure 5.20 are all consistent with the sample result of Figure 5.16. Maximum deviation of $2.34 \times 10^{-8} \text{sec}$ and $22.34 \times 10^{-8} \text{sec}$ was observed on measurement through profile “I” and “II”
respectively. Maximum standard deviation of $8 \times 10^{-9}$ sec and $8 \times 10^{-9}$ sec was observed for the temporal deviation distributions through profile “I” and “II” respectively. Both standard deviations are about 1.6% of the pulse period at 2MHz.

In some temporal deviation plots, a positive or negative constant offset was observed, e.g. focal point No.6 in Figure 5.19 or focal point No.2 in Figure 5.20. In the author’s opinion, this relates to the depth positioning error of the hydrophone via the 4D scanner. Such a shift can be better understood when a simple focusing scenario, such as the case for Figure 5.14, is considered. If the hydrophone is placed closer to the probe’s surface than the intended focus, the distribution will have a shift below the zero line in the corresponding temporal deviation plot. A depth positioning inaccuracy of $15 \mu m$ would result in a shift of up to 2% in the temporal deviation plots.

The quantitative assessment of the developed adaptive focusing method showed consistency between the simulation and experimental results. The overall order of measured and reported deviations, as a small percentage of the signal period, would lead to an intensity error of less than 0.06% in the final image. This small error is acceptable for the final beamforming purposes of this study.

### 5.4 Transmission mode adaptive beamforming – Matrix array

After the simulation results of the developed adaptive beamforming method in transmission mode were experimentally verified for linear arrays, the author conducted a series of laboratory experiments for the case of matrix arrays. The experimental setup of Figure 5.8 was almost unchanged, except the linear probe was replaced with a matrix array and the full-calvaria skull
phantom was used instead of the previous skull phantom with linear symmetry. The full-calvaria phantom is geometrically realistic and has thickness variation along both surface axes similar to real human skull. A picture of this setup is shown in Figure 5.21.

![Figure 5.21: A picture of the matrix probe and the full-calvaria phantom used in the experimental setup](image)

Since the current version of ULA-OP is a 64-channel system, only a combination of 64 elements (out of the 192 available elements, section 4.5) can be used as the active transmitting aperture. An arrangement of the middle 64 elements was chosen by the author and is shown in Figure 5.22.
Figure 5.22: Arrangement of the 64 middle elements on the matrix probe used as the transmitting aperture

Prior to the beamforming process, the skull profile was extracted using the method described in section 5.2. The extracted profile was then used as input to the simulation code to calculate a new timing pattern for the transmitting array elements for any specified focal point. The adapted timing pattern was then applied to the matrix probe via the ULA-OP. The hydrophone was used to collect intensity profiles of the ultrasonic field at any desired cross-section in front of the array.

Figure 5.23, as a representative example, shows three cross-sectional beam profiles experimentally measured in the water tank. Each image is a result of scanning the hydrophone with 0.5mm increments. The intended focus (white cross) has the coordinates \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\). Each image represents a 5cmx5cm area parallel to the plane of the 2D array and 5cm away from it.
Figure 5.23: Experimentally measured cross-section beam profiles: (a) original, with no skull phantom (b) distorted, with no correction, in presence of the skull phantom and (c) corrected, in presence of the skull phantom. White cross shows the intended focus.

Figure 5.23.a shows the experimentally measured beam profile in water without the phantom. Figure 5.23.b shows the experimentally measured beam pattern through the skull phantom when the conventional delay scheme (without correction) is applied. It can be seen that the beam is strongly deflected from its intended focal point, due to phase aberration induced by the portion of the phantom in contact with the probe. Figure 5.23.c shows the experimentally measured beam profile after applying the corrected delay pattern through ULA-OP. It can be seen that the misdirected beam of Figure 5.23.b in now brought back to the desired orientation and passes through the intended focus. In Figure 5.23.b and Figure 5.23.c, displacements of 8.74mm and 0.11mm were observed between the intended focal coordinates and the maxima of the focal spots respectively.

5.4.1 Discussion

Compared to the beam of Figure 5.23.a, lower pressure levels are observed in both Figure 5.23.b and Figure 5.23.c. A pressure amplitude of 1.213MPa was measured at the intended focus in Figure 5.23.a via the calibrated hydrophone. A pressure amplitude of 0.093MPa was measured at the intended focus in Figure 5.23.c. The expected loss of signal strength in Figure 5.23.b and
Figure 5.23.c, compared to Figure 5.23.a, is due to the presence of the skull phantom which leads to attenuation and interfacial transmission losses.

Compared to the previous case of the linear probe, overall pressure amplitudes were lower. This arises from the smaller active area of the transmitting aperture in the matrix probe and therefore lower final transmitting power. Elements of the linear array are long strips of piezo-material as opposed to small squares in the matrix probe (see section 4.4 and section 4.5 for the elements’ dimensions).

Different groups of 64 elements (out of the 192 available elements on the matrix probe) were considered for the active aperture in Transmission mode. Sparse, square, and rectangular apertures were tested. The use of a sparse aperture resulted in a narrower focal spot, but it also caused an increase of the sidelobe levels, which would result in lower contrast in a final image. This was an expected observation that is consistent with phased array theory, based on the increased ratio of center-to-center element pitch to wavelength (~0.75mm in water at 2MHz). Moreover, use of square or rectangular apertures resulted in nonsymmetrical focal spot which also was not desired. Therefore, the arrangement of Figure 5.22 was the final selected element pattern for the active aperture in transmission mode adaptive beamforming for the matrix array. This aperture showed an overall optimized performance in various focusing scenarios for the equipment in hand.
5.4.2 Quantitative assessment and error analysis

The beam profiles of Figure 5.23 successfully verified the simulation results of the developed adaptive beamforming method in transmission mode for matrix arrays (section 3.4). Yet, for more accurate evaluation of the calculated time delay patterns, similar to the linear array’s case, a quantitative assessment was conducted. The apparatus shown in Figure 5.11 was again used, this time with the matrix probe and the skull phantom of Figure 5.21.

