Advanced 3-D Ultrasound Imaging: 3-D Synthetic Aperture Imaging using Fully Addressed and Row-Column Addressed 2-D Transducer Arrays.

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Advanced 3-D Ultrasound Imaging

3-D Synthetic Aperture Imaging using Fully Addressed and Row-Column Addressed 2-D Transducer Arrays

Hamed Bouzari

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Kgs. Lyngby, Denmark 2017
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Summary

Compared with conventional 2-D ultrasound imaging, real-time 3-D (or 4-D) ultrasound imaging has several advantages, resulting in a significant progress in the ultrasound imaging instrumentation over the past decade. Viewing the patient’s anatomy as a volume helps physicians to comprehend the important diagnostic information in a noninvasive manner. Diagnostic and therapeutic decisions often require accurate estimates of, e.g., organ, cyst, or tumor volumes. 3-D ultrasound imaging can provide these measurements without relying on the geometrical assumptions and operator-dependent skills involved in such estimations using 2-D scans. Although the detail resolution of ultrasound cannot compete with 3-D imaging modalities such as CT and MRI, the combination of patient safety by using nonionizing radiation, cost-effectiveness, portability, and real-time imaging ability makes ultrasound the preferred choice in many clinical applications.

Real-time 3-D ultrasound imaging is still not as widespread in use in the clinics as 2-D ultrasound imaging. Two limiting factors have traditionally been the low image quality as well as low volume rate achievable with a 2-D transducer array using the conventional 3-D beamforming technique, Parallel Beamforming.

The first part of the scientific contributions of this Ph.D. project demonstrate that 3-D synthetic aperture imaging achieves a better sensitivity and a higher volume rate than the parallel beamforming technique. Data were obtained using both Field II simulations and measurements with the ultrasound research scanner SARUS and a 3.8 MHz 1024 element 2-D transducer array. In all investigations, 3-D synthetic aperture imaging achieved a better resolution, lower side-lobes, higher contrast, and better signal to noise ratio than parallel beamforming. This is achieved partly because synthetic aperture imaging removes the limitation of a fixed transmit focal depth and instead enables dynamic transmit focusing. Particularly, synthetic aperture imaging could increase the achievable volume rate compared with parallel beamforming, to almost 50 times.

Lately, the major ultrasound companies have produced ultrasound scanners using 2-D transducer arrays with enough transducer elements to produce high quality 3-D images. Because of the large matrix transducers with integrated custom electronics, these systems are extremely expensive. The relatively low price of ultrasound scanners is one of the factors for the widespread use of ultrasound imaging. The high price tag on the high quality 3-D scanners is limiting their market share.
Row-column addressing of 2-D transducer arrays is a low cost alternative to fully addressed 2-D arrays, for 3-D ultrasound imaging. Using row-column addressing, the number of transducer elements is dramatically reduced. This reduces the interconnection cost and removes the need to integrate custom made electronics into the probe. Two downsides of row-column addressing 2-D arrays are its lower lateral resolution due to its one-way focusing compared with two-way focusing in fully addressed 2-D arrays and also the inherent forward-looking imaging field of view.

In the second part of the scientific contributions of this Ph.D. project, row-column addressing of 2-D arrays was investigated to assess the possibilities and drawbacks associated with transducer arrays using this addressing scheme, when integrated into probe handles. For that reason, two in-house prototyped 62+62 row-column addressed 2-D array transducer probes were manufactured using capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technology. Based on a set of acoustical measurements the center frequency, bandwidth, surface pressure, sensitivity, and acoustical cross-talks were evaluated and discussed. The imaging quality assessments were carried out based on Field II simulations as well as phantom measurements. Moreover, an analysis on comparing the lateral resolution with a fully addressed array were presented. To improve the imaging sensitivity, spatial matched filter beamforming was used as well as delay-and-sum approach.

An analysis on increasing the inherent forward-looking achievable field of view of a flat row-column addressed 2-D array by using a double curved row-column addressed 2-D array was presented. A delay-and-sum beamforming approach suitable for a double curved row-column addressed 2-D array was introduced. Due to challenges on manufacturing double curved 2-D arrays, using a diverging acoustical lens was proposed and its imaging abilities were evaluated based on Field II simulations and measurements. Thereby, the inherent imaging limitation with flat row-column addressed 2-D arrays was overcome by using a diverging lens. Overall, having a low channel count and a large field of view, offers the potential to fabricate arrays with large aperture sizes, which is important for abdominal scans. Thus by using a curved row-column addressed 2-D array, 3-D imaging with equipment in the price range of conventional 2-D imaging could be possible.

The main part of the thesis consists of eight scientific papers submitted for international conferences and journals during the Ph.D. project.
Sammenlignet med konventionelle 2-D ultralydsbilleddannelse, real-time 3-D (eller 4-D) ultralydsbilleddannelse har flere fordele, som resulterer i et betydeligt fremskridt inden for ultralydsskanning instrumentering i det seneste årti. Skanning af et 3-D volumen giver lægen frihed til at undersøge den målte anatomi i ethvert ønskeligt snit, efter skanningen er afsluttet. Dette giver mulighed for præcise målinger af organernes størrelse, og gør skanningen mere operatør uafhængig. Selvom opløsning af ultralyd ikke kan konkurrere med 3-D billeddiagnostiske modaliteter såsom CT og MR, så gør kombinationen af patientsikkerhed ved at bruge ikke-ioniserede stråling, omkostningseffektiviteten, bærbaretheden, og billeder i real-time, at ultralyd er det foretrukne valg i mange kliniske situationer.

Real-time 3D-ultralydsskanning er stadig ikke så udbredt i brug i klinikkerne som 2-D ultralydsskanning. To begrænsende faktorer har været traditionelt, den lave billedkvalitet såvel som den lave volume/frame rate som kan opnås med en 2-D transducer array ved hjælp af den konventionelle 3-D beamforming teknik, Parallel Beamforming.

I den første del af dette videnskabelige bidrag i denne afhandling blev det vist, at 3-D syntetisk blænde billeddannelse opnår en bedre billedkvalitet end Parallel Beamforming. Undersøgelsen blev lavet ved hjælp af både Field II simuleringer og målinger med en eksperimentel ultralydskanner, SARUS, og et 3.8 MHz 1024 element 2-D transducerarray. I alle undersøgelser opnåede 3-D syntetisk apertur billeddannelse en bedre opløsning, lavere sidesløjfer, højere kontrast og bedre signal støjforhold end Parallel Beamforming. Dette blev opnået, til dels fordi syntetisk blænde billeddannelse fjerner begrænsningen af en fast sende fokus dybde og i stedet gør det muligt dynamisk at fokusere sendeskuddet. Især syntetisk apertur billeddannelse øger den opnåelige volumen/frame rate sammenlignet med Parallel Beamforming, med næsten 50 gange.

De seneste år har de store ultralydsselskaber produceret ultralydsskannere, der anvender 2-D transducerarrays med nok transducererelmente til at opnå en høj kvalitet i deres 3-D billeder. På grund af de store matrix transducere med specialdesignet, integreret elektronik er disse systemer ekstremt dyre. Den relativt lave pris på ultralydsskannere er en af de faktorer, der har hjulpet ultralydsbilleddannelse med at blive så udbredt. Den høje pris på kvalitets 3-D skannerne er en begrænsende faktor på deres markedsandel.
Row-Column adressering af 2-D transducerpaneler er et prisbilligt alternativ til fuldt adresserede 2-D arrays, til 3-D ultralydsskanning. Ved brug af Row-Column adressering, er antallet af transducerelementer dramatisk reduceret. Dette reducerer omkostningerne til integrering med transducerhovedet, og det fjerner helt behovet for at skulle integrere specialfremstillet elektronik. To ulemper ved Row-Column adressering af 2-D arrays er dels dens laveere laterale opløsning på grund af envejsfokuseringen i forhold til tovejsfokuseringen i fuldt adresserede 2-D arrays, og så er synsfeltet begrænset til direkte foran transduceren.

I den anden del af dette videnskabelige bidrag i denne afhandling, er Row-Column adressering af 2-D arrays blev undersøgt for at vurdere mulighederne og ulemperne, når de integreres i et probehåndtag. To 62x62 Row-Column-addressede 2-D array transducer prototype prober er blevet fremstillet baseret på Capacitive Micromachined Ultrasonic Transducer (CMUT) og Piezoelektrisk Tandsucer (PZT) teknologi. På baggrund af en række af akustiske målinger er center frekvens, båndbredde, overfladetryk, sensitivitet og akustisk cross-talk blevet evaluereet og diskuteret. Billedkvaliteten blev vurderet på baggrund af Field II simuleringer samt fantom målinger. Derudover blev den laterale opløsning analyseret og sammenlignet med et fuldt adresserede 2-D array. For at forbedre billeddannelse sensitiviteten, blev spatial matched filter beamforming anvendt såvel som delay-and-sum beamforming.


Den væsentligste del af afhandlingen består af otte videnskabelige artikler indsendt til internationale konferencer og tidsskrifter i løbet af Ph.D. projektet.
Preface

This Ph.D. thesis has been submitted to the Department of Electrical Engineering at the Technical University of Denmark in partial fulfillment of the requirements for acquiring the Ph.D. degree. The research providing the foundation for the thesis has been conducted over a period of three years from December 14th, 2013 to December 14th, 2016 at the Center for Fast Ultrasound Imaging (CFU), the Biomedical Engineering Group, Department of Electrical Engineering, Technical University of Denmark. The project was funded by grant 82-2014-4 from the Danish National Advanced Technology Foundation and BK Ultrasound Aps, Denmark. The thesis recapitulates the conducted research and included are four journal papers and four conference papers as well as two patents. During my work I have had the opportunity to attend conferences in Chicago, Orlando, Taipei, San Diego, and Tours to present my research, and it has led to many fruitful discussions. Traveling to these conferences has been a huge privilege, and the experiences have broadened my horizon of both the technical and the clinical side of medical ultrasound as well as within acoustics in general. Additionally, the conferences have given me the opportunity to nurture and expand my professional as well as social networks.

Hamed Bouzari
Copenhagen, December 2016
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### Abbreviations

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<tr>
<td>1-D</td>
<td>One Dimensional</td>
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<td>Two Dimensional</td>
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<td>3-D</td>
<td>Three Dimensional</td>
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<tr>
<td>ASICs</td>
<td>Application-specific Integrated Circuits</td>
</tr>
<tr>
<td>CMUT</td>
<td>Capacitive Micromachined Ultrasonic Transducer</td>
</tr>
<tr>
<td>CNR</td>
<td>Contrast to Noise Ratio</td>
</tr>
<tr>
<td>CR</td>
<td>Cystic Resolution</td>
</tr>
<tr>
<td>DAS</td>
<td>Delay-and-Sum</td>
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<tr>
<td>DRF</td>
<td>Dynamic Receive Focusing</td>
</tr>
<tr>
<td>FOV</td>
<td>Field of View</td>
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<tr>
<td>FWHM</td>
<td>Full Width at Half Maximum</td>
</tr>
<tr>
<td>IC</td>
<td>Integrated Circuit</td>
</tr>
<tr>
<td>IQ</td>
<td>In-phase and Quadrature-phase (A phase shift of: 0° and 90°.)</td>
</tr>
<tr>
<td>LSF</td>
<td>Line Spread Function</td>
</tr>
<tr>
<td>PB</td>
<td>Parallel Beamforming</td>
</tr>
<tr>
<td>PRF</td>
<td>Pulse Repetition Frequency</td>
</tr>
<tr>
<td>PSF</td>
<td>Point Spread Function</td>
</tr>
<tr>
<td>PZT</td>
<td>Piezoelectric Lead Zirconate Titanate Compound</td>
</tr>
<tr>
<td>RC</td>
<td>Row-column</td>
</tr>
<tr>
<td>RCA</td>
<td>Row-Column Addressed</td>
</tr>
<tr>
<td>RF</td>
<td>Radio Frequency</td>
</tr>
<tr>
<td>RI</td>
<td>Relative Intensity</td>
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<tr>
<td>RMS</td>
<td>Root Mean Square</td>
</tr>
<tr>
<td>SA</td>
<td>Synthetic Aperture</td>
</tr>
<tr>
<td>SAI</td>
<td>Synthetic Aperture Imaging</td>
</tr>
<tr>
<td>SARUS</td>
<td>Synthetic Aperture Real-time Ultrasound System</td>
</tr>
<tr>
<td>SASB</td>
<td>Synthetic Aperture Sequential Beamforming</td>
</tr>
<tr>
<td>SAR</td>
<td>Synthetic Aperture Radar</td>
</tr>
<tr>
<td>SMF</td>
<td>Spatial Matched Filter</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal to Noise Ratio</td>
</tr>
<tr>
<td>ToF</td>
<td>Time of Flight</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Full Form</td>
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<td>-------------</td>
<td>-----------------------------------------------</td>
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<tr>
<td>US</td>
<td>Ultrasound</td>
</tr>
<tr>
<td>VS</td>
<td>Virtual Source</td>
</tr>
<tr>
<td>SLURM</td>
<td>Simple Linux Utility for Resource Management</td>
</tr>
<tr>
<td>LSF</td>
<td>Load Sharing Facility</td>
</tr>
<tr>
<td>HPC</td>
<td>High Performance Computing</td>
</tr>
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</table>
CHAPTER 1

Introduction

Ultrasound refers to mechanical vibrations of particles that propagate through the medium with frequencies higher than 20 KHz. Usually, imaging is any representation or reproduction of an object’s form or layout. The term ultrasound imaging covers the area of remote sensing, where mechanical vibrations with known parameters are generated, sent through a medium, and consecutively recorded. The changes of the parameters introduced during the propagation are used to represent or characterize the medium. Medical ultrasound imaging is one of the most used medical imaging techniques for diagnosis, inspection, and guidance during surgery in many different application areas such as cardiac imaging, blood flow imaging, abdominal imaging, etc. It can image soft tissue in real-time, providing radiologists with dynamic view of the anatomy by noninvasive means and with no known side-effects. Ultrasound imaging is inexpensive and does not require special facilities compared with most of the other imaging modalities, such as X-ray, PET, SPECT, CT, and MRI. At the same time, most ultrasound scanners are portable and can run on batteries, which is useful in emergency situations.

Pulse-echo ultrasound imaging was first introduced in the late 1940’s (Wild 1950) and emerged from ideas developed in radar and sonar. In this way, an ultrasound pulse is sent into the tissue in a given direction. As the pulse propagates through the tissue, it passes through some inhomogeneities or reflective surfaces. Each inhomogeneity, depending on its size, causes part of the pulse energy to be scattered or reflected back to the transducer. The transducer receives these echoes and converts them into electrical voltage signals. Since the inhomogeneities are related to the transition between different tissues, a map of the tissue layout can be made from the received signal. In the beginning, only a single acoustic scan line was measured and presented to the operator as a function of depth. This is known as the A-mode, or amplitude mode. When the first ultrasound imaging scanners were developed, the probe consisted of a physically concave transducer with a single fixed focus (Ebina et al. 1967; Edler and Lindström 2004). The A-mode lines were stacked next to each other to create the B-mode or brightness mode image, by manually sliding the probe above the object of interest. Later the transducers were placed on rotating motors to automatize the process (Griffith and Henry 1974). B-mode imaging is a central principle in medical ultrasound and therefore it is the standard option in all today’s ultrasound scanners.

Ultimately, such convex transducers were replaced by an array of smaller piezoelectric transducers. Unlike before, the focus depth could now be controlled electronically by applying a set of individual delays to each element of the transducer array. On transmit,
the focal depth could be selected to improve the image quality around a preferred range. On reception, a series of delay curves could be applied such that the pulse propagation was followed at all depths, introducing a dynamic focusing capability, also known as Dynamic Receive Focusing (DRF). In addition, the set of delays applied to the transducer elements could be adjusted to steer the ultrasound beams in any direction, removing the need for motors in 2-D imaging (Somer 1968). In this way, in order to build a B-mode image, the desired field of view is first divided into a given number of directions known as image lines. Each of these image lines are then filled up by emitting an ultrasound pulse focused along this direction and by listening to the echoes propagating back to the surface of the transducer from any reflector located along the image line direction. Ultimately, in a clinical use, the B-mode image displays a slice of the anatomy below the transducer array.

Compared with conventional 2-D imaging, real-time 3-D (4-D) ultrasound imaging has several advantages, resulting in a significant progress in the ultrasound imaging instrumentation over the past decade. In 2-D imaging the location and orientation of B-mode images are determined by the position of the probe over the patient, making some of the interesting views inaccessible. However, with 3-D imaging, any 2-D view angle is available from the acquired volume data for diagnostic purposes. Viewing the patient’s anatomy as a volume helps physicians to comprehend the important diagnostic information in a noninvasive manner. This can be equally valuable for therapeutic procedures requiring follow-up studies that depend on the ability to acquire a view that has the same orientation and position as a previous examination. Furthermore, diagnostic and therapeutic decisions often require accurate estimates of e.g., organ, cyst, or tumor volumes. 3-D ultrasound imaging can provide these measurements without relying on the geometrical assumptions and operator-dependent skills involved in such estimations using 2-D scans. Although the detail resolution of ultrasound can not compete with 3-D imaging modalities such as CT and MRI, the combination of patient safety by using nonionizing radiation, cost-effectiveness, portability, and real-time imaging ability makes ultrasound the preferred choice in many clinical applications.

To understand the challenges involved in 3-D and 4-D ultrasound imaging we have to look back a few decades. In the 1980’s, a lot of research was carried out on improving the acquisition and visualization as well as demonstrating the clinical utilities of 3-D ultrasound. Free-hand scanning using 1-D transducer arrays was relatively a low-cost method to acquire volume data. In that way, a sensing system tracked the movement of the transducer, and all of the 2-D images, each with the corresponding transducer position and angle information. Afterwards, they were post-processed to produce a volumetric image of the scanned region. While this approach offers a flexible solution, image quality suffers from position inaccuracies and irregular scanning due to the hand-held manipulation, especially when imaging small structures. A method to better control and track the position of the 1-D transducer array is to mechanically move it in a precise and predefined way. That was the principle of mechanical probes, in which the 1-D transducer array is mounted on a mechanical stage inside the probe. A system of servo or stepper
motors is then used to control and track the movement of the transducer array. A spatially sampled volume could be acquired either by translating, angular fanning, or spinning a 1-D transducer array about an axis. Due to the larger accelerations involved in changing the direction of the array, the tracking error naturally increases with the displacement frequency, and therefore such probes are usually limited to producing images with frame rates below 10 Hz (Roh 2014). The technical issues that arose concerning the better performance of these transducers were a higher volume rate, larger field of view, and better long-term reliability and compliance.

In all of the previous methods, 3-D information is acquired by physically moving the 1-D transducer. However, 3-D information can also be acquired by steering the ultrasound beam electronically in both the azimuth and elevation directions using 2-D matrix transducer arrays. Such arrays were first introduced in the early 1990’s by researchers at Duke University (Smith et al. 1991; von Ramm et al. 1991). Since no moving parts are involved, they can achieve real-time 3-D images at a frame rate of more than approximately 20 Hz. High volume rate 3-D ultrasound imaging can be considerably beneficial for echocardiography, enabling detailed anatomical assessment of cardiac pathology. In principle, the concept is simple, but in practice it presents a formidable technical challenge. To achieve high quality images and steering to a desirable angle, 2-D matrix array must be designed with a pitch in the range of half of the wavelength of operating frequency. It means that for a 3 MHz transducer operating in water, a typical pitch of 250 µm is required. Such a small pitch forces the elements to become smaller in a 2-D matrix, which in fact results in lowered capacitance and hence an increased electrical impedance mismatch between the element and the cable connecting it to the ultrasound scanner. This calls for preamplifiers and matching circuits in the probe handle (Karadayi et al. 2009). On the other hand, to obtain the same spatial resolution as in 2-D imaging, the number of elements along each lateral dimension must be equal to that of a 1-D array. Yet, even a small 1-D array with 128 elements would translate into $128 \times 128 = 16,384$ elements in a 2-D matrix array. From a transducer fabrication viewpoint, the combination of small pitch and large number of elements generates few construction difficulties, particularly for interconnections and ground electrode distribution to each element. Nonetheless, the sheer number of wires results in an impractically large cable from the transducer to the scanner.

The issue of reducing channel count whilst maintaining the size of the array aperture was in the earlier versions of 2-D matrix arrays addressed by introducing sparse arrays, in which only a subset of the elements are active at the same time. Amongst these are Vernier arrays, random arrays, and Mills cross arrays, presenting each their benefits and drawbacks (Davidsen et al. 1994; Brunke and Lockwood 1997; Yen, Steinberg, et al. 2000; Austeng and Holm 2002; Karaman, Wygant, et al. 2009). However, in all of them focusing in transmit and receive was compromised, which resulted in reduced signal-to-noise ratio (SNR) and introduced higher sidelobes and/or grating lobes (Turnbull and Foster 1991). More recently, fully sampled arrays with reduced channel count have become available by splitting the beamforming in two stages through the so called µ-beamforming: Fine delays and summation between elements in close proximity are
carried out by electronics placed inside the transducer probe, and much fewer signals are then funnelled out to the ultrasound scanner, performing the final beamforming. An example of such a state-of-the-art fully sampled matrix transducer is the X6-1 PureWave xMATRIX Array from Phillips (Eindhoven, Netherlands), with 9,212 elements (Phillips 2015).

Aside from the practical challenges involved in the fabrication of 2-D matrix arrays, the beamforming process presents its own issues. In conventional ultrasound pulse-echo B-mode imaging, the maximum frame rate (FR) achievable is ultimately limited by the speed of sound in tissue. For 1-D imaging techniques, such as A-mode, the temporal sampling is determined by the maximum range being imaged. When extended to 2-D imaging, the field of view, or number of image lines acquired, must also be factored into the FR. For traditional pulse-echo B-mode imaging, the maximum achievable FR, and thus the maximum temporal sampling rate, is the inverse of the product of the time of flight for one transmit-receive operation and the total number of transmit-receive operations to generate one image. In conventional ultrasound imaging, FR can only be increased by decreasing the resolution, the range, or field of view of the image.

However, the conventional ultrasound imaging paradigm had to be adapted with the arrival of 3-D imaging. Conventional ultrasound imaging is a tedious process, as it requires to wait for the propagation of the ultrasound pulse back and forth in the body for each single image line. Scanning a 3-D volume requires squaring the number of image lines, and hence it imposes a quadratic reduction on the achievable frame rate. Indeed, considering the speed of sound in biological tissues to be around 1540 m/s, about 200 µs are required to acquire a single image line down to 15 cm depth. This is approximately 5000 lines per second which may be used to form either 50 volumes per second using 10×10 lines to cover a reasonably large sector, or 1 volume every 2 seconds using 100×100 image lines. In terms of volume rate, for 3-D cardiac imaging, this is very far from what can be achieved with a conventional approach. The scan rates of 25 to 50 Hz are adequate for many cardiac, anatomical and functional, diagnoses but are inadequate for studies of electromechanical coupling events in the heart. Electrical activity as measured by the ECG should be sampled at rates of 500 Hz or greater for diagnostic purposes (Kligfield et al. 2007). To study the interaction of electrical and contractile events with comparable temporal resolution imaging at 500 Hz would be required. Moreover, new clinical methods based on strain estimations also require very high frame rates (Yoshiara et al. 2007). Volumetric images of a beating heart can also be acquired consecutively through synchronization to ECG signals, which relies on the assumption of a repetitive heart beat pattern through a longer acquisition time.

A problem for conventional 2-D or 3-D imaging is that it only has one fixed transmit focal depth, compared to in receive where dynamic focusing can be performed. This reduces the image quality at other depths than the focal depth. It is always desirable to increase the image quality, which makes the diagnosing process for the medical doctors easier. Potentially, it could also enable the medical doctors to detect more diseases at an earlier state and reduce the cost of each investigation if less scans had to be completed on
the same patient.

A method that enables dynamic focusing also for the transmit beams, is the synthetic (transmit) aperture imaging (SAI) technique, originally developed for radar systems (Jensen et al. 2006). By coherently compounding, i.e., summing in phase, the data received from successive and spatially overlapping ultrasound pulse emissions, one may retrospectively recreate a transmit focus along each line of the final image. The quality of the transmit focus depends on several factors such as the number of overlapping transmissions and the lateral spatial bandwidth in k-space of the transmitted sound fields (Walker and Trahey 1998; Nikolov et al. 2010).

Several versions of SAI exist (Nock and Trahey 1992; Ylitalo and Ermert 1994; Karaman and O’Donnell 1998; Gammelmark and Jensen 2003). The original SAI used a single element excitation, but is now applied using virtual point sources generated with subset of the elements to increase the energy and therefore increase the SNR (Jensen et al. 2006). SAI can be used to recreate the transmit focus using conventional B-mode focused transmissions, in regions before and after the focal depth where they are overlapping (R. Zemp and Insana 2007). Another application of SAI is by transmitting plane waves with linear arrays or diverging waves with phased arrays in different directions to recreate the transmit focus (Tanter et al. 2002; Montaldo et al. 2009; Hasegawa and Kanai 2011; Tong, Gao, et al. 2012). Although, due to energy dispersion, generations of second harmonics are prevented (Hasegawa and Kanai 2012). However, both are suitable for very high frame rate imaging. Ideally, SAI with diverging waves would achieve full dynamic focusing in transmit with a frame rate high enough to be used for cardiac imaging. The gain in resolution and frame rate comes with the cost of some compromises such as side-lobes and penetration depth.

Using multiple transmit beams, either at the same time or in quick succession can also increase the frame rate (Mallart and Fink 1992). While this method provides an additional increase in FR, cross-talk between the simultaneous beams may lead to increased noise and potential artifacts in the resulting image. Recent work has described methods for reducing cross-talk between beams by various methods, including spatial separation (Mallart and Fink 1992), spectral separation or frequency multiplexing (Demi, Viti, et al. 2013; Demi, Ramalli, et al. 2015), and various apodization schemes (Tong, Gao, et al. 2012; Tong, Ramalli, et al. 2014). Cross-talk between beams may exist depending upon the shape (apodization) and separation of the beam. Such cross-talk may lead to image artifacts, such as bright targets appearing in multiple location in the image.

Another approach to increase the frame rate is by simultaneously beamforming a plurality of receive beams covering a small sub-volume around the broadened transmit beam. This principle was introduced in the late 1970’s by combining the output of receive electronic boards working in parallel in the scanner, and was designed as a way to achieve frame rates required in 3-D cardiac imaging (Delannoy et al. 1979; Shattuck et al. 1984; von Ramm et al. 1991; Thomenius 1996). The misalignment of the transmit and receive directions degrades the lateral shift invariance and produces some block-shaped artifacts (Hergum et al. 2007). Using SAI for two adjacent transmit lines with 50%
overlapped parallel received lines can reduce these artifacts. The topic of the first part of this thesis is to investigate if synthetic aperture 3-D imaging could increase the imaging quality compared to 3-D parallel beamforming and still achieve a frame rate suitable for cardiac imaging.

Despite the recent advances in real-time 3-D ultrasound imaging, the ultrasound systems supporting such imaging modalities are highly advanced and rely on cutting edge software, hardware, and manufacturing technology. This results in expensive equipment, impairing the low-cost advantage of ultrasound and thus limiting its more widespread use. Moreover, due to the constraints on transducer probe heating dictated by the standards for medical equipment, the thermal budget is becoming a consideration for modern probes with integrated electronics (Sampson et al. 2013).

Recently, an alternative to matrix arrays has been suggested. These are so-called row-column addressed 2-D arrays, which were first proposed in 2003 by Morton and Lockwood (Morton and Lockwood 2003). Here, the 2-D array is addressed via its row and column indices, effectively producing two 1-D arrays oriented orthogonal to one another. The number of connections needed to address an $N \times N$ array thereby becomes $2N$, as opposed to the $N^2$ connections required in a fully addressed matrix array. For example, the number of connections for a $128 \times 128$ array is consequently reduced from 16,384 to 256. This significant reduction decreases the complexity of 2-D arrays for real-time 3-D imaging considerably.

Realizations of row-column addressed arrays have previously been presented by several groups. The first experimental demonstration of row-column addressed arrays were presented in 2006 by Seo and Yen (Seo and Yen 2006). The array was a PZT piezoelectric in a 64+64 layout, fabricated using a 1-3 ceramic with the row and column electrodes defined on separate sides of the ceramic. This array was later surpassed by the same authors with a 256+256 array using the same fabrication technique (Seo and Yen 2007, 2008, 2009). In 2009 Yen et al. introduced a simplified process for fabrication of row-column addressed piezoelectric arrays using a dual layer structure (Yen, Seo, et al. 2009). Row-column arrays based on capacitive micromachined ultrasonic transducers (CMUT) technology were first presented in 2009 by Logan et al. (Logan, Wong, and Yeow 2009). They showed a 32+32 array fabricated using the wafer bonding process with a silicon nitride plate, and later they presented characterization of a similar array (Logan, Wong, Chen, et al. 2011). Zemp et al. (R. J. Zemp et al. 2011) and Sampaleanu et al. (Sampaleanu et al. 2014) presented row-column addressed arrays fabricated using the sacrificial release process and performed feasibility studies. More recently they have presented photoacoustic imaging using row-column addressed CMUT arrays (Chee et al. 2014).

Although, row-column addressed transducer arrays have only been sparsely investigated in the literature, however, further research is needed to assess the possibilities and drawbacks associated with transducer arrays using this addressing scheme. The topic of the second part of this thesis is to investigate the performance of synthetic aperture 3-D imaging with row-column addressed 2-D arrays and the possibility to increase the
inherent limited imaging field of view of these arrays from a rectilinear volume region to a curvilinear one using a double curved diverging lens.

1.1 Publications in the thesis

This thesis is based on the following publications. The publication letter refers to the appendix name containing the corresponding paper.

3-D Phased Array Synthetic Aperture Imaging

3-D Synthetic Aperture Imaging (SAI) was in one conference paper and in a draft journal paper, compared with the Parallel Beamforming (PB) technique.

Paper A
“In Vivo Real Time Volumetric Synthetic Aperture Ultrasound Imaging”.

Paper B
“Real-Time 3-D Synthetic Aperture Imaging”.

3-D Imaging with Row-Column Addressed 2-D Arrays

The pros and cons of row-column addressing 2-D ultrasound transducer arrays was described in three conference papers and three journal papers.

Paper C
“Volumetric ultrasound imaging with Row-column addressed 2-D arrays using spatial matched filter beamforming”.

Paper D
“Volumetric Synthetic Aperture Imaging with a Piezoelectric 2-D Row-Column Probe”. 
Paper E

Paper F

Paper G

Paper H

Patents on Imaging with Row-column 2-D Arrays
Two patents were taken on imaging with a flat as well as a double-curved row-column addressed 2-D arrays.

Patent A
H. Bouzari, T. L. Christiansen, S. Holbek, M. B. Stuart, E. V. Thomsen, and J. A. Jensen
“Row-column Addressed 2-D Array with a Double Curved Surface”. Filed on June 8, 2016, Number PCT/IB2016/053367.

Patent B
H. Bouzari, S. Holbek, M. Engholm, J. Jensen, M. B. Stuart, E. V. Thomsen, and J. A. Jensen
“3-D Imaging and/or Flow Estimation with a Row-Column Addressed 2-D Transducer
1.2 Publications not included in the thesis

Three co-authored conference papers and one co-authored journal paper are not included in this thesis.

External Paper I
The inherent nonlinear behavior of the CMUT, poses an issue for harmonic imaging as it is difficult to dissociate the harmonics generated in the tissue from the harmonic content of the transmitted signal. The generation of intrinsic harmonics by the CMUT can be minimized by decreasing the excitation signal. This, however, leads to lower fundamental pressure which limits the desired generation of harmonics in the medium. The harmonic to fundamental ratio of the surface pressures declines for decreasing excitation voltage and increasing bias voltage.


External Paper II
The paper presents the fabrication and assembly process to design a 3 MHz, $\lambda/2$-pitch $62+62$ channel row-column addressed 2-D CMUT array, which is mounted in a probe handle and connected to a commercial BK Medical scanner for real-time volumetric imaging.


External Paper III
This journal paper presented 3-D vector flow images obtained using the 3-D Transverse Oscillation (TO) method. The method employed a 2-D transducer and estimated the three velocity components simultaneously, which is important for visualizing complex flow patterns. Data were acquired using the experimental ultrasound scanner SARUS on a flow rig system with steady flow. It was shown that the ultrasound method is suitable for real-time data acquisition as opposed to magnetic resonance imaging (MRI). The results demonstrated that the 3-D TO method is capable of performing 3-D vector flow imaging.


**External Paper IV**
The paper presents a simple approach to reduce the parasitic capacitance is presented, which is based on depleting the semiconductor substrate. To reduce the parasitic capacitance by 80% the bulk doping concentration should be at most $10^{12} \text{cm}^{-3}$. Experimental results show that the parasitic capacitance can be reduced by 87% by applying a substrate potential of 6 V relative to the bottom electrodes. The depletion of the semiconductor substrate can be sustained for at least 10 minutes making it applicable for row–column-addressed CMUT arrays for ultrasonic imaging. Theoretically the reduced parasitic capacitance indicates that the receive sensitivity of the bottom elements can be increased by a factor of 2.1.


### 1.3 Other Contributions

Besides the papers published, several software projects were developed at the center for fast ultrasound imaging (CFU).

A collection of MATLAB and bash scripts that solves problems when working with high performance computing (HPC) clusters were programmed. The scripts are included in a larger collection called the “CFUtools”. Excluding all external scripts, CFUtools consists of more than 9000 lines of code that has to be maintained and debugged. The cluster scripts can be used for submitting, monitoring, and collecting the results of jobs over SLURM and LSF platforms through MATLAB command window.

A collection of scripts and \LaTeX{} files were gathered and maintained as “CFU \LaTeX{}”. The scripts define poster, pre-print, report and thesis layouts.

A delay-and-sum beamformer that can handle double-curved row-column addressed 2-D arrays was programmed.

### 1.4 Outline

The outline of the remaining part of this thesis is as follows. The next chapter contains a resume of the results achieved with 3-D synthetic aperture imaging in comparison with parallel beamforming using a fully addressed 2-D array. In Chapter 3 through Chapter 4, the research conducted on the subject of row-column addressing of 2-D arrays is presented. Two prototyped probes with row-column addressed 2-D transducer arrays based on PZT and CMUT technology were characterized from electro-acoustic and imaging perspectives. In Chapter 5, imaging with a double curved row-column addressed 2-D array is presented.
in general and more specifically, by using a diverging acoustic lens over a flat row-column addressed array, curvilinear imaging has been evaluated. This is followed by a conclusion and perspectives in Chapter 6.

The main part of this thesis consists of the papers published on the subject of 3-D synthetic aperture imaging and row-column addressing of 2-D arrays. The papers are included as appendices named Paper A through Paper H. At the very end are included two patent applications related to imaging with row-column addressed 2-D arrays.

References


References


Two major obstacles that have delayed the implementation of real-time 3-D imaging systems are the low frame rate often achievable when scanning a full volume as well as the large amount of channels on a 2-D array transducer to scan the volume.

Parallel beamforming (PB) was introduced to address the first issue. Compared with the conventional line by line imaging, PB could increase the frame rate by simultaneously beamforming a plurality of receive beams around the broadened transmit beam. According to Fraunhofer approximation for a rectangular aperture, the beam width in the focal plane is equal to $\lambda f_n$, where $f_n$ is the $f$-number in transmission or reception, and $\lambda$ is the wavelength. Using a broader transmit beam in PB requires to increase the transmit $f_t'$ compared with the receive $f_r'$. However, the loss in two-way lateral resolution due to broadening of the transmit beam, can not be recovered. In addition, the misalignment of transmit and receive directions introduces degradations in the lateral shift invariance, and thus results in block-like artifacts. To remove those block-like artifacts and using a lower transmit $f_t'$ a technique called synthetic transmit beamformation (STB) has been proposed to synthesize a transmit beam along each receive line by spatially interpolating the overlapped adjacent transmit beams (Hergum et al. 2007; Bjastad et al. 2009). Although it is an effective method to remove the block-like artifacts, it can not control the focal point of the synthesized transmit beam. However, by using synthetic aperture imaging (SAI), the received data can be delayed and summed to recover the focusing in every location in the image.

One of the goals of this project was to investigate if the results achieved in 2-D SAI could be extended to 3-D imaging and validate it based on a comparison with PB method. The analyses as well as in vivo results that are presented in the Paper A (included on page 129) performed a comparison study by limiting the number of active channels for both methods to only 256.

In this chapter, the extended research carried out on 3-D SAI is presented without imposing any limitation on the number of active channels being used. To perform a fair comparison the volume rates of both methods were set equal, however the 3-D SAI results, which have volume rates up to 50 times higher are presented as well. The setup and all of the results presented in this chapter have not been submitted before and should be considered as a draft paper. The draft Paper B is included on page 141.

The outline of the remaining part of this chapter is as follows. First, the pulse-echo imaging principle and its limitations when translating to 3-D imaging are discussed. This is followed by defining a set of imaging quality assessment measures and also an
introduction to the hardware being used. Then, after introducing the imaging application requirement, the basics of SAI and PB techniques as well as the analyses on choosing their parameters are given. Afterwards, the results of the simulations and the measurements on phantoms as well as on in vivo are presented. The final section concludes the chapter with a discussion.