First, aberration in the absence of the skull phantom was measured to estimate the baseline error imposed by the experimental setup. For this experiment, the hydrophone was placed in front of the centre of the probe at focal coordinates of \((R, \varphi, \psi) = (5\text{cm}, 0^\circ, 0^\circ)\). For accurate placement of the tip of the hydrophone at these coordinates, its position was calibrated both in depth and laterally. For depth calibration, the hydrophone was moved along the probe’s centre-line until time-of-flight corresponding to 5cm was observed on the oscilloscope. For lateral calibration, four corner elements of the probe were excited separately. The probe was then positioned at the point where the four received signals arrived at the same time. After calibration, the 64 elements of the aperture shown in Figure 5.22 were individually excited and the corresponding signals received at the hydrophone were saved in the oscilloscope memory. Relative time delays between the received signals were then calculated by finding the peak location of the envelope function of each signal. The results are shown in Figure 5.24.a (blue circles). A simulated delay pattern for the same focusing scenario was also calculated and shown on the same plot, Figure 5.24.b (Red squares).
Figure 5.24: Comparison of measured and calculated delay patterns in absence of the skull phantom

The temporal deviation between the simulated and measured data points is plotted in Figure 5.25. Please note that, for better clarity and comparison with the linear array results, the temporal deviation plots presented in this section are also shown in only two dimensions. The horizontal
axis still refers to the element numbers but based on the numbering scheme of Figure 5.22. This baseline deviation (i.e., for the case of no skull phantom) must be recognized and considered shortly in cases for which the skull phantom is present.

Figure 5.25: Plot of deviation between the two patterns of Figure 5.24

In the next step, the skull phantom of Figure 5.21 was added to the setup and again the deviation between the simulated and experimental time delay patterns was measured. The temporal deviation plots for six focal points are reported in this thesis for both of two different skull phantom profiles. The two skull phantom profiles are shown in Figure 5.26 while Figure 5.27 shows the arrangement of the six examined focal coordinates.
Figure 5.26: The two extracted skull phantom profiles used in the temporal deviation measurement experiments for the matrix probe
Figure 5.27: The focal coordinates used in the temporal deviation measurement experiments for the matrix probe.

Temporal deviation plots for the two skull phantom profiles of Figure 5.26.a and Figure 5.26.b are presented in Figure 5.28 and Figure 5.29 respectively.
Figure 5.28: Temporal deviation plots when focusing on representative focal points of Figure 5.27 through skull profile “1” in Figure 5.26.
Figure 5.29: Temporal deviation plots when focusing on representative focal points of Figure 5.27 through skull profile “II” in Figure 5.26.

5.4.3 Discussion

Similar to the linear array’s case, when a skull phantom is introduced to the setup, an overall higher temporal deviation was observed compared to the baseline temporal deviation of
Figure 5.25. Maximum temporal deviation values of $2.09 \times 10^{-8}$ sec and $2.51 \times 10^{-8}$ sec were observed in the measurements through profiles “I” and “II” respectively. Maximum standard deviation of $7.5 \times 10^{-9}$ sec (1.5% of the pulse period at 2MHz) and $8 \times 10^{-9}$ sec (1.6% of the pulse period at 2MHz) was observed for the temporal deviation distributions through profile “I” and “II” respectively. Besides that, an overall positive or negative constant offset was observed in some temporal deviation plots in the matrix probe case as well, e.g. results for focal point No.4 in Figure 5.28 or focal point No.3 in Figure 5.29. This again relates to the positioning and alignment errors of the hydrophone via the 4D scanner. A positioning inaccuracy of $15 \mu m$ would result in a shift of up to 2% in the temporal deviation plots.

The quantitative assessment of the developed adaptive focusing method showed overall consistency between the simulation and experimental results. The overall order of measured and reported deviations, as a small percentage of the signal period, would lead to an intensity error of less than 0.06% in the final image. This small error is acceptable for the final beamforming purposes of this study.

5.5 Chapter conclusion

Experimental verification of the developed adaptive beamforming method in transmission mode was the focus of this chapter. Description of a customized time-of-flight measurement technique specifically developed to obtain information about the geometry of the skull phantom prior to the beamforming process was presented. This mode was designed and programmed on ULA-OP to accurately measure the skull profile in front of the array. Results of skull phantom profile
readings in cases of both linear and matrix probes were provided along with discussion over the experimental results.

The simulation results of the developed adaptive beamforming method (chapter 3) were initially examined for the case of the linear array. Representative beam profiles were presented, Figure 5.9, showing a sample aberration (beam displacement from intended focus) induced by the skull phantom; the corrected beam profile upon applying an adapted delay pattern was also provided. Quantitative assessment and error analysis of the method in this case were reported based on the experimentally measured difference between calculated and measured delay patterns in a variety of temporal focusing scenarios. Following the linear array’s case, the simulation results of the developed adaptive beamforming method were also examined via the matrix probe. Similar to the former case, representative beam profiles were presented, Figure 5.23, to compare original, distorted and corrected beam profiles for cases of both absence and presence of a realistic skull phantom. Quantitative assessment and error analysis of the method in this case were also reported and discussed.

The single-point adaptive focusing experimental results of this chapter are also an indicator of the potential of the approach to correct for phase aberration effects of skull layers in the reception mode of a full beamforming process.
6.1 Introduction

In chapter 5, our new adaptive beamforming method was experimentally verified for the case of single-point focusing in transmission mode. Implementation of that adaptive focusing technique into an imaging process is the focus of the current chapter. As mentioned in Section 2.3, an ultrasonic imaging process via phased arrays is comprised of two major components: transmission mode and reception mode.

The chapter starts with a description of the adaptive imaging process for the case of linear arrays. Implementation of the transmission and reception modes into a final imaging system including ULA-OP is explained. Various examples of 2D sonograms generated through skull phantoms are presented. Qualitative and quantitative comparisons among original, distorted and corrected sonograms are made.

The chapter then continues with implementation of the developed adaptive beamforming process through ULA-OP for the case of matrix probes. Challenges associated with this extension are discussed. Since the current version of ULA-OP does not have 3D screening capabilities, a third-party 3D-viewing software was employed to screen the 3D sonograms. Various examples of final generated 3D sonograms through realistic skull phantoms are presented. Qualitative and
quantitative comparisons among original, distorted and corrected 3D sonograms are made and discussion of the results is provided.