2.1 Principle of Pulse-Echo Imaging and Its Limitations

To evaluate an imaging technique, first one has to understand the technological limitations and physical boundaries involved. In principle, the sensitivity in a pulse-echo imaging system is directly related to the characteristics of the emitted pulses. Thereby, the axial resolution is proportional to the pulse bandwidth (a larger bandwidth means a shorter pulse and thus a higher axial resolution). However, a higher bandwidth increases the thermal noise, which on the other hand lowers the sensitivity. At the same time, the penetration depth is also related to the pulse center frequency (a higher frequency means higher absorption and less reflection as well as higher attenuation). Thus, the sensitivity might be improved by maximizing the acoustic output by transmitting a more powerful pulse. Yet, there are technical and biological limitations on the amount of transmitted energy, which must be considered.

Using a powerful signal generator to drive a big amount of energy through the transducer may lead to over-heating of the probe surface. At the same time, any damage to the tissues caused by cavitational effects or over-heating has to be avoided. In practice, the acoustic output is adjusted such that both the peak and the temporal average intensities remain under given thresholds. Such safety guides are regulated by the United States Food and Drug Administration (FDA) (FDA 2008), and take the form of upper limits on given indexes: the mechanical index \((MI \leq 1.9)\), the derated spatial-peak-temporal-average intensity \((I_{spta} \leq 720\text{mW/cm}^2\) for peripheral vessel, \(I_{spta} \leq 430\text{mW/cm}^2\) for cardiac), and the derated spatial-peak-pulse-average intensity \((I_{sppa} \leq 190\text{mW/cm}^2\)). Increasing the transmitted energy by using a longer pulse results in a poorer axial resolution. Alternatively, it has been proposed to use linear frequency modulation (FM) excitations combined with match filtering on reception, to increase the energy level without sacrificing the axial resolution. Unfortunately, longer excitations increase the probe heating and may burn the transducer.

Unlike in conventional imaging, the lateral resolution in SAI is directly related to the steering angle span of the in-phase summed emissions, but a larger steering span increases the level of side-lobes. Using an apodization in transmit and receive may decrease the side-lobe level but at the same time degrades the spatial resolution. In terms of SNR, using SAI technique by summing \(n\) emissions in phase, noise will be suppressed and thereby the SNR will be improved by a factor of \(\sqrt{n}\) (Karaman et al. 1995; Oddershede and Jensen 2007).
2.2 Imaging Quality Assessment Measures

The imaging performance is computed using four measures:

2.2.1 Signal-to-noise ratio (SNR)

The SNR is the measure to distinguish soft tissue from electronic noise and is calculated from a number of B-mode images measured on a tissue mimicking phantom. The average of \( N \) B-mode images, and its difference to one of the B-mode images, are computed to yield the signal and electronic noise. The SNR is calculated by:

\[
\text{SNR}(x) = \frac{\left| \frac{1}{N} \sum_{n=1}^{N} s_n(x) \right|^2}{\left| \frac{1}{N} \sum_{m=1}^{N} \left( s_m(x) - \frac{1}{N} \sum_{n=1}^{N} s_n(x) \right) \right|^2},
\]

where \( x = (x,y,z) \) is the voxel coordinate, and \( s_n \) a single IQ-beamformed image frame with index \( n \). The point where SNR falls below 0 dB is the penetration depth.

2.2.2 Spatial resolution

The spatial resolution is calculated as the FWHM of the imaging system’s PSF.

2.2.3 Cystic resolution

The CR is the ability to detect an anechoic cyst in a uniform scattering medium (Vilkomerson et al. 1995; Ranganathan and Walker 2007; Guenther and Walker 2009). The relative intensity (RI) of the anechoic cyst was shown by Ranganathan and Walker (Ranganathan and Walker 2007), to be quantized as the clutter energy to total energy ratio,

\[
\text{RI}(R) = \sqrt{\frac{E_{\text{out}}(R)}{E_{\text{tot}}}} = \sqrt{1 - \frac{E_{\text{in}}(R)}{E_{\text{tot}}}},
\]

where \( E_{\text{in}} \) is the signal energy inside a circular region with radius, \( R \), centered on the peak of the point spread function, \( E_{\text{tot}} \) is the total PSF energy, and \( E_{\text{out}} \) is the PSF energy outside the circular region. The RI\((R)\) curve can be compressed to a single number by sampling the curve at e.g. 20 dB. The result is the required cyst radius at which the intensity at the center of the cyst is 20 dB lower than its surroundings, written as \( R_{20\text{dB}} \).

2.2.4 Contrast resolution

The contrast resolution in B-mode images, \( i.e. \), contrast-to-noise ratio (CNR) is defined as
\[ \text{CNR} = \frac{\left| \mu_{\text{bck}} - \mu_{\text{cyst}} \right|}{\sqrt{\sigma^2_{\text{bck}} + \sigma^2_{\text{cyst}}}}, \]  \hspace{1cm} (2.3)

where \( \sigma^2_{\text{bck}} \) and \( \sigma^2_{\text{cyst}} \) are variances, and \( \mu_{\text{bck}} \) and \( \mu_{\text{cyst}} \) are mean values of gray levels within the background and lesion, respectively.

### 2.3 Hardware

All measurements are carried out using a fully wired 32\( \times \)32 PZT matrix transducer probe connected to the 1024 channel research ultrasound scanner, SARUS:

##### 2.3.1 2-D probe

The 2-D probe used in both simulations and in the measurements is seen in Fig. 2.1. The probe consists of a fully wired 32\( \times \)32 PZT matrix transducer and is produced by Vermon S.A., Tours, France. The averaged pulse-echo impulse response of the transducer is shown in Fig. 2.2(a) on the next page. The spectrum of the averaged pulse-echo impulse response is shown in Fig. 2.2(b) on the facing page. The center frequency of the Vermon probe is 3.8 MHz and the pitch is 300 \( \mu \)m, corresponding to 0.74 \( \lambda \). To avoid grating lobes within the \( \pm 45^\circ \) beamformed volume, the pitch of the transducer array should not be larger than \( \lambda / 2 \). As a compromise between the transducer efficiency in converting electrical to mechanical energy, and grating-lobe levels, the center frequency of the emission is set to 3.0 MHz, corresponding to 0.58 \( \lambda \) pitch. The orientation of the elements are shown in Fig. 2.3 on the next page.

![Figure 2.1: The 32\( \times \)32 element phased array ultrasound probe used for the measurements and modeled in the simulations. The probe is produced by Vermon S.A. (Tours, France).](image-url)
2.3. Hardware

(a) Averaged impulse response of the Vermon 32 × 32 element phased array ultrasound probe.

(b) Averaged impulse response spectra of the Vermon 32 × 32 element phased array ultrasound probe.

Figure 2.2: Averaged impulse response and its spectrum of the Vermon 32 × 32 element phased array ultrasound probe. The center frequency of the probe is 3.8 MHz, which was calculated as a weighted mean of the spectra. The center frequency and the −6 dB fractional bandwidth are indicated in the plot.

Figure 2.3: The orientation of the transducer elements on the Vermon 300 µm-pitch 32 × 32 element phased array ultrasound probe.
2.3.2 SARUS
The volumetric data were acquired using the 1024 channel experimental ultrasound scanner, SARUS seen in Fig. 2.4 (Jensen, Holten-Lund, et al. 2013). It can sample RF data with a sampling frequency of 70 MHz with a precision of 12 bits.

![Figure 2.4: SARUS, the 1024 channel experimental ultrasound scanner used for all of the measurements.](image)

2.4 Application Requirements
The two imaging techniques are designed for cardiac imaging, which requires imaging down to 15 cm and a frame rate $f_r$ of at least 20 Hz. To be comparable with products from the medical ultrasound industry, a volume scan spanning 90° in both the azimuth and elevation direction is chosen, i.e., a field of view of $90° \times 90°$. With a maximum scan depth $r_{\text{max}}$ of 15 cm and a speed of sound $c$ equal to approximately 1540 m/s, the maximum pulse repetition frequency is

$$f_{prf} = \frac{c}{2r_{\text{max}}} = 5.13 \text{kHz} \ . \quad (2.4)$$

The possible number of emissions per frame then becomes

$$N_{\text{ems}} = \frac{5.13 \text{kHz}}{20 \text{Hz}} \approx 256 . \quad (2.5)$$

Using 256 emissions per frame allows for resolving the azimuth and elevation directions with $\sqrt{256} = 16$ emissions each.
2.5 Synthetic Aperture Imaging

When using synthetic transmit focusing, by taking advantage of superposition theorem, a virtual transmit aperture is synthesized for every location by delaying and summing a plurality of datasets acquired from successive transmissions. In other words, one virtual element is synthesized in the synthetic aperture for each transmission. The location of the virtual elements influences the distribution of the emitted energy and thereby the signal-to-noise ratio (SNR) within the imaged volume.

On a phased array, the synthetic aperture can be synthesized by applying beam steering to the transmissions as it is seen in Fig. 2.5. Placing the virtual sources in front of the array compromises the overlapping between transmit beams. To have a higher overlap between the transmit beams for the same number of emissions, the transmit \( f^t_0 \) should be low as is seen in Fig. 2.6 on the following page. However, even by lowering the transmit \( f^t_0 \), near the focal zone no overlapping occurs.

![Virtual sources](image)

**Figure 2.5:** Synthetic aperture with beam steering and no translation. The virtual source is located in front of the aperture. \( D \) is the active aperture, and \( F \) denotes the focal point distance of the middle emission to the the center of the active aperture.

Placing the virtual sources behind the transducer, as it is seen in Fig. 2.7 on the next page, can increase the overlapping between transmit beams. Increasing the steering angle of the defocused transmit beams increases the overlapped region and thereby the synthesized aperture becomes larger. Since the relations between aperture array design and the PSF also apply to the synthesized aperture array (Frazier and O’Brien 1998), to lower the side lobe levels and also to avoid grating lobes, the width of the synthesized array (synthesized apodization) and the pitch of the virtual elements must be considered. In the configuration shown in Fig. 2.7 on the following page, due to a smaller synthesized aperture, the achievable lateral resolution is worse or equal to the lateral resolution achievable by the physical aperture at the focal point. However, the advantage of SAI compared with the conventional imaging is that the transmit focus is not fixed and can be
Figure 2.6: Synthetic aperture with beam steering and no translation. The virtual source is located in front of and near to the aperture. $D$ is the active aperture, and $F$ denotes the focal point distance of the middle emission to the center of the active aperture.

maintained dynamically throughout the image. On a phased array with a limited number of active elements, the synthetic transmit aperture can be synthesized by both translating the transmit aperture as well as applying beam steering as it can be seen in Fig. 2.8 on the next page. To enable the translation of the active aperture, it needs to be small in order to fit on the physical array and still leave room for translation.

Figure 2.7: Synthetic aperture with beam steering and no translation. The virtual source is located behind the aperture. $D$ is the active aperture, and $F$ denotes the focal point distance of the middle emission to the center of the active aperture.

The placement of the virtual sources affects the imaging resolution achievable. The
best achievable lateral resolution for a given ultrasound system is defined by its two-way beam width at the focal depth using conventional focusing on both reception and transmission (Szabo 2014). For a rectangular aperture, using the Fraunhofer approximation, the beam width in the focal plane is equal to \( \lambda f_\# \), where \( f_\# \) is the \( f \)-number in transmission \( (f_\#^\text{t}) \) or reception \( (f_\#^\text{r}) \), and \( \lambda \) is the wavelength determined by the pulse center frequency. According to Fraunhofer approximation for a rectangular aperture, the transmit beam width can be written as

\[
\text{Transmit beam width} = \lambda f_\#^\text{t} = \frac{\lambda F}{a_t}, \tag{2.6}
\]

where \( a_t \) is the transmit (synthesized) aperture size, and \( F \) is the transmit focus depth. The receive beam width can be written similarly

\[
\text{Receive beam width} = \lambda f_\#^\text{r} = \frac{\lambda F}{a_r}. \tag{2.7}
\]

The lateral two-way beam width can be found using the convolution of the transmit and receive beam profiles, which is equal to

\[
\text{Two-way beam width} = \lambda f_\#^{\text{t}r} = \frac{\lambda F}{a_t + a_r} = \frac{\lambda f_\#^\text{t} \cdot f_\#^\text{r}}{f_\#^\text{t} + f_\#^\text{r}}, \tag{2.8}
\]
and therefore we also have

\[ f_{tr}^f = \left( \frac{1}{f_{tr}^t} + \frac{1}{f_{tr}^r} \right)^{-1}. \]  

(2.9)

As a side note, (2.9) states that when the transmit focal distance is approaching infinity \((f_{tr}^t \to \infty)\) in plane wave imaging, \(f_{tr}^f \approx f_{tr}^r\). Thereby, the field of view becomes limited and only focusing in receive is possible. However, still due to the in phase summation of the low resolution images the SNR increases.

To increase the transmitted energy as well as the spatial resolution, the active aperture needs to be larger, therefore a setup has been considered, where all the elements on the transducer are used both in transmit and receive. For this setup, the active aperture could either be a \(32 \times 32\) square or a circle with a radius of approximately 16 elements. To increase the circular symmetry of the PSF, the circular shape is chosen as the active aperture. The transmit aperture, which is static during all emissions, is shown in Fig. 2.9(a). The receive aperture, which is also static during all emissions is shown in Fig. 2.9(b).

![Synthetic aperture imaging transmit and receive apertures](image)

Figure 2.9: The synthetic aperture imaging transmit and receive apertures as implemented on the \(32 \times 32\) element transducer array. The transmit aperture is static for all emissions. The receive aperture is also static during all 256 emissions.

The resulting synthesized aperture is shown in Fig. 2.10 on the facing page. The transmit beam for the shown emission is illustrated with an arrow. The source of the beam is the active virtual source, shown with a circle. For each emission, a low resolution volume is beamformed. Each point in the low resolution volume is then weighted by a virtual source apodization, similar to what is shown in Fig. 2.11. The virtual source apodization has the shape of a cone centered around the transmit beam and with its apex
2.6. Parallel Beamforming

Based on the 90° imaging field of view and the number of emissions in each direction, the transmit aperture width that resulting in a beam width of approximately $90°/16 = 5.63°$ was determined to be equal to 5.2 mm at 3 MHz or approximately 17 transducer elements on the Vermon probe. A circular aperture with a diameter of 16 elements, as shown in Fig. 2.12(a) on page 29, was chosen as the transmit aperture. For 3-D imaging using

Figure 2.10: The setup for a single emission is shown, synthesizing one virtual element (shown as a circle) in the synthetic aperture. The remaining virtual elements of the sequence are shown as squares raised above the physical aperture. The virtual sources are located behind the aperture and the sound is emitted downwards, in the direction of the arrow. The cross marks the center of the active aperture and the colors of the physical elements represent their apodization value.
parallel beamforming \(N \times N\) receive lines per emission have to be beamformed. This can be derived from the ratio between (2.6) and (2.8)

\[
N = \frac{\text{Transmit beam width}}{\text{Two-way beam width}} = \left(\frac{f_t^2}{f_r^2} + 1\right) = \left(\frac{a_r}{a_t} + 1\right).
\] (2.10)

Using the full aperture in reception, which has 32 active elements in each dimension, \(N = 3\) scan lines per dimension were determined as an adequate spatial sampling frequency to represent the PSF at the focal point. As determined earlier, the maximum \(f_{prf}\) allows for 16 emissions per steering angle, leading to \(16 \times 3 = 48\) scan lines to be beamformed per steering angle per emission. To beamform the \(3 \times 3\) lines per emission, the area of the receive aperture should be almost four times as wide as the transmit aperture. The receive aperture is shown in Fig. 2.12(b) on the facing page. A transmit focal depth of 30 mm was chosen based on trial and error. The transmit beam for the shown emission is illustrated with an arrow in Fig. 2.13 on page 30. For each emission, a volume consisting of nine receive lines are beamformed. Each point in the low resolution volume is then weighted by an hourglass shape apodization, similar to what is shown in Fig. 2.14. In this work, the hourglass angular width is 7°. Although different advanced methods have previously...
been proposed to minimize the block-like artifacts with PB (Augustine 1987; Liu et al. 2002; Hergum et al. 2007), in this study a 50% overlap between the receive lines of two adjacent transmit lines, and coherently compounding those lines, was used to compensate for those artifacts and therefore, in that regard it is different from the transitional PB technique.

![TX Aperture](image1)

![RX Aperture](image2)

(a) (b)

Figure 2.12: The parallel beamforming transmit and receive apodization implemented on the $32 \times 32$ element array. The receive aperture is the widest symmetrical area on the $32 \times 32$ element array. Transmit apodization contains 256 active elements and is used for all 256 emissions.

### 2.7 Simulation and Measurement Setups

Table 2.1 on page 32 lists the measurement configuration parameters. The RF-data were beamformed using the beamformation toolbox 3 (BFT 3) (Hansen et al. 2011).

### 2.8 Results

In this section, the results of the comparison between synthetic aperture imaging and parallel beamforming are presented.

#### 2.8.1 The Simulated Point Spread Function

In Fig. 2.15 on page 38, three cross-planes (azimuth, elevation, and C-plane) of two 3-D PSFs imaged using synthetic aperture imaging as well as parallel beamforming at a depth of 62 mm are shown with a dynamic range of 60 dB. Both PSFs are normalized...
Figure 2.13: The setup for a single emission is shown. For every transmit line shown in orange, nine receive lines, shown in blue, are beamformed. Every adjacent transmit line has 50% overlap over the receive lines.

to their maximum values. The −6 dB and −20 dB main-lobes of the PB technique are seen to be smaller in both the elevation and azimuth directions, except the −40 dB. The side-lobe levels are lower for synthetic aperture imaging than for parallel beamforming, specifically at −40 dB. The side-lobes are seen to be asymmetrical, as they are wider in the elevation direction than in the azimuth direction. The asymmetric PSF is due to the asymmetry of the transducer array used. The discontinuities in the probe cause the increased side-lobe levels in the elevation direction. In the elevation direction it is hard to separate the side-lobes from grating-lobes. The FWHM and cystic resolution calculated values for both of the simulated 3-D PSFs are listed in Table 2.2 on page 32.

2.8.2 The Measured Point Spread Function

In Fig. 2.16 on page 39, three cross-planes (azimuth, elevation, and C-plane) of two 3-D PSFs imaged using both parallel beamforming and synthetic aperture imaging techniques, at a depth of 62 mm are shown with a dynamic range of 60 dB. A needle with a diameter of 300 μm facing towards the transducer was used as a point scatterer, therefore the secondary lobe after the main lobe in axial direction is a result of the scattered echoes from the needle shaft. The −6 dB and −20 dB contours of the main-lobes of the PB technique are seen to be smaller in both the elevation and azimuth directions, except the
2.8. Results

Apodization

(a) Virtual source apodization in azimuth-range plane.

(b) Virtual source apodization in azimuth-elevation plane at depth of 80 mm.

Figure 2.14: Dynamic transmit apodization for the virtual sources. The cross marks the center of the active aperture, the circle mark denotes the virtual source, and the colors represent the apodization value.

−40 dB contours. The side-lobe levels are clearly lower for synthetic aperture imaging than for parallel beamforming. For the same reason as described earlier, the side-lobes are seen to be asymmetrical, as they are wider in the elevation direction than in the azimuth direction. The FWHM and cystic resolution calculated values for both of the measured 3-D PSFs are listed in Table 2.2 on the following page. Considering the needle diameter of 300 µm, the simulation and measurement results are quite similar.

2.8.3 SNR

The estimated SNR is calculated from stochastic data and a limited amount of data is available due to the depth dependent SNR. Therefore, averaging has to be employed to reduce the variance of the estimates. To measure the SNR of both imaging methods, a region of a tissue mimicking phantom without any cyst has been imaged 20 times. The noisy estimates are low-pass filtered with a 3-D FIR filter. The measured SNR for both methods are shown in Fig. 2.17 on page 40. The SNR of synthetic aperture imaging is higher than the SNR of parallel beamforming. Two SNR profiles along the indicated dashed lines in Fig. 2.17 on page 40 are shown separately in Fig. 2.18 on page 41. The penetration depth, where the SNR crosses 0 dB, is by linear regression estimated to be
Table 2.1: Setup configuration

<table>
<thead>
<tr>
<th></th>
<th>SAI</th>
<th>PB</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frame rate</td>
<td>20</td>
<td>20</td>
</tr>
<tr>
<td>Pulse repetition frequency</td>
<td>5.13</td>
<td>5.13</td>
</tr>
<tr>
<td>Emissions per frame</td>
<td>256</td>
<td>256</td>
</tr>
<tr>
<td>Number of active elements</td>
<td>1024</td>
<td>1024</td>
</tr>
<tr>
<td>Scan depth (max range)</td>
<td>15</td>
<td>15</td>
</tr>
<tr>
<td>Number of Sinusoidal cycles</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>Focus in transmit</td>
<td>-6</td>
<td>30</td>
</tr>
<tr>
<td>Sampling frequency</td>
<td>12</td>
<td>12</td>
</tr>
<tr>
<td>Transmit voltage</td>
<td>±100</td>
<td>±100</td>
</tr>
<tr>
<td>Field-of-view</td>
<td>90° × 90°</td>
<td>90° × 90°</td>
</tr>
<tr>
<td>Beamformed lines per emission</td>
<td>64 × 64</td>
<td>3 × 3</td>
</tr>
<tr>
<td>Sound speed in vivo</td>
<td>1540</td>
<td>1540</td>
</tr>
<tr>
<td>Sound speed in 24 °C water</td>
<td>1494</td>
<td>1494</td>
</tr>
</tbody>
</table>

Table 2.2: FWHM and cystic resolution measurements

<table>
<thead>
<tr>
<th></th>
<th>PB Measurement</th>
<th>PB Simulation</th>
<th>SAI Measurement</th>
<th>SAI Simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_{6dB}$</td>
<td>1.43</td>
<td>0.89</td>
<td>1.33</td>
<td>0.74</td>
</tr>
<tr>
<td>$R_{12dB}$</td>
<td>2.25</td>
<td>1.16</td>
<td>1.99</td>
<td>0.97</td>
</tr>
<tr>
<td>Axial FWHM</td>
<td>0.65</td>
<td>0.22</td>
<td>0.62</td>
<td>0.25</td>
</tr>
<tr>
<td>Azimuth FWHM</td>
<td>2.59</td>
<td>2.15</td>
<td>2.59</td>
<td>2.68</td>
</tr>
<tr>
<td>Elevation FWHM</td>
<td>2.41</td>
<td>2.18</td>
<td>3.64</td>
<td>2.68</td>
</tr>
</tbody>
</table>
149 mm for parallel beamforming and 183 mm for synthetic aperture imaging. In other words, synthetic aperture imaging increases the penetration depth by approximately 23%.

2.8.4 Cysts Embedded in Tissue Mimicking Phantom
In Fig. 2.19 on page 42, three cross-planes (azimuth, elevation, and C-plane) of the cyst phantom are shown. Fig. 2.24(c), 2.19(d), and 2.19(f) on page 42 are made with parallel beamforming and Fig. 2.24(d), 2.19(c), and 2.19(e) on page 42 with synthetic aperture imaging. The cysts are clearly more apparent when imaged with synthetic aperture imaging than with parallel beamforming.

The cyst statistics are measured from a sphere with a radius of 6 mm located at the center of each cyst. The speckle statistics are estimated on the exact same spheres but on the tissue mimicking phantom containing only random scatterers. The CNR as shown in Fig. 2.20 on page 43 was better for synthetic aperture imaging at all depths compared with parallel beamforming. When ignoring the first cyst, the CNR for parallel beamforming cysts decreased approximately linearly however, for synthetic aperture imaging it almost maintained throughout the imaging depth.

2.8.5 Abdominal Anatomic 3-D Phantom
A volumetric region of an abdominal anatomic phantom (Model 057A, CIRS, Virginia, USA) containing the liver has been imaged with both SAI and PB methods. The phantom simulates the abdomen from approximately the thorax vertebrae (T9/T10) to the lumbar vertebrae (L2/L3) using simplified anthropomorphic geometry. The materials provide similar acoustic features to human body. In Fig. 2.21 on page 44, two cross-planes (azimuth and elevation) of the liver imaged with SAI technique are shown in 60 dB dynamic range. In Fig. 2.22 on page 45 for the same region, two cross-planes (azimuth and elevation) of the liver imaged with PB technique are shown. Both methods are able to image the hepatic veins inside the liver of similar quality, with SAI having slightly better contrast. However, the number of emissions for SAI technique can be reduced to 64 instead of 256 for the same field of view as shown in Fig. 2.23(a) and Fig. 2.23(b) on page 46, which corresponds to a volume rate of 80 Hz. Moreover, in Fig. 2.23(c) and Fig. 2.23(d) on page 46 using only 16 emissions, corresponding to a volume rate of 320 Hz, the SAI technique could visualize the anatomy of comparable quality to Fig. 2.21. Using only five emissions, SAI technique was able to image the anatomy as shown in Fig. 2.23(e) and Fig. 2.23(f) on page 46, corresponding to a volume rate of 1020 Hz. Although, lowering the number of emissions increases the temporal resolution, the SNR goes down.

2.8.6 Intensity Measurements
Before any in vivo measurements, the ultrasound imaging technique on the scanner has to fulfill all the requirements regarding the intensity levels and safety limits. Any
damage to the tissues caused by cavitational effects or over-heating has to be avoided. In practice, the acoustic output is adjusted such that both the peak and the temporal average intensities remain under given thresholds. As of today, such safety guides are regulated by the the FDA (FDA 2008), and take the form of upper limits on given indexes: the mechanical index \( (MI \leq 1.9) \), the derated spatial-peak-temporal-average intensity \( (I_{spta} \leq 720\,\text{mW}/\text{cm}^2 \text{ for peripheral vessel}, \ I_{spta} \leq 430\,\text{mW}/\text{cm}^2 \text{ for cardiac} \)), and the derated spatial-peak-pulse-average intensity \( (I_{sppa} \leq 190\,\text{mW}/\text{cm}^2) \). (FDA 2008). This requires to measure the emitted pressure of the transducer as a function of spatial position. The intensity measurements have been carried out using the experimental ultrasound scanner SARUS and the AIMS III intensity measurement system (Onda Corporation, Sunnyvale, California, USA)(Jensen, Rasmussen, et al. 2016). The measured mechanical index and the intensity as a function of depth and lateral position are shown in Fig. 2.24 on page 47 for a \( f_{prf} = 100\,\text{Hz} \). The frame rate is lowered to decrease the effect of reverberations and therefore the intensity values should be scaled by a factor of 51.33 for the actual measurements with \( f_{prf} = 5.133\,\text{KHz} \). The measured MI and \( I_{spta} \) before scaling the excitation signal for PB are 0.77 and 168.8\,\text{mW}/\text{cm}^2, and for SAI are 0.14 and 4\,\text{mW}/\text{cm}^2, accordingly. The lower intensity values for SAI is due to the placement of the virtual sources behind the array, and as long as the probe can handle the increment of the temperature, the excitation voltage can increase before reaching the FDA limits.

\section*{2.8.7 Probe Temperature Measurements}

Another criterion that has to be measured before any \textit{in vivo} scans, is the probe sole temperature. Based on the safety guidances provided by FDA, the temperature of the probe running in still air should be lower than the body’s temperature. Also, the temperature rise while using on a patient should be below ten degrees Celsius for approximately half an hour. The measured temperature of the probe sole using both imaging methods are shown in Fig. 2.25 on page 48. Due to a larger active transmit area of the SAI technique, the probe temperature increases faster compared with the PB technique. However, both techniques satisfy the FDA safety requirements for \textit{in vivo} measurements.

\section*{2.8.8 \textit{In Vivo} Measurement}

Two cross-planes (azimuth and elevation) of an \textit{in vivo} volumetric data of a healthy male’s gallbladder, imaged with SAI and PB techniques, are illustrated in Fig. 2.26 on page 49 and Fig. 2.27 on page 50. Although both methods can visualize the gallbladder, they both suffer from a low clinical value, attributed to the limited size of the used 2-D probe. Based on the Fraunhofer approximation, to increase the lateral resolution, the transmit and receive \( f \)-numbers should become smaller, corresponding to increasing the aperture size. However, increasing the aperture size and keeping a \( \lambda/2 \)-pitch result in a dramatic increase on the number of elements and thereby a large number of channels. The next upcoming chapters will try to investigate alternative ways to lower the number of channels required for 3-D imaging using row-column addressed 2-D arrays.
2.9 Discussion and Conclusions

The imaging quality of SAI was investigated in comparison with PB technique, which used to be the gold standard of 3-D ultrasound imaging. The comparison was based on Field II simulations, phantom measurements, as well as in vivo measurements with a λ/2-pitch 3.8 MHz 32×32 2-D transducer connected to the experimental ultrasound scanner SARUS. Two sequences with both SAI and PB techniques were designed for imaging a volume region with 90°×90° field of view down to 15 cm at a 20 Hz volume rate. Using both simulations and measurements, it was shown that 3-D synthetic aperture imaging increases the imaging sensitivity compared with parallel beamforming. An iron needle facing towards the transducer inside a water tank used as a point target and was imaged with both techniques to characterize measured PSFs. The point target measurements were carried out at 0° steering angle and showed the same tendency as the simulations. Synthetic aperture imaging increased the contrast and had similar resolution. Measurements on a tissue mimicking phantom indicated that the penetration depth is deeper for synthetic aperture imaging compared with parallel beamforming. Synthetic aperture had a higher SNR than parallel beamforming at all depths and the increased SNR resulted in a penetration depth increase of 23%. The CNR was improved by 50% at 70 mm depth. The penetration depth reached the design goal of 15 cm for both methods. Although both methods were able to reach the application requirements, the image quality was not comparable with conventional 2-D imaging. This is due to the small transducer array surface area as well as the limitation on the acoustic output.

Based on the acoustic intensity and temperature measurements, in the SAI configuration with virtual sources behind the transducer, the measured acoustic intensity and probe temperature are far below the FDA safety limits. This indicates that, as long as these safety limits are satisfied, we are allowed to increase the acoustic output. The setup used for the measurement limits the maximum value for the excitation voltage, however the transmitted acoustic energy can be boosted by using coded excitation, increasing the contrast and penetration depth. On the other hand, using a long excitation waveform increases the transducer temperature very rapidly, which might burn the transducer. Therefore, a fine tuning is required on the coded excitation waveform, which only can be done by knowing an accurate model for the transducer heat transfer model. This study presented some of the promising potentials of 3-D synthetic aperture imaging in terms of image quality and volume rate.

One conference paper was published on the subject of 3-D synthetic aperture imaging (included on page 129), in which a comparison of real-time 3-D synthetic aperture imaging and parallel beamforming using only 256 active channels was presented. The comparison was based on both Field II simulations as well as in vivo measurements. The setup and all of the results presented in this chapter have not been submitted before and should be considered as a draft paper.
References


Figure 2.15: SAI simulated 3-D point spread function for the 1024 active elements setup, sliced into three 2-D planes. The point spread functions are observed at 62 mm depth and 0° azimuth and elevation tilt angle. The PSFs are normalized to their own maximum values.
Figure 2.16: PB and SAI measured 3-D point spread function sliced into three 2-D planes. The point spread functions are observed at 62 mm depth and 0° azimuth and elevation tilt angle. The left column is SA and the right column is PB. The PSFs are normalized to their own maximum values.
Figure 2.17: The SNR of the research scanner for both SAI (top) and PB (bottom) imaging methods in a tissue mimicking phantom.
Figure 2.18: The SNR profile along dashed line in Fig. 2.17 for both SAI (left) and PB (right) imaging methods.
Figure 2.19: Synthetic aperture imaging (a, c, and e) and parallel beamforming (b, d, and f) of anechoic cysts embedded in a tissue-mimicking phantom. The dynamic range is 60 dB. The large cysts have a diameter of 8 mm, the small cysts a diameter of 4 mm. C-scans are at a constant distance of 50 mm to the array center. The cysts are water-filled pipes aligned 45° to the vertical scan plane.
Figure 2.20: The estimated contrast to noise ratios of the 8 mm cysts are shown. When ignoring the cyst at 10 mm, the contrast for parallel beamforming decreases faster than for synthetic aperture imaging. The cyst at 10 mm depth is an outlier. This is probably because of reverberations from the phantom to transducer interface.
Figure 2.21: Two cross-planes (azimuth and elevation) of the liver over an anatomic phantom are shown for the SAI method. The dynamic range is 60 dB.
Figure 2.22: Two cross-planes (azimuth and elevation) of the liver over an anatomic phantom are shown for the PB imaging method. The dynamic range is 60 dB.
Figure 2.23: Synthetic aperture imaging of hepatic veins of liver in an abdominal anatomic phantom with different number of emissions. The dynamic range is 60 dB.
Figure 2.24: Synthetic aperture imaging (a and c) and parallel beamforming (b and d) derated measured intensities. The $I_{spta}$ intensities are measured for an $f_{prf} = 100$Hz and have to be scaled by a factor of 51.33 for the actual measurements with $f_{prf} = 5.133$KHz.
Figure 2.25: The probe sole temperature measurements for both SAI (left) and PB (right) imaging methods.
Figure 2.26: Two cross-planes (azimuth and elevation) imaged in vivo of the gallbladder are shown for SAI method. The dynamic range is 60 dB.
Figure 2.27: Two cross-planes (azimuth and elevation) imaged \textit{in vivo} of the gallbladder are shown for PB method. The dynamic range is 60 dB.
Chapter 3

Row-Column Addressed Arrays

One of the goals of this research project was to investigate 3-D imaging using row–column-addressed (RCA) 2-D arrays. For that reason, two in-house prototyped 62+62 RCA 2-D array transducer probes were manufactured using capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technology. The transducers are designed with similar acoustical features, i.e., dimensions, center frequency, and packaging. The probes are fully integrated RCA 2-D arrays equipped with integrated hardware apodization. This gives the unique possibility of evaluating the two probes relative to each other and comparing the row–column addressing scheme based on two different technologies. The scope of this chapter is therefore to display the capabilities of RCA transducers, when integrated into probe handles, and to evaluate their performance. This chapter describes the design, fabrication, and characterization of the two probes based on acoustical measurements. From these measurements the center frequency, bandwidth, surface pressure, sensitivity, and acoustical cross-talks are evaluated and discussed.

Chapter 4 on page 75 investigates the rectilinear volumetric imaging performance of these two probes based on simulation and phantom measurements.

The analyses and results included in this chapter are based on the work presented in the papers D on page 163 and E on page 175. First, an introduction to RCA arrays and the previous literature are briefly described, afterwards the probe design is detailed. Then, the results of the electro-acoustical measurements are presented, followed by a description of the published papers and a conclusion.

3.1 Introduction

Based on Fraunhofer diffraction theory, the upper limit of achievable imaging quality scales linear with wavelength as well as the f-number, which is the focal length divided by the aperture diameter. To increase the imaging quality of the 2-D transducer array, either the frequency, the aperture diameter, or both, therefore has to be increased. Increasing the frequency has the disadvantage of decreasing the penetration depth. Increasing the aperture width would increase both the imaging quality and the transducer surface area, which could be used to increase the penetration depth.

The disadvantage of increasing the side length is a dramatic increase in the number of transducer elements in the array. Often 1-D arrays feature 128 or more transducer elements. A 2-D array with $128 \times 128$ elements would contain 16,384 transducer elements. From a transducer fabrication perspective, this poses a great challenge for providing
electrical connections to all the elements while maintaining a high element yield. The interconnecting wires between the 16,384 elements and the ultrasonic system result in a large, heavy cable, excluding it from any practical use. From beamformation point of view, it per se is a tremendous challenge to process this amount of information.

The issue of reducing channel count, whilst maintaining the size of the array aperture, was addressed in the earlier versions of 2-D matrix arrays by introducing sparse arrays. Here only a subset of elements are active at the same time. Amongst these are Mills cross arrays, random arrays, and Vernier arrays, each presenting their benefits and drawbacks (Davidsen et al. 1994; Brunke and Lockwood 1997; Yen, Steinberg, et al. 2000; Austeng and Holm 2002; Karaman et al. 2009). However, all of them suffer from reduced SNR, due to the reduced active area, and introduce higher sidelobes and/or grating lobes (Turnbull and Foster 1991).

Recently, fully populated arrays with reduced channel count have become available by integrating electronic pre-beamformers (µ-beamformers) inside the transducer probe (Savord and Solomon 2003; Halvorsrod et al. 2005; Blaak et al. 2009). This results in much fewer signals to be funneled out to the ultrasound scanner. An example, of such a state-of-the-art fully populated matrix transducer, is the X6-1 PureWave xMATRIX Array from Phillips (Eindhoven, Netherlands), with 9,212 elements (Phillips 2015). Despite the recent advances in real-time 3-D ultrasound imaging, the ultrasound systems supporting such imaging modalities are highly advanced and rely on cutting edge software, hardware, and manufacturing technology. This results in expensive equipment that impairs the low-cost advantage of ultrasound, thus limiting its more widespread use. Moreover, the thermal budget starts to become a consideration for modern probes with integrated electronics, due to the constraints on transducer probe heating dictated by the standards for medical equipment (Sampson et al. 2013; IEC 2015).

A low cost alternative to µ-beamforming has been suggested that maintains the aperture width and the surface area: The RCA 2-D arrays, first proposed in 2003 by Morton and Lockwood (Morton and Lockwood 2003). Row–column-addressing of 2-D arrays is a scheme to reduce the number of active channels needed for contacting the elements in the array. The idea is to contact the elements in the 2-D array either by their row or column index. Each row or column thereby acts as one large element. This effectively turns the array into two orthogonal 1-D arrays as shown in Fig. 3.1 on the facing page. The imaging principle relies on using one of the 1-D arrays as the transmit array, creating a line focus of the transmit pulse. The perpendicular 1-D array is used to receive, enabling receive focus in the orthogonal dimension. The combination of transmit and receive focus provides focusing on a point in the volume, hence a volumetric image can be created. Whereas an $N \times N$ fully addressed array needs $N^2$ connections, an RCA array only needs $2N$ connections. The RCA array can therefore have a larger aperture compared to the fully addressed array, having the same number of connections. A simulation study by Rasmussen and Jensen (Rasmussen and Jensen 2013b) and a measurement study (Rasmussen and Jensen 2013a), both compared the two different addressing schemes. With the same number of connections, a superior image quality is
3.1. Introduction

Vertical Array
Horizontal Array
Connection Sub-Element
Row element Column element

Figure 3.1: A row-column addressed 2-D array can be interpreted as two orthogonal 1D arrays: One array consisting of row-elements and one array consisting of column elements. The Figure is taken from paper (Rasmussen, Christiansen, et al. 2015).

obtained using the RCA array.

3.1.1 Literature review

Realizations of RCA arrays have previously been presented by several groups. The first experimental demonstration of RCA arrays were presented in 2006 by Seo and Yen (Seo and Yen 2006). The array was a PZT in a 64+64 layout, fabricated using a 1-3 ceramic with the row and column electrodes defined on separate sides of the ceramic. This array was later surpassed by the same authors with a 256+256 array using the same fabrication technique (Seo and Yen 2007, 2008, 2009). In 2009 Yen et al. introduced a simplified process for fabrication of RCA PZT arrays using a dual layer structure (Yen, Seo, et al. 2009). The dual layer structure was composed of a piezoelectric 2-2 composite for the transmit array, and a single sheet of undiced copolymer was used as the receive array. Row-column arrays based on CMUT technology were first presented in 2009 by Logan et al. (Logan, Wong, and Yeow 2009). They showed a 32+32 array fabricated using the wafer bonding process with a silicon nitride plate, and later they presented characterization of a similar array (Logan, Wong, Chen, et al. 2011). Zemp et al. (Zemp et al. 2011) and Sampaleanu et al. (Sampaleanu et al. 2014) presented RCA arrays fabricated using the sacrificial release process and performed feasibility studies. More recently they have presented photoacoustic imaging using RCA CMUT arrays (Chee et al. 2014). In 2015 Rasmussen et al. (Rasmussen, Christiansen, et al. 2015) and Christiansen et
al. (Christiansen, Rasmussen, et al. 2015) presented a two-part paper presenting an RCA CMUT array with integrated apodization. The apodization was added as a static roll-off apodization region located at the ends of the line elements. They showed that the main lobe was unaffected by integrating this type of apodization. Part II showed experimental results of an CMUT RCA 2-D array with this roll-off apodization. The CMUT array was a 62+62 layout with four apodization regions fabricated using the wafer bonding technique, two SOI wafers and a plate of highly doped silicon.

3.2 Array design and fabrication

The general design of the RCA arrays is based on the findings by Rasmussen et al. (Rasmussen, Christiansen, et al. 2015) and Christiansen et al. (Christiansen, Rasmussen, et al. 2015). The arrays consist of 62 row elements and 62 column elements, and four apodization regions. Only the 62+62 elements are connected to beamformer channels. The design of the RCA array can therefore be divided into two parts: The central region and the apodization region.

The central part of the array may be considered as a conventional RCA array, a 3-D diagram of a corner of such an array is shown in Fig. 3.2 on page 56. The diagram includes four top and four bottom electrodes placed orthogonal to each other and colored orange and blue, respectively. Between the top and bottom electrodes is the "active" material, which is either the CMUTs or the piezoelectric material. The element contacts are placed alternately on each side of the array as showed in the figure. The top elements can be used as a 1-D array by grounding all of the bottom elements, and the bottom elements can be used as an orthogonal 1-D array by grounding all of the top elements.

The four apodization regions are located outside the central part of the array and are added to avoid the abrupt truncation of the elements, giving rise to the ghost echoes (Rasmussen, Christiansen, et al. 2015). The layout of the array including the apodization is shown in Fig. 3.3 on page 58. The central part is showed within the dashed line and has an apodization value of 1. The apodization regions are placed on each side of the central region and the apodization value follows a Hann function from the edge of the central part to the edge of the array, where the apodization is 0. The apodization was originally developed for the CMUT array (Christiansen, Rasmussen, et al. 2015), but has been adapted to have the same dimension and roll-off characteristic for the PZT array.

Two arrays are fabricated using the design introduced above, one based on CMUT technology and one based on PZT technology. The pitch, number of elements, active footprint, center frequency, and excitation voltage are designed to be identical for the two arrays. This makes it possible to evaluate/compare the row–column-addressing scheme based on two different technologies. A center frequency of 3 MHz was chosen with lambda half pitch and an excitation voltage of $\pm 75 \text{ Vac}$. The dimensional parameters of both arrays are given in Table 3.1.
### Table 3.1: Transducer dimensional parameters (The Table is taken from paper E.)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>CMUT</th>
<th>PZT</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements</td>
<td>62+62</td>
<td>62+62</td>
<td>–</td>
</tr>
<tr>
<td>Number of apodization region electrodes</td>
<td>4</td>
<td>4</td>
<td>–</td>
</tr>
<tr>
<td>Element pitch</td>
<td>270</td>
<td>270</td>
<td>µm</td>
</tr>
<tr>
<td>Element width</td>
<td>265</td>
<td>245</td>
<td>µm</td>
</tr>
<tr>
<td>Kerf</td>
<td>5</td>
<td>25</td>
<td>µm</td>
</tr>
<tr>
<td>Element length</td>
<td>24.84</td>
<td>24.84</td>
<td>mm</td>
</tr>
<tr>
<td>Acoustic window thickness</td>
<td>1.5</td>
<td>1.27</td>
<td>mm</td>
</tr>
<tr>
<td>Acoustic window velocity</td>
<td>1.0</td>
<td>1.0</td>
<td>mm/µs</td>
</tr>
<tr>
<td>Length of apodization regions</td>
<td>4.05</td>
<td>4.05</td>
<td>mm</td>
</tr>
<tr>
<td>Array outer dimensions (square)</td>
<td>26.3</td>
<td>26.3</td>
<td>mm</td>
</tr>
<tr>
<td>Cell side length (square)</td>
<td>56</td>
<td>–</td>
<td>µm</td>
</tr>
<tr>
<td>Kerf between cells</td>
<td>7</td>
<td>–</td>
<td>µm</td>
</tr>
<tr>
<td>Plate thickness</td>
<td>1.85</td>
<td>–</td>
<td>µm</td>
</tr>
<tr>
<td>Al electrode thickness</td>
<td>400</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Vacuum gap height</td>
<td>448</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Nitride thickness</td>
<td>56</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Insulation oxide thickness</td>
<td>410</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Post oxide thickness</td>
<td>1346</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>PZT layer thickness</td>
<td>–</td>
<td>500</td>
<td>µm</td>
</tr>
<tr>
<td>PZT volume fraction</td>
<td>–</td>
<td>66</td>
<td>%</td>
</tr>
<tr>
<td>Electrode thickness</td>
<td>–</td>
<td>640</td>
<td>nm</td>
</tr>
<tr>
<td>Matching layers net thickness</td>
<td>–</td>
<td>0.433</td>
<td>mm</td>
</tr>
<tr>
<td>Matching layers average velocity</td>
<td>–</td>
<td>1.95</td>
<td>mm/µs</td>
</tr>
<tr>
<td>Backing round-trip attenuation at $f_c/2$</td>
<td>–</td>
<td>100</td>
<td>dB</td>
</tr>
</tbody>
</table>
3.3 Probe assembly

The assembly of the two probes is almost identical after the fabrication of the arrays, and both were assembled at the facilities of Sound Technology Inc. (State College, PA, USA). The probes are composed of four parts in addition to the transducer array itself, see Fig. 3.4 on page 59. These are a flexible printed circuit board (PCB) for connecting the transducer to the electronics, the integrated electronics containing buffer amplifiers, a cable for connecting the probe to the scanner, and a 3-D printed probe handle.

3.4 Transducer characterization

This section describes the characterization of the two probe, while their performance is evaluated concurrently and is described below. Table 3.2 taken from paper E summarizes the main results of the characterization, allowing easy location of specific parameters.

3.4.1 Impulse response

The pulse-echo response of each element was measured using an XCDR II Pulse Echo Test System, by emitting and receiving with one element at a time against a planar stainless steel reflector. The planar reflector was placed in deionized (DI) water 25 mm from the face of the probe. With the available setup, the elements were actuated with a square unipolar pulse with a duration of 100 ns for the PZT and 150 ns for the CMUT and having an amplitude of 50 V. The CMUT elements were biased at 200 Vdc. The system was set
<table>
<thead>
<tr>
<th>Parameter</th>
<th>CMUT</th>
<th>PZT</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Row</td>
<td>Column</td>
</tr>
<tr>
<td>Capacitance [pF]</td>
<td>339.4 ± 0.8</td>
<td>136.1 ± 1.4</td>
</tr>
<tr>
<td>Center frequency [MHz]</td>
<td>2.97 ± 0.07</td>
<td>3.05 ± 0.09</td>
</tr>
<tr>
<td>Bandwidth [%]</td>
<td>111 ± 3</td>
<td>106 ± 4</td>
</tr>
<tr>
<td>Phase delay [°]</td>
<td>0 ± 5.0</td>
<td>0 ± 3.1</td>
</tr>
<tr>
<td>Surface Pressure [MPa]</td>
<td>0.56 ± 0.05</td>
<td>0.54 ± 0.07</td>
</tr>
<tr>
<td>Sensitivity [µV/Pa]</td>
<td>12.9 ± 0.7</td>
<td>4.3 ± 0.7</td>
</tr>
<tr>
<td>Insertion loss [dB]</td>
<td>−26.4 ± 0.9</td>
<td>−36.5 ± 2.5</td>
</tr>
<tr>
<td>Nearest neighbor crosstalk [dB]</td>
<td>−30.5 ± 0.8</td>
<td>−26.3 ± 1.4</td>
</tr>
<tr>
<td>Transmit-receive elements crosstalk [dB]</td>
<td>−39.9 ± 0.2</td>
<td>−40.2 ± 0.6</td>
</tr>
</tbody>
</table>
Figure 3.3: A sketch of the device layout, including apodization roll-off region. The colors represent the apodization value. $a_x$ denotes the apodization profile along the $x$-axis, which follows a Hann window at the edges. The Figure is taken from paper E.

to sample at a sampling frequency of 500 MHz. The received signal was de-convolved with the excitation pulse to yield the two-way element-element impulse response.

The signals received from the impulse response measurements were de-convolved with the excitation pulse to yield the two-way element-element impulse response. Fig. 3.5 on page 60 shows the average two-way element-element impulse response and the associated envelope of the CMUT (a) and the PZT probe (b). The solid line represents the impulse response and the dashed line the envelope with black and orange representing the rows and columns, respectively.

Two extra lobes after the main lobe, around $-30$ dB, are observed for the CMUT at starting times of 3.2 $\mu$s and 4.7 $\mu$s. These extra lobes also exist for the PZT probe, but are not as easily recognized as they coincide with the ringing of the transducer itself. The time difference between the two lobes is 1.5 $\mu$s corresponding to the time difference between the main lobe and the first secondary lobe. This suggests that these extra lobes originate from reflections within the probes. Both arrays are encapsulated in RTV, which
3.4. Transducer characterization

Figure 3.4: The five main components of the probe. The components of the two probes are identical except for the transducers. The Figure is taken from paper E.

has an acoustic velocity of 1.0 mm/µs, indicating that a reflecting structure is present at a distance of 0.75 mm from the transducer surfaces. This corresponds to the shielding foils that cover the arrays.

The received signals of the columns of the CMUT probe are observed to be roughly half of the signal measured for the rows. This is believed to be due to the capacitive coupling to the substrate as discussed in paper E. It should, however, be noted that even though the signal amplitude is different of the rows and columns, the envelope shape is identical.

The frequency spectra are calculated by computing the Fourier transform of the impulse response and are shown in Fig. 3.6 on page 61. The spectra shows that the internal reflections of the probe are roughly –30 dB, which is the acceptable limit for ultrasound transducers. From the spectrum of the impulse response, the center frequency, and bandwidth for each element is found and are presented in the following.

3.4.2 Center frequency

The center frequency is calculated as a weighted mean of the frequencies present in the received signal. It is determined by summing all frequencies from zero to half of the sampling frequency, \( f_s \), weighted by its spectrum amplitude at each frequency, \( S \), and dividing by the sum of the spectrum amplitudes:
Figure 3.5: Average impulse response (solid) and normalized envelope (dashed) of the probe elements (a): CMUT. (b): PZT. The Figure is taken from paper E.
3.4. Transducer characterization

Figure 3.6: The mean impulse response spectra are shown for the rows and columns for both probes. The center frequency and the fractional bandwidth are indicated on the plot. (a): CMUT. (b): PZT. The Figure is taken from paper E.
Figure 3.7: Center frequency across the array elements of both probes. The center frequency is calculated as a weighted mean of the frequencies present in the signal. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.

\[
f_c = \frac{\sum_{i=0}^{N/2} S(if_s/N) \cdot if_s/N}{\sum_{i=0}^{N/2} S(if_s/N)},
\]

where \(N\) is the number of samples in the two-sided spectrum.

Fig. 3.7 shows the uniformity of the center frequency across the arrays. Both probes have a center frequency of 3 MHz as they were designed for, and only a small smooth variation across the arrays is observed. This smooth variation indicates that the variations are mostly caused by non-uniformities of the silicon plates of the CMUT probe, and thickness-variations of the piezoelectric material of the PZT probe.

### 3.4.3 Bandwidth

The −6 dB bandwidth was determined from the difference in frequency between the −6 dB points in the frequency spectrum. A mean bandwidth of 3.26 MHz and 2.39 MHz is found for the CMUT probe and the PZT probe, respectively. The fractional bandwidth are calculated from the bandwidth relative to the weighted center frequency, and a mean value
Figure 3.8: Bandwidth across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.

of 109% and 80% are found for the CMUT and the PZT, respectively. The uniformity across the array for both probes is shown in Fig. 3.8. The probes have a high uniformity with a standard deviation of the fractional bandwidth of 4% for the CMUT probe and 3% for the PZT probe.

### 3.4.4 Phase delay

The phase delay was found by cross-correlating the impulse response for each element with the mean impulse response and interpolating to find the lag of the maximum of the cross-correlation. Correction for any linear slope due to misalignment between the transducer and the plane reflector was done, and the mean was set to zero. The phase delay was then calculated by dividing the time it takes the wave to travel one wavelength at 3 MHz, and multiplying it by 360°, to obtain the phase delay in degrees. Fig. 3.9 on the following page shows the phase delay across the array for the CMUT and the PZT in top and bottom, respectively.

No curvature is seen of the CMUT, however the PZT is observed to curve. The bottom/column elements phase delays are seen to have a concave profile, whereas the top/row elements have a convex profile. This saddle shape is believed to originate from stress build up during the assembly.

### 3.4.5 Surface pressure

The pressure was measured using an HGL-0400 hydrophone connected to an AC-2010 pre-amplifier (Onda Corporation, CA, USA). The hydrophone was placed in front of the transducer surface and scanned over each element using the position system of the intensity measurement AIMS-3 (Onda Corporation, CA, USA), while transmitting a 3 MHz, 4-cycle sinusoidal signal on the element being measured. An amplitude of 75 Vac
Figure 3.9: Phase delay across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.

Figure 3.10: Cross-correlation based. Phase delay across the probe. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.
3.4. Transducer characterization

Figure 3.11: Surface pressure across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.

was used and the CMUT probe was biased with 200 Vdc. The pressure was recorded at 5.8 mm and 5.9 mm for the PZT and CMUT, respectively.

The recorded pressure was compensated to find the emitted pressure at the transducer surface. This compensation factor was calculated by simulating a single element in Field II (Jensen 1996). The element was set to emit a 3 MHz, 4-cycle sinusoidal wave, and the pressure magnitude relative to the pressure magnitude at the element surface was simulated. The compensation factor for the PZT and CMUT was 9.14 and 8.83, respectively. The difference was caused by the different locations of the hydrophone during the two measurements. The surface pressure across the arrays is shown in Fig. 3.11. The mean values for the CMUT and PZT are 0.55 ± 0.06 MPa and 1.68 ± 0.09 MPa, respectively.

Notice that there is no difference between the pressures emitted by the CMUT columns (elements 1-62) and the CMUT rows (elements 63-124). One would expect a lower emitted pressure from the columns due to the increased parasitic capacitance, hence a lower coupling coefficient. This is, however, not the case, since the power source during the emission is not limited in the amount of energy it can supply to the transducer.

3.4.6 Receive sensitivity

A plane stainless steel reflector was positioned in a water tank at a distance of 7.3 cm from, and parallel to, the transducer surface of the probe being characterized. The transmit signals were generated using the experimental ultrasound system, SARUS (Jensen et al. 2013), which also recorded the received signals. Twenty realizations of a 3 MHz, 4-cycle sinusoidal excitation pulse was transmitted on one element at a time and received on all elements, both rows and columns. The 20 realizations were averaged to minimize the
noise. The system was set to sample, at a sampling frequency of 70 MHz, down to a depth of 10 cm. A second measurement was performed using the same setup, but without the planar reflector, to assess the acoustical cross-talk.

The receive sensitivity is calculated by combining the result from hydrophone measurement with the result from pulse echo measurement. It is defined as the ratio of the output voltage to the received sound pressure in the surrounding fluid and is given as a linear value in $\mu V/\text{Pa}$. The receive sensitivity of the transducers is found by dividing the received voltage signal after a pulse-echo event with the incident pressure. The incident pressure was deduced using the pressure measured from the hydrophone setup. The pressure drop was compensated using the same Field II model described in previous section. Besides compensating the incident pressure for the diffraction loss, the non-ideality of the plane reflector is also compensated for. The reflection coefficient for a normal incident wave is solely determined from the acoustic impedance discontinuity in the transmission medium. In water, the reflection coefficient for a stainless steel reflector is 0.93 (Szabo 2014). The receive sensitivity for each element across the two probes is shown in Fig. 3.12. The mean values of the CMUT and the PZT probe are $8.5 \pm 4.4 \mu V/\text{Pa}$ and $14.4 \pm 1.9 \mu V/\text{Pa}$, respectively.

The sensitivity of the CMUT bottom/column elements is 67% lower than the top/row elements. This is due to the capacitive coupling to the substrate discussed in paper E. When imaging with RCA arrays, either the rows or columns are used as transmitters and the orthogonal elements as receivers, by choosing the bottom/column elements as the emitters and the top/row elements as receivers, the imaging is not affected by the lower sensitivity. However, determining 3-D vector flow might be affected since the sequence uses both rows and columns as emitters and receivers (Holbek, Christiansen, Rasmussen, et al. 2015; Holbek, Christiansen, Engholm, et al. 2016).
3.4.7 Acoustical crosstalk

There are two main types of crosstalk in a transducer array: Electrical crosstalk due to capacitive and inductive coupling, and acoustical crosstalk due to vibrations being transmitted to neighboring elements. The electrical crosstalk happens much faster than the acoustical crosstalk, since the latter is limited by the relatively low speed of sound. The acoustical crosstalk is more severe than the electrical cross-talk for the beamformed image due to the slow decay of the former. The reference measurements described in previous section are used for evaluating the crosstalk. The first $3\mu s$ of the received data are disregarded because the receivers are saturated due to the transmit pulse (electrical crosstalk) and the ring-down of the electronics.

Two different types of acoustical crosstalk can be evaluated when using RCA arrays: Nearest neighbor crosstalk and transmit-to-receive crosstalk (Christiansen, Jensen, et al. 2015). Emitting with one element at the time and extracting the maximum of the signal from its neighbor yielded the nearest neighbor crosstalk for every element. To provide a relative measure, the signal was normalized to the transmit voltage after the latter was corrected for the insertion loss of the emitting element. The insertion loss is reported in paper E. The correction corresponds to a normalization of the neighbor’s signal to the signal that the emitting element would have received if the transmitted pulse was reflected right at the transducer surface and subsequently received by the emitting element. Thus, it yields the relative acoustical coupling from one element to its neighbor. The nearest neighbor crosstalk across the probes is shown in Fig. 3.13 on the next page and the mean values are $-28.4 \pm 2.4 \, \text{dB}$ and $-30.0 \pm 2.2 \, \text{dB}$ for the CMUT and PZT probe, respectively. The nearest neighbor crosstalk of the CMUT is roughly 5 dB lower than what have earlier been reported in literature (Bayram et al. 2007; Christiansen, Jensen, et al. 2015). The lower crosstalk could be due to the RTV on top of the array. The amount of crosstalk for the PZT probe is in the limit of what is usually accepted for ultrasound probes. Ideally for phased arrays, one would dice into the piezoelectric ceramic during fabrication and fill it up with the RTV to reduce the crosstalk. This is however not possible with row-column arrays.

To provide a measure of the cross-talk in an imaging setup, the transmit-to-receive cross-talk are estimated. This is calculated as the average of the maximum signal received on all elements orthogonal to the emitting element. The average is normalized to the transmit voltage of the emitting element and corrected for the insertion loss. The transmit-to-receive elements crosstalk is shown in Fig. 3.14 on the following page for both arrays and the mean values are $-40.0 \pm 0.5 \, \text{dB}$ and $-53.7 \pm 0.9 \, \text{dB}$ for the CMUT and PZT probe, respectively. This is consistent with results in literature for the CMUT (Christiansen, Jensen, et al. 2015) and has not been previously reported for PZT RCA arrays.
Figure 3.13: Nearest neighbor crosstalk across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.

Figure 3.14: Transmit-receive elements crosstalk across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows. The Figure is taken from paper E.
3.5 Discussion

The development and transducer performance of two RCA probes for real-time volumetric imaging based on two competing technologies: PZT and CMUT, are presented. The central goal of this chapter has been to characterize the two developed transducers. The characterization should not be seen as a comparison of the two technologies, but as a display of the capabilities of the row–column-addressing scheme using these two technologies. However, since these two technologies are evaluated next to each other, one cannot avoid comparing them. The strengths and weaknesses of the emerging technology, CMUT, will now be discussed in relation to the traditional technology, PZT.

One of the most highlighted advantages of the CMUT is its higher bandwidth relative to the PZT technology. The mean $-6\text{dB}$ bandwidth is 29 percentage point higher for the CMUT probe compared to the PZT. The higher bandwidth is caused by the fact that the CMUT act as an overdamped system, due to the low impedance of the vibrating plate in immersion. One consequence of the high bandwidth is an increased axial resolution, which has been assessed in chapter F. Another interesting advantage of the high bandwidth is tissue harmonic imaging. Usually one will emit at $2/3$ of the center frequency, $f_c$, and receive at $4/3f_c$. Having a higher bandwidth results in a relative higher emitted pressure and receive sensitivity at those frequencies.

A current limitation of the CMUT technology is the lower emitted pressure. This is a result of the low inertia of the plate (thin plate, low mass). The surface pressure of the PZT probe is consequently 3 times higher than the CMUT probe. Contrary to expectations, the mean receive sensitivity of the PZT probe is 11% higher than the top/row elements of the CMUT probe. The sensitivity of the CMUT array can be improved by optimizing the structure, plate design, layout, and driving conditions. Packing the cells closer will increase the effective area. The CMUT structure can be designed to decrease the parallel parasitic capacitance originating from the bonding area between the cells. This could be implemented by incorporating a bump in the cavity as introduced by Park et al. (Park et al. 2008). This facilitates the possibility of having a high ratio of post oxide thickness to gap height. Improving the driving conditions also makes it possible to increase performance of the CMUT probe. The bias voltage is closely related to the electro-mechanical coupling coefficient describing the efficiency at which the mechanical energy (vibrations) is converted to electrical energy and vice versa. The coupling coefficient approaches unity at the pull-in voltage (Yaralioglu et al. 2003). Increasing the bias voltage will result in a higher receive sensitivity and emitted pressure. The bias voltage of the probe is in this study limited to 200 V because of the integrated electronics. As a result, the probe is operated at a maximum of 83% of the pull-in voltage. The optimal driving conditions and the gain hereof will be investigated in future research. If both the emitted pressure and the receive sensitivity is taken into account, one will expect the penetration depth of the PZT probe is 3.4 times higher compared to the CMUT probe at these driving conditions. A potential way of increasing the pressure, hence the penetration depth, is by emitting with more than one element. However, the pressure generated by the transducer
is usually limited by both the mechanical index and the temperature of the probe itself. All of these aspects is investigated in next chapter 4.

Even though this PZT probe is superior on most parameters evaluated in this chapter, except for a lower bandwidth, the CMUT probe will benefit by increasing the center frequency. The reason for this is the manufacturing process. When the center frequency is increased the width of the elements is decreased to be able to keep the $\lambda/2$-pitch. The width of the PZT element, hence the area, is limited by the kerf. Modern manufacturing processes enable kerfs down to 25 $\mu$m, whereas it is possible to fabricate kerf-less transducers with the CMUT technology. Since both the pressure and receive sensitivity scales linearly with the area, halving the area means that the pulse-echo sensitivity is decreased by a factor of four. Take, for example, the case of a 15 MHz transducer with $\lambda/2$-pitch (50 $\mu$m). The active area of the PZT element would be half of the CMUT element. It then follows that the pulse-echo sensitivity (combined surface pressure and receive-sensitivity) would be 18% higher for the CMUT probe. Together with the higher bandwidth, this emphasizes the use of CMUT transducers for high frequency, high resolution applications. This is consistent with findings by Savoia et al. (Savoia et al. 2012). They demonstrated a 1-D CMUT probe with a center frequency of 13.6 MHz having a higher two-way sensitivity compared to a similar commercial PZT probe.

### 3.6 Conclusion

This chapter presented the development and characterization of two 62+62 RCA ultrasound probes based on CMUT and PZT technology. The objective has been to show the capabilities of the RCA transducer implemented using the two specific technologies. Both transducers have integrated apodization to reduce ghost echoes and are designed with similar acoustical features. They were designed to be used with a commercial scanner made for conventional 2-D imaging, due to the low channel count of the row-column addressed probes, the probes and scanner can be directly interfaced. A solution with a flexible mounting PCB and two rigid amplifier PCBs was used to mount the array and interface it to the scanner cable via buffer amplifiers. The array and electronics were electrically shielded with a metal foil, and the array was covered with a silicone coating before the entire probe was encapsulated in a 3-D printed handle. The reliability and performance of the probes were assessed through electrical and acoustical measurements. Four different measurement setups were used and the probes electrical capacitances, center frequencies, bandwidths, phase delays, surface pressures, receive sensitivities, insertion loss, and acoustical crosstalks were evaluated. The weighted center frequency is exactly 3.0 MHz for both probes, as they were designed for. The $−6$ dB fractional bandwidth were 29 percentage point higher for the CMUT probe than the PZT. The surface pressure of the PZT probe is a factor 3 times higher relative to the CMUT probe, and the expected penetration depth is 3.4 times higher in favor of the PZT. The authors emphasize that the driving conditions of the CMUT probe was limited by the integrated electronics in the
probe handle, which could otherwise have improved its performance.

In the following chapter 4 on page 75, the imaging performances are evaluated. The quality assessments of the B-mode images acquired with both probes, i.e., spatial resolution, contrast resolution, and SNR, are carried out based on simulations and also the measurements over several phantoms using synthetic aperture imaging (SAI) technique.

References


Holbek, S., T. L. Christiansen, M. Engholm, A. Lei, M. B. Stuart, C. Beers, L. N. Moesner, J. P. Bagge, E. V. Thomsen, and J. A. Jensen (2016). “3-D Vector Flow Using a Row-


CHAPTER 4

Imaging Performance Assessment with Row-Column Addressed Arrays

Previous chapter, Chapter 3 described the design, fabrication, and characterization of two in-house prototyped 62+62 row–column-addressed (RCA) 2-D array transducer probes manufactured using capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technology. This chapter investigates the rectilinear volumetric imaging performance of these two probes based on simulation and phantom measurements.

The analyses and results included in this chapter are based on the work presented in the papers C on page 155, D on page 163, and F on page 189. First, an overview of beamforming with RCA 2-D arrays is presented, afterwards a theoretical comparison between imaging with RCA and fully addressed 2-D arrays is given. Then, the results of the measurements are presented, followed by a discussion and a conclusion.

4.1 Introduction

Previous chapter detailed on the fabrication process, assembly, and the electro-acoustical evaluation of two in-house prototyped 62+62 RCA 2-D array transducer probes using CMUT and PZT technologies. In this chapter, the rectilinear volumetric imaging performance of these two probes is investigated based on simulations and phantom measurements. Therefore, the scope of this study is to evaluate the focusing ability of RCA 2-D arrays in general and more specifically the imaging performance of two fully integrated CMUT and PZT RCA 2-D array transducer probes quantitatively and comparatively. Both delay-and-sum (DAS) and spatial matched filter (SMF) beamformation (Jensen and Gori 2001; Kim et al. 2006; Yen 2013) methods are used for processing the IQ-modulated RF data generated with a synthetic aperture imaging (SAI) technique (Jensen, Nikolov, et al. 2006). The quality assessments of the B-mode images acquired with both probes, i.e., spatial resolution, contrast resolution, and signal-to-noise ratio (SNR) are carried out based on simulations and measurements on several phantoms. Two different SAI sequences were designed for imaging down to 14 cm of depth at a volume rate of 88 Hz. The first sequence uses 62 virtual sources behind the array, and the second sequence utilizes 62 single element transmissions. In both sequences the echoes are collected with all column elements.
4.2 Synthetic Aperture Imaging and Beamforming

For a speed of sound of 1540 m/s, 182 µs is required to acquire data from a single emission to a depth of 14 cm. For 62 emissions this is equivalent to a volume rate of 88 Hz. The first sequence utilizes 62 virtual line sources behind the array by adjusting the transmit delays of all the row elements. The second sequence utilizes 62 single element transmissions on the row elements. In both sequences the echoes are collected with all the column elements.

IQ-modulated RF data are used for beamforming a low-resolution volume for every emission and finally, by summing all the low-resolution volumes, a high-resolution volume is generated. This section introduces the two beamforming methods employed for generating the low-resolution volumes: a modified DAS scheme in section 4.2.1 and a spatial matched filtering approach in section 4.2.2.

4.2.1 Delay-and-sum

The details on DAS beamforming method with RCA 2-D arrays are presented in (Rasmussen et al. 2015). This method approximates each row or column element with a line-segment instead of a point, and therefore calculates the delays based on the geometrical distance between an imaging point and a line-segment. Fundamentally due to the flat design of RCA 2-D arrays only rectilinear volumetric imaging is achievable. Despite the fact that it is possible to focus the ultrasound wavefronts curvilinearly in transmit and receive, their perpendicular orientation limits the pulse-echo field to a rectilinear region in front of the transducer. During the DAS beamforming, the RF signals from each receive channel were temporally matched filtered by the transmit excitation pulse and IQ-modulated using Hilbert transform, before getting delayed and summed at all imaging points. Furthermore, a spline interpolation has been used for calculating the amplitude at any time instance.

4.2.2 Spatially matched filters

Normal DAS focusing assumes that the spatial impulse response of the transducer is a delta function, and that the alignment can be done by merely delaying the responses. This is appropriate in the far-field for small element arrays and at the focus for single element transducers. However, in the near-field, the pulse-echo spatial impulse responses are different from a delta function (Jensen and Gori 2001).

As an alternative to dynamic receive focusing using DAS beamforming method, the signal from each channel of an array can be spatially matched filtered to align its output with that from the other channels (Jensen and Gori 2001; Kim et al. 2006; Yen 2013). When the RF signal is matched filtered, the output will have zero phase and all frequencies thereby add constructively to give the maximum SNR. The impulse response of the matched filter is the time inverted version of the pulse-echo spatial impulse response spectrum at a specific point. Since each element’s received signal is dependent on the
element location and the scatterer’s position, a new matched filter must be used depending on the element and the scatterer’s position. The SMF impulse response \( m_p(\vec{r}_{trn}, \vec{r}_{rcv}, t) \) is then given by (Jensen and Gori 2001):

\[
m_p(\vec{r}_{trn}, \vec{r}_{rcv}, t) = p_r(\vec{r}_{trn}, \vec{r}_{rcv}, -t)
\]

\[
p_r(\vec{r}_{trn}, \vec{r}_{rcv}, t) = v_{pe}(t) * h_t(\vec{r}_{trn}, \vec{r}_{rcv}, t) * h_r(\vec{r}_{rcv}, \vec{r}_{trn}, t),
\]

which is dependent on the transmitter location \( \vec{r}_{trn} \), the receiver element at \( \vec{r}_{rcv} \), and the electro-mechanical impulse response of the transducer \( v_{pe}(t) \). The impulse responses during transmission and reception are \( h_t(\vec{r}_{trn}, \vec{r}_{rcv}, t) \) and \( h_r(\vec{r}_{rcv}, \vec{r}_{trn}, t) \) for the combined response for all of the array elements including their focusing and apodization (Jensen 1991). The focusing is then performed by adding the matched filtered signals from all the elements for the different locations

\[
r_s(\vec{r}_i) = \sum_{j=1}^{M} \int_{t_{ij}}^{t_{ij}+\Delta T_{ij}} v_r(\vec{r}_j, t) p_r(\vec{r}_i, \vec{r}_j, t) dt,
\]

where \( r_i \) designates the relative distance of the imaging point to the transmitting element \( i \), \( r_j \) is the relative distance of the imaging point to the receiving element \( j \) of the transducer, \( t_{ij} \) is the start of the response, and \( \Delta T_{ij} \) is the duration of the matched filter. The convolution integral is replaced by a correlation, since the time reversal of the response is replaced by the time reversal in the convolution.

It should be noticed that (4.2) can be used for any image point, and that it is only necessary to process the point in the image that must be displayed on the screen, when using IQ-modulated RF data. This approach does not put any restrictions on the transducer geometry, excitation, focusing, apodization, or impulse response. In this work, Field II Pro (Jensen and Svendsen 1992; Jensen 1996, 2014) is used for calculations of the SMF coefficients, providing the arrays dimensions, excitation pulse, and the measured impulse response. During the SMF beamforming, the RF signals from each receive channel are Hilbert transformed, before getting match filtered.

4.3 Imaging with RCA 2-D arrays

To evaluate the volumetric imaging performance of the two prototyped CMUT and PZT RCA 2-D array transducer probes, the focusing ability of RCA 2-D arrays has to be studied compared to fully addressed 2-D arrays. Thereby, it is possible to investigate whether the two CMUT and PZT RCA probes attained those performance criteria or not. Principally, the best achievable lateral resolution of a given ultrasound system is defined by its two-way beam width at the focal depth using conventional focusing on both reception and transmission (Szabo 2014). However, in imaging with an RCA 2-D array, the focusing in transmit direction is independent from the receive direction, thus,
the spatial resolution in each direction can differ from the other direction depending on how well the focus lines are generated in each direction. In RCA 2-D arrays due to the perpendicular orientation of the transmit and receive apertures, only one-way focusing is possible in each lateral direction (Démoré et al. 2009; Yen 2013; Rasmussen et al. 2015).

The Fresnel approximation states that in the far-field of the transducer array, and at the focal distance of a focused transducer, the pressure field may be described by the Fourier transform of the transducer aperture. A finite array of transducer elements has an aperture \( A \), described by a simple rectangular window function along one lateral dimension. The Fourier transform of a rectangular function is the sinc function, which therefore describes the pressure field in that dimension. Denoting the size of this array along the \( x \)-dimension \( L_x \), the position along the array \( x \) (\( x = 0 \) being the center of the array), the wavelength of the ultrasound wave \( \lambda \), and the mass density of the medium \( \rho_a \), the pressure field at the depth \( z \) becomes (Szabo 2014):

\[
p_{x,\text{one-way}} = \mathfrak{F}[A] = \frac{L_x \sqrt{\rho_a}}{\sqrt{\lambda z}} \text{sinc} \left( \frac{L_x x}{\lambda z} \right),
\]

(4.3)

where \( \mathfrak{F} \) denotes the Fourier transform. It is assumed here that \( z \) is either at the focus of the transducer or in the far-field. The full-width at half-maximum (FWHM) of the sinc function is

\[
\text{FWHM}_{\text{one-way}} = \frac{1.208 \lambda z}{L_x} = 1.208 \lambda f_\theta.
\]

(4.4)

This shows that the lateral detail resolution for a given wavelength and depth scales with the array size. The subscript “one-way” is to emphasize that the FWHM is for focusing of only the transmit aperture (or only the receive aperture due to acoustic reciprocity). An RCA array can only perform one-way focusing in each lateral dimension, if conventional DAS beamforming is used, and its detail resolution is therefore defined by (4.4). As opposed to this, a 2-D matrix array can focus in each lateral dimension both in transmit and receive. The resulting pulse-echo field is proportional to the Fourier transform of the convoluted transmit and receive apertures (Karaman et al. 2009). If the same aperture is used for transmitting and receiving, the pulse-echo field along one dimension thereby becomes:

\[
p_{x,\text{two-way}} \propto \mathfrak{F}[A \ast A] = \mathfrak{F}[A] \mathfrak{F}[A] = (\mathfrak{F}[A])^2.
\]

(4.5)

The FWHM of two-way focusing is:

\[
\text{FWHM}_{\text{two-way}} = \frac{0.886 \lambda z}{L_x} = 0.886 \lambda f_\theta.
\]

(4.6)

The ratio between the resolution of one-way focusing and the resolution of two-way focusing is therefore
4.3. Imaging with RCA 2-D arrays

Thus, for the same aperture size, the FWHM of an RCA array is 36% larger than the FWHM of a two-way focused 2-D array. Based on the FWHM, the detail resolution for the two types of arrays will consequently be equal, if the side-length of the RCA array is increased by 36% relative to the fully addressed 2-D array. In Fig. 4.1, the resulting field from the one-way focused array (black), the two-way focused array (red), and the one-way focused array with a 36% larger side-length (orange) is plotted. It is seen that the FWHM of the two latter are indeed identical.