6.2 2D adaptive imaging with linear probes

A software package was developed by the author to prepare inputs for ULA-OP prior to the adaptive imaging process with linear probes. The software accepts user-specified parameters and desired specifications of the 2D sonogram to be generated. It then implements the new single-point adaptive focusing method into transmission and reception mode beamforming processes. The resulting software outputs are then transferred to ULA-OP and the imaging process starts.

Default values for the 2D image were selected to be 40 degrees angle-of-view (-20 to 20 degrees from the normal of the probe with 1 degree increments) covering a total depth range of 2.0 - 12.0cm. These parameters can be altered to any desired range as long as they do not exceed the available inspection area of the linear probe (see section 4.4 for characteristics of the linear probe). Based on these default parameters, preparation of the transmission and reception modes in the developed software is explained in the following subsections.

6.2.1 Transmission mode – linear array

The 64 middle elements of the linear array, i.e. elements 33 to 96, were employed in the transmission mode. As specified in the previous section, the sonogram will comprise 40 lines. Each of these 40 lines will be referred to as a transmission line in this context. All time delay patterns, each responsible for steering the beam of the linear array along one of the transmission lines, are calculated in the software. As an example, transmission mode time delay patterns to be
used for image generation in absence of a skull phantom are shown in Figure 6.1. The current version of ULA-OP is capable of accommodating only one focusing depth in transmission mode; that focusing depth is a user-defined value, and was set to 20cm in the example shown in Figure 6.1. It is notable that unless inspection at a specific depth is intended, choice of a very long focusing depth in transmission mode (beyond the feasible focusing range of the phase array) is normally used. In more recent studies, use of multi-depth focusing in transmission mode has been suggested, mainly to achieve better SNR and lateral resolution in the final image. Use of multi-depth focusing in transmission mode comes at the cost of higher processing time and lower frame rate in the final imaging process.

Figure 6.1: Example of calculated time delay patterns in the transmission mode when a sonogram with 40 degrees angle-of-view (one degree resolution) via the linear probe is desired. Each of the 40 curves in this plot is shown with a different color and represents the time delay pattern for steering along a single transmission line. The focal depth in this example was set to 20cm (infinite focus). The time delay patterns are plotted relative to the center element delay value.
When a skull phantom is present in front of the linear phased array, the skull phantom profile is first extracted via the method described in Section 5.2. The extracted profile is then input to the developed software to calculate all transmission mode time delay patterns adapted to the acoustical and local geometrical properties of the phantom. Results of this process for the sample skull phantom profile of Figure 6.2.a are shown in Figure 6.2.b. In this figure, the same imaging specifications as used to generate Figure 6.1 (defined in section 6.2) were used.

Figure 6.2: (a) Example of an extracted skull phantom profile with the linear probe, and (b) transmission mode time delay patterns adapted to this profile. Same sonogram specifications as Figure 6.1 were desired. Each of the 40 curves in this plot represents the time delay pattern for steering along a single transmission line. The focal depth in this example was set to 20cm (infinite focus). The time delay patterns are plotted relative to the center element delay value.
The timing patterns of Figure 6.1 are distributed symmetrically around the center of the transmitting aperture. This is due to the symmetry of the propagation medium in the absence of the skull phantom. The timing patterns of Figure 6.2.b are not symmetric as they are adapted to the asymmetric geometry of the skull phantom shown in Figure 6.2.a.

The next step is to define the spatial apodization over the transmitting elements. The effect of spatial apodization in sidelobe level reduction in the far-field response of phased arrays was discussed in Section 3.2. Our tests indicated that the “Cosine”, “Hamming”, “Sinc” and “10% truncated Gaussian” weighting functions gave similar results and were optimal in regard to image quality. All these functions are available in the developed software. The “Cosine” apodization function was set as the default in transmission mode.

6.2.2 Reception mode

Beamforming in reception mode of an imaging process is dynamic and is accomplished by passive focusing at various depths along each transmission line. Passive focusing is the term used in reception mode for focusing on a particular focal point on a transmission line. It refers to applying compensating time delays on the receiving elements to align the phase spectra of the RF signals received at all the array elements, before the summation process. After their phase spectra are “synchronized”, the summation process over the RF signals defines a final signal for that particular focal point. This final signal is referred to as post-beamform data as opposed to the originally received RF signals (referred to as pre-beamform data). The amplitude of a post-beamform signal defines the brightness of the corresponding pixel in the final image.
Figure 6.3 shows reception mode time delay patterns for the same sectorial imaging specifications as defined in section 6.2, when imaging in the absence of a skull phantom. In this figure only a small portion of all delay patterns is shown for clarity (the full scale goes from $2.6 \times 10^{-5}$ sec to $16.4 \times 10^{-5}$ sec). Each bundle of curves (shown with a different color) is responsible for focusing at a specific depth during the reception mode focusing process. The curves in each bundle (40 curves) are again responsible for steering from -20 to 20 degrees with one degree increments.

Unlike the transmission mode where only 64 elements could comprise the active aperture, all 128 elements of the linear array were used in the reception mode. Use of more elements than the available channels on the probe’s controller (ULA-OP: 64-channel controller, section 4.3) is possible in reception mode via the compound beamforming concept explained in section 6.3.2 for the matrix probe. Use of all 128 elements in the reception mode for the linear array increases both lateral resolution and SNR of the final 2D sonogram.

In the presence of a skull phantom, reception mode time delay patterns are corrected and adapted to the local skull phantom profile. Corrected reception mode time delay patterns adapted to the same extracted skull phantom profile of Figure 6.2.a are shown in Figure 6.4. Again, the asymmetrical nature of patterns in Figure 6.4 is noticeable compared to the patterns of Figure 6.3. The timing patterns of Figure 6.4 are adapted to the asymmetric geometry of the extracted skull phantom profile of Figure 6.2.a.
Figure 6.3: Example of calculated time delay patterns in absence of a skull phantom for dynamic focusing in the reception mode. The final sonogram generated via these patterns will have a 40 degrees angle-of-view (1 degree lateral resolution) and covers depth range of 2-12cm. Only a small portion of all delay patterns is shown for clarity (the full scale goes from $2.6\times10^{-5}$ sec to $16.4\times10^{-5}$ sec).