![Figure 4.1: Plot of the resulting field at the focal distance or in the far field originating from a one-way focused array (black), a one-way focused array with 36% larger aperture side-length (orange), and a two-way focused array (red). The two former are plotted using a normalized (4.3), while the latter uses (4.5). The Figure is taken from paper F.](image)

The ratio between the number of elements in an RCA array and a fully addressed 2-D matrix array with equal detail resolution is from the above derivation found to be

\[
\frac{\text{No. elem. in fully addr. array}}{\text{No. elem. in RCA array}} = \frac{N^2}{1.36 \cdot 2 \cdot N} = \frac{N}{2.72}.
\] (4.8)

To have the same lateral resolution for both fully addressed and RCA 2-D arrays, the number of row or column elements on an RCA array has to get increased only by a factor of \(1.208/0.886 = 1.36\), i.e., by a factor of \(2 \times 1.36 = 2.72\) for the total number of elements. For instance for a 2-D array with \(256 \times 256\) elements, row–column addressing corresponds to a reduction in the total number of channels by 99.6%, i.e., from 65,536
channels to 512 channels. Any \( N + N \) channel RCA array with \( N \geq 3 \) will, thus, achieve a better detail resolution than a fully addressed 2-D array with the same total number of channels. However, changing the aperture size will only affect the argument in the sinc function in (4.3), not the shape of the function. Hence, the normalized amplitudes remain unchanged, and so do the side-lobe levels relative to the main lobe level. This is seen in Fig. 4.1, where the two one-way focused arrays have side-lobe levels of equal magnitude. As a consequence of the squaring of the Fourier transform of the apertures given in (4.5), the amplitudes of the side-lobes are halved by a factor of two in dB when two-way focusing is performed. One measure of contrast is the side-lobe level (Szabo 2014). Therefore, a fully addressed 2-D array will have superior contrast performance relative to an RCA 2-D array.

Note that the above calculations are only strictly valid for a continuous wave emission at a single frequency (Szabo 2014), and as such they should only be seen as estimates of the imaging performance. Also, the estimates are made using an un-apodized aperture, and does therefore not take into account the effects of applying different apodization functions.

### Table 4.1: Transducer dimensional parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>CMUT</th>
<th>PZT</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements</td>
<td>62+62</td>
<td>62+62</td>
<td>–</td>
</tr>
<tr>
<td>Number of apodization region electrodes</td>
<td>4</td>
<td>4</td>
<td>–</td>
</tr>
<tr>
<td>Element pitch</td>
<td>270</td>
<td>270</td>
<td>µm</td>
</tr>
<tr>
<td>Element width</td>
<td>265</td>
<td>245</td>
<td>µm</td>
</tr>
<tr>
<td>Kerf</td>
<td>5</td>
<td>25</td>
<td>µm</td>
</tr>
<tr>
<td>Element length</td>
<td>24.84</td>
<td>24.84</td>
<td>mm</td>
</tr>
<tr>
<td>Length of apodization regions</td>
<td>4.05</td>
<td>4.05</td>
<td>mm</td>
</tr>
<tr>
<td>Array outer dimensions (square)</td>
<td>26.3</td>
<td>26.3</td>
<td>mm</td>
</tr>
</tbody>
</table>

### 4.4 Equipment and Measurement Setup

The dimensional parameters of both arrays are found in Table 4.1. The probes are plugged into the experimental ultrasound scanner, SARUS (Jensen, Holten-Lund, et al. 2013). The measured IQ-modulated RF signals are beamformed using two MATLAB (MathWorks Inc., Massachusetts, USA) implementations of the DAS and SMF beamformers described in Section 4.2.

To evaluate the imaging performance of both probes, several ultrasound phantoms are used. An iron needle with diameter of 300 µm facing towards the transducer along its central axis, was used as a point target in a water bath for characterizing the 3-D point spread function (PSF). To evaluate the FWHM and the cystic resolution (CR) as a function
of depth, a geometrical copper wire phantom was used as line targets, where wires were located at different depths with 1 cm spacing. The wire grid phantom has three columns separated by 1 cm and each has 13 rows of wire.

A tissue mimicking phantom with cylindrical anechoic targets, model 571 from Danish Phantom Design (Frederikssund, Denmark) with attenuation of 0.5 dB/(cm MHz) was used for SNR and contrast measurements.

The transmit pressure intensity and temperature measurements of the probes were carried out using the AIMS III intensity measurement system (Onda Corporation, Sunnyvale, California, USA) connected to the experimental research scanner SARUS (Jensen 2016; Jensen, Rasmussen, et al. 2016). The mechanical index (MI) and the spatial-peak-temporal-average intensity ($I_{spta}$) are both measured for each of the transducers and for both imaging sequences. The current prototyped probes do not have the required permissions to be used on humans, therefore no in vivo data have been acquired.

### Table 4.2: Setup configuration for the simulations and measurements

<table>
<thead>
<tr>
<th>SAI sequences</th>
<th>single element</th>
<th>$f_# = -1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pulse repetition frequency</td>
<td>5</td>
<td>5</td>
</tr>
<tr>
<td>No. of active elements in Tx</td>
<td>1</td>
<td>62</td>
</tr>
<tr>
<td>Scan depth (max range)</td>
<td>14</td>
<td>14</td>
</tr>
<tr>
<td>Emission center frequency</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Sinusoid emission cycles</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Focus in transmit</td>
<td>0</td>
<td>-16.8</td>
</tr>
<tr>
<td>Focus in receive</td>
<td>Dynamic</td>
<td>Dynamic</td>
</tr>
<tr>
<td>Tx electronic apodization</td>
<td>-</td>
<td>Hann.</td>
</tr>
<tr>
<td>Rx electronic apodization</td>
<td>Rect.</td>
<td>Rect.</td>
</tr>
<tr>
<td>Sampling frequency</td>
<td>70</td>
<td>70</td>
</tr>
<tr>
<td>Tx voltage for PZT &amp; CMUT</td>
<td>±75</td>
<td>±75</td>
</tr>
<tr>
<td>DC bias voltage for CMUT</td>
<td>190</td>
<td>190</td>
</tr>
</tbody>
</table>

### 4.5 Defocused SAI Sequence Choice of Parameters

To accomplish the best performance, the location and number of virtual sources have to be optimized in a trade-off between spatial resolution, field-of-view, and SNR. A parameter study is carried out over the maximum steering angle for placing the 62 virtual line sources
and the transmit $f_{#}$ of the defocused SAI sequence to image a point scatterer at a depth of 20 mm in front of the array. The lateral FWHM and CR values of the beamformed PSFs for maximum steering angles in range of $\pm 10^\circ$ to $\pm 60^\circ$ and transmit $f_{#}$s from $-3$ to $-0.5$ are shown in Fig. 4.2. The criteria to choose the best parameters is to have the best contrast and spatial resolutions for the lowest steering angle and transmit $f_{#}$. As a trade-off between contrast and spatial resolutions, the maximum steering angle of $\pm 30^\circ$ and transmit $f_{#} = -1$ are chosen for the defocused SAI sequence. The parameters of both SAI sequences are listed in Table 4.2.

![Figure 4.2](image)

Figure 4.2: The figures are illustrating the lateral FWHM and CR as function of steering angle and the transmit $f_{#}$ for a point target located along the central axis at depth of 20 mm. The simulated RF data are generated using Field II Pro (Jensen and Svendsen 1992; Jensen 1996, 2014) and beamformed with DAS method. As a trade-off between contrast and spatial resolution, a steering angle of $\pm 30^\circ$ and transmit $f_{#} = -1$ (indicated with a blue marker) are chosen for the defocused SAI sequence. The Figure is taken from paper F.
4.6 Imaging performance assessment

Fig. 4.3 illustrates three cross-planes (azimuth, elevation, and C-plane) of the volumetric pulse-echo beam patterns measured with both probes in comparison with Field II simulations using the DAS beamformation method. The iron needle faces towards the transducer and it is imaged with the single element transmissions SAI sequence. A Hanning apodization is applied over the receive and synthesized transmit apertures. The measured pulse-echo impulse responses of both probes (Engholm et al. 2016) as well as...
the diameter of the needle are taken into account for the simulations by imaging a disk consisting of 500 point targets to represent the tip of the needle.

Table 4.3: FWHM and CR of simulated and measured 3-D PSFs

<table>
<thead>
<tr>
<th>CR</th>
<th>Simulation</th>
<th>Measurement</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>CMUT</td>
<td>PZT</td>
</tr>
<tr>
<td></td>
<td>mm</td>
<td>mm</td>
</tr>
<tr>
<td><strong>DAS</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>R(_{6\text{dB}})</td>
<td>0.37</td>
<td>0.35</td>
</tr>
<tr>
<td>R(_{12\text{dB}})</td>
<td>1.03</td>
<td>0.86</td>
</tr>
<tr>
<td>R(_{20\text{dB}})</td>
<td>3.06</td>
<td>2.9</td>
</tr>
<tr>
<td>R(_{35\text{dB}})</td>
<td>4.17</td>
<td>4.01</td>
</tr>
<tr>
<td><strong>SMF</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>R(_{6\text{dB}})</td>
<td>0.37</td>
<td>0.36</td>
</tr>
<tr>
<td>R(_{12\text{dB}})</td>
<td>0.7</td>
<td>0.64</td>
</tr>
<tr>
<td>R(_{20\text{dB}})</td>
<td>2.7</td>
<td>2.52</td>
</tr>
<tr>
<td>R(_{35\text{dB}})</td>
<td>3.7</td>
<td>3.61</td>
</tr>
<tr>
<td><strong>FWHM</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Axial</td>
<td>0.44</td>
<td>0.45</td>
</tr>
<tr>
<td>Azimuth</td>
<td>0.77</td>
<td>0.8</td>
</tr>
<tr>
<td>Elevation</td>
<td>0.78</td>
<td>0.8</td>
</tr>
</tbody>
</table>

Note the secondary lobes after the main lobe along the axial direction for both probes at the range of 20 mm to 22 mm of depth, which are due to the internal reflections from the RF shielding foils covering the arrays. These secondary reflections are visible in the impulse responses of the probes (Engholm et al. 2016). Among these secondary echoes are the edge echoes, originating from the either ends of the line elements and due to imperfect static roll-off apodization they were not suppressed below 40 dB (Christiansen et al. 2015; Rasmussen et al. 2015). The way to classify them is only based on the timing of their occurrences. The same RF data acquired in Fig. 4.3 are used for the SMF beamforming method. FWHM and the CR of the simulated and measured 3-D PSFs are
listed in Table 4.3 for both DAS and SMF beamforming methods.

Due to the similar dimensions of the probes, the lateral and elevation FWHM values are close to each other, however the axial FWHM values of the CMUT probe are 17% smaller than the PZT probe. That is due to the higher bandwidth of the CMUT probe compared with the PZT probe, which also causes the CMUT probe to have larger CR values compared with the PZT probe. Although both DAS and SMF beamformation techniques have resulted in similar lateral and elevation FWHM values, CR values have improved using SMF method. During SMF beamforming with broadband RF signals, a 2-D spatio-temporal matched filtering is applied, which involves not only a lateral convolution in space but also a temporal convolution in the axial direction. The temporal convolution elongates the pulse-echo response which, results in worse axial spatial resolution, consequently leading to a larger speckle size in the axial direction.

To study the focusing abilities of both probes as a function of depth, a wire grid phantom is used to quantify the line spread function (LSF) characteristics of both probes, since the echoed signals from the needle are too low at higher depths. The diameter of the wires is 200 µm, which is smaller than a wavelength, considering the speed of sound in water and therefore can be used as line targets. The phantom has three columns of wire, which are separated by 10 mm in axial and lateral directions. Both of the SAI sequences are used to image the wire phantom with the PZT and CMUT probes by placing them centered around the middle column. A volume region of 26×10×5 mm$^3$ centered around each beamformed wire in the middle column is used for the LSF characteristics evaluation as a function of depth. Fig. 4.4(a) and Fig. 4.4(b) illustrate the calculated FWHM in the lateral and axial directions for both of the SAI sequences with the PZT and CMUT probes at different depths. Using the SAI sequence with transmit $f_# = -1$ has increased the axial FWHM values compared with the SAI sequence with single element at a time. Likewise what was observed previously with the 3-D PSFs, here for the LSFs, the CMUT probe has smaller FWHM values in the axial direction comparing with the PZT probe. Although the lateral FWHM values increase linearly by increasing the depth similar to (4.4), the SMF beamforming has lowered the FWHM values for the SAI sequence with transmit $f_# = -1$. As it is illustrated in Fig. 4.5(a) and Fig. 4.5(b), CR values increase by increasing the depth almost linearly for both probes.

To calculate the SNR of the SAI sequences for each of the probes, a volume region of a tissue mimicking phantom with no cyst was imaged 20 times. The measured SNRs for the SAI sequences with both probes are illustrated in Fig. 4.8. It is worth to mention that by using only the SAI sequence with single element transmissions, the PZT probe has a penetration depth of around 14 cm, whereas the CMUT probe can only penetrate down to 10 cm. However by using all the elements in transmit in the SAI sequence with transmit $f_# = -1$, the CMUT probe has a penetration depth of around 14 cm, whereas the PZT can penetrate down to almost 25 cm. Using a SAI sequence with transmit $f_# = 1$ has increased the penetration depth for the CMUT probe down to 15 cm, whereas the PZT can penetrate down to almost 30 cm.

Due to the perpendicular orientation of the transmit and the receive directions, the
field-of-view of RCA arrays is limited to the forward looking region in front of the array, e.g., $26 \times 26 \text{ mm}^2$ for these probes. Two cross-planes (azimuth and C-plane) are shown in Fig. 4.6 and Fig. 4.7 at a dynamic range of 40 dB from a volume of $26 \times 26 \times 80 \text{ mm}^3$ acquired with both probes using a cyst phantom and beamformed with the DAS and SMF beamforming methods. Only down to 80 mm are shown, due to the limited space in the paper. The origin corresponds to the center of the transducer surface. Data were acquired with both probes using the SAI sequence with a transmit $f_h = -1$. The hollow cysts are located along a $10^\circ$ tilted plane and therefore, the lower hollow cysts are not visible clearly at regions farther from the array. See the videos in the supplementary materials for the full sized DAS beamformed volumes of the cyst phantom using the SAI sequence with transmit $f_h = -1$ from paper F. The CNR measure is calculated in a cylindrical region centered at each of those large hollow cysts with a diameter of 8 mm. The calculated CNR values for each imaging sequence with each probe are shown in the Fig. 4.9. Due to the higher generated pressure with the PZT probe compared with the CMUT probe, the CNR of the PZT probe is more than 3 times larger than the CMUT probe.

4.7 Intensity and Temperature Measurement

MI and $I_{spta}$ are measured for the SAI sequence with a transmit $f_h = -1$ at a pulse repetition frequency of 5 kHz, since it uses all row elements in transmit and thereby has the largest emitted energy. For the PZT probe they are $\text{MI} = 0.67$ and $I_{spta} = 0.53 \text{ mW/cm}^2$. The measured $\text{MI} = 0.06$ and $I_{spta} = 0.18 \text{ mW/cm}^2$ of the CMUT probe and the PZT probe are both within the FDA safety limits (FDA 2008; Jensen 2016; Jensen, Rasmussen, et al. 2016). For the SAI sequence with a transmit $f_h = 1$ at a pulse repetition frequency of 5 kHz, the measured $\text{MI} = 0.88$ and $I_{spta} = 5.5 \text{ mW/cm}^2$ of the PZT probe, and $\text{MI} = 0.13$ and $I_{spta} = 0.55 \text{ mW/cm}^2$ of the CMUT probe are also both within the FDA safety limits. Therefore, both sequences can be scaled to a larger excitation voltage for the in vivo measurements before reaching the FDA limits, thus achieving a higher penetration depth.

Another criteria that has to be addressed is the heating of the probe, which has to be within the FDA safety limits (FDA 2008). The linear voltage regulators used for the amplifiers in the prototype probes are dissipating power and generating waste heat. Therefore, the temperature rise in the probe is a combination of heating produced by the linear regulators as well as the transducer arrays. To separate the heating caused by the amplifiers in the handle from the transducers themselves, two spots on the probes were measured for temperature changes in a still air environment. One sensor located at the sole of the probe and the other sensor was located on the body of the probe, where the amplifiers are located. These two sensor measurements are shown in Fig. 4.10 for situations when only the amplifiers are turned on, but without transmitting any excitation pulse, and also when the transducers are transmitting the excitation pulses with amplifiers turned on. Fig. 4.10 illustrates the temperature rise for both probes using all of the imaging sequences firing sequentially after each other without any delay between each sequence.
4.8. Discussion

It is important to note that the effect of roll-off apodization is outside of the rectilinear imaging field-of-view of the array and therefore will not affect the lateral resolution and the FWHM measurements should be comparable to (4.3), since no electronic apodization has been applied. For the 3-D PSF measurement at the depth of 19.8 mm as shown in Fig. 4.3, but without using electronic apodization, the simulated lateral FWHM values listed in Table 4.3 were slightly larger than 0.71 mm that is estimated from (4.3). Using SMF increased the lateral and axial FWHM values for both CMUT and PZT probes, instead the CR has improved compared to DAS. It was expected from the SMF beamforming method to improve the contrast while worsening the spatial resolution. The effect of using SMF was more pronounced with the CMUT probe compared to the PZT probe, which can be attributed to the higher bandwidth of the CMUT probe for the spatio-temporal convolution with the spatial impulse response in the SMF beamformation method. Using DAS, the measured lateral FWHM values were larger compared to the simulations by 15% for the PZT probe and by 20% for the CMUT probe, yet through using SMF, the increments compared to simulation were by 17% for both probes.

The measured CR values were larger than the simulation values for both CMUT and PZT probes, using DAS and SMF methods. No significant difference between DAS and SMF has been observed for CR values for the PZT probe. However, for the CMUT probe,
SMF outperformed DAS by 22% for $R_{6\text{dB}}$, and there was no significant difference for $R_{12\text{dB}}$ and $R_{20\text{dB}}$.

In Fig. 4.4(a) the larger axial FWHM values for the defocused SAI sequence compared to the single element transmissions can be attributed to the phase delay difference of the elements in the PZT probe discussed in Part I (Engholm et al. 2016), which leads to imperfect transmit and receive focusing. Although the receive delay can be compensated after the measurements, the transmit delays can only be compensated during the measurements. The measured lateral FWHM values are close to the estimated lateral FWHM values from (4.3), as indicated by a red dotted line in Fig. 4.4(a) and Fig. 4.4(b). The best lateral resolution is achieved with the single element transmissions SAI sequence using DAS. In Fig. 4.5(a) and Fig. 4.5(b) by using SMF, the CR values have been improved for both CMUT and PZT probes for the single element transmission SAI sequence. However, for the CMUT probe, using the defocused SAI sequence, no significant difference between DAS and SMF has been observed. For the PZT probe, on the other hand, using SMF for the defocused SAI sequence has worsened the CR compared to DAS. For both probes, by using the SMF method, the seventh wire at the depth of 154\(\lambda\) had a lower main-lobe to side-lobe ratio compared to DAS, thus increased the CR value at that depth.

Theoretically, transmitting with row element and receiving with column elements should image exactly the same rectilinear volume as transmitting with column element and receiving with row elements. However, because of the manufacturing process, the sensitivity of row and column elements might be slightly different. For the PZT probe the difference is small, since transmitting with row elements and receiving the echoes with column elements or vice versa, are similar as shown in Fig. 4.11. For the CMUT probe, because of a capacitive substrate coupling of the bottom electrodes, as discussed in the previous chapter, the receive sensitivity of the bottom elements is lower and therefore in our imaging set-up, the elements with higher receive sensitivity, i.e., the top elements, are chosen for receiving, while the bottom elements are used for transmitting. As a result, for the CMUT probe by coherently compounding the two volumes, assuming no movement, the spatial resolution and contrast will degrade.

Using single element for transmission, the CMUT probe has a lower penetration depth on a tissue mimicking phantom compared to the PZT probe. On the contrary by using all the elements in transmit, both probes penetrate down to 15 cm, with PZT probe even down to 30 cm. In the same way, the CNR values increased, when using all the elements in transmit. Placing the virtual focus lines in front of the transducer would increase the penetration depth further, but larger transmit $f_{\#}$ values will degrade the spatial resolution. For SAI sequences with the transmit focus in front of the array, the transmitted acoustic energy is focused along a line in contrast to a point by using fully addressed 2-D arrays and therefore for the same size of the 2-D arrays, the MI and the $I_{\text{spta}}$ are quite lower for RCA 2-D arrays.
4.9 Conclusion

In this chapter, the imaging performance of two in-house prototyped 62+62 RCA 2-D array probes fabricated in CMUT and PZT transducer technologies were demonstrated quantitatively and comparatively. Using SAI technique both probes were able to image down to 14 cm at a volume rate of 88 Hz. DAS and SMF beamforming methods were both able to perform dynamic transmit-receive focusing throughout the rectilinear field-of-view. The performance of both probes was evaluated through simulation and experiments. Results show that both probes can image a rectilinear volume in front of the transducer successfully. Integrated hardware apodization along each line-element effectively removed the ghost echoes without altering the main echo’s beam width. It was demonstrated that volumetric imaging with equipment in the price range of conventional 2-D imaging is possible. Both probes were prototypes and not optimized, which limited the imaging performance. Future work will focus on configuring the probes for better performance through adjusting the DC bias voltage for the CMUT probe for achieving higher penetration and using a better shielding method for both probes to eliminate the reflections within the probes.

The results of this study have demonstrated the promising potentials of RCA 2-D arrays, which are their low channel count, low MI and \( I_{spta} \) values, and high penetration depth compared with fully addressed 2-D arrays. However, there are still a number of challenges that must be addressed for the technology to be a realistic alternative to the existing matrix probes. In RCA 2-D arrays only one-way focusing is possible in each dimension however, a strategy that might be employed to remedy this is by using advanced signal processing algorithms in SMF beamforming.

The second major issue is that the RCA arrays can only emit energy directly below the array and in a cross-shape to the sides. For applications such as cardiac imaging, it is relevant to have a probe with a small foot-print capable of phased array imaging, such that the heart can be visualized through the ribs. True volumetric phased array imaging is possible with RCA arrays, provided that the array is double curved to spread the energy during transmit. An in-depth study of the possibilities in this approach will be presented on the next chapter.

Due to the low channel count of the RCA 2-D arrays, it is possible to fabricate 2-D arrays with larger aperture size, which might be beneficial for abdominal scans. This is only a small selection of the challenges involved in developing a commercially viable RCA probe. However, the results so far demonstrate that the technology is a realistic alternative to state-of-the-art matrix probes, especially as a low-cost alternative.

References


DOI: http://scitation.aip.org/content/asa/journal/jasa/120/2/10.1121/1.2214393 (cit. on pp. 75, 76).
Figure 4.4: Axial and lateral FWHM of the PZT (a) and CMUT (b) probes as a function of depth for DAS beamformed images of the wire phantom. The solid lines correspond to lateral FWHM (left axis) and the dashed lines correspond to axial FWHM (right axis). $\lambda$ was calculated for soft tissue. The red dotted line shows the estimated lateral FWHM based on the Fresnel approximation in (4.4). The Figure is taken from paper F.
Figure 4.5: Cystic resolution for $R_{6\text{dB}}$ and $R_{12\text{dB}}$ radius of the PZT (a) and CMUT (b) probes as a function of depth. Calculated over beamformed images of the wire phantom. The solid lines correspond to $R_{6\text{dB}}$ (left axis) and the dashed lines correspond to $R_{12\text{dB}}$ (right axis). $\lambda$ was calculated for soft tissue. The Figure is taken from paper F.
Figure 4.6: Volumetric imaging of a tissue mimicking phantom using both PZT and CMUT probes and beamformed with the DAS method. Two cross-planes (azimuth and C-plane) are shown from a volume of $26 \times 26 \times 85$ mm$^3$ at a dynamic range of 40 dB. Data were acquired with the PZT probe using the SAI sequence with 62 single element emissions (PZT: (a) and (e)), (CMUT: (c) and (g)), and also using the SAI sequence with transmit $f_{\theta} = -1$ (PZT: (b) and (f)), (CMUT: (d) and (h)). The C-planes are at the depth of 30 mm. For the full-sized volumes of the SAI sequence with transmit $f_{\theta} = -1$ see the videos in the supplementary materials from paper F.
Figure 4.7: Volumetric imaging of a tissue mimicking phantom using both PZT and CMUT probes and beamformed with the SMF method. Two cross-planes (azimuth and C-plane) are shown from a volume of $26 \times 26 \times 85 \text{ mm}^3$ at a dynamic range of 40 dB. Data were acquired with the PZT probe using the SAI sequence with 62 single element emissions (PZT: (a) and (e)), (CMUT: (c) and (g)), and also using the SAI sequence with transmit $f_\# = -1$ (PZT: (b) and (f)), (CMUT: (d) and (h)). The C-planes are at the depth of 30 mm.
Figure 4.8: SNR of the PZT and the CMUT probes on a region of a tissue mimicking phantom with no cyst and acoustical attenuation of 0.5 dB/(cm MHz). The dashed lines are linearly fitted to each curve. $\lambda$ was calculated in soft tissue. The blue line indicates the maximum depth of the measured data. The Figure is taken from paper F.
Figure 4.9: CNR values measured with both PZT and CMUT probes as a function of depth. Calculated over DAS beamformed images of the cyst phantom acquired with both PZT and CMUT probes. $\lambda$ was calculated in soft tissue. The Figure is taken from paper F.
Figure 4.10: Temperature values of the PZT and the CMUT probes as a function of time for both cases, when only the amplifiers in the handle are turned on, and when both the amplifiers in the handle and the ±75 volts amplifiers are turned on. The whole imaging sequences has been running in still air while the temperature sensors were mounted on the center of the probes soles and also on the probes handles. The Figure is taken from paper F.
Figure 4.11: Two azimuth planes from volumes of $26 \times 26 \times 15 \text{ mm}^3$ acquired with the PZT probe using single element transmission SAI sequence at a dynamic range of 40 dB are shown when: (a) row elements transmit and column elements receive, and (b) column elements transmit and row elements receive. The Figure is taken from paper F.
CHAPTER 5

Curvilinear Imaging with Row-Column Addressed Arrays

Previous chapters E and F evaluated row–column addressing scheme based on simulation and measurements. The issue with row–column addressing scheme is that it can only emit acoustic energy directly below the array and in a cross-shape to the sides. Although enlarging the aperture size would cover a larger volume, however for applications such as cardiac imaging, it is relevant to have a probe with a small foot-print capable of phased array imaging, such that the heart can be visualized through the ribs.

This chapter investigates if true volumetric phased array imaging is possible with row–column-addressed (RCA) arrays, provided that the array is double curved to spread the energy during transmit. An alternative to manufacturing a double curved array would be to use a diverging lens instead. An in-depth study of the possibilities in this approach will be presented in this chapter.

The analyses and results included in this chapter are based on the work presented in the papers G on page 203 and H on page 209. First, an overview of beamforming with RCA 2-D arrays is presented, afterwards a theoretical comparison between imaging with RCA and fully addressed 2-D arrays is given. Then, the results of the measurements are presented, followed by a discussion and a conclusion.

5.1 Introduction

An $N \times N$ element 2-D array can be operated utilizing only $2N$ connections, when a row–column or cross-electrode addressing scheme is used (Morton and Lockwood 2003; Démoré et al. 2009; Seo and Yen 2009; Zemp et al. 2011; Chee et al. 2014; Sampaleanu et al. 2014; Rasmussen et al. 2015). This is contrary to the $N^2$ connections needed, when conventionally addressing the elements. In general, a RCA array is a 2-D matrix array, which is addressed via its row- and column indices. Effectively, it consists of two 1-D arrays arranged orthogonal to each other as shown in Fig. 5.1. As an example, for a 256 $+$ 256 RCA array, a 2-D matrix array of equivalent size would have 65,536 elements, over a factor of 7 more than the current state-of-the-art X6-1 PureWave xMATRIX probe from Phillips (Eindhoven, Netherlands) having 9212 elements (Phillips 2015). This exhibits the potential of having very large RCA 2-D arrays with low channel count and real-time capabilities.
It has been demonstrated in several studies that row–column technology is a realistic alternative to the state-of-the-art matrix probes, especially as a low-cost alternative (Morton and Lockwood 2003; Démoré et al. 2009; Seo and Yen 2009; Zemp et al. 2011; Chee et al. 2014; Sampaleanu et al. 2014; Christiansen et al. 2015; Rasmussen et al. 2015). However, one major issue with the RCA arrays is that they can only emit acoustic energy...
directly below the array and in a cross-shape to the sides. Therefore, imaging can only be done in a rectilinear region in front of the array. For applications such as cardiac imaging, it is relevant to have a probe with a small foot-print capable of phased array imaging, such that the heart can be visualized through the ribs. True volumetric phased array imaging is possible with RCA arrays, provided that the array is double curved to spread the energy during transmit (Démoré et al. 2009). However, manufacturing curved transducer elements is challenging for both capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technologies. Another approach to spread the acoustic energy is by using a double curved diverging acoustic lens on top of the RCA array (Joyce and Lockwood 2014). Using a lens makes it easier to fabricate curved arrays, as it is not needed to manufacture curved elements, and also making a lens is a well-tested technology. An in-depth study of the possibilities in this approach is therefore the main goal of this study.

In this chapter, the curvilinear volumetric imaging performance of an RCA array equipped with a diverging lens is investigated based on Field II (Jensen and Svendsen 1992; Jensen 1996) simulations. The quality assessments of the B-mode images, i.e., spatial resolution and contrast resolution, are carried out based on the simulations using synthetic aperture imaging (SAI) technique. The SAI sequence is designed for imaging down to 14 cm of depth.

This chapter is organized as follows: The current limitations with flat RCA arrays are discussed in Section 5.2.1, and different approaches to disperse the acoustic energy are introduced in Section 5.2.2. Section 5.2.3 presents an overview of the delay-and-sum (DAS) beamformation with a double curved RCA, and the utilized SAI sequence is explained in Section 5.3.1. In Section 5.3.2, the imaging quality assessment measures are explained. In Section 5.3.3, a detailed overview of the simulation setup is presented. Section 5.4 explains and discusses the simulation results with an RCA 2-D array equipped with a diverging lens. The final section concludes the chapter with suggestions for future work.

5.2 RCA 2-D Arrays

5.2.1 Flat RCA 2-D Arrays

In 3-D ultrasound imaging with flat RCA 2-D arrays, the two orthogonal 1-D transmit and receive arrays are both used for focusing in the lateral and elevation directions separately. Each of the two 1-D arrays can electronically focus in one lateral dimension, when delays are applied to the elements in the array. One of the 1-D arrays is used to transmit ultrasound into the object of interest. For example, the transmit array is able to focus the beam in the x- and z-directions, whilst no electronic transmit focusing can be performed in the y-direction. As a result, the emitted ultrasound is focused along a line parallel to the y-direction. By adjusting the delays on the transmit elements, this focal line may be translated to any position in the xz-plane. The orthogonal 1-D array then receives the
Figure 5.2: The static roll-off apodization layout is applied to either ends of the line-elements of the array, e.g., here a 16+16 RCA array with roll-off apodization regions is shown. The central region, shown in black, has an apodization value of one. Each roll-off region is connected to each line-element.

DAS beamformers usually assume the geometry of the sound sources and receivers to be points. However, by row–column addressing the elements on a 2-D matrix array, each row and column is acoustically equivalent to a line-element. Furthermore, the emitted wavefront of a line-element has the shape of a cylinder, i.e., it is a plane wave in the plane aligned along the line-element and a circle arc in the plane orthogonal to the line-element. Assuming the geometry of the line-elements to be points is therefore a poor approximation. A more accurate approximation assumes the line-elements to be line segments instead of points, and the beamformer should calculate the distances between line-elements and the point (Rasmussen et al. 2015). However, the long length of the line-elements results in prominent edge effects (Démoré et al. 2009; Rasmussen et al. 2015). These edge effects are due to the limited size of the aperture and originates from both ends of the volume.
5.2. RCA 2-D Arrays

Figure 5.3: Relative transmit and receive pressure fields at radial distance of 80 mm for azimuth and elevation steering angles from $-45^\circ$ to $45^\circ$. The imaging area is the intersection of these two fields, which is (a) the rectilinear forward-looking box, and (b) the curvilinear forward-looking region in front of the transducer using a lens with $f_\# = -1$.

The pulse-echo field for the flat RCA 2-D array is limited to the forward looking rectilinear region in front of the transducer as shown in Fig. 5.3(a), due to the perpendicular orientation of the transmit and receive fields. The relative orthogonal transmit and receive pressure fields at the depth of 80 mm are shown when steering the beam to the sides. Both transmit and receive beams were steered by $\pm 45^\circ$. When the horizontal array is used as a transmit array, it can steer the transmit beam in the $z$-$x$ plane, and at the same time the vertical array is receiving in the $z$-$y$ plane. Only the region indicated by white dashed lines, which is the intersection of transmit and receive pressure fields, can be imaged at any depth with an acceptable dynamic range.

5.2.2 Curved RCA 2-D Arrays

Using a double-curved RCA 2-D array can extend the volumetric imaging field-of-view (FOV) to a curvilinear region. To spread the acoustic energy of a line-element curvilinearly along its larger dimension, it has to be curved to generate a diverging wave. The defocusing of the waves can be made by using a fixed electronic delay profile along each flat line-element, similar to a fixed first stage in micro-beamforming with 2-D arrays (Savord and
Figure 5.4: Reduction in the pulse-echo energy using a diverging lens relative to a flat transducer. The points are located on a line at 80 mm away from the surface of the transducer.

Solomon 2003). Another approach is to use a double curved diverging acoustic lens on top of the flat RCA array (Joyce and Lockwood 2014).

A concave diverging lens could be designed with a material, which has lower speed of sound compared to the human tissue. It will have a higher thickness around the corners and the sides of the array, and less thickness close to center of the array. Alternatively, a convex diverging lens can be made from a material with a higher speed of sound compared to the human tissue, which is preferred for a better contact surface. A flat diverging lens also can be made by using a combination of two different materials, one with higher and other one with lower speed of sound compared to the human tissue.

In Fig. 5.4 the pulse-echo energy as a function of lateral position for different lens f-numbers ($f_#$) is illustrated on an RCA array. The $f#$ is defined as a ratio between the focal distance to the lens diameter. The pulse-echo energy drops by moving away from the forward looking region of the array. Almost at around $8^\circ$ steering angle the pulse-echo energy drops by 40 dB, when no lens is used. However, by using a diverging acoustic lens on top of an RCA 2-D array, a larger FOV can be illuminated. The FOV can be adjusted by using different $f#$ values for the lens. By using a diverging lens with $f# = -1$, the overlapped transmit and receive region increases to about ±30° in both directions as shown in Fig. 5.3(b) compared to Fig. 5.3(a).
Note that, for the same aperture size, lower $f_{\#}$ values for the lens corresponds to larger thicknesses of the lens and therefore the attenuation becomes higher through the lens material. Thus, there is a trade-off between FOV and attenuation. For example the delay profile can be in a range of $0 \mu s$ to $3.5 \mu s$ for a lens with $f_{\#} = -0.7$ and a speed of sound of $1400 \text{ m/s}$, which corresponds to a thickness range of $0 \text{ mm}$ to $5 \text{ mm}$. A suitable material for a lens could be Sylgard 160 (PDMS) with a density of $1580 \text{ kg/m}^3$ and a speed of sound of $950 \text{ m/s}$ and attenuation of $0.4 f^{1.4} \text{ dB/(cm MHz)}$, where $f$ is the operating frequency in MHz. Therefore, for an operating frequency of $3 \text{ MHz}$ the maximum attenuation is $6.14 \text{ dB}$ at the largest thickness (Chang et al. 2014). This might be compensated by doubling the amplitude of the excitation pulse.

Figure 5.5: The time-of-flight of a wavefront is given by the shortest distance from the source $s_m$ to the point being focused $p$ and back to the receiving element $r_n$, divided by the speed of sound.

### 5.2.3 DAS Beamforming with Curved RCA 2-D Arrays

The time-of-flight (ToF) of a wavefront is given by the shortest distance from the arc source $s_m$ to the point being focused $p$, and back to the receiving element $r_n$, divided by the speed of sound. Using the notations from Fig. 5.5 this can be written as:
where \( c \) is the speed of sound in the medium, \( n \) is an index from 1 to the number of receive line elements \( N \), and \( m \) is the emission index. The function \( d(.,.) \) calculates the shortest distance between an arc and a point in space, which will be defined in the remainder of this section.

The arc segment from point \( a \) to point \( b \) with center \( c \) is termed \( \widehat{ab} \) assuming the center at the origin. This is illustrated in Fig. 5.6. The projection of point \( p \) onto the plane passing through the arc \( \widehat{ab} \) is termed \( p' \) and is determined by the usual equation for projection. To determine if the vector \( \hat{c}p' \) is in between vector \( \hat{c}a \) and vector \( \hat{c}b \), we define the normalized cross products \( \hat{l}_a \) and \( \hat{l}_b \) as

\[
\hat{l}_a = \frac{cp' \times ca}{\|cp'\|\|ca\|}, \\
\hat{l}_b = \frac{cp' \times cb}{\|cp'\|\|cb\|}.
\]

Depending on the location of the point \( p \), vectors \( \hat{l}_a \) and \( \hat{l}_b \) can be either \( \hat{j} \) or \( -\hat{j} \), where \( \hat{j} \) is the unit vector of the \( z \)-axis. \( \hat{l}_a \) and \( \hat{l}_b \) have different signs, when \( \alpha_1 \leq \phi \leq \alpha_2 \) and same sign, when \( \alpha_2 \leq \phi \) or \( \phi \leq \alpha_1 \), where \( \alpha_1 \), \( \alpha_2 \), and \( \phi \) are the angles between the \( x \)-axis and vectors \( ca \), \( cb \), and \( cp \), respectively, as shown in Fig. 5.6.

When \( \hat{l}_a \) and \( \hat{l}_b \) have different signs, i.e. \( \hat{l}_a = \hat{j} \) and \( \hat{l}_b = -\hat{j} \), or \( \hat{l}_a = -\hat{j} \) and \( \hat{l}_b = \hat{j} \), the standard formula for the distance between an arc and a point can be used:

\[
d = \sqrt{\|pp'\|^2 + (\|cp'\| - R)^2},
\]

where \( R \) is the curvature of the arc and equals to \( \|ca\| \) or \( \|cb\| \).

When \( \hat{l}_a \) and \( \hat{l}_b \) have the same signs, i.e. \( \hat{l}_a = \hat{j} \) and \( \hat{l}_b = \hat{j} \), or \( \hat{l}_a = -\hat{j} \) and \( \hat{l}_b = -\hat{j} \), the shortest distance from the arc segment to the point is the distance from the closest end of the arc segment (\( a \) or \( b \)) to the point \( p \). The following therefore determines the minimum distance between the point \( p \) and the arc segment \( \widehat{ab} \):

\[
d(\widehat{ab}, p) = \begin{cases} \sqrt{\|pp'\|^2 + (\|cp'\| - R)^2} & \text{if } \hat{l}_b = -\hat{j} \text{ and } \hat{l}_a = \hat{j} \\ \|ap\| & \text{if } \hat{l}_b = -\hat{j} \text{ and } \hat{l}_a = -\hat{j} \\ \|bp\| & \text{if } \hat{l}_b = \hat{j} \text{ and } \hat{l}_b = \hat{j} \end{cases}
\]

(5.5)
Using (5.5), the distances \( d(\mathbf{s}_m, \mathbf{p}) \) and \( d(\mathbf{r}_n, \mathbf{p}) \) can now be determined. The focused signal at point \( \mathbf{p} \) is calculated by summing all receive signals at the time instances given by (5.1):

\[
z_m(\mathbf{p}) = \sum_{n=1}^{N} a_{\text{elec}}(n, \mathbf{p}) y_{m,n}(\text{ToF}_m(n, \mathbf{p})), \tag{5.6}
\]

where \( N \) is the number of receive elements, \( a_{\text{elec}} \) is the electronic receive apodization, and \( y_{m,n}(t) \) is the measured signal from emission \( m \) on the receive element \( n \) at time \( t \).

The synthetic transmit aperture (STA) focused signal at point \( \mathbf{p} \) is calculated by summing the focused signals from all emissions:

\[
l_{\text{STA}}(\mathbf{p}) = \sum_{m=1}^{M} b_{\text{elec}}(m, \mathbf{p}) z_m(\mathbf{p}), \tag{5.7}
\]

where \( M \) is the number of transmissions, \( b_{\text{elec}} \) is the electronic transmit apodization, and \( z_m(\mathbf{p}) \) is the focused received signal from emission \( m \) at point \( \mathbf{p} \). In general, both \( a_{\text{elec}} \) and \( b_{\text{elec}} \) are dependent on the imaging point \( \mathbf{p} \) so that, a dynamic apodization can be achieved in transmit and receive. In this study however, they are fixed to an apodization window, e.g., Hanning, for all imaging points.

Figure 5.6: Distance between a point \( \mathbf{p} \) and an arc \( \widehat{ab} \) is calculated using (5.5).
Table 5.1: Transducer and simulation parameters.

<table>
<thead>
<tr>
<th>Parameter name</th>
<th>Notation</th>
<th>Value</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements</td>
<td>–</td>
<td>62+62</td>
<td>–</td>
</tr>
<tr>
<td>Center frequency</td>
<td>(f_0)</td>
<td>3.0</td>
<td>MHz</td>
</tr>
<tr>
<td>Speed of sound</td>
<td>(c)</td>
<td>1480</td>
<td>m/s</td>
</tr>
<tr>
<td>Wave length</td>
<td>(\lambda)</td>
<td>493.3</td>
<td>µm</td>
</tr>
<tr>
<td>Array pitch -x</td>
<td>(d_x)</td>
<td>(\lambda/2 = 246.6)</td>
<td>µm</td>
</tr>
<tr>
<td>Array pitch -y</td>
<td>(d_y)</td>
<td>(\lambda/2 = 246.6)</td>
<td>µm</td>
</tr>
<tr>
<td>Sampling frequency</td>
<td>(f_s)</td>
<td>120</td>
<td>MHz</td>
</tr>
<tr>
<td>Emission pulse</td>
<td>–</td>
<td>2-cycles, Hann-weighted</td>
<td>–</td>
</tr>
<tr>
<td>Lens focal ratio</td>
<td>(f_#)</td>
<td>-1</td>
<td>–</td>
</tr>
</tbody>
</table>

5.3 Methods

5.3.1 Synthetic Aperture Imaging Technique
In conventional ultrasound imaging it will be a tedious method to transmit for each steering angle so many times to cover the whole volume. However, this will not be a problem, if a SAI technique is used (Jensen, Nikolov, et al. 2006). Thereby, all the transmit delay calculations can be done after the acquisition. A SAI sequence is designed for imaging down to 14 cm of depth. It utilizes single element transmissions on the row elements and the echoes are collected with all the column elements. For a speed of sound of 1540 m/s, 182 µs is required to acquire a single image line to a depth of 14 cm. For 62 emissions this is equivalent to a volume rate of 88 Hz. IQ-modulated RF data are used for beamforming a low-resolution volume for every emission and finally, by summing all the low-resolution volumes, a high-resolution volume is generated.

5.3.2 Imaging Quality Assessment Measures
The imaging performances of a double-curved RCA 2-D array is computed using the two measures described below:

5.3.2.1 Spatial Resolution
The spatial resolution is calculated as the full-width at half-maximum (FWHM) of the imaging system’s point spread function (PSF).

5.3.2.2 Contrast resolution
The spatial resolution is calculated as the cystic resolution (CR), which is the ability to detect an anechoic cyst in a uniform scattering medium (Vilkomerson et al. 1995; Ranganathan and Walker 2007; Guenther and Walker 2009).
5.3. Methods

Figure 5.7: Lens (a): circumscribes the array and (b): inscribes in the array. In case (a) the effective FOV is less than the f-number of the lens. In case (b) the FOV is equal to that of the lens. The lens material is shown in gray and the array is shown in blue.

Figure 5.8: The image is the envelope of the received signals from a single element emission reflected by a scatterer located at \((x, y, z) = (0, 0, 20)\) mm and the overlaid blue line is the predicted time-of-flight calculated using (5.6).
Chapter 5. Curvilinear Imaging with Row-Column Addressed Arrays

Figure 5.9: Three cross-planes (azimuth, elevation, and C-plane) of 3-D PSFs are shown at a dynamic range of 40 dB. The origin corresponds to the center of the transducer surface aligned with a point target positioned at \((x, y, z) = (0, 0, 30)\) mm for PSF 1 ~ PSF 4, and a point target positioned at \((x, y, z) = (0, -15, 25.9)\) mm for PSF 5. The C-planes are at depth of 30 mm.

5.3.3 Simulation Setup
In this study, Field II (Jensen and Svendsen 1992; Jensen 1996) is used for all simulations. A MATLAB (MathWorks Inc., Massachusetts, USA) beamformer that implements (5.7) was programmed to beamform data from curved RCA arrays and produce the PSFs included in this study. The simulation parameters of a RCA 62+62 element 2-D array are shown in Table 5.1. The receive array is rotated 90° with respect to the transmit array. Field II is set up to use lines to describe the apertures and each line-element is divided into square mathematical sub-elements with a side length of \(\lambda/4\). To remove the otherwise apparent secondary echoes originating from the either ends of line-elements, two roll-off apodization regions are placed at both ends of each element (Christiansen et al. 2015; Rasmussen et al. 2015). The length of each apodization region was equal to 15 times the pitch of the array. Each mathematical sub-element in both transmit and receive arrays
is delayed according to the lens delay profile and no attenuation is assumed for the lens. Theoretically, transmitting with row elements and receiving with column elements should image exactly the same curvilinear volume as transmitting with column elements and receiving with row elements. Thus, no preference is considered in transmitting with row elements and receiving the echoes with column elements, or vice versa.

Fig. 5.7 illustrates two different ways to integrate a diverging lens over the array. The lens shown in Fig. 5.7(a) circumscribes the whole underlying array. On the other hand, the lens shown in Fig. 5.7(b) does not cover the whole array, instead the lens is inscribed in the array. In this configuration, essentially there is no diverging focusing applied to the end-most elements, and all elements between the end and the middle have compromised divergence. Thus, the defocusing is applied inconsistently across the array.

Both inscribed and circumscribed cases can provide apodization from lens attenuation as the lens becomes thicker toward the edges. The circumscribed case actually provides more apodization because in this case the lens gets thicker in the corners than the inscribed lens. The inscribed lens is advantageous because it has a smaller lens arc height and shorter chord length than the circumscribed lens. This reduced arc height improves patient contact possibilities, but, more importantly, the shorter chord length enables lower $f_\#$ defocusing. Fresnel lens could be another configuration as a diverging lens, however in this study the configuration shown in Fig. 5.7(a) has been chosen for the simulations.

### 5.4 Results and Discussion

The beamformer can IQ-beamform 250,000 voxels from a complex data set of 1.5 MiB from 62 receive line elements in approximately 14.1 s on a PC with a 3.4-GHz Intel Core i7-4770 CPU (Intel Corp., Santa Clara, CA, USA) and 32 GiB of RAM. The proof-of-concept Matlab implementation of the beamformer can therefore not achieve a frame rate useful for real-time applications, but the frame rate is adequate for research purposes.

Fig. 5.8 shows the received echoes that are generated from a single scatterer located at $(x, y, z) = (0, 0, 20)$ mm. The secondary echoes after the main echo are suppressed below $-40$ dB by using the static roll-off apodization regions. The overlaid blue line is the predicted time-of-flight using (5.6).

Fig. 5.9 shows five 3-D PSFs simulated with Field II (Jensen and Svendsen 1992; Jensen 1996) using SAI technique and beamformed for both flat and curved RCA 2-D array. The point targets are located at $(x, y, z) = (0, 0, 30)$ mm and $(x, y, z) = (0, -15, 25.9)$ mm and a Hanning electronic apodization is applied over the received RF data. A Hanning apodization is applied over the low-resolution volumes before summing in the SAI technique. The PSFs are normalized to their maximum value and shown in a dynamic range of 40 dB. For the PSF 1 and PSF 2, the roll-off apodization is disabled. The secondary lobes located slightly after 30 mm depth in PSF 1 and PSF 2 are the apparent edge echoes and cannot be suppressed by using the electronic apodization. On the other hand, for the PSF 3 ~ PSF 5, the roll-off apodization is activated. It can be noticed that the
Table 5.2: FWHM and CR of simulated 3-D PSF 3 ∼ PSF 5 shown in Fig. 5.9

<table>
<thead>
<tr>
<th>CR</th>
<th>PSF 3</th>
<th>PSF 4</th>
<th>PSF 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_{6dB}$</td>
<td>0.68</td>
<td>0.68</td>
<td>0.97</td>
</tr>
<tr>
<td>$R_{12dB}$</td>
<td>1.11</td>
<td>1.12</td>
<td>1.55</td>
</tr>
<tr>
<td>$R_{20dB}$</td>
<td>1.61</td>
<td>1.64</td>
<td>2.8</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>FWHM</th>
<th>PSF 3</th>
<th>PSF 4</th>
<th>PSF 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radial</td>
<td>0.37</td>
<td>0.37</td>
<td>0.38</td>
</tr>
<tr>
<td>Azimuth</td>
<td>1.88</td>
<td>1.9</td>
<td>1.9</td>
</tr>
<tr>
<td>Elevation</td>
<td>1.88</td>
<td>1.9</td>
<td>1.64</td>
</tr>
</tbody>
</table>

apparent secondary echoes after each main echo in both PSF 3 and PSF 4 are suppressed by using the static roll-off apodization regions. Using the roll-off apodization does not change the lateral resolution of the main echo, this can be seen by comparing PSF 1 and PSF 2 with PSF 3 and PSF 4. The effect of roll-off apodization is mostly outside of the curvilinear imaging FOV of the array, and therefore will not affect the lateral resolution within the imaging FOV. Table 5.2 lists the FWHM and the CR of the simulated 3-D PSF 3 ∼ PSF 5 shown in the Fig. 5.9.

To study the PSF characteristics as a function of lateral angular position and radial distance, a point scatterer is imaged by sweeping it from 0° to 40° in the lateral plane with steps of 10° at radial distances from 10 mm to 60 mm from the center of the array. At each radial distance and angular position the FWHM and CR values are calculated over a volume of 10 mm × 10 mm × 10 mm surrounding the point target. Fig. 5.10 is illustrating the measured FWHM and CR values as a function of depth and angular position in lateral plane.

Using multiple elements in the transmit or receive and by adjusting their delays, the flat RCA 2-D array generates a focal line, however, with a curved RCA 2-D array, focusing in transmit or receive generates two intersection points instead of a focal line as shown in Fig. 5.11. Similar to flat RCA 2-D array, either of these two intersection points can be focused in receive and in this way suppressing the secondary intersection points in both transmit and receive. Looking at Fig. 5.11, it can be noticed that the characteristics of the focused intersections can be different at different angles. Moving away from the center of the elements towards the edges, the transmit wavefronts contact each other at a sharper angle compared with the contact point at the center. It can be observed in Fig. 5.10
Figure 5.10: CR and FWHM values calculated for point targets located at radial distances from 10 mm to 60 mm as a function of different azimuth steering angle away from the central forward-looking axis of the array.
Figure 5.11: Focusing the wavefronts at a fixed distance with arc-shaped elements $s_1$ and $s_2$ generates two intersection points $t_1$ and $t_2$. In conventional row-column imaging either of these intersections can be identified in receive.

that by moving away from the center towards the higher angular position in lateral plane, the elevation FWHM values become smaller while the CR values become larger. On the other hand, the lateral FWHM values stay constant, this is due to the intersection of the wavefronts in the receive direction which is at the center of the receive elements and therefore the elevation FWHM values stay constant for all angular positions in the lateral plane.

Fig. 5.12 illustrates three cross-planes (azimuth, elevation, and C-plane) of a phantom with point targets in water simulated with and without a diverging lens. The point targets are distributed in both lateral and elevation dimensions from $-60$ mm to $60$ mm with step size of $10$ mm, as well as in an axial range of $5$ mm to $95$ mm with step size of $4.5$ mm. It can be seen from Fig. 5.12 that, by using a diverging lens the FOV is extended compared to the flat RCA array.

Fig. 5.13 illustrates three cross-planes (azimuth, elevation, and C-plane) of an anechoic cyst vessel phantom simulated with and without a diverging lens. The phantom with volume size of $40\times40\times20$ mm$^3$ contains an anechoic cyst vessel with radius of $5$ mm
5.4. Results and Discussion

Figure 5.12: Three cross-planes (azimuth, elevation, and C-plane) of a phantom with point scatterers axial and lateral spacing of 5 mm and 10 mm imaged with and without a diverging lens ($f_\# = -1$), are shown at a dynamic range of 40 dB. The C-planes are at depth of 47 mm.
along the azimuth dimension located at a 20 mm depth. The phantom is simulated in water with average scatterer density of 8 per mm$^3$. The FOV is extended compared to the flat RCA array. In Fig. 5.14, the cyst phantom is located deeper at a 60 mm depth and beamformed with the proposed DAS beamforming method, with and without a diverging lens. Similar to Fig. 5.13 here also using a diverging lens extends the FOV compared to the flat RCA array.

Figure 5.13: Three cross-planes (azimuth, elevation, and C-plane) of a hollow tube with a diameter of 10 mm inside a rectangular box imaged with an RCA 2-D array with a diverging lens ($f_# = -1$) are shown in 40 dB dynamic range. The cyst box dimensions are $40 \times 40 \times 20$ mm$^3$. The C-planes are at a depth of 20 mm.
Figure 5.14: Three cross-planes (azimuth, elevation, and C-plane) of a hollow tube with a diameter of 10 mm inside a rectangular box imaged with an RCA 2-D array with a diverging lens ($f_\# = -1$) are shown in 40 dB dynamic range. The cyst box dimensions are $40 \times 40 \times 20 \text{mm}^3$. The C-planes are at depth of 60 mm.

Diverging the wavefronts has the negative effect of lowering the pulse-echo energy as it is shown in Fig. 5.4 compared with the conventional row–column imaging using flat arrays. This loss of the energy can be compensated for by using all the elements in transmit and placing the transmit focus in front of the array.

Using a diverging lens, the elements at the middle of the array can be represented as an arc, but the 3-D focusing characteristics of the off-center elements should make their
Figure 5.15: If each arc-shaped elements $s_1$ and $s_2$ is divided into two sub-elements, by activating each sub-element, only one intersection can be produced, $t_1$ or $t_2$ depending on which side has been activated. Thus, it is possible to accurately calculate the time-of-flight using only either row or column elements independently. Thereby, a two-way focusing profile can be achieved.

...representations more complicated. It requires to formulate the trigonometric functions that are used for delay calculations in a spherical geometry. It is also possible to have a different curvature in transmit and receive, however that requires to formulate the delay calculations in bispherical coordinates, which was beyond the scope of this study.

5.4.1 Experimental Results
Two diverging acoustic lenses with different curvatures have been manufactured at STI (Sound Technology Inc, PA, USA), which are mountable on the RCA probes using a holder as shown in Fig. 5.16. The manufactured lenses are made of RTV silicone casted inside a thermoplastic frame. The lenses have $f_\# = 1.4$ and $f_\# = 2.8$. A wire grid phantom has been imaged using both of these lenses mounted on a flat 62+62 PZT RCA probe. Considering the correct speed of sound for the RTV silicone rubber (1 mm/μs), the beamformed cross planes (azimuth and elevation) are shown in Fig. 5.17. A lower
value corresponds to a thicker lens which results in a higher reverberations between
the probe surface and the lens. This can be seen for the lens with \( f_# = 1.4 \) in Fig. 5.17(a)
and Fig. 5.17(b). Using the lens with \( f_# = 2.8 \), the beamformed cross planes are shown in
Fig. 5.17(c) and Fig. 5.17(d). In both azimuth planes in Fig. 5.17(a) and Fig. 5.17(c), the
wires are not completely visible, which is due to the reflection of sound waves away from
the transducer. The experimental results should be considered as preliminary results and
further investigation on the imaging performance needs to be performed.

5.5 Conclusion

In this study the qualitative imaging performance of a curved 62+62 RCA 2-D array
was evaluated. The capabilities of a curved RCA 2-D array to effectively focus in both
transmit and receive were investigated, and a suitable DAS beamformer introduced and
implemented. Using SAI technique it was possible to image down to 14 cm at a volume
rate of 88 Hz. To validate the performance simulations of the imaging performance with a
curved RCA 2-D array at several different situations was evaluated. Results confirm that
using a diverging lens with \( f_# = -1 \) can increase the imaging FOV to 60° \( \times \) 60°, and it is
also possible to perform dynamic transmit-receive focusing throughout the curvilinear
FOV. Thereby, the inherent imaging limitation with flat RCA 2-D arrays, i.e., its forward
looking rectilinear FOV, is overcome using a diverging lens. Overall, having a low channel
count and a large FOV offers the potential to fabricate arrays with large aperture sizes,
which is important for abdominal scans. Thus, by using a curved RCA 2-D array, 3-D
imaging is possible with equipment in the price range of conventional 2-D imaging. These
advantages might contribute to an increased use of real-time 3-D ultrasound imaging in
medical diagnostics, and to the development of new clinical applications.

If each line-element can be divided into two equal sub-elements as shown in Fig. 5.15,
by activating each sub-element of the row elements it is possible to eliminate either of
those intersection points. The advantage of doing so is that, if the echoed signals are
collected with the same transmitting elements, a two-way focusing profile can be produced,
which is not possible with traditional row-column imaging since the transmit and receive
apertures are perpendicular to each other. Therefore, by dividing the curvilinear FOV into
2 sub-volumes, each sub-volume can be beamformed with only row or column elements.
Although this was not the main focus of this study, it could be interesting to investigate
the focusing abilities using only the curved row or column elements.

References

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ultrasonic transducers”. In: Journal of Micromechanics and Microengineering 24.8,
Figure 5.16: Two diverging acoustic lenses with $f_\# = 1.4$ and $f_\# = 2.8$ were manufactured. The lenses are made of RTV silicone casted inside a thermoplastic frame.
Figure 5.17: Synthetic aperture imaging using diverging lenses on a wire grid phantom. No wires are installed at −20 mm elevation distance for orientation purposes.


CHAPTER 6

Conclusion and Perspectives

The work presented in this thesis explored two different approaches to acquire 3-D information using 2-D arrays: fully addressing each individual transducer element in contrast with a row-column addressing scheme. However, both of these approaches face technical challenges concerning the performance, volume rate, and probe manufacturing.

In the first part, 3-D synthetic aperture imaging was compared with 3-D parallel beamforming, which used to be the “gold standard” of real-time 3-D ultrasound imaging. On that subject a conference paper was published, however results presented in Chapter 2 were not published yet. Using both simulations and measurements, it was shown that 3-D synthetic aperture imaging increases the imaging sensitivity compared with parallel beamforming. Measurements on a tissue mimicking phantom indicated that the penetration depth is deeper for synthetic aperture imaging compared with parallel beamforming. Synthetic aperture had a higher SNR than parallel beamforming at all depths and the increased SNR resulted in a penetration depth increase of 23%.

The theory of synthetic aperture imaging using 2-D phased arrays as well as its different configurations were explained, which ease the optimization of the synthetic aperture imaging setup. The setup of synthetic aperture imaging was made using a parameter study. Some of the questions still to be answered are: Can the image quality be predicted using only the synthesized transmit aperture and the receive aperture as well as the field of view? And how should the tissue 3-D motion be estimated in 3-D synthetic aperture imaging?

The leading 2-D matrix probes contain a very large number of transducer elements, and still only a standard amount of cables connects the probe to the scanner. The beamforming technique they use is called µ-beamforming (Savord and Solomon 2003). With µ-beamformers a group of neighboring elements is pre-beamformed (µ-beamformed) before being sent to the ultrasound scanner as one signal for final beamforming. This way, the number of cables connecting the probe and scanner is greatly reduced. However, this clearly indicates that only one scan line per emission is beamformed. Therefore, for applications requiring a large field of view, the volume rate is low. The µ-beamformer technique could therefore be implemented as a future reference for investigations of 3-D synthetic aperture imaging. Combining µ-beamformers with the synthetic aperture technique would also be interesting, perhaps increasing the volume rate and the image quality of the large 2-D arrays even further.

An interesting research subject is to increase the frame rate of 3-D synthetic aperture imaging by using multiple transmit beams. Multiple virtual elements would then be
synthesized per emission, thereby increasing the frame rate, or possibly the image quality.

The subject of the second part was row-column addressing of 2-D arrays. On that subject three conference papers and three journal papers were submitted. The capabilities of row-column addressed transducers, when integrated into probe handles were investigated. For that reason, two prototyped 62+62 RCA 2-D array transducer probes were manufactured using CMUT and PZT technology. The transducers were designed with similar acoustical features, i.e. dimensions, center frequency, and packaging. That gave the unique possibility of evaluating the two probes relative to each other and comparing the row-column addressing scheme based on two different technologies. The design, fabrication, and characterization of the two probes were described. Based on acoustical measurements the center frequency, bandwidth, surface pressure, sensitivity, and acoustical cross-talks were evaluated and discussed.

With simulations it was demonstrated that a double curved RCA 2-D array increases the inherent rectilinear field of view to a curvilinear volume region and has the potential to increase the image quality. It was also shown, based on simulations as well as measurements, that using a diverging lens as a cheaper alternative to fabrication of double curved arrays can also increase the field of view. Imaging quality using a large RCA 2-D array with a diverging lens has the advantages of having a high volume rate as well as a large field of view, and at the same time keeping the channel count low. A future research subject could be to demonstrate that large row-column addressed arrays using a diverging lens can achieve a 3-D image of clinically relevant quality.
Papers
In Vivo Real Time Volumetric Synthetic Aperture Ultrasound Imaging

Authors: Hamed Bouzari\textsuperscript{a}, Morten Fischer Rasmussen\textsuperscript{a}, Andreas Hjelm Brandt\textsuperscript{b}, Matthias Bo Stuart\textsuperscript{a}, Svetoslav Nikolov\textsuperscript{c} and Jørgen Arendt Jensen\textsuperscript{a}.


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In Vivo Real Time Volumetric Synthetic Aperture Ultrasound Imaging

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ABSTRACT

Synthetic aperture (SA) imaging can be used to achieve real-time volumetric ultrasound imaging using 2-D array transducers. The sensitivity of SA imaging is improved by maximizing the acoustic output, but one must consider the limitations of an ultrasound system, both technical and biological. This paper investigates the \textit{in vivo} applicability and sensitivity of volumetric SA imaging. Utilizing the transmit events to generate a set of virtual point sources, a frame rate of 25 Hz for a 90° × 90° field-of-view was achieved. Data were obtained using a 3.5 MHz 32×32 elements 2-D phased array transducer connected to the experimental scanner (SARUS). Proper scaling is applied to the excitation signal such that intensity levels are in compliance with the U.S. Food and Drug Administration regulations for \textit{in vivo} ultrasound imaging. The measured Mechanical Index and spatial-peak-temporal-average intensity for parallel beamforming (PB) are 0.83 and 377.5 mW/cm\textsuperscript{2}, and for SA are 0.48 and 329.5 mW/cm\textsuperscript{2}. A human kidney was volumetrically imaged with SA and PB techniques simultaneously. Two radiologists for evaluation of the volumetric SA were consulted by means of a questionnaire on the level of details perceivable in the beamformed images. The comparison was against PB based on the \textit{in vivo} data. The feedback from the domain experts indicates that volumetric SA images internal body structures with a better contrast resolution compared to PB at all positions in the entire imaged volume. Furthermore, the autocovariance of a homogeneous area in the \textit{in vivo} SA data, had 23.5\% smaller width at the half of its maximum value compared to PB.

Keywords: Real-time volumetric ultrasound imaging, 2-D phased array transducer, synthetic aperture (SA)

1. INTRODUCTION

Volumetric ultrasound enables imaging of the whole volume in one acquisition similar to x-ray computed tomography (x-ray CT) and magnetic resonance imaging (MRI). In conventional 2-D ultrasound imaging, it is required to wait for the propagation of the ultrasound pulse back and forth in the body for each single image line. In volumetric imaging on the other hand, the number of image lines is squared, and hence a quadratic reduction on the achievable frame rate is imposed. Imaging the complex kinematics of organs such as the beating heart requires an increased frame rate, which is far from achievable on large imaging volumes using the conventional approach. Indeed, considering the speed of sound in biological tissues to be around 1540 m/s, about 155 μs are required to acquire a single image line with a 14 cm depth. This is approximately 6400 lines per second which may be used to form a volume of 80×80×80 image lines, in every second. Von Ramm and Smith\textsuperscript{1,2} introduced the first true volumetric ultrasound system, which allowed real-time 3-D scanning at acceptable volume rates. The system applied a parallel beamforming (PB) technique that permitted the formation of a plurality of adjacent lines surrounding the transmit beam direction. To achieve higher volume rates, i.e., higher temporal resolution, broadened transmit beams can be used to illuminate the desired field-of-view resulting in a reduced number of emissions.\textsuperscript{3} However, reverberations and aberrations of the ultrasonic wave fronts caused by fatty subcutaneous tissues tend to destroy the focusing capabilities of the system in PB.\textsuperscript{4}

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To recover the focusing capability, while using broad illuminations for a higher temporal resolution, it has been proposed to use synthetic aperture (SA) imaging, originally developed for radar detection systems.\(^5\)\(^,\)\(^6\) By coherently combining the data acquired from successive and spatially overlapping ultrasound pulse emissions, one may retrospectively recreate a dynamic transmit focus along each line of the final image. The original SA application used single element excitations, but is now applied using virtual point sources\(^7\) generated with small subsets of transducer elements to increase the energy transferred into the tissue and thereby increase the SNR.

Although SA imaging may allow the spatial resolution to be similar or even improved compared to single line focused ultrasound, it does not ensure a good sensitivity in the sense that, neither the SNR, the penetration nor the contrast are ensured to be similar to what is achieved in conventional ultrasound imaging. The sensitivity is directly related to the characteristics of the emitted pulses: 1) the frequency band and focus depth give an indication of how attenuated a pulse will be, 2) the peak intensity is an indicator of the non-linear effects build-up during the propagation, 3) the aperture size and pulse duration describe the amount of energy transmitted. Thus the sensitivity can be improved by maximizing the acoustic output, but the limitations of the ultrasound system, both technical and biological must be considered. A powerful signal generator to drive a big amount of energy through the piezo-electric transducer is required, which may lead to over-heating of the probe surface. Any damage to the tissues caused by cavitational effects or over-heating has to be avoided. In practice, the acoustic output is adjusted such that both the peak and the temporal average intensities remain under given thresholds.

Previously it has been shown that SA can be used to achieve real-time volumetric imaging based on simulation and phantom studies.\(^8\) However, as of today, it has not yet been successfully adapted to \textit{in vivo} volumetric 2-D phased array imaging. Provost et al.\(^9\) have done volumetric \textit{in vivo} SA measurements, however they did not provide any other ultrasound volumetric imaging method to compare their results with. In this study, the imaging quality of SA is investigated using \textit{in vivo} measurements. Simulations are used to optimize both techniques before comparison and also to ensure the FDA limits. The optimization is done for a channel limited 3-D ultrasound system with 256 active channels.\(^9\)

The structure of the paper is as follows: First, a brief description of the measurement and simulation setups is given. The point spread functions (PSFs) for the PB and SA are studied both based on simulation and measurement data. The intensity measurements for PB and SA are presented. Then \textit{in vivo} results and the qualitative assessments by experts in the field are presented. Finally, the perspectives of the volumetric SA technique are discussed.

\section*{2. MEASUREMENT AND SIMULATION SETUPS}

The volumetric data were acquired using the 1024 channel experimental ultrasound scanner, SARUS.\(^10\) To estimate the quality of the SA images compared to PB, the subjects were imaged with both techniques simultaneously. The RF-data were beamformed using the beamformation toolbox \(^3\)\(^,\)\(^1\)\(^1\) Table 1 lists the measurement configuration parameters. The centers of all translated transmit apertures for SA are shown with a dot in Fig. 1(left). The cross is the center of the shown active aperture. The receive aperture which is static during all 256 emissions, is illustrated in Fig. 1 (right). As a trade-off between emitted energy and side-lobe performance, the 24 element wide cross array, seen in Fig. 2 (left), is chosen as the transmit aperture.\(^8\) To get a wide receive aperture and thereby a narrow receive beam main-lobe, the cross array is also used in receive. The widest possible array, a cross array along the diagonals, is chosen as receive aperture. Because the receive aperture is too narrow, it is apodized with a Tukey function with a \(\Psi\) parameter close to zero value. The receive aperture is shown in Fig. 2 (right). In Fig. 3, a simulated point spread function example of PB and SA is shown, and can be visually inspected and compared. A point target is located at 62 mm depth and 0\(^\circ\) azimuth and elevation tilt angle. In the azimuth plane the sidelobes are larger than in the elevation plane. This is due to three inactive rows of elements on the transducer, all orthogonal to the elevation plane. SA and PB appear to have approximately the same main lobe size. Figure 4, shows the measured 3-D point spread function of PB and SA in a water bath imaging a tip of an iron needle facing toward the transducer and parallel to the center line of the transducer (speed of sound in water 1480 m/s). The apparent noise is due to the low SNR of the research scanner. Figure 5 shows the SNR of the research scanner for both SA and PB in a tissue mimicking phantom. The full width
Table 1. Setup configuration

<table>
<thead>
<tr>
<th></th>
<th>SA</th>
<th>PB</th>
</tr>
</thead>
<tbody>
<tr>
<td>Center frequency</td>
<td>3 MHz</td>
<td></td>
</tr>
<tr>
<td>Pitch x</td>
<td>300 µm</td>
<td></td>
</tr>
<tr>
<td>Pitch y</td>
<td>300 µm</td>
<td></td>
</tr>
<tr>
<td>Number of elements in x</td>
<td>32</td>
<td></td>
</tr>
<tr>
<td>Number of elements in y</td>
<td>35 (3 inactive rows)</td>
<td>-</td>
</tr>
<tr>
<td>Techniques</td>
<td>SA</td>
<td>PB</td>
</tr>
<tr>
<td>Frame rate</td>
<td>25 Hz</td>
<td>25 Hz</td>
</tr>
<tr>
<td>Pulse repetition frequency</td>
<td>5.133 kHz</td>
<td>5.133 kHz</td>
</tr>
<tr>
<td>Emissions per frame</td>
<td>256</td>
<td>256</td>
</tr>
<tr>
<td>Number of active elements</td>
<td>256</td>
<td>256</td>
</tr>
<tr>
<td>Scan depth (max range)</td>
<td>14 cm</td>
<td>14 cm</td>
</tr>
<tr>
<td>Emission cycles</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>Focus in transmit</td>
<td>-6 mm</td>
<td>60 mm</td>
</tr>
<tr>
<td>Sampling frequency</td>
<td>12 MHz</td>
<td>12 MHz</td>
</tr>
<tr>
<td>Transmit voltage</td>
<td>±100 V</td>
<td>±100 V</td>
</tr>
<tr>
<td>Field-of-view</td>
<td>$90^\circ \times 90^\circ$</td>
<td>$90^\circ \times 90^\circ$</td>
</tr>
<tr>
<td>Beamformed lines per emission</td>
<td>64×64</td>
<td>4×4</td>
</tr>
</tbody>
</table>

at half maximum (FWHM) and the side lobe energy metrics, for the simulated and measured PSF are listed in table 2.

Figure 1. The synthetic aperture imaging transmit and receive apodization implemented on the 32×32 element array. The transmit aperture translates between emission. The center of the shown aperture is illustrated with a green cross. The receive aperture is static during all 256 emissions.

3. INTENSITY MEASUREMENTS

Before any in vivo measurements, the ultrasound imaging technique on the scanner has to fulfill all the requirements regarding the intensity levels and safety limits. Any damage to the tissues caused by cavitation or over-heating has to be avoided. In practice, the acoustic output is adjusted such that both the peak and the
Figure 2. The parallel beamforming transmit and receive apodization implemented on the 32×32 element array. The receive aperture is the widest possible cross array implementable on the 32×32 element array. Both apodizations contain 256 active elements and are used for all 256 emissions.

Table 2. FWHM and side lobe measurements

<table>
<thead>
<tr>
<th></th>
<th>SA</th>
<th>PB</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Cystic resolution (6 dB)</strong></td>
<td>Simulation</td>
<td>Measurement</td>
</tr>
<tr>
<td></td>
<td>1.55</td>
<td>2.10</td>
</tr>
<tr>
<td><strong>Cystic resolution (12 dB)</strong></td>
<td>2.22</td>
<td>3.45</td>
</tr>
<tr>
<td><strong>Cystic resolution (20 dB)</strong></td>
<td>3.47</td>
<td>12.25</td>
</tr>
<tr>
<td><strong>FWHM (at 62 mm depth)</strong></td>
<td>Axial</td>
<td>0.76</td>
</tr>
<tr>
<td></td>
<td>Azimuth</td>
<td>4.32</td>
</tr>
<tr>
<td></td>
<td>Elevation</td>
<td>4</td>
</tr>
</tbody>
</table>

temporal average intensities remain under given thresholds. As of today, such safety guides are regulated by the the FDA, and take the form of upper limits on given indexes: the mechanical index ($MI \leq 1.9$), the derated spatial-peak-temporal-average intensity ($I_{spta} \leq 720 \text{ mW/cm}^2$ for peripheral vessel, $I_{spta} \leq 430 \text{ mW/cm}^2$ for cardiac), and the derated spatial-peak-pulse-average intensity ($I_{sppa} \leq 190 \text{ mW/cm}^2$). This requires to measure the emitted pressure of the transducer as a function of spatial position. The intensity measurements have been carried out using the experimental ultrasound scanner SARUS and the AIMS III intensity measurement system (Onda Corporation, Sunnyvale, California, USA). The measured $MI$ and $I_{spta}$ before scaling the excitation signal for PB are 0.83 and 377.5 mW/cm$^2$, and for SA are 0.48 and 329.5 mW/cm$^2$, accordingly (Table 3).