Figure 6.4: Example of corrected reception mode time delay patterns adapted to the skull phantom profile of Figure 6.2.a. The final sonogram generated via these patterns will have a 40 degrees angle-of-view (1 degree lateral resolution) and covers depth range of 2-12cm. Only a small portion of all delay patterns is shown for clarity (the full scale goes from $2.6\times10^{-5}$ sec to $16.4\times10^{-5}$ sec).
Calculation of spatial apodization patterns is the next step in the software package. Unlike the transmission mode, apodization in the reception mode is depth-dependent. The “Sinc” apodization function was set as default for reception mode.

### 6.2.3 2D adaptive imaging via the linear array

In our 2D adaptive imaging system, the probe is first held in contact with the skull phantom. The skull phantom profile is extracted and the corrected time delay patterns in both transmission and reception modes are calculated. Our software then communicates with ULA-OP’s built-in software and transfers the corrected timing patterns along with the calculated apodizations to ULA-OP. ULA-OP then implements the desired beamforming process via the control module programmed on it by the author and generates a final 2D sectorial image.

During the imaging process, the frame rate is defined by Pulse Repetition Frequency (PRF), which is the inverse of the Pulse Repetition Interval (PRI). The PRI is the time interval between two consecutive transmission events, representing the transmission/reception time required to generate a single line in the final sonogram. Therefore, forty such intervals are required to refresh all forty lines of our sectorial scan. Within each PRI, the 64 elements of the active aperture transmit their excitation signals based on the timing pattern corresponding to that line. Then in reception mode, the analog signal from each individual probe element is first amplified (Analog Gain) to ensure optimal use of the dynamic range of the Analog to Digital (A/D) converters. The analog gain factor varies with depth to amplify signals from deep regions most (TGC: Time Gain Compensation). The received sampled signals are then time delayed based on the imported reception mode delay patterns signals, and weighted to obtain the desired
apodization. Finally, the weighted and delayed signals are summed to generate the post-beamform data corresponding to that line.

Although the pos-beamform data could be used to generate the final sonogram directly, an In-phase/Quadrature (IQ) demodulation process is performed before final image generation. The main purpose of IQ demodulation is to down-sample the post-beamform signals when a higher than required sampling frequency is used. ULA-OP has a fixed-in-hardware sampling frequency of 50MHz which is much higher than the necessary Nyquist limit for the frequency range of this study (2MHz-55%BW).

An experimental immersion setup, Figure 6.5, was prepared for the 2D adaptive imaging. Three randomly shaped reflectors (made of Lead) with characteristic dimensions of the order of ~0.5mm to ~4.5mm were used as targets. Approximate size of each target is shown in Figure 6.5 as the average of maximum and minimum distance across the target. The targets were randomly placed in front of the linear probe and our 4D scanner was used to position various sections of the skull phantom in front of the array, as shown in Figure 6.5. The linear probe itself was connected to ULA-OP and the developed software was used to communicate with the sectorial imaging routine programmed on ULA-OP based on the same imaging specifications as used in section 6.2.
Results of imaging for 20 different locations of the array on the skull phantom are discussed in this thesis. Three representative examples of image distortion by the skull phantom (‘a: severe distortion’, ‘b: moderate distortion’ and ‘c: mild distortion’) are shown in Figure 6.6. Figure 6.6.a-original shows the sonogram produced via the conventional sectorial imaging routine of ULA-OP when the skull phantom is not present in the setup. The “original” positions of the three reflectors can be seen in this diagram; this sonogram can be used as a reference to gauge the ability of our new beamforming algorithm to locate reflectors when skull bone is present. This sonogram is therefore repeated in each row of Figure 6.6 to facilitate comparison between “distorted” and “corrected” images in each example. The “ringing” effect seen below the main image of each target is due to multiple reflections and reverberation within each target. The extra echoes near the top of the distorted and corrected images are due to multiple reflection and reverberation effects of the skull phantom.

The “distorted” image in each of the three examples of Figure 6.6 shows the sonogram with same three reflectors, but with the skull phantom positioned in contact with the probe; conventional timing patterns are used with no correction for the skull’s distortional effects. As
can be seen, the targets are either missing or sometimes severely displaced due to phase aberration caused by the skull phantom. These displacements are in terms of both lateral position and depth. Schematic displacement arrows are shown in the first distorted image as an example. This distortion can all be traced to acoustical mismatch between the skull phantom and brain tissue (represented by water in our experiments).

When the corrected timing patterns are applied, the beamforming process is adapted to the skull phantom properties. As a result, there is proper compensation for the phase aberration effects of the skull in the “corrected” image, which contains accurate information regarding the position of the targets.
Figure 6.6: Three representative examples of applying the developed adaptive imaging method through the skull phantom of Figure 6.5. These examples show different levels of distortion induced by the skull phantom.
Quantitative results of the examples of Figure 6.6 are reported in Table 6.1. For each example, a comparison between the location of original, distorted and corrected images was made for each of the three targets. Coordinates of the center point in each reflector’s image, i.e. geometric center of each target’s image excluding the multiple reflections and reverberation tail, was used for this assessment. The author developed a MATLAB code which defined the geometric center coordinates of each target’s image via the IQ signals of the corresponding sonogram.

Table 6.1: Quantitative comparison among original, distorted and corrected images for the examples shown in Figure 6.6

<table>
<thead>
<tr>
<th>Example</th>
<th>Reflecting target No.</th>
<th>Displacement from original location</th>
<th>Re-positioning error</th>
</tr>
</thead>
<tbody>
<tr>
<td>(a)</td>
<td>1</td>
<td>Missing echo</td>
<td>0.1 mm</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>12.2 mm</td>
<td>0.3 mm</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>20.1 mm</td>
<td>0.7 mm</td>
</tr>
<tr>
<td>(b)</td>
<td>1</td>
<td>5.3 mm</td>
<td>0.1 mm</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>8.4 mm</td>
<td>0.2 mm</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>11.1 mm</td>
<td>0.5 mm</td>
</tr>
<tr>
<td>(c)</td>
<td>1</td>
<td>6.0 mm</td>
<td>0.0 mm</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>4.2 mm</td>
<td>0.3 mm</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>5.1 mm</td>
<td>0.5 mm</td>
</tr>
</tbody>
</table>
Among all 20 examples, including the three representative examples of Table 6.1, a minimum distortional displacement of 2.2mm was observed when uncompensated delay times were used for the array elements. This minimum value for displacement in the distorted images was observed for target No.1 when the skull phantom in front of the probe had a very flat and thin profile (~4.1mm thick). Cases of maximum distortion, on the other hand, resulted in total elimination of reflecting targets No.1 and No.2 in two separate distorted sonograms. A maximum displacement of 25.3mm was observed for target No.3 in one of the distorted sonograms. By contrast, the positioning errors ranged from 0.0 to 0.9 mm when our new algorithm to compensate for skull-induced aberration was employed.