Table 3. Intensity measurement results

<table>
<thead>
<tr>
<th></th>
<th>SA</th>
<th>PB</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak MI in water</td>
<td>0.48</td>
<td>0.83</td>
</tr>
<tr>
<td>Peak MI derated</td>
<td>0.46</td>
<td>0.76</td>
</tr>
<tr>
<td>Peak $I_{spta}$ in water</td>
<td>329.5</td>
<td>377.5</td>
</tr>
<tr>
<td>Peak $I_{spta}$ derated</td>
<td>260</td>
<td>312.5</td>
</tr>
</tbody>
</table>
4. RESULTS AND DISCUSSION

Cut sections of an in vivo volume data of a healthy male’s kidney, imaged with SA and PB techniques, are illustrated in Fig. 6. Figure 7 illustrates the magnified areas of the Fig. 6 top row. Two radiologists were consulted for evaluation of volumetric SA by means of a questionnaire and compared SA against PB in terms of the pathological features presented in the in vivo data. The PB technique suffers from the block-like artifact. The spatial resolution achieved with SA imaging is dependent on the characteristics of successive transmissions. To achieve higher sensitivity with SA imaging, the energy of each transmitted pulse and also the number of data-sets, which are coherently compounded have to get maximized. But these two constraints imply a trade-off: to maximize the number of combined beams, spatial overlaps between transmitted pulses has to be ensured, and hence use broad beam transmissions where the energy is likely to spread out and the intensity will be reduced. To increase the sensitivity of SA, one may want to increase the energy per pulse by increasing the pulse amplitude or its duration. Increasing the amplitude has an upper limit via MI and \( I_{pta} \). Unfortunately, increasing the pulse duration also will result in a poorer axial resolution. As an alternative, it has been proposed to use linear frequency modulated (FM) excitations combined with match filtering on reception, to increase the energy level without sacrificing the axial resolution.\(^{14}\) The lateral (elevational) resolution is only defined by the capability of the SA to synthesize a narrow synthetic transmit beam profile.

5. CONCLUSION AND PERSPECTIVES

In this study, a comparison between real-time 3-D synthetic aperture imaging and parallel beamforming using only 256 active channels was made with both Field II simulations and in vivo measurements from the experimental ultrasound scanner SARUS. The contrast resolution was improved by synthetic aperture imaging at all positions in the entire imaged volume. The autocovariance of a homogeneous area in the in vivo SA data, has 23.5% smaller width compared to PB at the half of its maximum value. Based on the feedbacks from domain experts, the in vivo imaging quality of synthetic aperture and parallel beamforming has been investigated. It was shown that using synthetic aperture imaging on a channel limited 3-D ultrasound system can achieve a high image quality at a low cost. Both techniques can volumetrically visualize internal body structures. Visualizing in 3-D gives the clinician a better insight for possible pathology and medical treatments. A novel data visualization tool will become very beneficial in clinical studies, as it is difficult to visualize the ultrasound volume in 3-D.

ACKNOWLEDGMENTS

This work was financially supported by grant 82-2012-4 from the Danish Advanced Technology Foundation and from BK Medical ApS, Herlev, Denmark.

REFERENCES


Figure 3. PB and SA simulated 3-D point spread function sliced into three 2-D planes. The point spread functions are observed at 62 mm depth and 0° azimuth and elevation tilt angle. The left column is SA and the right column is PB.
Figure 4. PB and SA measured 3-D point spread function sliced into three 2-D planes. The point spread functions are observed at 62 mm depth and 0° azimuth and elevation tilt angle. The left column is SA and the right column is PB.
Figure 5. The SNR of the research scanner for both SA (left) and PB (right) imaging methods in a tissue mimicking phantom.

Figure 6. Cut sections of an in vivo volume of a human kidney imaged with PB and SA techniques. The left column is SA and the right column is PB.
Figure 7. The magnified areas of the in vivo image of a human kidney imaged with PB and SA techniques shown in top row of Fig. 6. The left one is SA and the right one is PB.
Real-Time 3-D Synthetic Aperture Imaging

Authors: Hamed Bouzari, Matthias Bo Stuart, Svetoslav Ivanov Nikolov, and Jørgen Ar- endt Jensen.


This research was financially supported by grant 82-2012-4 from the Danish Advanced Technology Foundation and from BK Medical ApS (Herlev, Denmark).

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Abstract—Synthetic aperture imaging (SAI) can be used to achieve real-time volumetric ultrasound imaging using 2-D array transducers, which is very beneficial for functional analyses, e.g., on the heart. Although SAI imaging may allow the spatial resolution to be similar or even improved compared to single line focused ultrasound, it suffers lower SNR, penetration depth and contrast comparing to conventional ultrasound imaging. However, the sensitivity of SAI can be improved by maximizing the acoustic output, but one must consider the limitations of an ultrasound system, both technical and biological. This paper investigates the in vivo applicability and sensitivity of volumetric SAI technique in comparison with 3-D parallel beamforming (PB) technique. Utilizing the transmit events to generate a set of virtual point sources, a frame rate of 20 Hz for a 90° × 90° field-of-view was achieved. Data were obtained using a λ/2-pitch 3.8 MHz 32 × 32 elements 2-D phased array transducer connected to the experimental scanner, SARUS. The intensity levels as well as probe temperature are in compliance with the U.S. Food and Drug Administration regulations for in vivo ultrasound imaging. The measured Mechanical Index and spatial-peak-temporal-average intensity for PB are 0.77 and 168.8 mW/cm², and for SAI are 0.14 and 4 mW/cm². On each measurement, a same volume region was volumetrically imaged with SAI and PB techniques simultaneously. The results indicate that volumetric SAI images structures with a better contrast resolution compared to PB at all positions in the entire imaged volume.

I. INTRODUCTION

Two major obstacles that have delayed the implementation of real-time 3-D imaging systems are the low frame rate often achievable when scanning a full volume as well as the large amount of channels on a 2-D array transducer to scan the volume. Parallel beamforming (PB) was introduced to address the first issue. Compared with the conventional line by line imaging, PB could increase the frame rate by simultaneously beamforming a plurality of receive beams around the broadened transmit beam. According to Fraunhofer approximation for a rectangular aperture, the beam width in the focal plane is equal to λ/λ₂, where λ₂ is the f-number in transmission or reception, and λ is the wavelength. Using a broader transmit beam in PB requires to increase the transmit f₂ compared with the receive f₂. However, the loss in two-way lateral resolution due to broadening of the transmit beam, can not be recovered. In addition, the misalignment of transmit and receive directions introduces degradations in the lateral shift invariance, and thus results into block-like artifacts. To remove those block-like artifacts and using a lower transmit f₂, it has been proposed a technique called synthetic transmit beamformation (STB) to synthesize a transmit beam along each receive line by spatially interpolating the overlapped adjacent transmit beams [1], [2].

Although it is an effective method to remove the block-like artifacts, it can not control the focal point of the synthesized transmit beam. However by using synthetic aperture imaging (SAI), the received data can be delayed and summed to recover the focusing in every location in the image.

The goal of this study was to investigate if the results achieved in 2-D SAI could be extended to 3-D imaging and validate it based on a comparison with PB method. In this paper, the extended research carried out on 3-D SAI is presented without imposing any limitation on the number of active channels being used. To perform a fair comparison the volume rates of both methods were set equal, however the 3-D SAI results, which have volume rates up to 50 times higher are presented as well.

The outline of the remaining part of this paper is as follows. First, the pulse-echo imaging principle and its limitations, when translating to 3-D imaging are discussed. This is followed by defining a set of imaging quality assessment measures and also an introduction to the hardware being used. Then, after introducing the imaging application requirement, the basics of SAI and PB techniques as well as the analyses on choosing their parameters are given. Afterwards, the results of the simulations and the measurements on phantoms as well as on in vivo are presented. Final section concludes the chapter with a discussion.

II. PULSE-ECHO IMAGING: PRINCIPLE AND LIMITATIONS

To evaluate an imaging technique, first one has to understand the technological limitations and physical boundaries involved. In principle the sensitivity in a pulse-echo imaging system is directly related to the characteristics of the emitted pulses. Thereby, the axial resolution is proportional to the pulse bandwidth (a larger bandwidth means a shorter pulse and thus a higher axial resolution). However, a higher bandwidth increases the thermal noise, which on the other hand lowers the sensitivity. At the same time, the penetration depth is also related to the pulse center frequency (a higher frequency means higher absorption and less reflection as well as higher attenuation). Thus, the sensitivity might be improved by maximizing the acoustic output by transmitting a more powerful pulse. Yet, there are technical and biological limitations on the amount of transmitted energy, which must be considered.

Using a powerful signal generator to drive a big amount of energy through the transducer may lead to over-heating of the probe surface. At the same time, any damage to the tissues caused by cavitational effects or over-heating has to be avoided. In practice, the acoustic output is adjusted such that both the peak and the temporal average intensities remain under given thresholds. Such safety guides are regulated by the the United States Food and Drug Administration (FDA) [3], and take the form of upper limits on given indexes: the mechanical...
index ($MI \leq 1.9$), the derated spatial-peak-temporal-average intensity ($I_{sppta} \leq 720 \text{ mW/cm}^2$ for peripheral vessel, $I_{sppta} \leq 430 \text{ mW/cm}^2$ for cardiac), and the derated spatial-peak-pulse-average intensity ($I_{sppat} \leq 190 \text{ mW/cm}^2$). Increasing the transmitted energy by using a longer pulses results in a poorer axial resolution. Alternatively, it has been proposed to use linear frequency modulation (FM) excitations combined with match filtering on reception, to increase the energy level without sacrificing the axial resolution. Unfortunately, longer excitations increase the probe heating and may burn the transducer.

Unlike in conventional imaging, the lateral resolution in SAI is directly related to the steering angle span of the in-phase summed emissions, but a larger steering span increases the level of side-lobes. Using an apodization in transmit and receive may decrease the side-lobe level but at the same time degrades the spatial resolution. In terms of SNR, using SAI technique by summing $n$ emissions in phase, noise will be suppressed and thereby the SNR will be improved by a factor of $\sqrt{n}$ [4], [5].

III. IMAGING QUALITY ASSESSMENT MEASURES

The imaging performance is computed using four measures:

A. signal-to-noise ratio (SNR)

The SNR is the measure to distinguish soft tissue from electronic noise and is calculated from a number of B-mode images, and its difference to one of the B-mode images, are computed to yield the signal and electronic noise. The SNR is calculated by:

$$\text{SNR}(x) = \frac{\left| \sum_{n=1}^{N} s_n(x) \right|^2}{\left( \frac{1}{N} \sum_{n=1}^{N} s_n(x) \right)^2},$$

where $x = (x, y, z)$ is the voxel coordinate, and $s_n$ a single IQ-beamformed image frame with index $n$. The point where SNR falls below 0 dB is the penetration depth.

B. Spatial resolution

The spatial resolution is calculated as the full-width at half-maximum (FWHM) of the imaging system’s point spread function (PSF).

C. Cystic resolution

The cystic resolution (CR) is the ability to detect an anechoic cyst in a uniform scattering medium [6]–[8]. The relative intensity (RI) of the anechoic cyst was shown by Ranganathan and Walker [7], to be quantized as the clutter energy to total energy ratio,

$$\text{RI}(R) = \sqrt{\frac{E_{\text{cl}}(R)}{E_{\text{tot}}}} = \sqrt{1 - \frac{E_{\text{out}}(R)}{E_{\text{tot}}}},$$

where $E_{\text{in}}$ is the signal energy inside a circular region with radius, $R$, centered on the peak of the point spread function, $E_{\text{tot}}$ is the total PSF energy, and $E_{\text{out}}$ is the PSF energy outside the circular region. The RI($R$) curve can be compressed to a single number by sampling the curve at e.g. 20 dB. The result is the required cyst radius at which the intensity at the cyst’s center is 20 dB lower than its surroundings, written as $R_{20\text{dB}}$.

D. Contrast resolution

The contrast resolution in B-mode images, i.e., contrast-to-noise ratio (CNR) is defined as

$$\text{CNR} = \frac{\mu_{\text{bck}} - \mu_{\text{cyst}}}{\sqrt{\sigma_{\text{bck}}^2 + \sigma_{\text{cyst}}^2}},$$

where $\sigma_{\text{bck}}^2$ and $\sigma_{\text{cyst}}^2$ are variances, and $\mu_{\text{bck}}$ and $\mu_{\text{cyst}}$ are mean values of gray levels within the background and lesion, respectively.

IV. HARDWARE

All measurements are carried out using a fully wired $32 \times 32$ PZT matrix transducer probe connected to the 1024 channel research ultrasound scanner, SARUS:

A. 2-D probe

The 2-D probe used in both simulations and in the measurements is seen in Fig. 1. The probe consists of a fully wired $32 \times 32$ PZT matrix transducer and is produced by Vermon S.A., Tours, France. The averaged pulse-echo impulse response of the transducer is shown in Fig. 2a. The spectrum of the averaged pulse-echo impulse response is shown in Fig. 2b. The center frequency of the Vermon probe is 3.8 MHz and the pitch is 300 µm, which corresponds to $0.74 \lambda$. To avoid grating lobes within the ±45° beamformed volume, the pitch of the transducer array should not be larger than $\lambda/2$. As a compromise between the transducer efficiency in converting electrical to mechanical energy, and grating-lobe levels, the center frequency of the emission is set to 3.0 MHz, which corresponds to 0.58λ pitch. The orientation of the elements are shown in Fig. 3.

Figure 1. The $32 \times 32$ element phased array ultrasound probe used for the measurements and modeled in the simulations. The probe is produced by Vermon S.A. (Tours, France).
Figure 2. Averaged impulse response and its spectrum of the Vermon 32 × 32 element phased array ultrasound probe.

Figure 3. The orientation of the transducer elements on the Vermon 300 µm-pitch 32×32 element phased array ultrasound probe.

B. SARUS

The volumetric data were acquired using the 1024 channel experimental ultrasound scanner, SARUS, which is seen in Fig. 4 [9]. It can sample RF data with a sampling frequency of 70 MHz with a precision of 12 bits.

Figure 4. SARUS, the 1024 channel experimental ultrasound scanner used for all of the measurements.

V. APPLICATION REQUIREMENTS

The two imaging techniques are designed for cardiac imaging, which requires imaging down to 15 cm and a frame rate \( f_s \) of at least 20 Hz. To be comparable with products from the medical ultrasound industry, a volume scan spanning 90° in both the azimuth and elevation direction is chosen, i.e., a field of view of 90° × 90°. With a maximum scan depth \( r_{\text{max}} \) of 15 cm and a speed of sound \( c \) equal to approximately 1540 m/s, the maximum pulse repetition frequency is

\[
f_{prf} = \frac{c}{2r_{\text{max}}} = 5.13 \text{ kHz}.
\]

Using 256 emissions per frame allows for resolving the azimuth and elevation directions with \( \sqrt{256} = 16 \) emissions each.

VI. SYNTHETIC APERTURE IMAGING

When using synthetic transmit focusing, by taking advantage of superposition theorem, a virtual transmit aperture is synthesized for every location by delaying and summing a plurality of datasets acquired from successive transmissions. In other words, one virtual element is synthesized in the synthetic aperture for each transmission. The location of the virtual elements influences the distribution of the emitted energy and thereby the signal-to-noise ratio (SNR) within the imaged volume.

On a phased array, the synthetic aperture can be synthesized by applying beam steering to the transmissions as it is seen in Fig. 5. Placing the virtual sources in front of the array compromises the overlapping between transmit beams. To have a higher overlap between the transmit beams for the same number of emissions, the transmit \( f_{\text{prf}} \) should be low as is seen in Fig. 6. However, even by lowering the transmit \( f_{\text{prf}} \), near the focal zone no overlapping occurs.

Figure 5. Synthetic aperture with beam steering and no translation. The virtual source is located in front of the aperture. \( D \) is the active aperture, and \( F \) denotes the focal point distance of the middle emission to the center of the active aperture.

Placing the virtual sources behind the transducer as it is seen in Fig. 7, can increase the overlapping between transmit beams. Increasing the steering angle of the defocused transmit beams, increases the overlapped region and thereby the synthesized aperture becomes larger. Since, the relations between aperture array design and the PSF also apply to the synthesized aperture array [10], to lower the side lobe levels and also to avoid grating lobes, the width of the synthesized array (synthesized apodization), and the pitch of the virtual elements must be considered. In the configuration shown in Fig. 7, due to a smaller synthesized aperture, the achievable lateral resolution is worse or equal to the lateral resolution achievable by the

\[
prf = \frac{c}{2r_{\text{max}}} = 5.13 \text{ kHz}.
\]

The possible number of emission per frame then becomes

\[
N_{\text{ems}} = \frac{5.13 \text{ kHz}}{20 \text{ Hz}} \approx 256.
\]
The placement of the virtual sources affects the imaging resolution achievable. The best achievable lateral resolution for a given ultrasound system is defined by its two-way beam width at the focal depth using conventional focusing on both reception and transmission [11]. For a rectangular aperture, using the Fraunhofer approximation, the beam width in the focal plane is equal to \( \lambda f_0 \), where \( f_0 \) is the number in transmission \( (f_0^t) \) or reception \( (f_0^r) \), and \( \lambda \) is the wavelength determined by the pulse center frequency. According to Fraunhofer approximation for a rectangular aperture, the transmit beam width can be written as

\[
\text{Transmit beam width} = \lambda f_0^t = \frac{\lambda F}{\alpha_t}, \quad (6)
\]

where \( \alpha_t \) is the transmit (synthesized) aperture size, and \( F \) is the transmit focus depth. The receive beam width can be written similarly

\[
\text{Receive beam width} = \lambda f_0^r = \frac{\lambda F}{\alpha_r}. \quad (7)
\]

The lateral two-way beam width can be found using the convolution of the transmit and receive beam profiles, which is equal to

\[
\text{Two-way beam width} = \lambda f_0^{tr} = \frac{\lambda F}{\alpha_t + \alpha_r} = \frac{\lambda f_0^t f_0^r}{\lambda f_0^t + \lambda f_0^r}. \quad (8)
\]

and therefore we also have

\[
f_0^{tr} = \left( \frac{1}{f_0^t} + \frac{1}{f_0^r} \right)^{-1}. \quad (9)
\]

As a side note, (9) states that when the transmit focal distance is approaching infinity \( (f_0^t \to \infty) \) in plane wave imaging, \( f_0^{tr} \approx f_0^r \). Thereby, the field of view becomes limited and only focusing in receive is possible. However, still due to the in phase summation of the low resolution images the SNR increases.

To increase the transmitted energy as well as the spatial resolution, the active aperture needs to be larger, therefore a setup has been considered, where all the elements on the transducer are used both in transmit and receive. For this setup, the active aperture could either be a 32 \( \times \) 32 square or a circle with radius of approximately 16 elements. To increase the circular symmetry of the PSF, the circular shape is chosen as the active aperture. The transmit aperture, which is static during all emissions is shown in Fig. 9a. The receive aperture, which is also static during all emissions is shown in Fig. 9b.

The resulting synthesized aperture is shown in Fig. 10. The transmit beam for the shown emission is illustrated with an arrow. The source of the beam is the active virtual source, shown with a circle. For each emission, a low resolution volume is beamformed. Each point in the low resolution volume is then weighted by a virtual source apodization, similar to what is shown in Fig. 11. The virtual source apodization has the shape of a cone centered around the transmit beam and with its apex located at the active virtual source. The angular width of
The cone depends on the focal distance of the active transducer array. The closer the virtual source is to the transducer surface, the wider is the cone. In this work, the cone angular width is 70°. The beamformed points located outside of the cone is weighted by 0. The points inside of the cone is weighted by a Hann window, centered on the transmit beam. The weight of a point within the cone then depends on its angular distance to the transmit beam. This procedure removes beamformed points located outside of the cone is weighted by an hourglass shape apodization, similar to what was determined as an adequate spatial sampling frequency to represent the PSF at the focal point. As determined earlier, the maximum \( f_{max} \) allows for 16 emissions per steering angle, leading to 16 \( \times 3 = 48 \) scan lines to be beamformed per steering angle per emission. To beamform the 3 \( \times 3 \) lines per emission, the area of the receive aperture should be almost four times as wide as the transmit aperture. The receive aperture is shown in Fig. 12b. A transmit focal depth of 30 mm was chosen based on trial and error. The transmit beam for the emission, a volume consisting of 9 receive lines are beamformed. Each point in the low resolution volume is then weighted by an hourglass shape apodization, similar to what is shown in Fig. 14. In this work, the hourglass angular width is 7°. Although different advanced methods have previously proposed to minimize the block-like artifacts with PB [1],[12],[13], in this study a 50% overlap between the receive lines of two adjacent transmit lines, and coherently compounding those lines, was used to compensate for those artifacts and therefore, in that regard it is different from the transitional PB technique.

VII. PARALLEL BEAMFORMING

Based on the 90° imaging field of view and the number of emissions in each direction, the transmit aperture width that results in a beam width of approximately 90°/16 = 5.63° was determined to be equal to 5.2 mm at 3 MHz or approximately 17 transducer elements on the Vernom probe. A circular aperture with a diameter of 16 elements, as shown in Fig. 12a, was chosen as the transmit aperture. For 3-D imaging using parallel beamforming \( N \times N \) receive lines per emission have to be beamformed. This can be derived from the ratio between (6)

\[
N = \frac{\text{Transmit beam width}}{\text{Two-way beam width}} = (f_{max} + 1) = \left(\frac{a_{tx}}{a_{rl}} + 1\right). \tag{10}
\]

Using the full aperture in reception, which has 32 active elements in each dimension, \( N = 3 \) scan lines per dimension were determined as an adequate spatial sampling frequency to represent the PSF at the focal point. As determined earlier, the maximum \( f_{max} \) allows for 16 emissions per steering angle, leading to 16 \( \times 3 = 48 \) scan lines to be beamformed per steering angle per emission. To beamform the 3 \( \times 3 \) lines per emission, the area of the receive aperture should be almost four times as wide as the transmit aperture. The receive aperture is shown in Fig. 12b. A transmit focal depth of 30 mm was chosen based on trial and error. The transmit beam for the emission is illustrated with an arrow in Fig. 13. For each emission, a volume consisting of 9 receive lines are beamformed. Each point in the low resolution volume is then weighted by an hourglass shape apodization, similar to what is shown in Fig. 14. In this work, the hourglass angular width is 7°. Although different advanced methods have previously proposed to minimize the block-like artifacts with PB [1],[12],[13], in this study a 50% overlap between the receive lines of two adjacent transmit lines, and coherently compounding those lines, was used to compensate for those artifacts and therefore, in that regard it is different from the transitional PB technique.

VIII. SIMULATION AND MEASUREMENT SETUPS

Table I lists the measurement configuration parameters. The RF-data were beamformed using the beamformation toolbox 3 (BFT 3) [14].

IX. RESULTS

In this section, the results of the comparison between synthetic aperture imaging and parallel beamforming are presented.
A. The Simulated Point Spread Function

In Fig. 15, three cross-planes (azimuth, elevation, and C-plane) of two 3-D PSFs imaged using synthetic aperture imaging as well as parallel beamforming are shown with a dynamic range of 60 dB. Both PSFs are normalized to their maximum values. The −6 dB and −20 dB main-lobes of the PB technique are seen to be smaller in both the elevation and azimuth directions, except the −40 dB. The side-lobe levels are lower for synthetic aperture imaging than for parallel beamforming, specifically at −40 dB. The side-lobes are seen to be asymmetrical, as they are wider in the elevation direction than in the azimuth direction. The asymmetric PSF is due to the asymmetry of the transducer array used. The discontinuities in the probe cause the increased side-lobe levels in the elevation direction. In the elevation direction it is hard to separate the side-lobes from grating-lobes. The FWHM and cystic resolution calculated values for both of the simulated 3-D PSFs are listed in Table II.

B. The Measured Point Spread Function

In Fig. 16, three cross-planes (azimuth, elevation, and C-plane) of two 3-D PSFs imaged using both parallel beamforming and synthetic aperture imaging techniques, at a depth of 62 mm are shown with a dynamic range of 60 dB. A needle with diameter of 300 µm facing towards the transducer was used as a point scatterer, therefore the secondary lobe after the main lobe in axial direction are a result of the scattered echoes from the needle shaft. The −6 dB and −20 dB main-lobes of the PB technique are seen to be smaller in both the elevation and azimuth directions, except the −40 dB. The side-lobe levels are clearly lower for synthetic aperture imaging than for parallel beamforming. For the same reason as described earlier, the side-lobes are seen to be asymmetrical, as they are wider in the elevation direction than in the azimuth direction. The FWHM and cystic resolution calculated values for both of the measured 3-D PSFs are listed in Table II. Considering the needle diameter of 300 µm, the simulation and measurement results are quite similar.

C. SNR

The estimated SNR is calculated from stochastic data and a limited amount of data is available due to the depth dependent SNR. Therefore averaging has to be employed to reduce the variance of the estimates. To measure the SNR of both imaging methods, a region of a tissue mimicking phantom without any cyst has been imaged 20 times. The noisy estimates are low,
pass filtered with a 3-D FIR filter. The measured SNR for both methods are shown in Fig. 17. The SNR of synthetic aperture imaging is higher than the SNR of parallel beamforming. Two SNR profiles along the indicated dashed lines in Fig. 17 are shown separately in Fig. 18. The penetration depth, where the SNR crosses 0 dB, is by linear regression estimated to be 149 mm for parallel beamforming and 183 mm for synthetic aperture imaging. In other words, synthetic aperture imaging increases the penetration depth by approximately 23%.

D. Cysts Embedded in Tissue Mimicking Phantom

In Fig. 19, three cross-planes (azimuth, elevation, and C-plane) of the cyst phantom are shown. Fig. 24c, 19c, and 19e are made with parallel beamforming and Fig. 24d, 19d, and 19f with synthetic aperture imaging. The cysts are clearly more apparent when imaged with synthetic aperture imaging than with parallel beamforming.

The cyst statistics are measured from a sphere with a radius of 6 mm located at the center of each cyst. The speckle statistics are estimated on the exact same spheres but on the tissue mimicking phantom containing only random scatterers. The CNR as shown in Fig. 20 was better for synthetic aperture imaging at all depths compared with parallel beamforming. When ignoring the first cyst, the CNR for parallel beamforming cysts decreased approximately linearly however, for synthetic aperture imaging it almost maintained throughout the imaging depth.

E. Abdominal Anatomic 3-D Phantom

A volumetric region of an abdominal anatomic phantom (Model 057A, CIRS, Virginia, USA) containing the liver has been imaged with both SAI and PB methods. The phantom simulates the abdomen from approximately the thorax vertebrae (T9/T10) to the lumbar vertebrae (L2/L3) using simplified anthropomorphic geometry. The materials provide similar acoustic features to human body. In Fig. 21, two cross-planes (azimuth and elevation) of the liver imaged with SAI technique are shown. Both methods are able to image the hepatic veins inside the liver of similar quality, with SAI having slightly better contrast. However, the number of emissions for SAI technique was able to image the anatomy of comparable quality to Fig. 21. Using only 5 emissions, SAI technique was able to image the anatomy as shown in Fig. 23e and Fig. 23f, which corresponding to a
volume rate of 1020 Hz. Although, lowering the number of emissions increases the temporal resolution, the SNR goes down.

F. Intensity Measurements

Before any in vivo measurements, the ultrasound imaging technique on the scanner has to fulfill all the requirements regarding the intensity levels and safety limits. Any damage to the tissues caused by cavitational effects or over-heating has to be avoided. In practice, the acoustic output is adjusted such that both the peak and the temporal average intensities remain under given thresholds. As of today, such safety guidelines are regulated by the the FDA [3], and take the form of upper limits on given indexes: the mechanical index \( MI \leq 1.9 \), the derated spatial-peak-temporal-average intensity \( I_{rpt,sa} \leq 720 \text{ mW/cm}^2 \) for peripheral vessel, \( I_{rpt,ca} \leq 430 \text{ mW/cm}^2 \) for cardiac, and the derated spatial-peak-pulse-average intensity \( I_{rpp,sa} \leq 190 \text{ mW/cm}^2 \), [3]. This requires to measure the emitted pressure of the transducer as a function of spatial position. The intensity measurements have been carried out using the experimental ultrasound scanner SARUS and the AI MS III intensity measurement system (Onda Corporation, Sunnyvale, California, USA)[15]. The measured mechanical index and the intensity as a function of depth and lateral position are shown in Fig. 24 for a \( f_{prf} = 100 \text{Hz} \). The frame rate is lowered to decrease the effect of reverberations and therefore the intensity values should be scaled by a factor of 51.33 for the actual measurements with \( f_{prf} = 5.133 \text{KHz} \). The measured MI and \( I_{rpt,sa} \) before scaling the excitation signal for PB are 0.77 and 109.8 mW/cm², and for SAI are 0.14 and 4 mW/cm², accordingly. The lower intensity values for SAI is due to the placement of the virtual sources behind the array, and as long as the probe can handle, the excitation voltage can increase before reaching the FDA limits.
When ignoring the cyst at 10 mm, the contrast for parallel beamforming increases faster compared with the PB technique. The cysts are water-filled pipes aligned 45° to the vertical scan plane. Due to a larger active transmit area of the SAI technique, the probe running in still air should be lower than the body’s temperature and also the temperature rise while using on a patient should be below 10 degrees Celsius for approximately half an hour. The measured temperature of the probe sole using both imaging methods are shown in Fig. 25. Due to a larger active transmit area of the SAI technique, the probe temperature increases faster compared with the PB technique.

However, both techniques satisfy the FDA safety requirements for in vivo measurements.

**H. In Vivo Measurement**

Two cross-planes (azimuth and elevation) of an in vivo volumetric data of a healthy male’s gallbladder, imaged with SAI and PB techniques, are illustrated in Fig. 26 and Fig. 27. Although both methods can visualize the gallbladder, however they both suffer from a low clinical value, attributed to the limited size of the used 2-D probe. Based on the Fraunhofer approximation, to increase the lateral resolution, the transmit and receive f-numbers should become smaller, which corresponds to increasing the aperture size. However, increasing the aperture size and keeping a λ/2-pitch result in a dramatic increase on the number of elements and thereby a large number of channels. The next upcoming chapters will try to investigate alternative ways to lower the number of channels required for 3-D imaging using a large 2-D arrays.

**X. Discussion and Conclusions**

The imaging quality of SAI was investigated in comparison with PB technique, which used to be the gold standard of 3-D ultrasound imaging. The comparison was based on Field II simulations, phantom measurements, as well as in vivo measurements with a λ/2-pitch 3.8 MHz 32×32 2-D transducer connected to the experimental ultrasound scanner SARUS. Two sequences with both SAI and PB techniques...
were designed for imaging a volume region with 90° × 90° field of view down to 15 cm at a 20 Hz volume rate. Using both simulations and measurements, it was shown that 3-D synthetic aperture imaging increases the imaging sensitivity compared with parallel beamforming. An iron needle facing towards the transducer inside a water tank used as a point target and was imaged with both techniques to characterize measured PSFs. The point target measurements were carried out at 0° steering angle and showed the same tendency as the simulations. Synthetic aperture imaging increased the contrast and had similar resolution. Measurements on a tissue mimicking phantom indicated that the penetration depth is deeper for synthetic aperture imaging compared with parallel beamforming. Synthetic aperture had a higher SNR than parallel beamforming at all depths and the increased SNR resulted in a penetration depth increase of 23%. The CNR was improved by 50% at 70 mm depth. The penetration depth reached the design goal of 15 cm for both methods. However, although both methods were able to reach the application requirements, the image quality was not comparable to conventional 2-D imaging. This is due to the small transducer array surface area as well as the limitation on the acoustic output.

Based on the acoustic intensity and temperature measurements, in the SAI configuration with virtual sources behind the transducer, the measured acoustic intensity and probe temperature are far below the FDA safety limits. This indicates that, as long as these safety limits are satisfied, we are allowed to increase the acoustic output. The setup used for the measurement limits the maximum value for the excitation voltage, however the transmitted acoustic energy can be boosted by using coded excitation, which increases the contrast and penetration depth. On the other hand, using a long excitation waveform increases the transducer temperature very rapidly, which might burn the transducer. Therefore, a fine tuning is required on the coded excitation waveform, which only can be done by knowing an accurate model for the transducer heat transfer model. This study presented some of the promising potentials of 3-D synthetic aperture imaging in terms of image quality and volume rate.

ACKNOWLEDGMENT

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REFERENCES

Figure 23. Synthetic aperture imaging of hepatic veins of liver in an abdominal anatomic phantom with different number of emissions. The dynamic range is 60 dB.


Derated Ita Ispta.3 = 0.08 mW/cm² at (x,y,z) = (0.00, 0.00, 18.00) mm

Derated MI = 0.14 at (x,y,z) = (0.00, 0.00, 14.00) mm

Derated MI = 0.77 at (x,y,z) = (0.00, 0.00, 10.00) mm

Derated Ita
Ispta.3 = 3.29 mW/cm² at (x,y,z) = (-1.00, 0.00, 10.00) mm

Figure 24. Synthetic aperture imaging (a and c) and parallel beamforming (b and d) derated measured intensities. The $I_{spta}$ intensities are measured for an $f_{pr,f} = 100$KHz and has to be scaled by a factor of 51.33 for the actual measurements with $f_{pr,f} = 5.133$KHz.

Figure 26. Two cross-planes (azimuth and elevation) imaged in vivo of the gallbladder are shown for SAI method. The dynamic range is 60 dB.

Figure 25. The probe sole temperature measurements for both SAI (left) and PB (right) imaging methods.

Figure 27. Two cross-planes (azimuth and elevation) imaged in vivo of the gallbladder are shown for PB method. The dynamic range is 60 dB.
Volumetric Ultrasound Imaging with Row-Column Addressed 2-D Arrays Using Spatial Matched Filter Beamforming

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Volumetric Ultrasound Imaging with Row-Column Addressed 2-D Arrays Using Spatial Matched Filter Beamforming

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Abstract—For 3-D ultrasound imaging with row-column addressed 2-D arrays, the two orthogonal 1-D transmit and receive arrays are both used for one-way focusing in the lateral and elevation directions separately and since they are not in the same plane, the two-way focusing is the same as one-way focusing. However, the achievable spatial resolution and contrast of the B-mode images in Delay and Sum (DAS) beamforming are limited by the aperture size and by the operating frequency. This paper, investigates Spatial Matched Filter (SMF) beamforming on row-column addressed 2-D arrays to increase spatial resolution. The performance is investigated on both simulated and experimentally collected 3-D data by comparing the Point Spread Functions (PSFs) and the phantom images obtained with standard DAS and with SMF. Results show that the SMF beamformer outperforms DAS in both simulated and experimental trials and that a higher contrast resolution can be achieved by SMF beamforming (i.e., narrower main lobe and lower side lobes).

Real-time 3-D ultrasonic imaging requires, 2-D array transducers [1]. The number of elements in a fully addressed $N \times N$ 2-D array scales with $N^2$. A 1-D array with a penetration depth of around 300 - 400 $\lambda$ will for an $f_s = 2$ at a depth of 200 $\lambda$ require an aperture size of 64 - 100 $\lambda$. This translates to 64 - 100 elements for a $\lambda$-pitch array, and 128 - 200 elements for a $\frac{1}{2}\lambda$-pitch array. Using a fully addressed 2-D array, this would ideally correspond to an array with more than 256$^2$ elements. Moreover to control the individual elements in the array, a connection has to be made to each element. However, addressing each element individually results in a vast amount of interconnections and offers a great challenge in acquiring and processing the large amount of data. Reducing the number of transducer elements by using sparse arrays has therefore attracted a great amount of interest in the last couple of decades. One of the drawbacks of sparse arrays, however, is the lower emitted energy from the reduced number of elements leading to a lower signal to noise ratio (SNR) in the recorded ultrasound image. The sparse arrays also have higher side-lobes and can introduce grating lobes in the field [2].

2-D row-column addressed arrays have recently received some attention [3]–[5]. In a row-column addressed array, the elements are accessed by their row or column index. Each row and column in the array thereby acts as one large element. This effectively transforms the dense 2-D array into two orthogonal 1-D arrays. This reduces the number of elements in an $N \times N$ 2-D array from $N^2$ to $2N$. Since fewer interconnections are needed, the cost of the system design is reduced.

By row-column addressing the elements on a 2-D matrix array, each row or column is acoustically equivalent to a line-element. The length of the line-elements results in prominent edge effects, which are due to the limited size of the aperture. It was shown that using hardware apodization along each row and column element, reduced those edge effects [6], [7]. In 3-D ultrasound imaging with row-column addressed 2-D arrays, the two orthogonal 1-D transmit and receive arrays are both used for focusing in the lateral and elevation directions separately. Even though it enables focusing in a 3-D volume, the spatial resolution of the two-way or transmit-receive focusing is equal to the one-way focusing in transmit or receive.

Spatial matched filter (SMF) beamforming is an algorithm, in which the impulse responses in transmission and in reception are considered for every point, unlike the DAS which assumes the impulse responses to be like delta functions [4], [8], [9]. Due to acoustically equivalent line-elements via row-column addressing the 2-D matrix array, SMF has the potential to improve the spatial resolution and the SNR. In [4], based on simulations, it was shown that SMF beamforming provides comparable results like synthetic aperture imaging with DAS beamforming. However this paper investigates SMF beamforming using synthetic aperture imaging.

In this study, Field II [10], [11] simulations are used to calculate the matched filter coefficients at all imaging points...
instead of the analytical solution. This work also involves measurement of the impulse response of a prototype row-column array. Implementation of the SMF method is highly dependent on accurate measurements of the overall system impulse response to provide appropriate spatial filters.

This paper is organized as follows: First an introduction to the spatial matched filter beamforming is given. Afterwards, the measurement and simulation setups are explained. Finally, the B-mode images with both SMF and DAS algorithms are shown for the simulated and measured data. The last section concludes the paper.