6.2.4 Discussion

As an important and promising milestone in the project, the above results verified the applicability of the developed adaptive beamforming method for 2D adaptive imaging via linear arrays. The presented results also demonstrated the potential of the developed method to be further enhanced and extended for 3D adaptive image generation.

It was mentioned that a maximum re-positioning error of 0.9mm (for target No.3) was observed once our new beamforming algorithm was employed. The main sources of error causing displacements of targets in the corrected images compared to their original positions are:

- Error in the measured sound speed for skull phantom: The uncertainty in sound speed measurement for the skull phantom is 10m/s as specified in Table 4.3. This inaccuracy could result in a repositioning error of up to 0.1mm in the imaging area of Figure 6.6 (2-12cm depth range and 40° angle of view).
- Error in the skull profile extraction system: Inaccuracy of 0.1mm in skull thickness measurement would result in repositioning error of up to 0.1 mm in the defined imaging area.

- Error in the measured sound speed for water: Inaccuracy of 10m/s would result in repositioning error of up to 0.05mm in the above imaging area.

- Error associated with defining the geometric center of each target’s image: The image of a target in the generated sonograms is not an indicator of its actual shape and dimensions. Finding the location of the reflecting targets was the true goal of this study. Although applying suitable image processing algorithms would help to find a correlation between the image of a target and its physical dimensions, this was out of the scope of this study.

Compared to the original 2D sonogram, lower intensity levels and consequently lower contrast were observed on the distorted and corrected 3D sonograms in Figure 6.6. This was expected, as the presence of the skull phantom leads to extra attenuation and interfacial transmission losses.

### 6.3 3D adaptive imaging via matrix arrays

Implementation of the 3D adaptive beamforming method on ULA-OP was accomplished in a similar manner to the previous case of the linear probe. The 2D software was extended to prepare input parameters for ULA-OP prior to the 3D imaging process. The software accepts user-specified parameters and desired specifications of the final 3D sectorial image. It then implements the new single-point adaptive focusing method into complete transmission and
reception mode beamforming processes. The resulting outputs of the software are then transferred to ULA-OP and the imaging process starts.

A new control module was programmed on ULA-OP to accommodate a 3D sectorial imaging scenario. Generation of final 3D sonograms with 40 degrees angle of view (-20 to 20 degrees from the normal of the probe along both surface axes - one degree resolution) covering a total depth range of 2.0-6.0cm were set as default values in the software. These parameters where selected to use the maximum capacity of ULA-OP; due to the limited internal memory space of the current version of ULA-OP, the lateral and depth range of the imaging volume is limited. Therefore a compromise between angle-of-view and depth range was necessary.

### 6.3.1 Transmission mode

The software first prepares the transmission mode time delay patterns, each responsible for steering along one line of the 3D sonogram. The 64 central elements of the matrix array in the arrangement shown in Figure 6.7 were used in this mode. The default focusing depth in transmission mode for the matrix array was set to 20cm, similar to the linear array’s case. Illustration of all 2D time delay patterns (equivalent of Figure 6.1) in a 3D plot is not readily feasible due to the multi-dimensional complexity of the system.
When a skull phantom is present in front of the phased array the skull phantom profile is first extracted via the method explained in Section 5.2. The extracted profile is then input to the developed software to calculate the transmission mode time delay patterns adapted to the acoustical and geometrical properties of the phantom. Similar to the linear array’s case, a two-dimensional “Cosine” apodization function was set as the default in transmission mode.
6.3.2 Reception mode

In reception mode, all 192 (= 64x3) available elements of the matrix array can be accommodated as the receiving aperture based on the *compound beamforming* concept. In compound beamforming, each transmission mode focusing scenario is repeated (three times in our case). For each of these three transmissions, reception mode dynamic focusing is performed in reception mode with a separate 64-element aperture. Arrangements of the receiving elements in this mode are graphically shown in Figure 6.8.a. In this figure, each of the three receiving sparse apertures is shown with a different color. Again, illustration of an example of all 2D time delay patterns in reception mode (equivalent of Figure 6.3) in a 3D plot is too complex to be informative.

The received RF signals from all three apertures are combined to mimic a single 192-element aperture. These pre-beamform data are then processed, as described in Section 6.2.2, to generate post-beamform data and IQ demodulation. The IQ data are used to generate a 3D sonogram with the desired specifications.
In the presence of a skull phantom, reception mode time delay patterns are corrected and adapted to the local skull phantom profile through the new software. Spatial apodization patterns are then calculated; the two-dimensional “Sinc” apodization function was set as the default in reception mode, just as was done for the linear array.

### 6.3.3 3D adaptive imaging via the matrix array

In a 3D adaptive imaging setup, the matrix probe is initially held in contact with the full-calvaria skull phantom. The skull phantom profile is extracted and the corrected time delay patterns in
both transmission and reception modes are calculated through the developed software as described in Sections 6.3.2 and 6.3.3. The software then transfers the corrected timing patterns and calculated apodizations to ULA-OP. ULA-OP then implements the desired beamforming process to generate a 3D sectorial image. During the imaging process, starting from one corner of the array the 40x40 lines of the final sonogram are made one by one. The data acquisition and conversion processes (Pre-beamform to post-beamform to IQ data) remain the same as described in section 6.2.3.