II. SPATIAL MATCHED FILTER BEAMFORMING

The signal from each channel of an array should be spatially matched filtered to align its output with that from the other channels so to add them constructively in phase. The received element signals are dependent on the element location and the scatterer’s position, and a new matched filter must be used depending on the element and on the scatterer’s position. The spatial matched filter \( m_p(\vec{r}_{tx}, \vec{r}_{rv}, t) \) is then given by [12]:

\[
\begin{align*}
    m_p(\vec{r}_{tx}, \vec{r}_{rv}, t) & = p_r(\vec{r}_{tx}, \vec{r}_{rv}, -t) \\
p_r(\vec{r}_{tx}, \vec{r}_{rv}, t) & = v_{pe}(t) + h_i(\vec{r}_{tx}, \vec{r}_{rv}, t) \ast h_j(\vec{r}_{rv}, \vec{r}_{tx}, t),
\end{align*}
\]

which is dependent on the transmitter location \( \vec{r}_{tx} \), the receiver element at \( \vec{r}_{rv} \), and the electro-mechanical impulse response of the transducer \( v_{pe}(t) \). The impulse responses during transmission and reception are \( h_i(\vec{r}_{tx}, \vec{r}_{rv}, t) \) and \( h_j(\vec{r}_{tx}, \vec{r}_{rv}, t) \), for the combined response for all of the array elements including their focusing and apodization. The focusing is then performed by adding the matched filtered signals from all the elements for the different locations

\[
r_i(\vec{r}) = \sum_{j=1}^{M} \int_{t_{ij}}^{t_{ij} + \Delta T_{ij}} v_i(\vec{r}_j, t) p_i(\vec{r}_j, \vec{r}_i, t) dt,
\]

where \( i \) designates the point in the image, \( j \) is the element number of the transducer, \( t_{ij} \) is the start of the response, and \( \Delta T_{ij} \) is the duration of the matched filter. The convolution integral in the equation is replaced by a correlation, since the time reversal of the response is replaced by the time reversal in the convolution.

It should be noticed that (2) can be used for any image point, and that it is only necessary to process the point in the image that must be displayed on the screen. The approach does not put any restrictions on the transducer geometry, excitation, focusing, apodization or impulse response. The approach can both be used for multi-element arrays and single element transducers, as long as the single element is moved compared to the scattering points during the imaging process in e.g. a polar scan. The approach improves on the focusing, if the pulse-echo spatial impulse responses are significantly different from a delta function. Normal delay focusing assumes that the geometric impulse response of the transducer is a delta function, and that the alignment can be done by merely delaying the responses. This is appropriate in the far-field for small element arrays and

![Fig. 1. Beamformed simulated PSFs of a scatterer positioned at (0,0,10) mm, using both SMF and DAS techniques. Roll-off apodization is applied along rows and columns (single cycle excitation, synthetic aperture imaging over elevation direction, i.e. transmit direction. (Note that SMF is applied in lateral direction, i.e. receive direction))](attachment:image.png)
at the focus for single element transducers. The approach will, thus, work best in the near field, where long spatial impulse responses are found. Specifically in row-column addressed 2-D arrays, where due to rather large elements the assumption of point sources for beamforming is inappropriate.

### III. Simulation and Measurement Setup

In this work, Field II is used for all simulations and also calculations of the spatial matched filters. The simulated receive signals are beamformed using two MATLAB (MathWorks Inc., Massachusetts, USA) specifically implemented DAS [7] and SMF beamformers for row-column addressed arrays. The transducer arrays used in the simulations are row-column addressed 62+62 element 2-D arrays using the parameters shown in Table I. The receive array is rotated 90° with respect to the transmit array. Field II is set up to use lines to describe the apertures and each line-element is divided into square mathematical sub-elements with a side length of $\lambda/4$.

Measurements are made with an in-house produced 62+62 element row-column addressed piezo array. An iron needle with a diameter of 300 µm facing towards the transducer and along its center line, was used as a point target in a water bath.

### IV. Results

In Fig. 1, the simulated beamformed PSFs are shown for both SMF and DAS beamforming algorithms for a single

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Table I: Transducer and Simulation Parameters

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</table>
cycle excitation with a roll-off apodization applied along each row and column (scatterer positioned at (0,0,10) mm). The secondary echoes are suppressed more efficiently with SMF comparing to DAS beamforming. Although the contrast has clearly improved, the spatial resolution still stays the same as DAS. In Fig. 2, the simulated beamformed PSFs are shown for a scatterer positioned at (0,0,30) mm. Fig. 4 illustrates the beamformed PSF images of measured RF-data of an iron needle positioned at (0,0,32.5) mm in front of the transducer. The probe had roll-off hardware apodization turned on, therefore the edge echoes are not visible on the final beamformed images. However, those secondary echoes are due to the reflection of shielding foil, which was covered the array. The measured pulse-echo impulse response of the row-column probe, which is shown in Fig. 3, has been used for the SMF beamforming. Fig. 4 also illustrates the performance of DAS and SMF beamforming on a cyst phantom measured data.

V. CONCLUSION

In this paper we demonstrate that the SMF beamforming algorithm is successfully employed for ultrasound B-mode image formation. Results of both simulated and experimental B-mode scans show that an increased contrast resolution, higher dynamic range and, consequently, better quality of the obtained images is achieved when using the SMF compared to standard DAS. This technique could be very promising for those applications which suffer from limited image contrast and resolution.

ACKNOWLEDGMENT

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Volumetric Synthetic Aperture Imaging with a Piezoelectric 2-D Row-Column Probe

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Volumetric Synthetic Aperture Imaging with a Piezoelectric 2-D Row-Column Probe

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ABSTRACT

The synthetic aperture (SA) technique can be used for achieving real-time volumetric ultrasound imaging using 2-D row-column addressed transducers. This paper investigates SA volumetric imaging performance of an in-house prototyped 3 MHz $\lambda/2$-pitch 62+62 element piezoelectric 2-D row-column addressed transducer array. Utilizing single element transmit events, a volume rate of 90 Hz down to 14 cm deep is achieved. Data are obtained using the experimental ultrasound scanner SARUS with a 70 MHz sampling frequency and beamformed using a delay-and-sum (DAS) approach. A signal-to-noise ratio of up to 32 dB is measured on the beamformed images of a tissue mimicking phantom with attenuation of $0.5 \text{ dBcm}^{-1} \text{MHz}^{-1}$, from the surface of the probe to the penetration depth of 300 $\lambda$. Measured lateral resolution as Full-Width-at-Half-Maximum (FWHM) is between 4 $\lambda$ and 10 $\lambda$ for 18\% to 65\% of the penetration depth from the surface of the probe. The averaged contrast is 13 dB for the same range. The imaging performance assessment results may represent a reference guide for possible applications of such an array in different medical fields.

Keywords: 3-D ultrasound imaging, 2-D row-column addressed transducer, synthetic aperture (SA)

1. INTRODUCTION

3-D ultrasound imaging using 2-D matrix array transducers enables acquiring volumetric images of soft tissue similar to X-ray computed tomography (X-ray CT) and magnetic resonance imaging (MRI), however with a better temporal resolution. In conventional 2-D ultrasound imaging, it is required to wait for the propagation of the ultrasound pulse back and forth in the medium for each single image line. In conventional 3-D imaging, on the other hand, the number of image lines is squared, hence a quadratic reduction on the achievable frame rate is imposed. To achieve higher volume rates, i.e., higher temporal resolution, broadened (defocused) transmit beams can be used, resulting in a reduced number of emissions.\textsuperscript{1} However, to recover the focusing capability in transmit, while using broad illuminations for a higher temporal resolution, it has been proposed to use synthetic aperture (SA) imaging, originally developed for radar detection systems.\textsuperscript{2} By coherently combining the data acquired from successive and spatially overlapping ultrasound pulse emissions, one may retrospectively recreate a dynamic transmit focus for the full volume. Despite the intrinsic imaging difficulties that exist in 3-D ultrasound imaging with 2-D arrays, there is still a practical challenge that remains to be addressed. The number of elements in a fully addressed $N \times N$ 2-D array scales with $N^2$. To control the individual elements in the array, a direct connection has to be made to each element. Hereby, any delay or apodization scheme can be applied, offering maximum control and flexibility in beamforming and image processing.\textsuperscript{3, 4} To give an example, a 1-D array with a penetration depth of around 300 $\lambda$ to 400 $\lambda$ will for an $f_0 = 2$ at a depth of 200 $\lambda$ require an aperture size of 64 $\lambda$ to 100 $\lambda$. This translates to 64 to 100 elements for a $\lambda$-pitch array, or 128 to 200 elements for a $\lambda/2$-pitch array. Using a fully addressed 2-D phased array, this would ideally correspond to an array with more than 200$^2$ elements. Addressing

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each element individually leads to both a great practical challenge in producing the interconnections from the fabrication point of view and in sampling and real-time processing the substantial amount of data. Likewise, the small size of the elements results in lowered capacitance and henceforth increased electrical impedance mismatch between the element and the cable connecting it to the ultrasound scanner requiring preamplifiers and matching circuits in the probe handle. Nevertheless, that numerous number of wires results in an impractically large cable from the transducer to the scanner.

Unlike the 1-D transducer array, where the focusing of the ultrasound wavefronts can be accomplished in only the lateral direction, in row-column addressed arrays, both orthogonal 1-D transmit and receive arrays are used for focusing separately. This enables focusing in a 3-D volume, since every point in the space can be addressed via the two-way focusing, i.e., the product of two orthogonal transmit and receive focus lines. By emulating the RF data of a 32 × 32 row-column addressed 2-D array, which was acquired with a 32 × 32 fully addressed 2-D matrix array, the imaging performance of a row-column addressed 2-D array, has been previously compared against a fully addressed 2-D array based on both simulations and measurements. In this study an in-house prototyped 62 × 62 row-column addressed 2-D probe with integrated hardware apodization, has been used to assess the imaging performance, based on measurements on phantoms, i.e., an iron needle in a water bath facing towards the transducer as a point source, a tissue mimicking phantom, and a wire grid phantom. The imaging performance assessment results may represent a reference guide for possible applications of such an array in different medical fields.

The remainder of the paper is organized as follows: In the next section, the characterization results are presented. Section 3 presents an overview of the row-column beamforming, its advantages and disadvantages. In Section 4, a detailed overview of the equipments and the measurement setup is presented. Section 5 explains the experimental results. The final section concludes the paper with suggestions and future work.

2. PROBE MANUFACTURING AND CHARACTERIZATION

The piezoelectric (PZT) row-column array was designed and fabricated with materials and processes commonly used in commercial medical ultrasound probes. Fig. 1a shows the orientation of the row and column elements on the array. The active layer consists of a 1-3 composite of high-dielectric PZT-5H and epoxy, and to achieve a favorable ceramic aspect ratio, the composite pitch was half the array pitch. The 1-3 composite was manufactured using conventional dice and fill technique, and was ground to obtained the desired thickness of 500 μm. A metal stack of titanium tungsten, nickel vanadium, and gold in that order were then sputtered on the top and bottom surfaces of the composite for electrodes. Row and column elements were defined by scribing the top electrode in one direction and scribing the bottom electrode in the orthogonal direction. This scribe was a shallow cut made with a dicing saw resulting in a kerf of 25 μm. Separately, a high-attenuation, mechanically rigid backing block and a stack of three quarter-wavelength matching layers were fabricated. All these components were glued together, with a thin electrical interconnect layer sandwiched between the composite and the backing block. The matching layers were then diced, and the dicing was aligned with the electrode grid to reduce the mechanical coupling between adjacent elements. An initial layer of planar room temperature vulcanized (RTV) silicone was used to fill the matching layer kerfs and an electrical shield (metalized polypropylene) was then applied. The array was then mounted into a 3-D printed plastic nose piece, after which a final layer of RTV silicone was applied and leveled such that no lens effect occurred. A flexible PCB from the array was then connected to two rigid PCBs with pre-amplifiers, and a 192-channel coaxial cable was attached to the latter. The entire probe electronics and the front of the array was wrapped in a shielding foil. This shield was then connected to the shield of the coaxial cable. Finally, the two 3-D printed parts constituting the handle were assembled to complete the row-column addressed probe. The fully assembled probe is shown in Fig. 1b.

The assembled row-column probe was characterized acoustically. The pulse-echo impulse response of each element was measured at Sound Technology, Inc. (State College, PA, USA) using a XCDR II Pulse Echo Test System by emitting with one element at a time against a plane stainless steel reflector placed in DI water 25 mm from the face of the probe. Due to the available setup, the elements were actuated with a square uniform pulse having an amplitude of 50 V. The received signal was then deconvolved with the excitation pulse to yield the impulse response. In Figs. 2a and 2b, the average impulse response and its envelope are shown for the rows and columns, respectively. In this setup, the columns are the top electrodes of the PZT array, while the rows are
the bottom electrodes. Note the two extra lobes after the main lobe around $-30$ dB starting at times $3.1 \mu s$ and $4.5 \mu s$. The time difference between the two is thus $1.4 \mu s$, which corresponds to the time difference between the main lobe and the first secondary lobe. This therefore suggests reflections within the probe. The reflections correspond to the shielding foil covering the array. In Figs. 2c and 2d, the impulse response spectra are shown for the rows and columns, respectively. The center frequency and $-6$ dB fractional bandwidth for rows are 2.99 MHz and 82%, and for columns are 2.99 MHz and 84%.

![Figure 1](image)

**Figure 1:** (a) Column electrodes are shown in blue and row electrodes in orange. The PZT material is shown in gray between the top and bottom electrodes. (b) In-house prototyped $3 \text{ MHz } \lambda/2$-pitch 62+62 element piezoelectric 2-D row-column addressed probe.

3. ROW-COLUMN BEAMFORMING

Delay-and-sum beamformers usually assume the geometry of the sound sources and receivers to be points. However, by row-column addressing the elements on a 2-D matrix array, each row or column is acoustically equivalent to a line-element. Furthermore, the emitted wavefront of a line-element has the shape of a cylinder surface: it is a plane wave in the plane aligned along the line-element and a circle arc in the plane orthogonal to the line-element.Assuming the geometry of the line-elements to be points is therefore a poor approximation. A better approximation assumes the line-elements to be line-segments. When an array of line-elements is focused, the geometry of the focal zone is also a line-segment. Calculating the distances between the line-elements and a given point should therefore be calculated as the distance between a line segment and a point. For beamforming with line-element sources, the time-of-flight for the sound propagating through the media has to be calculated as (See Ref. 9 for more details):

$$t_{TOP} = \frac{|\vec{r}_{p\text{-target}} - \vec{r}_{\text{point}}| \pm d (\vec{r}_{\text{point}} - \vec{r}_{p\text{-front})} + d (\vec{r}_{\text{receiver}} - \vec{r}_{p})}{c},$$

where $\vec{r}_{\text{target}}$ and $\vec{r}_{p\text{-front}}$ are the vectors for each transmit line-element (along the center of the element from one end to the other) and the focal line-segment (along the focal line from one end to the other). The $\vec{r}_{\text{point}}$ is the position vector of the beamforming point $p$. The $\vec{r}_{\text{receiver}}$ is the vector for each receive line-element. Note that the distance between the point $p$ and each of the transmit or receive line-elements, follows the method to find the minimum distance between a point and a line-segment. Moreover, $\pm$ in Eq. 1, refers to whether the focal line-segment is above or below a plane orthogonal to the center line of the beam. The minimum distance between the point $p$
Figure 2: Average impulse response normalized to excitation voltage and its envelope for row elements (a) and column elements (b) of the probe are shown. Note the two lobes after the main lobe, indicated by dashed black rectangles, which are due to reflections from the shielding foil. The impulse response spectra are shown for the row elements in (c) and for the column elements in (d). The center frequency and \(-6\) dB fractional bandwidth are indicated in the plot.

and the line segment \(\vec{a} \vec{b}\) is calculated as:

\[
d(\vec{a} \vec{b}, p) = \begin{cases} 
\frac{\|\vec{a} \times \vec{b}\|}{\|\vec{a}\|} & \text{if } 0 \leq \frac{\vec{a} \vec{b}}{\|\vec{a}\|} \leq 1 \\
\frac{\|\vec{a}\|}{\|\vec{b}\|} & \text{if } \frac{\vec{a} \vec{b}}{\|\vec{a}\|} < 0 \\
\frac{\|\vec{b}\|}{\|\vec{a}\|} & \text{if } \frac{\vec{a} \vec{b}}{\|\vec{a}\|} > 1.
\end{cases}
\]

However, the long length of the line-elements results in prominent edge effects. These edge effects are due to the limited size of the aperture and originating from both ends of the line-element. Therefore, any excitation to a line-element generates three wavefronts: the main wavefront and two edge waves. Considering the reciprocity principle in ultrasound, every received echo also generates three pulses on the receive element.\(^{11}\) Altogether there will be nine wavefronts in the pulse-echo response of a line-element. The secondary echoes after the main echo, will increase the uncertainty in the DAS beamforming process. It was shown that using hardware apodization
along each row and column element, which is different from the electronic apodization, will reduce those edge effects without altering the main echo response. An alternative approach to decrease the amplitude of the edge echoes and also increase the spatial resolution, is beamforming by using spatial matched filtering.

4. MEASUREMENT SETUPS

The volumetric data were acquired using the experimental ultrasound research scanner, SARUS. The measured RF signals were beamformed using a MATLAB (MathWorks Inc., Massachusetts, USA) implemented delay-and-sum beamformer for row-column addressed arrays. Table 1 lists the measurement configuration parameters. To assess the imaging performance of the PZT row-column array, several ultrasound phantoms are used. A geometrical copper wire phantom, where wires (or line targets) located at different depths have been used. A tissue mimicking phantom of comparable acoustical properties to human soft tissues also has been used. An iron needle with diameter of 300 µm facing toward the transducer and parallel to the center line of the transducer, was used as a point target in a water bath.

Table 1: Transducer’s parameters and setup configuration

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Center frequency</td>
<td>3 MHz</td>
</tr>
<tr>
<td>Pitch row (column)</td>
<td>270 µm</td>
</tr>
<tr>
<td>Number of rows (columns)</td>
<td>62 -</td>
</tr>
<tr>
<td>SA sequence (single-element emission)</td>
<td>90 Hz</td>
</tr>
<tr>
<td>Frame rate</td>
<td>90 Hz</td>
</tr>
<tr>
<td>Pulse repetition frequency</td>
<td>5 kHz</td>
</tr>
<tr>
<td>Emissions per frame</td>
<td>62 -</td>
</tr>
<tr>
<td>Number of active elements</td>
<td>124 -</td>
</tr>
<tr>
<td>Scan depth (max range)</td>
<td>14 cm</td>
</tr>
<tr>
<td>Emission cycles</td>
<td>2 -</td>
</tr>
<tr>
<td>Tx apodization</td>
<td>Hamming -</td>
</tr>
<tr>
<td>Rx apodization</td>
<td>Hamming -</td>
</tr>
<tr>
<td>Sampling frequency</td>
<td>70 MHz</td>
</tr>
</tbody>
</table>

Table 2: FWHM and cystic resolution measurements

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Measurement</th>
<th>Simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_{6dB}$</td>
<td>0.51</td>
<td>0.38 mm</td>
</tr>
<tr>
<td>$R_{12dB}$</td>
<td>0.82</td>
<td>0.61 mm</td>
</tr>
<tr>
<td>$R_{20dB}$</td>
<td>1.33</td>
<td>0.9 mm</td>
</tr>
<tr>
<td>Axial FWHM</td>
<td>532.5</td>
<td>325 µm</td>
</tr>
<tr>
<td>Azimuth FWHM</td>
<td>1375</td>
<td>1030 µm</td>
</tr>
<tr>
<td>Elevation FWHM</td>
<td>1312</td>
<td>1030 µm</td>
</tr>
</tbody>
</table>

5. RESULTS AND DISCUSSION

Figure 3 shows the measured and simulated 3-D point spread function (PSF) of SA in a water bath. In this study, Field II simulations are used to validate the measured results. The full-width at half-maximum (FWHM)
and the side lobe energy metrics (cystic resolution\(^6\)), for the measured and simulated PSFs are listed in Table 2. Lateral FWHM values are larger compared to 1-D transducers with the same lateral size. For row-column addressed 2-D arrays, the transmit focusing and receive focusing are not taking place in the same plane, but instead in planes perpendicular to each other and this affects the PSF size. Figs. 4 and 5 show the volumetric images of a tissue mimicking cyst phantom and a wire grid phantom. It is worth mentioning that in the wire grid image, the size of the line spread functions (LSF) is getting larger and therefore around the depth of 90 mm, it is not possible to discriminate individual wires from each other. Once more, due to one-way focusing, the lateral and elevation resolution as a function of depth, decreases faster than 1-D transducers of equivalent size. The cylindrical cyst regions are visible in the tissue mimicking phantom in Fig. 4. The contrast to noise ratio (CNR) over an anechoic cylindrical cyst region with diameter of 1 mm at the depth of 80 mm, (Fig. 4) on a tissue mimicking phantom with attenuation coefficient of 0.5 dB MHz\(^{-1}\) cm\(^{-1}\), was 0.54, compared to the background. The CNR was calculated as \(\mu_{\text{bck}} - \mu_{\text{cyst}} / \sqrt{(\sigma_{\text{bck}}^2 + \sigma_{\text{cyst}}^2)/2}\), where \(\mu_{\text{bck}}\) and \(\mu_{\text{cyst}}\) are mean gray level of background and cyst, \(\sigma_{\text{bck}}^2\) and \(\sigma_{\text{cyst}}^2\) are variances of gray levels within background and lesion, respectively. Fig. 6 shows the SNR of the SA imaging single-element sequence on a tissue mimicking phantom with an attenuation coefficient of 0.5 dB MHz\(^{-1}\) cm\(^{-1}\). A maximum SNR of 32 dB is measured and the penetration depth is about 300 \(\lambda\) for SNR=0 dB.

6. CONCLUSION AND PERSPECTIVES

In this study, the imaging performance of a piezoelectric 2-D row-column addressed probe for real-time 3-D synthetic aperture imaging was presented based on phantom measurements from the experimental ultrasound
Figure 4: Volumetric imaging of a hollow cyst phantom using the developed probe. Three cross-planes (elevation, azimuth, and C-plane) are shown from a beamformed volume of $30 \text{ mm} \times 30 \text{ mm} \times 100 \text{ mm}$ at a dynamic range of 50 dB. The origin corresponds to the center of the transducer surface. The C-plane was at $30 \text{ mm}$ depth.

Figure 5: Volumetric imaging of a wire phantom using the developed probe. Three cross-planes (elevation, azimuth, and C-plane) are shown from a beamformed volume of $30 \text{ mm} \times 30 \text{ mm} \times 100 \text{ mm}$ at a dynamic range of 50 dB. The origin corresponds to the center of the transducer surface. The spacing between wires is $10 \text{ mm}$. The C-plane was at $26 \text{ mm}$ depth.
scanner SARUS. It was shown that using synthetic aperture imaging on a 2-D row-column addressed transducer array a high volume rate imaging at a low cost could be achieved. Although 3-D visualizing gives the clinician a better insight for possible pathology and medical treatments, having a high volume rate makes it possible to capture dynamics, which otherwise cannot be detected by other 3-D imaging modalities. On the other hand, by row-column addressing, larger arrays are possible to build, which seemed impractical until recently. The direct benefit of a larger aperture size is to have a higher resolution, which makes it suitable for abdominal scans. The imaging performance results, which presented in this paper may represent a reference guide for possible applications of row-column addressed arrays in different medical fields.

REFERENCES


CMUT and PZT Row–Column Addressed 2-D Array Probes
Part I: Transducer Characterization

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CMUT and PZT Row–Column-Addressed 2-D Array Probes—Part I: Transducer Characterization

Mathias Engholm, Hamed Bouzari, Thomas Lehrmann Christiansen, Christopher Beers, Jan Peter Bagge, Lars Nordahl Moesner, Søren Elmin Diederichsen, Matthias Bo Stuart, Anders Lei, Jørgen Arendt Jensen, and Erik Vilain Thomsen

Abstract—This paper presents the characterization of two in-house prototyped fully integrated 62×62 row-column-addressed (RCA) 2-D array transducer probes based on capacitive micro-machined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technologies. Both probes have integrated apodization to reduce ghost echoes and are designed with similar acoustical features i.e. 3 MHz center frequency, λ/2-pitch and 24.8 × 24.8 mm² active footprint. The probes are assembled in a 3-D printed probe handle with electromagnetic shield and integrated electronics for driving the 128-channel coaxial cable to the scanner. The electronics are designed to allow all elements, both rows and columns, to be used interchangeably as either transmitters or receivers. The transducer characterization i.e. bandwidth, phase delay, surface pressure, sensitivity, insertion loss, and acoustical crosstalk are based on several single element measurements including pressure and pulse-echo and are evaluated quantitatively and comparatively. The calculated weighted center frequency is 3.0 MHz for both probes and the measured -6 dB bandwidth is 109 ± 4% and 80 ± 3% for the CMUT and the PZT probe, respectively. The surface pressures of the CMUT and PZT are 0.55 ± 0.06 MPa and 1.68 ± 0.09 MPa, respectively, and the receive sensitivities of the rows (receiving elements) are 12.9 ± 0.71 V/Pa and 13.7 ± 2.1 V/Pa. The expected penetration depth is therefore 3.4 times higher for the PZT than the CMUT.

I. INTRODUCTION

For the last 30 years, time resolved 3-D (4-D) imaging has received considerable interest, since it offers several advantages over conventional 2-D imaging. Images acquired using a traditional 2-D probe are dependent on the positioning and scan angle, making some imaging planes inaccessible due to the anatomy of the human body. Volumetric imaging does not have the same drawback, since any view angle is possible from the volume data. It also offers more accurate size estimations of organs, cysts, and tumors for diagnostics without relying on the assumptions and the operator skills needed for 2-D imaging estimations.

To obtain real time-resolved volumetric imaging with frame rates higher than 20 Hz, 2-D transducer arrays are necessary [1], [2]. Such transducers were first seen in the early 1990s [3]. By placing the elements in a rectangular grid, the beam can be steered electronically in two perpendicular directions (azimuth and elevation) and hereby acquire data from a volume. To obtain an image quality similar to that of a 1-D transducer, the same number of elements in both lateral dimensions is needed. A 1-D array of 128 elements would translate into 128 × 128 = 16,384 elements in a 2-D matrix array. From a transducer fabrication perspective, this poses a great challenge for providing electrical connections to all the elements while maintaining a high element yield. The interconnecting wires between the 16,384 elements and the ultrasonic system result in a large, heavy cable, which excludes it from any practical use.

The issue of reducing channel count, whilst maintaining the size of the array aperture, was addressed in the earlier versions of 2-D matrix arrays by introducing sparse arrays. Here only a subset of elements are active at the same time. Amongst these are Mills cross arrays, random arrays, and Vernier arrays, each presenting their benefits and drawbacks [4]–[8]. However, all of them suffer from reduced signal-to-noise ratio (SNR), due to the reduced active area, and introduce higher sidelobes and/or grating lobes. Recently, fully populated arrays with reduced channel count have become available by integrating electronic pre-beamformers inside the transducer probe [9]–[11]. Approaches to include the integrated circuit (IC) directly on a capacitive micromachined ultrasonic transducer (CMUT) has also been investigated, both by flip-chip bonding the CMUT to the IC [12]–[14] and monolithically integrating the CMUT on the CMOS [15], [16]. Integrating the electronics in the handle can result in much fewer signals to be funneled out to the ultrasound scanner. An example, of such a state-of-the-art fully populated matrix transducer, is the X6-1 PureWave xMATRIX Array from Phillips (Eindhoven, Netherlands), with 9,212 elements [17]. Despite the recent advances in real-time 3-D ultrasound imaging, the ultrasound systems supporting such imaging modalities are highly advanced and rely on cutting edge software, hardware, and manufacturing technology. This results in expensive equipment that impairs the low-cost advantage of ultrasound, thus limiting its more widespread use. Moreover, the thermal budget starts to become a consideration for modern probes with integrated electronics, due to the constraints on transducer probe heating dictated by the standards for medical equipment [18], [19].

Recently, an alternative to fully addressed matrix arrays has been suggested: The row–column-addressed (RCA) 2-D arrays,
first proposed in 2003 by Morton and Lockwood [20]. Row–
column-addressing of 2-D arrays is a scheme to reduce the 
number of active channels needed for contacting the elements 
in the array. The idea is to contact the elements in the 2-D 
array either by their row or column index. Each row or column 
thereby acts as one large element. This effectively turns the 
array into two orthogonal 1-D arrays. The imaging principle 
relies on using one of the 1-D arrays as the transmit array, 
creating a line focus of the transmit pulse. The perpendicular 
1-D array is used to receive, enabling receive focus in the 
orthogonal dimension. The combination of transmit and receive 
focus provides focusing on a point in the volume, hence a 
volumetric image can be created. Whereas an $N \times N$ fully 
addressed array needs $N^2$ connections, an RCA array only needs $2N$ connections. The RCA array can therefore have a 
larger aperture compared to the fully addressed array, having the 
same number of connections. A simulation study by Rasmussen 
and Jensen [21] and a measurements study [22], both compared 
the two different addressing schemes. With the same number 
of connections, a superior image quality is obtained using the 
RCA array.

An inherent drawback of the row–column-addressing, is that 
the long elements produce considerable edge effects, leading to 
ghost echoes in the beamformed image. Since the elements do 
not allow electronic control along their length, the ghost echoes 
cannot be removed with conventional electronic apodization. 
This issue was first addressed by Demoré et al. [23] and 
later investigated in detail by Rasmussen and Jensen [21]. 
Both studies concluded that integrating the apodization in the 
transducer itself, was an effective way of solving the issue. 
Several ways of realizing the integrated apodization have been 
suggested, including a variable polarization of the piezo ceramic 
material [24] and varying the density of CMUT cells [25].

Realizations of RCA arrays have previously been presented 
by several groups. The first experimental demonstration of 
RCA arrays were presented in 2006 by Seo and Yen [26]. 
The array was a piezoelectric transducer (PZT) in a 64x64 
layout, fabricated using a 1-3 ceramic with the row and column 
electrodes defined on separate sides of the ceramic. This array 
was later surpassed by the same authors with a 256x256 array 
using the same fabrication technique [27]–[29]. In 2009 Yen 
et al. introduced a simplified process for fabrication of RCA 
PZT arrays using a dual layer structure [30]. The dual layer 
structure was composed of a piezoelectric 2-2 composite for 
the transmit array, and a single sheet of undiced copolymer was 
used as the receive array. Row-column arrays based on CMUT 
technology were first presented in 2009 by Logan et al. [31]. 
They showed a 32x32 array fabricated using the wafer bonding 
process with a silicon nitride plate, and later they presented 
characterization of a similar array [32]. Zemp et al. [33] and 
Sampaleanu et al. [34] presented RCA arrays fabricated using 
the sacrificial release process and performed feasibility studies. 
More recently they have presented photoacoustic imaging using 
RCA CMUT arrays [35]. In 2015 Rasmussen et al. [36] and 
Christiansen et al. [37] presented a two-part paper presenting 
a RCA array with integrated apodization. The apodization was 
added as a static roll-off apodization region located at the 
ends of the line elements. They showed that the main lobe 
was unaffected by integrating this type of apodization. Part II 
showed experimental results of an CMUT RCA 2-D array 
with this roll-off apodization. The CMUT array was a 62x62 
layout with four apodization regions fabricated using the wafer 
bonding technique, two SOI wafers and a plate of highly doped 
silicon. In 2016 Zeshan et al. [38] presented a 32x32 RCA 
CMUT array fabricated using an anodizing bonding process and 
it was designed to provide a solution for macro-particle trapping 
and handling.

This paper is the first of a two-part paper describing the 
experimental results of two RCA 2-D array probes, one based 
on CMUT technology and one based on PZT technology. 
The probes are fully integrated RCA 2-D arrays equipped 
with integrated electronic apodization. Both of the transducers 
are designed with similar acoustical features, i.e. dimensions, 
center frequency, and packaging, and plugged into the research 
ultrasound scanner, SARUS [39]. This gives the unique 
possibility of evaluating the two probes relative to each other 
and comparing the row–column addressing scheme based on 
two different technologies. The scope of this paper is therefore 
to display the capabilities of RCA transducers, when integrated 
into probe handles, and to evaluate their performance. Part I 
describes the design, fabrication, and characterization of the 
two probes based on acoustical measurements. From these 
measurements the center frequency, bandwidth, phase delay, 
surface pressure, sensitivity, insertion loss, and acoustical cross-
talks are evaluated and discussed. Part II investigates the 
rectilinear volumetric imaging performance of these two probes 
based on phantom measurements [40]. The quality assessments 
of the B-mode images acquired with both probes, i.e. spatial 
resolution, contrast resolution, and SNR, are carried out based 
on the measurements over several phantoms using synthetic 
aperture imaging (SAI) technique.

Initial results of this study has been published in a conference 
paper [41] describing the development of the CMUT RCA 
probe. The array design, fabrication, electronics and probe 
assembly are partially included in this paper and extended 
as another probe was developed based on PZT technology 
using the same hardware architecture as the CMUT probe. 
This study extends the characterization of the CMUT probe 
by reporting the bandwidth, phase delay, surface pressure, 
sensitivity, insertion loss, and acoustical cross-talks across the 
array, thereby presenting a full characterization of both probes.

This paper is organized as follows: Section II covers the 
general design of the RCA transducer followed by the fabrication 
of both the CMUT and the PZT array. Section III describes 
the assembly of the probe and the integrated electronics. Section IV 
introduces the five different measurement setups and Section 
V contains the results from the characterization of the arrays. 
Section VI features a discussion of the results and a comparison 
of the two technologies, and finally a conclusion is presented 
in Section VII.

II. Array design and fabrication
The general design of the RCA arrays is based on the findings 
by Rasmussen et al. [36] and Christiansen et al. [37]. The arrays 
consist of 62 row elements and 62 column elements, and four
apodization regions. Only the 62+62 elements are connected to beamformer channels. The design of the RCA array can therefore be divided into two parts: The central region and the apodization region.

The central part of the array may be considered as a conventional RCA array, a 3-D diagram of a corner of such an array is shown in Fig. 1. The diagram includes four top and four bottom electrodes placed orthogonal to each other and colored orange and blue, respectively. Between the top and bottom electrodes is the "active" material, which is either the CMUTs or the piezoelectric material. The element contacts are placed alternately on each side of the array as showed in the figure. The top elements can be used as a 1-D array by grounding all of the bottom elements, and the bottom elements can be used as an orthogonal 1-D array by grounding all of the top elements.

The four apodization regions are located outside the central part of the array and are added to avoid the abrupt truncation of the elements, which gives rise to the ghost echoes [36]. The layout of the array including the apodization is shown in Fig. 2. The central part is showed within the dashed line and has an apodization value of 1. The apodization regions are placed on each side of the central region and the apodization value follows a Hann function from the edge of the central part to the edge of the array, where the apodization is 0. The apodization was originally developed for the CMUT array [37], but has been adapted to have the same dimension and roll-off characteristic for the PZT array.

Two arrays are fabricated using the design introduced above, one based on CMUT technology and one based on PZT technology. The pitch, number of elements, active footprint, center frequency, and excitation voltage are designed to be identical for the two arrays. This makes it possible to evaluate/compare the row–column-addressing scheme based on two different technologies. A center frequency of 3 MHz was chosen with half wavelength pitch and an excitation voltage of ±75 Vac. The dimensional parameters of both arrays are given in Table I.

### Table I

<table>
<thead>
<tr>
<th>Parameter</th>
<th>CMUT</th>
<th>PZT</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Array number of elements</td>
<td>62+62</td>
<td>62+62</td>
<td>–</td>
</tr>
<tr>
<td>Number of apodization region electrodes</td>
<td>4</td>
<td>4</td>
<td>–</td>
</tr>
<tr>
<td>Element pitch</td>
<td>270</td>
<td>270</td>
<td>µm</td>
</tr>
<tr>
<td>Element width</td>
<td>265</td>
<td>245</td>
<td>µm</td>
</tr>
<tr>
<td>Kerf</td>
<td>5</td>
<td>25</td>
<td>µm</td>
</tr>
<tr>
<td>Element length</td>
<td>24.84</td>
<td>24.84</td>
<td>mm</td>
</tr>
<tr>
<td>Acoustic window thickness</td>
<td>1.5</td>
<td>1.27</td>
<td>mm</td>
</tr>
<tr>
<td>Acoustic window velocity</td>
<td>1.0</td>
<td>1.0</td>
<td>mm/µs</td>
</tr>
<tr>
<td>Length of apodization regions</td>
<td>4.05</td>
<td>4.05</td>
<td>mm</td>
</tr>
<tr>
<td>Array outer dimensions (square)</td>
<td>26.3</td>
<td>26.3</td>
<td>mm</td>
</tr>
<tr>
<td>CMUT cell</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cell side length (square)</td>
<td>56</td>
<td>–</td>
<td>µm</td>
</tr>
<tr>
<td>Kerf between cells</td>
<td>7</td>
<td>–</td>
<td>µm</td>
</tr>
<tr>
<td>Plate thickness</td>
<td>1.85</td>
<td>–</td>
<td>µm</td>
</tr>
<tr>
<td>Al electrode thickness</td>
<td>400</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Vacuum gap height</td>
<td>448</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Nitride thickness</td>
<td>56</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Insulation oxide thickness</td>
<td>410</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Post oxide thickness</td>
<td>1346</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>PZT element</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PZT layer thickness</td>
<td>500</td>
<td>–</td>
<td>µm</td>
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<tr>
<td>PZT volume fraction</td>
<td>66</td>
<td>–</td>
<td>%</td>
</tr>
<tr>
<td>Electrode thickness</td>
<td>640</td>
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<td>nm</td>
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<tr>
<td>Matching layer net thickness</td>
<td>0.433</td>
<td>–</td>
<td>mm</td>
</tr>
<tr>
<td>Layer 1 (Closest to acoustic stack)</td>
<td>0.190</td>
<td>–</td>
<td>mm</td>
</tr>
<tr>
<td>Layer 2</td>
<td>0.125</td>
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<td>mm</td>
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<tr>
<td>Layer 3</td>
<td>0.118</td>
<td>–</td>
<td>nm</td>
</tr>
<tr>
<td>Matching layer bulk acoustic impedance</td>
<td>12 MRayl</td>
<td>–</td>
<td>MRayl</td>
</tr>
<tr>
<td>Layer 1 (Closest to acoustic stack)</td>
<td>4</td>
<td>–</td>
<td>MRayl</td>
</tr>
<tr>
<td>Layer 2</td>
<td>2.8</td>
<td>–</td>
<td>MRayl</td>
</tr>
<tr>
<td>Layer 3</td>
<td>1.95</td>
<td>–</td>
<td>mm/µs</td>
</tr>
<tr>
<td>Matching layers average velocity</td>
<td>100</td>
<td>–</td>
<td>dB</td>
</tr>
<tr>
<td>Backing round-trip attenuation at fc/2</td>
<td>100</td>
<td>–</td>
<td>dB</td>
</tr>
</tbody>
</table>
A. CMUT

The CMUT array layout is identical to the design introduced by Christiansen et al. [37]. This design employs 16 square cells in each sub element (crossing of a row and a column element) and the apodization regions were implemented by decreasing the amount of cells. An apodization value of one corresponds to 16 cells in a sub-element, and an apodization value of zero to 9 cells. Each CMUT cell is a square with a side length of 56 µm and with a silicon plate thickness of 1.85 µm with 400 nm aluminum on top. A gap of 410 nm silicon dioxide, 56 nm silicon nitride and 448 nm vacuum was employed to obtain a pull-in voltage of approximately 240 V.