A challenge in the case of imaging via the matrix probe was to employ suitable 3D-viewing software that is compatible with ULA-OP. Although the current version of ULA-OP has 3D beamforming capabilities, it does not have 3D screening software. Therefore a third-party 3D-viewer was required to receive the final data from ULA-OP and display a volumetric image of the inspection area. For this purpose, “3D-Slicer” open source viewer (Version 4.3.0) was chosen. The open-source software was programmed by one of our team members, under general supervision of the author, to interact with ULA-OP’s software. The current customized version of the 3D-Slicer reads the data from ULA-OP’s external memory and updates the 3D image at a frame rate of 5 frames/sec.

An experimental immersion setup, Figure 6.9, was prepared to be used for 3D adaptive imaging. Five randomly shaped reflectors (made of lead and skull phantom bone material) of different sizes from ~1.5mm to ~5mm were used as imaging targets. The targets were randomly placed in front of the matrix probe and our 4D scanner was used to position different sections of the full-calvaria skull phantom in front of the matrix array. The matrix array itself was connected to ULA-OP and the final developed software was used to communicate with the 3D sectorial
imaging routine programmed on ULA-OP. The imaging specifications remained the same as specified in section 6.3.1.

Figure 6.9: The experimental setup used for adaptive imaging via the matrix array

Similar to the linear array’s case, results of imaging through 20 different locations on the full-calvaria skull phantom are discussed. Three representative examples (‘a: severe distortion’, ‘b: moderate distortion’ and ‘c: mild distortion’) are shown in Figure 6.10. Figure 6.10.a-original shows the 3D sonogram produced via the conventional 3D sectorial imaging routine programmed on ULA-OP when the skull phantom is absent. The true (“Original”) positions of the five reflectors can be observed from this 3D sonogram for future reference and comparison. This sonogram is therefore repeated in each row of Figure 6.10 for better comparison with distorted and corrected images in each example.

The Distorted image in each example shows the 3D sonogram when the skull phantom is positioned in contact with the probe and a conventional array element timing pattern is used (no
compensation for aberration caused by skull). The targets are either missing entirely, or severely displaced compared to their original positions due to phase aberration induced by the skull phantom. As was seen with the linear array, the errors in the target locations include both lateral and depth displacements.

When the corrected timing patterns are applied, the beamforming process is adapted to compensate for the skull phantom. The corrected image in each example accurately depicts the positions of the targets.
Figure 6.10: Three representative examples of applying the developed 3D adaptive imaging method through the full-calvaria skull phantom of Figure 6.9. These examples show different levels of distortion caused by the skull phantom.

Quantitative information for the examples of Figure 6.10 is given in Table 6.2. For each example, comparison between the location of original, distorted and corrected images was made for each of the five targets. Coordinates of the center point in each reflector’s image were used for this assessment, and were automatically located using the IQ signals of the corresponding 3D sonogram.
Table 6.2: Quantitative comparison among original, distorted and corrected images for the examples shown in Figure 6.10

<table>
<thead>
<tr>
<th>Example</th>
<th>Reflecting target No.</th>
<th>Displacement from original location</th>
<th>Re-positioning error</th>
</tr>
</thead>
<tbody>
<tr>
<td>(a)</td>
<td>1</td>
<td>Missing echo</td>
<td>0.1 mm</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>9.8 mm</td>
<td>0.1 mm</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>12.0 mm</td>
<td>0.2 mm</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>14.3 mm</td>
<td>0.4 mm</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>Missing echo</td>
<td>0.5 mm</td>
</tr>
<tr>
<td>(b)</td>
<td>1</td>
<td>7.8 mm</td>
<td>0.2 mm</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>7.9 mm</td>
<td>0.2 mm</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>6.5 mm</td>
<td>0.3 mm</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>11.1 mm</td>
<td>0.3 mm</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>Missing echo</td>
<td>0.5 mm</td>
</tr>
<tr>
<td>(c)</td>
<td>1</td>
<td>3.1 mm</td>
<td>0.1 mm</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>3.2 mm</td>
<td>0.2 mm</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>4.0 mm</td>
<td>0.3 mm</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>5.0 mm</td>
<td>0.4 mm</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>6.3 mm</td>
<td>0.4 mm</td>
</tr>
</tbody>
</table>

Among all 20 examples, including the three representative examples of Figure 6.10 and Table 6.2, a minimum distortional displacement of 3.3mm was observed when a conventional beamforming technique was implemented with no compensation for the skull phantom. This minimum value for displacement was observed on target No.2 when the skull phantom in front
of the probe had an almost flat profile, with uniform thickness of ~6.5mm. Other sonograms featured much larger distortions and even complete echo loss for three reflecting targets. A maximum displacement of 15.1mm was observed on target No.4 in one of the distorted sonograms. On the corrected sonograms where adaptive beam forming was used to compensate for the skull phantom, minimum and maximum re-positioning errors of 0.0mm and 0.6mm were observed, respectively.

6.3.4 Discussion

The results of the new 3D adaptive imaging method via the matrix probe showed consistency with the previous results for the linear array. For the 20 examples considered, a maximum re-positioning error of 0.6mm when imaging a target 5.4mm away from the surface of the probe was observed for the matrix array. The re-positioning error on the corrected images is attributed to the same sources of errors discussed in Section 6.2.4 for the linear array. Compared to the original 3D sonogram, lower intensity levels and consequently lower contrast were observed on the distorted and corrected 3D sonograms; this is due to the presence of the attenuative full-calvaria skull phantom.

In all generated 3D sonograms, a high level of background noise was observed in the central regions. This is manifested as the darker background in central regions compared to the sides of the volumetric images. The author and his team made several attempts to determine the source of such noise, e.g. by examining the grounding connection of the probe’s casing, but the noise source could not be established. The array manufacturer was able to reduce (but not eliminate) the noise when the probe was sent for repair. In the case of ultrasound propagation in real brain tissue, speckle noise will add even further to the overall background noise. Noisy 3D sonograms
could cause serious operational difficulties, such that this issue will need to be addressed before future clinical trials.
A new adaptive beamforming method was developed for ultrasonic imaging via small-aperture phased arrays through highly aberrating layers, such as human skull. The objective was to overcome the major disadvantages of existing phase aberration correction methods when their application for ultrasonic transcranial imaging of certain types of head injuries is considered. A noticeable number of head injuries involve foreign objects of various sizes lodged into the brain tissue. If conventional imaging techniques via phased arrays are used for ultrasonic transcranial imaging of such objects, strong phase aberration and refractive effects of the human skull could result in significant distortion of beam patterns causing defocusing of the ultrasonic field. These effects, along with human skull's natural strong attenuation, lead to significant degradation of the image quality, loss of spatial resolution, and possibly large displacement or total disappearance of reflecting targets in the resulting image.

The proposed method was based on a custom-designed predetermination of acoustical and geometrical properties of the segment of the skull layer in contact with the array. The procedure is in fact a construction of a matched filter that automatically adapts the transmission and reception patterns of the phased array to the local geometry and acoustical properties of the skull and cancels its distorting effects.
7.1 **Major contributions**

The major contributions of this work are as follows:

- The background theory and a new algorithm were developed for single-point adaptive focusing via small aperture ultrasonic phased arrays. The proposed method was optimized for the case of human skull as the aberrating layer. The theory and algorithms were initially developed for the case of linear phased arrays for 2D adaptive beamforming. The proposed theory was then extended to the case of matrix phased arrays for 3D adaptive beamforming.

- Numerical simulations with wave propagation modeling were developed in MATLAB to accommodate and verify the proposed theory for single-point adaptive focusing through simulated skull layers. Simulation results for cases of both linear and matrix arrays were presented and discussed. Although the physical theory of the proposed method remained the same for both cases of linear and matrix probes, simple extension of the proposed algorithm from linear to matrix phased arrays was not possible. This mainly stems from the presence of more complicated boundary conditions in the case of matrix arrays and also the high processing time of the original approach. Therefore, a more accurate, efficient, and robust approach was designed in transition to matrix arrays. The author believes that to date no commercial simulators are capable of these particular computations. This in fact was the main motivation to develop the independent simulation software used in this study. The numerical simulation results validated the proposed theory for single-point adaptive focusing in transmission mode. The single-point adaptive focusing simulation results indicated the potential of the proposed approach to correct for phase aberration effects of skull layers in experimental setups.
To check the validity of the developed theory and numerical simulations for single-point adaptive focusing, a series of laboratory experiments were designed and conducted. For this experimental verification:

- Three generations of skull phantoms were developed by our team, with the development of the first generation phantoms by the author himself. The final generation of our phantoms has its main geometrical and acoustical properties, i.e. sound speed, attenuation and density, well within the ranges reported in the literature for the human skull.

- A custom-designed time-of-flight measurement technique was specifically developed to obtain information about the geometry of the skull phantom prior to the beamforming process. This experimental mode was designed and programmed on ULA-OP to accurately measure the skull profile in front of the array.

- Two separate experimental setups were designed and prepared for qualitative and quantitative verification of the single-point adaptive focusing simulation results.

The simulation results were then examined for cases of both linear and matrix arrays. For qualitative assessment, ultrasonic beam profiles were collected and presented to compare original, distorted and corrected beam patterns in both absence and presence of skull phantoms. Quantitative assessment and error analysis of the method were also conducted based on the experimentally measured deviation between calculated and measured delay patterns in a variety of temporal focusing scenarios. The single-point adaptive focusing experimental results confirmed the potential of the approach to correct for phase aberration effects of skull layers in an adaptive imaging process.
The experimentally verified single-point adaptive focusing method was implemented into an imaging process for cases of both linear and matrix arrays. For this purpose, ULA-OP was fully programmed to integrate the developed beamforming algorithms; a software package was also developed to prepare inputs for ULA-OP prior to the imaging process. The software accepts user-specified parameters and desired specifications of the 2D or 3D sonograms to be generated. It then implements the new single-point adaptive focusing method into complete transmission and reception mode beamforming processes. The resulting software outputs are then transferred to ULA-OP to start the imaging process.

- **2D adaptive imaging**- Various examples of 2D sonograms (2-12cm depth range, 40 degrees angle-of-view) generated with a 2MHz linear array (section 4.4) were presented in section 6.2.3. Qualitative and quantitative comparisons among original, distorted and corrected 2D sonograms were made. The targets had characteristic dimensions of 0.5-4.5mm. A maximum re-positioning uncertainty of 0.8mm in the locations of reflecting targets was observed in a number of trials when the new adaptive beamforming scheme was implemented. Current processing time to generate an adapted 2D sonogram on our laboratory desktop (dual-core-i5 intel-processor, 4Mbytes of RAM) is about 14 seconds.

- **3D adaptive imaging**- Various examples of 3D sonograms (2-6cm depth range, 40 degrees angle-of-view along both surface axes) generated with the 2MHz matrix probe (section 4.5) were presented in section 6.3.3. Qualitative and quantitative comparisons among original, distorted and corrected 3D sonograms were made. The targets had characteristic dimensions of 1.5-5.0mm. A maximum re-positioning uncertainty of 0.6mm was observed in 20 trials when the new adaptive beamforming
scheme was implemented. Current processing time to generate an adapted 3D sonogram on our laboratory desktop (dual-core-i5 intel-processor, 4Mbytes of RAM) is about 12 minutes.

- A table-top prototype of an ultrasound imaging instrument, including the biomedical matrix array (section 4.5), an imaging screen, and ULA-OP was delivered.

7.2 **Recommendations for future work**

A. The processing time of the delivered imaging system must be reduced for practical applicability of the developed method. As a proof-of-concept study, all software codes were developed in MATLAB. The processing time can be dramatically reduced by applying appropriate code optimization techniques, translation to a compiling language (e.g. C or C++), use of multi-processor system, etc.

B. Finding the location of the reflecting targets was the primary goal of this study. However, the image of a target in the generated sonograms is not an indicator of its actual shape and dimensions. Suitable image processing algorithms can be applied to find a correlation between the image of a target and its physical shape and true dimensions.

C. More advanced signal processing and cross-correlation techniques can be investigated and employed in the profile extraction software. This must help to detect the inner boundary profile at even high porosity regions where multiple reflections and reverberation effects from the outer boundary and diplo’e severely overlap with the reflection from the inner boundary. This would make the adaptive beamforming
technique presented in this thesis applicable to a larger percentage surface area of a typical skull.

D. The delivered table-top laboratory prototype must be miniaturized into a portable adaptive imaging system for practical applicability of the system in emergency scenes.
**APPENDIX A**

Main features of ULA-OP are summarized in Table A.1.

**Table A.1: Main features of ULA-OP [50]**

| **General Features** | Open platform  
|                     | 64 independent TX/RX channels  
|                     | Size: 33x23x18 cm; Weight: 5.5 kg + 4 kg (power supply).  
|                     | Power consumption < 90W  
| **Transmitter** | 64 Arbitrary waveform generators  
|                  | Max output voltage: 24 Vpp  
|                  | Frequency: 1 to 16 MHz  
| **Receiver** | Input Noise: 2 nV/√Hz  
|                | Bandwidth: 1 to 16 MHz  
|                | Analog gain: 6 – 46 dB with programmable TGC  
|                | 12bit @ 50 MSPS ADCs  
| **Beamformer** | Programmable apodization and delays (dynamic focusing)  
| **Processing Modules** | Coherent demodulation, band-pass filtering, decimation, B-mode,  
|                        | Multi-gate Spectral Doppler, Vector Doppler… (open to new, custom modules).  
| **Storage capabilities** | Up to 1 GB for RF data (pre- or post-beamformed)  
|                        | Up to 512 MB for baseband data  

Figure A.1 illustrates the system architecture [50]. The operation sequence during a scanning session is supervised by the ‘system manager’, which coordinates the transmission (TX) and reception (RX) sections. In a typical Pulse Repetition Interval (PRI), after the system manager issues a PRI start, the TX section generates 64 independent custom-designed waveforms that are
sent to linear power amplifiers. A programmable multiplexer-matrix maps the waveforms to a group of elements which can be arbitrarily selected out of the available 192 probe channels. During the RX phase, the multiplexer-matrix composed by MAX312L (Maxim Integrated Products Inc., Sunnyvale, CA) can be programmed to receive the backscattered echoes either from the same transducer elements excited in TX or from a different element group. These signals are conditioned by 16 AD8335 (Analog Devices, Norwood, MA) that includes Low Noise Amplifiers (LNA). The amplified signals feed a bank of 8 ADS5281 (Texas Instruments, Dallas, TX), each featuring 8 Analog-to-Digital converters (AD), working at 50 Msps with 12 bit resolution. The DB is based on 5 FPGAs from the Stratix family (Altera, San Jose, CA, USA) and one DSP from the C6455 family (Texas Instruments, Austin, TX, USA). Four FPGAs are devoted to beamforming. The last FPGA, referred to as the coprocessor FPGA, supports the DSP in the beamformed signal elaboration. The processed data are finally transferred to the host PC through the USB 2.0 channel. The software running on the PC performs signal processing for video reproduction, and finally presents the results on display windows. The same software includes the control interface that allows the user to set the preferred operating environment, to change parameters affecting real-time operations and to download the acquired data.
Figure A.1: Simplified architecture of ULA-OP [50]
The custom-designed cross-correlation algorithm used for skull profile extraction is described in this appendix. The algorithm is explained when it is applied to the matrix array which is the more comprehensive case:

After signal acquisition (196 IQ signals from the 14x14-element aperture on the matrix probe), the first step is to extract the outer boundary of the skull profile. This is simply achieved by finding the depth corresponding to maximum amplitude on each of the IQ signals (strongest reflection). The mesh-grid of these depths defines the profile of the outer boundary. The next step is to extract the inner boundary. Use of the second strongest reflection of the IQ signals is not a reliable method for detection of the inner boundary. The reason is that at many areas across a typical skull profile, multiple reflections and reverberation effects from the outer boundary and diplo’e overlap with the reflection from the inner boundary. Simple use of second strongest reflection leads to incorrect profile extraction is such areas. Therefore after examining some of the conventional cross-correlation techniques, the following method was custom-designed and developed for inner boundary detection:

The 196 IQ signals are first gated for a range of 4-14mm deeper than the outer boundary. Reflection from the inner boundary is assumed to fall in this depth range. The algorithm then needs to find a suitable point on the inner boundary to start the cross-correlation. To find this starting point, first the signal corresponding to the highest peak among the previously gated 196 signals is picked. Let us assume this gated IQ signal is from the 3x3-element sub-aperture at \( (x_0, \)
The coordinates \((x_0, y_0)\) refer to the location of the center element of the sub-aperture on the surface of the matrix probe. The temporal sample corresponding to the highest peak on this signal, \(S_{z00}\), is picked in the next step and the starting point of the cross correlation is set as \((x_0, y_0, S_{z00})\).

After the starting point is picked, the cross-correlation is applied by comparing two temporal sample sub-sets from adjacent signals. The first subset of temporal samples is chosen on the \((x_0, y_0)\) signal from \((S_{z00} - 12)\) to \((S_{z00} + 12)\) making a 25-samples subset. This subset is called \(A_k\). A similar subset, \(B_k\), is then taken from the adjacent line at \((x_0, y_1)\). The two subsets are assumed to be similar except for a slight misalignment, \(m\), along the \(z\)-axis (if present). The cross-correlation of the two subsets is then calculated as

\[
CC = \sum A_k B_k^* \tag{B.1}
\]

where * is the conjugate operator. The misalignment, \(m\), of the two subsets is finally calculated via [1]

\[
m = \frac{\theta c}{4\pi k F_0} \tag{B.2}
\]

where

\[
\theta = \tan^{-1} \frac{\text{imag}(cc)}{\text{real}(cc)} \tag{B.3}
\]

\(F_0\) is the central frequency of the transmitted burst (2MHz), \(c\) is the speed of sound in the skull phantom, and \(k\) is an empirical correction factor (=0.85 in the code). This factor is required because of the non-symmetric bandwidth of the probe; and also since the mean frequency of the received signal is not necessarily the same as the transmitted signal, mostly due to attenuation in
skull. The inner boundary point on the adjacent signal is now located at \((x_0, y_1, S_{z01})\) with \(S_{z01}\) calculated as \(S_{z01} = S_{z00} + m\).

Starting from the signal of \((x_0, y_0)\), and repeating the procedure for all adjacent signals, it is possible to scan across the surface of the probe and find the \(S_z\) value for all other 195 lines.
REFERENCES