The fabrication is based on the direct wafer bonding technique and the well-known LOCOS process (LOCal Oxidation of Silicon) inspired by Park et al. [42]. It is a nine step process using four lithography masks and two SOI wafers. A complete description of the fabrication is found in [41].

No backing was applied for the CMUT array, since it was calculated that the substrate ringing would not occur within the frequency band of the transducer itself [43], when a center frequency of 3 MHz and a substrate thickness of 500 µm were used.

B. PZT

The PZT RCA array was designed and fabricated with materials and processes commonly used in commercial medical ultrasound probes and were fabricated by Sound Technology, Inc. (State College, PA, USA). The active layer consists of a 1-3 composite of high-dielectric PZT-5H (CTS 3257 HD, CTS Corporation, Albuquerque, NM, USA) and epoxy. To achieve a favorable ceramic aspect ratio, the composite pitch was half the array pitch with a volume fraction of 66%. The apodization region has been integrated into the composite structure, but due to present intellectual property limitations the details of how it is integrated are not included in this paper.

An illustration of the individual fabrication steps is shown in Fig. 3. The 1-3 composite was manufactured using conventional dice and fill technique, and was ground and polished to obtain the desired thickness of 500 µm, as shown in Fig. 3(a). A 640 nm metal stack of titanium tungsten (TiW), nickel vanadium (NiV) and gold (Au), in that order, were sputtered on the top and bottom surfaces of the composite for electrodes. TiW functions as an adhesion promoter for the NiV electrode material with gold as a passivating layer on top. Row and column elements were defined by scribing the top electrodes in one direction and the bottom electrodes in the orthogonal direction. This scribe was a shallow cut made with a dicing saw resulting in a kerf of 25 µm, see Fig. 3(b). Separately, a high-attenuation, mechanically rigid backing block and a stack of three quarter-wavelength matching layers were fabricated. The backing material was glued on to the backside with a thin electrical interconnect layer sandwiched between the composite and the backing block, and the matching layers were glued on to the front side of the composite, see Fig. 3(c). The matching layers were then diced, and the dicing was aligned with the electrode grid to reduce the mechanical coupling between adjacent elements, see Fig. 3(d).

Figure 3. The process flow of the PZT fabrication. (a): 1-3 composite. To achieve a favorable ceramic aspect ratio, the composite pitch was half the array pitch. (b): Electrodes defined. Top electrodes run perpendicular to the page; bottom electrodes run parallel to the page. (c): Components glued together. Bottom to top: backing, interconnect, composite, matching layers. (d): Matching layers diced. The dicing is aligned with the electrode pitch.

III. PROBE ASSEMBLY

The assembly of the two probes is almost identical after the fabrication of the arrays, and both were assembled at the facilities of Sound Technology Inc. (State College, PA, USA). The probes are composed of four parts in addition to the transducer array itself, see Fig. 5. These are a flexible printed circuit board (PCB) for connecting the transducer to the electronics, the integrated electronics containing buffer amplifiers, a cable for connecting the probe to the scanner, and a 3-D printed probe handle.

A. Assembly

The arrays are mounted on a flexible PCB and are rigidly supported directly beneath the arrays. The PZT array connection to the PCB involves soldering, whereas the CMUT array uses wire-bonding. Both arrays are mounted in such a way that the rows are the top electrodes and the columns are the bottom electrodes. A glob-top dam was applied around the CMUT array to ensure protection of the wire-bonds against mechanical stresses, when pressing the probe against an object. Both arrays then underwent an encapsulation step in which an initial layer...
arrays have flexible printed circuit interconnect components with identical connectors and pin maps, so they were attached to identical amplifier PCBs. These amplifier PCBs were attached to a 128-channel coaxial cable and wrapped with copper tape to provide an electromagnetic shield. The copper shield was connected to both the shielding foil applied to the front of the array and the shield of the coaxial cable, and hereby, also the reference ground on the scanner. Finally, the two 3-D printed plastic handle halves were glued to each other, the nose piece, and the strain relief to complete the probe assembly.

B. Electronics

The interconnect electronics are the same as used by Christiansen et al. [37], however the MAX4805A amplifier (Maxim Integrated, San Jose, CA, USA) has been exchanged with the ADHV1301 amplifier (Analog Devices, Norwood, Ma, USA). The two electronics PCBs in the probe contain preamplifiers with a 0dB voltage gain, and they work as high impedance buffer amplifiers that remove the electrical load of the cable on the arrays. The PCBs also contain passive high pass filters for separating the DC bias voltage from the AC voltage signals. The electronics are designed to allow all elements to be used interchangeably as either transmitters or receivers. Both probes are equipped with the exact same integrated electronics, but no bias voltage is supplied to the PZT probe.

The ADHV1301 is an application specific standard product (ASSP) integrated circuit with a similar performance to the MAX4805A, i.e. a high-voltage-protected, low-noise operational amplifier designed to be used as an amplifier for in-probe buffering and amplification. The device has 16 input channels, a 30MHz − 3dB bandwidth and the small-signal output impedance of amplifiers is 18Ω suitable for matching the cable impedance. The voltage and current noise is 1.7nV/√Hz and 2.1pA/√Hz, respectively. The high voltage protection circuit is activated for voltages larger than 1000mV and the maximum allowed voltage is ±125V. The recovery time after a transmit burst is less than 800ns.

IV. MEASUREMENT SETUP

Characterization of the two developed RCA probes was performed using five different measurement setups. Two measurements were carried out during, and directly after, the assembly at the facilities of Sound Technology Inc. (State College, PA, USA): One measurement dedicated to measuring the element capacitance and one for ascertaining the two-way impulse response. The last three measurements were performed using the experimental ultrasound system, SARUS [39], one to obtain the emitted pressure, the second is a pulse-echo measurement, and third is to asses the acoustical crosstalk. The measurement setups are introduced below.

A. Capacitance (STI)

The element capacitances were measured using a HP 4263B LCR meter (Hewlett-Packard Company, CA, USA). Because the amplifiers in the probe handle have protection circuitry
that is incompatible with the LCR meter, the measurement was performed before the final assembly. The transducers were connected to a test cable, which was plugged into a multiplexer box connected to the LCR meter. In order to neglect the parasitic capacitance and inductance of the setup, all channels of the multiplexer and the test cable were measured independently.

B. Impulse Response (STI)

The pulse-echo response of each element was measured using an XCDR II Pulse Echo Test System, by emitting and receiving with one element at a time against a planar stainless steel reflector. The planar reflector was placed in deionized (DI) water 25 mm from the face of the probe. With the available setup, the elements were actuated with a square unipolar pulse with a duration of 100 ns for the PZT and 150 ns for the CMUT and having an amplitude of 50 V. The CMUT elements were biased at 200 Vdc. The system was set to sample at a sampling frequency of 500 MHz. The received signal was de-convolved with the excitation pulse to yield the two-way element-element impulse response.

From the impulse response and the corresponding spectrum, the center frequency, the bandwidth, and the phase delay of each element were found.

C. Hydrophone (SARUS)

The pressure was measured using an HGL-0400 hydrophone connected to an AC-2010 pre-amplifier (Onda Corporation, CA, USA). The hydrophone was placed in front of the transducer surface and scanned over each element using the position system of the intensity measurement AIMS-3 (Onda Corporation, CA, USA), while transmitting a 3 MHz, 4-cycle sinusoidal excitation pulse on the element being measured. An amplitude of 75 Vac was used and the CMUT probe was biased with 200 Vdc. The pressure was recorded at 5.8 mm and 5.9 mm for the PZT and CMUT, respectively.

D. Pulse-Echo and Crosstalk (SARUS)

A plane stainless steel reflector was positioned in a water tank at a distance of 7.3 cm from, and parallel to, the transducer surface of the probe being characterized. The transmit signals were generated using the experimental ultrasound system, SARUS [39], which also recorded the received signals. Twenty realizations of a 3 MHz, 4-cycle sinusoidal excitation pulse was transmitted on one element at a time and received on all elements, both rows and columns. The 20 realizations were averaged to minimize the noise. The system was set to sample, at a sampling frequency of 70 MHz, down to a depth of 10 cm. A second measurement was performed using the same setup, but without the planar reflector, to assess the acoustical crosstalk.

V. TRANSDUCER CHARACTERIZATION

This section describes the characterization of the two probes based on the measurements introduced in the previous section. The performance of the two probes is evaluated concurrently and is described below. Table II summarizes the main results of the characterization, allowing easy location of specific parameters.

A. Capacitance

The capacitance measured using the LCR meter also includes a contribution from the multiplexer and the test cable. To isolate the element capacitances from the LCR measurements, the independent measurements of all the channels of the multiplexer and the test cable were subtracted from the overall measurements (multiplexer + test cable + array). The element capacitances of both probes are shown in Fig. 6. Some of the element capacitances appear to be missing; this is due to an artifact of the measurement setup. A high series resistance results in a faulty detection of the capacitance and is, therefore, omitted. This might be the effect of inferior connections to the specific elements. There is a high level of uniformity between the probes reflecting good control of the fabrication methods.

The bottom elements (1-62) of the CMUT probe have a higher capacitance than the top elements (63-124). This is due to a capacitive coupling to the substrate of the bottom SOI wafer [32], [37], [41], [44]. When a top electrode is probed, all bottom electrodes are grounded. As seen in Fig. 7(a), only the capacitance from the top electrode to the grounded bottom electrodes, denoted $C_{CMUT}$, is seen in this configuration. The situation when a bottom electrode is probed is illustrated in Fig. 7(b). In this case, the substrate will appear grounded, since the signal may follow a path through the substrate and couple to ground via the neighboring bottom electrodes as shown in the figure. Although the substrate has a non-negligible impedance $Z_s$, the parallel coupling of the bottom electrodes results in a relatively low-impedance path to ground. Therefore, the capacitance measured, when probing a bottom electrode, will have a contribution from both $C_{CMUT}$ and $C_{BOX}$. The capacitance of the BOX, $C_{BOX}$, is approximately 230 pF, in agreement with the 1 µm silicon oxide between the bottom electrode and the substrate, and the dimensions of the electrode.

B. Impulse response

The signals received from the impulse response measurements, described in Section IV-B, were de-convolved with the excitation pulse to yield the two-way element-element impulse response. Fig. 8 shows the average two-way element-element impulse response and the associated envelope of the CMUT (a) and the PZT probe (b). The solid line represents the impulse
Table II

<table>
<thead>
<tr>
<th>Parameter</th>
<th>CMUT</th>
<th>PZT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Capacitance [pF]</td>
<td>339.4±0.8 136.1±1.4</td>
<td>339.6±102.1</td>
</tr>
<tr>
<td>Center frequency [MHz]</td>
<td>2.97±0.07 3.05±0.09</td>
<td>3.01±0.09</td>
</tr>
<tr>
<td>Bandwidth [%]</td>
<td>111±3 106±4</td>
<td>109±4</td>
</tr>
<tr>
<td>Phase delay [°]</td>
<td>0±5.0 0±3.1</td>
<td>0±4.5</td>
</tr>
<tr>
<td>Surface Pressure [MPa]</td>
<td>0.56±0.05 0.54±0.07</td>
<td>0.55±0.06</td>
</tr>
<tr>
<td>Sensitivity [µV/Pa]</td>
<td>129±0.7 4.3±0.7</td>
<td>8.5±4.4</td>
</tr>
<tr>
<td>Insertion loss [dB]</td>
<td>−26.4±0.9 −36.5±2.5</td>
<td>−31.5±5.4</td>
</tr>
<tr>
<td>Nearest neighbor crosstalk [dB]</td>
<td>−30.5±0.8 −26.3±1.4</td>
<td>−28.4±2.4</td>
</tr>
<tr>
<td>Transmit-receive elements crosstalk [dB]</td>
<td>−39.9±0.2 −40.2±0.6</td>
<td>−40.0±0.4</td>
</tr>
</tbody>
</table>

C. Center frequency

The center frequency is calculated as a weighted mean of the frequencies present in the received signal as:

\[ f_c = \frac{\sum_{i=1}^{N} S(f_i/N) \cdot f_i/N}{\sum_{i=1}^{N} S(f_i/N)} \]  

where \( N \) is the number of frequency bins in the two-sided spectrum.

Fig. 10 shows the uniformity of the center frequency across the arrays. The both probes have a center frequency of 3MHz as they were designed for, and only a small smooth variation across the arrays is observed. This smooth variation indicates that the variations are mostly caused by non-uniformities of the silicon plates of the CMUT probe, and thickness-variations of the piezoelectric material of the PZT probe.

D. Bandwidth

The −6dB bandwidth was determined from the difference in frequency between the −6dB points in the frequency spectrum. A mean bandwidth of 3.26±0.02MHz and 2.39±0.02MHz is found for the CMUT probe and the PZT probe, respectively. The fractional bandwidth are calculated from the bandwidth relative to the weighted center frequency, and a mean value of 109% and 80% are found for the CMUT and the PZT, respectively. The uniformity across the array for both probes is shown in Fig. 11. The probes have a high uniformity with a standard deviation of the fractional bandwidth of 4% for the CMUT probe and 3% for the PZT probe.

E. Phase delay

The phase delay was found by cross-correlating the impulse response for each element with the mean impulse response and interpolating to find the lag of the maximum of the cross-correlation. Correction for any linear slope due to misalignment between the transducer and the plane reflector was done, and the mean was set to zero. The phase delay was then calculated by dividing the time it takes the wave to travel one wavelength at 3MHz, and multiplying it by 360°, to obtain the phase delay in degrees. Fig. 12 shows the phase delay across the array for the CMUT and the PZT in top and bottom, respectively.

No curvature is seen of the CMUT, however the PZT is observed to curve. The bottom/column elements phase delays

![Figure 7 Illustration of the electrical circuit seen by the LCR meter when probing a top electrode (a) and a bottom electrode (b). The figures show a cross-section, such that the top electrodes are oriented perpendicular to the cut in (a) while the array is rotated 90° in (b).](image-url)
are seen to have a concave profile, whereas the top/row elements have a convex profile. This saddle shape is believed to originate from stress build up during the assembly.

F. Surface pressure

The surface pressure was derived from the hydrophone measurement with the setup described in section IV-C. The recorded pressure was compensated to find the emitted pressure at the transducer surface. This compensation factor was calculated by simulating a single element in Field II [45], [46]. The element was set to emit a 3 MHz, 4-cycle sinusoidal wave, and the pressure magnitude relative to the pressure magnitude at the element surface was simulated. The compensation factor
for the PZT and CMUT was 9.14 and 8.83, respectively. The difference was caused by the different locations of the hydrophone during the two measurements. The surface pressure across the arrays is shown in Fig. 13. The mean values for the CMUT and PZT are 0.55 ± 0.06 MPa and 1.68 ± 0.09 MPa, respectively.

Notice that there is no difference between the pressures emitted by the CMUT columns (elements 1-62) and the CMUT rows (elements 63-124). One would expect a lower emitted pressure from the columns due to the increased parasitic capacitance, hence a lower coupling coefficient. This is, however, not the case, since the power source during the emission is not limited in the amount of energy it can supply to the transducer.

Figure 12. Phase delay across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows.

Figure 13. Surface pressure across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows.

Figure 14. Sensitivity across the array elements of both probes. Element number from 1-62 corresponds to the columns and 63-124 to the rows.

G. Receive sensitivity

The receive sensitivity is calculated by combining the result from hydrophone measurement (Section IV-C) with the result from pulse echo measurement (Section IV-D). The receive sensitivity of the transducers is found by dividing the received voltage signal after a pulse-echo event with the incident pressure. The incident pressure was deduced using the pressure measured from the hydrophone setup. The pressure drop was compensated using the same Field II model described in the previous section. Besides compensating the incident pressure for the diffraction loss, the non-ideality of the plane reflector is also compensated for. The reflection coefficient for a normal incident wave is solely determined from the acoustic impedance discontinuity in the transmission medium. In water, the reflection coefficient for a stainless steel reflector is 0.93 [47]. The receive sensitivity for each element across the two probes is shown in Fig. 14. The mean values of the CMUT and the PZT probe are 8.5 ± 4.4 μV/Pa and 14.4 ± 1.9 μV/Pa, respectively.

The sensitivity of the CMUT bottom/column elements is 67% lower than the top/row elements. This is due to the capacitive coupling to the substrate discussed in Section V-A. The two parallel coupled capacitors (C_{CMUT} and C_{BOX}, Fig. 7(b)) act as a current divider, resulting in the lower detected voltage (lower sensitivity). When imaging with RCA arrays, either the rows or columns are used as transmitters and the orthogonal elements as receivers. Choosing the bottom/column elements as the emitters and the top/row elements as receivers, the imaging is not affected by the lower sensitivity. However, determining 3-D vector flow might be affected since the sequence uses both rows and columns as emitters and receivers [48], [49].

H. Insertion loss

The insertion loss is the loss of signal power resulting from the "insertion" of the device and is a measure of the overall round-trip efficiency (round-trip loss of signal power). It is calculated as the ratio of voltage received by an element after a pulse echo event, V_R, to the transmit voltage used to excite the element, V_T [47]. The received signal is compensated to exclude the loss of signal due to diffraction and for the non-ideality of the planar reflector. A log-compression of the ratio yields the insertion loss in dB:

\[ \text{Insertion loss (dB)} = 20 \log_{10} \frac{V_R}{V_T} \] (2)

The insertion loss across the array is shown in Fig. 15. The mean value of the PZT probe is −15.9 ± 1.5 dB and for the CMUT probe the rows and columns are −26.4 ± 0.9 dB and −36.5 ± 2.5 dB, respectively. Since the receive sensitivities of the two arrays are similar, the lower insertion loss of the CMUT probe is mainly an effect of the lower transduction efficiency from the mechanical domain to the acoustic domain i.e. due to the lower surface pressure. The insertion loss is mainly a parameter used in the next section when estimating the acoustical crosstalk.
I. Acoustical crosstalk

The second measurement setup described in Section IV-D is used for evaluating the crosstalk. The first 3 μs of the received data are disregarded because the receivers are saturated due to the transmit pulse (electrical crosstalk) and the ring-down of the electronics.

Two different types of acoustical crosstalk can be evaluated when using RCA arrays: Nearest neighbor crosstalk and transmit-to-receive crosstalk [50]. Emitting with one element at the time and extracting the maximum of the signal from its neighbor yielded the nearest neighbor crosstalk for every element. To provide a relative measure, the signal was normalized to the transmit voltage after the latter was corrected for the insertion loss. The insertion loss is reported in Section V-H. The correction corresponds to a normalization of the neighbor’s signal to the signal that the emitting element would have received, if the transmitted pulse was reflected right at the transducer surface and subsequently received by the emitting element. Thus, it yields the relative acoustical coupling from one element to its neighbor. The nearest neighbor crosstalk across the probes is shown in Fig. 16 and the mean values are \(-28.4 ± 2.4\) dB and \(-30.0 ± 2.2\) dB for the CMUT and PZT probe, respectively. The nearest neighbor crosstalk of the CMUT is roughly 5 dB lower than what have earlier been reported in literature [50], [51]. The lower crosstalk of the CMUT is roughly 5 dB lower than what have earlier been reported in literature [50], [51]. The lower crosstalk is caused by the RTV on top of the array. The amount of crosstalk for the PZT probe is in the limit of what is usually accepted for ultrasound probes. Ideally for phased arrays, one would dice into the piezoelectric ceramic during fabrication and fill it up with the RTV to reduce the crosstalk. This is, however, not possible with row-column arrays.

To provide a measure of the cross-talk in an imaging setup, the transmit-to-receive cross-talk are estimated. This is calculated as the average of the maximum signal received on all elements orthogonal to the emitting element. The average is normalized to the transmit voltage of the emitting element and corrected for the insertion loss. The transmit-to-receive elements crosstalk is shown in Fig. 17 for both arrays and the mean values are \(-40.0 ± 0.5\) dB and \(-53.7 ± 0.9\) dB for the CMUT and PZT probe, respectively. This is consistent with results in literature for the CMUT [50] and has not been previously reported for PZT RCA arrays.

VI. Discussion

We have presented the development and transducer performance of two RCA probes for real-time volumetric imaging based on two competing technologies: PZT and CMUT. The central part of this paper has been to characterize the two developed transducers. The characterization should not be seen as a comparison of the two technologies, but as a display of the capabilities of the row-column-addressing scheme using these two technologies. However, since these two technologies are evaluated next to each other, one cannot avoid comparing them. The strengths and weaknesses of the emerging technology, CMUT, will therefore be discussed in relation to the traditional technology, PZT.

One of the most highlighted advantages of the CMUT is its higher bandwidth relative to the PZT technology. The mean –6 dB bandwidth is 29 percentage pointa higher for the CMUT probe compared to the PZT. The higher bandwidth is caused by the CMUT acting as an overdamped system, due to the low impedance of the vibrating plate in immersion. One consequence of the high bandwidth is an increased axial resolution, which will be assessed in the companion paper, Part II [40]. Another interesting advantage of the high bandwidth is tissue harmonic imaging. Usually one will emit at 2/3 of the center frequency, \(f_c\), and receive at 4/5\(f_c\). Having a higher bandwidth results in a relative higher emitted pressure and receive sensitivity at those frequencies.

A current limitation of the CMUT technology is the lower emitted pressure. This is a result of the low inertia of the plate (thin plate, low mass). The surface pressure of the PZT probe is...
consequently 3 times higher than the CMUT probe. Contrary to expectations, the mean receive sensitivity of the PZT probe is 11% higher than the top/row elements of the CMUT probe. The sensitivity of the CMUT array can be improved by optimizing the structure, plate design, layout, and driving conditions. Packing the cells closer will increase the effective area. The CMUT structure can be designed to decrease the parallel parasitic capacitance originating from the bonding area between the cells. This could be implemented by incorporating a bump in the cavity as introduced by Park et al. [52]. This facilitates the possibility of having a high ratio of post oxide thickness to gap height. Improving the driving conditions also makes it possible to increase performance of the CMUT probe. The bias voltage is closely related to the electro-mechanical coupling coefficient describing the efficiency at which the mechanical energy (vibrations) is converted to electrical energy and vice versa. The coupling coefficient approaches unity at the pull-in voltage [53]. Increasing the bias voltage will result in a higher receive sensitivity and emitted pressure. The bias voltage of the probe is in this study limited to 200 V because of the integrated electronics. As a result, the probe is operated at a maximum of 83% of the pull-in voltage. The optimal driving conditions and the gain hereof will be investigated in future research. If both the emitted pressure and the receive sensitivity is taken into account, one will expect the penetration depth of the PZT probe is 3.4 times higher compared to the CMUT probe at these driving conditions. A potential way of increasing the pressure, and hence the penetration depth, is by emitting with more than one element. However, the pressure generated by the transducer is usually limited by both the mechanical index and the temperature of the probe itself. All of these aspects are investigated in the companion paper, part II [40].

VII. CONCLUSION

This paper presented the development and characterization of two 62×62 RCA ultrasound probes based on CMUT and PZT technology. The objective of the paper has been to show the capabilities of the RCA transducer implemented using the two specific technologies. Both transducers have integrated apodization to reduce ghost echoes and are designed with similar acoustical features. They were designed to be used with a commercial scanner made for conventional 2-D imaging, due to the low channel count of the RCA probes, the probes and scanner can be directly interfaced. A solution with a flexible mounting PCB and two rigid amplifier PCBs was used to mount the array and interface it to the scanner cable via buffer amplifiers. The array and electronics were electrically shielded with a metal foil, and the array was covered with a silicone coating before the entire probe was encapsulated in a 3-D printed handle. The reliability and performance of the probes were assessed through electrical and acoustical measurements. Four different measurement setups were used and the probes electrical capacitances, center frequencies, bandwidths, phase delays, surface pressures, receive sensitivities, insertion loss, and acoustical crosstalks were evaluated. The weighted center frequency is exactly 3.0MHz for both probes, as they were designed for. The −6dB fractional bandwidth were 29 percentage point higher for the CMUT probe than the PZT. The surface pressure of the PZT probe is a factor 3 times higher relative to the CMUT probe, and the expected penetration depth is 3.4 times higher in favor of the PZT. The authors emphasize that the driving conditions of the CMUT probe was limited by the integrated electronics in the probe handle, which could otherwise have improved its performance.

In part II of this work, the imaging performances are evaluated [40]. The quality assessments of the B-mode images acquired with both probes, i.e. spatial resolution, contrast resolution, and SNR, are carried out based on the measurements over several phantoms using SAI technique.

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CMUT and PZT Row–Column Addressed 2-D Array Probes Part II: Imaging Performance Assessment

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CMUT and PZT Row–Column-Addressed 2-D Array Probes—Part II: Imaging Performance Assessment

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Abstract—This study evaluates the volumetric imaging performance of two prototyped 62–62 row–column-addressed (RCA) 2-D array transducer probes using synthetic aperture imaging (SAI) techniques. The probes are fabricated using capacitive micromachined ultrasonic transducer (CMUT), and piezoelectric transducer (PZT) technology. Both have similar dimensions and center frequency. Raw RF data are obtained using an experimental research ultrasound scanner, SARUS. Two SAI sequences, a defocused and a single element transmissions, are designed for imaging down to 14 cm at a volume rate of 88 Hz. Spatial matched filtering as well as traditional delay-and-sum are used for beamforming the IQ-modulated RF data. The imaging quality of the probes is investigated through simulations and phantom measurements. Both probes on average have similar lateral full-width at half-maximum (FWHM) values, but the PZT probe has 20% smaller cystic resolution values and 70% larger contrast-noise ratio compared to the CMUT probe. The CMUT probe can penetrate down to 15 cm, and the PZT probe down to 30 cm. The CMUT probe has 17% smaller axial FWHM values. Both probes and imaging sequences are within the FDA safety limits for abdominal imaging. The results of this study demonstrate the potentials of RCA 2-D arrays against fully addressed 2-D arrays, which are low channel count (e.g. 124 instead of 3,844), low acoustic intensity (MI ≤ 0.88 and I_{ppto} ≤ 5.5 mW/cm^2), and high penetration depth (down to 30 cm), which makes 3-D imaging at high volume rates possible with equipment in the price range of conventional 2-D imaging.

I. INTRODUCTION

The companion paper [1] details the fabrication process, assembly, and the electro-acoustical evaluation of two prototype 62–62 row–column-addressed (RCA) 2-D arrays using capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technologies. In this paper, the rectilinear volumetric imaging performance is investigated based on simulations and phantom measurements. The two probes are equipped with hardware static roll-off apodization connected to both ends of each row and column element [2], [3]. Both of the transducers are designed with similar acoustical features, i.e., dimensions, center frequency, and packaging, and plugged into the ultrasound research scanner, SARUS [4].

Imaging with RCA 2-D arrays has been investigated based on simulations [2], [5]–[7] as well as measurements with arrays fabricated in CMUT [3], [8]–[11] and PZT [12], [13] technologies separately. The purpose of this paper is to evaluate the focusing ability of RCA 2-D arrays in comparison with fully addressed 2-D arrays and more specifically the imaging performance of the two fully integrated CMUT and PZT RCA 2-D arrays quantitatively and comparatively. Both delay-and-sum (DAS) and spatial matched filter (SMF) beamformation [14]–[16] methods are used for processing the IQ-modulated RF data generated with a synthetic aperture imaging (SAI) technique [17]. The quality assessments of the B-mode images acquired with both probes, i.e., spatial resolution, contrast resolution, and signal-to-noise ratio (SNR) are carried out based on simulations and measurements on several phantoms. Two different SAI sequences were designed for imaging down to 14 cm of depth at a volume rate of 88 Hz. The first sequence uses 62 virtual sources behind the array, and the second sequence utilizes 62 single element transmissions. The echoes are collected with all column elements.

Initial results of this study have been published as conference papers [18]–[20] for the PZT and CMUT technologies using a SAI sequence with single element transmission. This study extends the DAS and SMF beamformation methods for the focused and defocused SAI sequences as well as reporting on the resolution, probe temperature, and acoustic intensities. Furthermore, the focusing ability of RCA 2-D arrays are compared with fully addressed 2-D arrays.

The remainder of the paper is organized as follows: Section II presents an overview of beamforming with RCA 2-D arrays. Theoretical imaging comparison between RCA and fully addressed arrays is studied in Section III. In Section IV-A the imaging quality assessment measures are explained. The utilized SAI sequences are explained in Section IV-C. Section IV-B gives a detailed overview of the equipment and the measurement setup. In Section V the experimental results with both of the probes are presented. The performance of the probes in comparison with each other based on phantom studies are discussed in Section VII. The final Section VIII concludes the paper with suggestions for future work.
II. SYNTHETIC APERTURE IMAGING AND BEAMFORMING

Two different SAI [17] sequences were designed for imaging down to 14 cm. The first sequence utilizes 62 virtual line sources behind the array by adjusting the transmit delays of all the row elements. A virtual line source emits cylindrical pressure waves, which propagate as plane wave in one lateral dimension and as arc in the perpendicular lateral dimension. The second sequence utilizes 62 single element transmissions on the row elements. In both sequences the echoes are collected with all the column elements.

IQ-modulated RF data are used for beamforming a low-resolution volume for every emission and finally, by summing all the low-resolution volumes, a high-resolution volume is generated. This section introduces the two beamforming methods employed for generating the low-resolution volumes: a modified DAS scheme in Section II-A and a spatial matched filtering approach in Section II-B.

A. Delay-and-sum

The details on the DAS beamforming method with RCA 2-D arrays are presented in [2]. This method approximates each row or column element with a line-segment instead of a point, and therefore calculates the delays based on the geometrical distance between an imaging point and a line-segment. Fundamentally due to the flat design of RCA 2-D arrays only rectilinear volumetric imaging is achievable. Despite the fact that it is possible to focus the ultrasound wavefronts curvilinearly in transmit and receive, their perpendicular orientation limits the pulse-echo field to a rectilinear region in front of the transducer. During the DAS beamforming, the RF signals from each receive channel were temporally matched filtered by the transmit excitation pulse and IQ-modulated using Hilbert transform, before being delayed and summed at all imaging points. Furthermore, a spline interpolation has been used for calculating the amplitude at any subsampled time instance.

B. Spatially matched filters

Normal DAS focusing assumes that the spatial impulse response of the transducer is a delta function, and that the alignment can be done by merely delaying the responses. This is appropriate in the far-field for small element arrays and at the focus for single element transducers. However, in the near-field, the pulse-echo spatial impulse responses are different from a delta function [14].

As an alternative to dynamic receive focusing using DAS beamforming, the signal from each channel of an array can be spatially matched filtered to align its output with that from the other channels [14]–[16], [19]. This was suggested by Yen [16] for use on RCA arrays. When the RF signal is matched filtered, the output will have zero phase and all frequencies thereby add constructively maximizing the SNR. Analytically, the filters remove the quadratic phase factor associated with out-of-focus beams bringing the beam into focus [14], [16]. The impulse response of the matched filter is the time inverted version of the pulse-echo spatial impulse response at a specific point. Since each element’s received signal is dependent on the element location and the scatterer’s position, a new matched filter must be used depending on the element and the scatterer’s position. The SMF impulse response \( m_p(\vec{r}_{rcv},\vec{r}_{trn},t) \) is then given by [14]:

\[
\begin{align*}
  m_p(\vec{r}_{rcv},\vec{r}_{trn},t) &= p_v(\vec{r}_{rcv},\vec{r}_{trn},-t) \\
  p_v(\vec{r}_{rcv},\vec{r}_{trn},t) &= v_{pe}(t) * h_t(\vec{r}_{rcv},\vec{r}_{trn},t) * h_s(\vec{r}_{rcv},\vec{r}_{trn},t),
\end{align*}
\]

which is dependent on the transmitter location \( \vec{r}_{trn} \), the receiver element at \( \vec{r}_{rcv} \), and the electro-mechanical impulse response of the transducer \( v_{pe}(t) \). The spatial impulse responses during transmission and reception are \( h_t(\vec{r}_{rcv},\vec{r}_{trn},t) \) and \( h_s(\vec{r}_{rcv},\vec{r}_{trn},t) \) for the combined response for all of the array elements including their focusing and apodization [21]. The focusing is then performed by adding the matched filtered signals from all the elements for the different locations

\[
r_i(\vec{r}) = \sum_{j=1}^{M} \int_{t_i}^{t_i+\Delta T_{ij}} v_x(\vec{r}_{rcv},\vec{r}_{trn},t) p_v(\vec{r}_{rcv},\vec{r}_{trn},t) dt,
\]

where \( r_i \) designates the relative distance of the imaging point to the transmitting element \( i \), \( r_j \) is the relative distance of the imaging point to the receiving element \( j \) of the transducer, \( t_i \) is the start of the response, and \( \Delta T_{ij} \) is the duration of the matched filter. The convolution integral is replaced by a correlation, since the time reversal of the response is replaced by the time reversal in the convolution.

It should be noticed that (2) can be used for any image point, and that it is only necessary to process the point in the image that must be displayed on the screen, when using IQ-modulated RF data. This approach does not put any restrictions on the transducer geometry, excitation, focusing, apodization, or impulse response. SMF beamforming can be used for both multi-element arrays and single element transducers, as long as the single element is moved compared to the scattering points during the imaging process in e.g. a polar scan. Studies have shown that the lateral resolution and contrast achievable by SMF beamforming is comparable to DAS beamforming using SAI technique [16]. SMF method requires experimental measurement of the system’s pulse-echo field to perform filtering. In this work, Field II Pro [22]–[24] is used for calculations of the SMF coefficients, based on the arrays dimensions, excitation pulse, and the measured impulse response. During the SMF beamforming, the RF signals from each receive channel are Hilbert transformed, before getting match filtered.

III. IMAGING WITH RCA 2-D ARRAYS

To evaluate the volumetric imaging performance of the two prototyped CMUT and PZT RCA 2-D array transducer probes, the focusing ability of RCA 2-D arrays has to be studied compared to fully addressed 2-D arrays. Thereby, it is possible to investigate whether the two CMUT and PZT RCA probes attained similar performance to fully addressed 2-D arrays. Principally, the achievable lateral resolution of a given ultrasound system is defined by its two-way beam width at the focal depth using conventional focusing on both reception and transmission [25]. However, in imaging with an RCA 2-D
array, the focusing in the transmit direction is independent from the receive direction, thus, the spatial resolution in each direction can differ from the other direction depending on how well the focus lines are generated in each direction. In RCA 2-D arrays due to the perpendicular orientation of the transmit and receive apertures, only one-way focusing is possible in each lateral direction [2], [7], [16].

The Fresnel approximation states that in the far-field of the transducer array, and at the focal distance of a focused transducer, the pressure field may be described by the Fourier transform of the transducer aperture. A finite array of transducer elements has an aperture $A$, described by a simple rectangular window function along one lateral dimension. The Fourier transform of a rectangular function is the sinc function, which therefore describes the pressure field in that dimension. Denoting the size of this array along the $x$-dimension $L_x$, the position along the array $x (x = 0$ being the center of the array), the wavelength of the ultrasound wave $λ$, and the mass density of the medium $ρ$, the continuous wave (CW) pressure field at depth $z$ becomes [25]:

$$p_{x,\text{one-way}} = \tilde{g}[A] = \frac{L_x}{L_x} \sqrt{\frac{\rho}{\lambda}} \sin{\left( \frac{L_x x}{\lambda z} \right)}.$$  (3)

where $\tilde{g}$ denotes the Fourier transform. It is assumed here that $z$ is either at the focus of the transducer or in the far-field. The full-width at half-maximum (FWHM) of the sinc function is

$$\text{FWHM}_{\text{one-way}} = \frac{1.208\lambda_z}{L_x} = 1.208\lambda_z f_s.$$  (4)

This shows that the lateral detail resolution for a given wavelength and depth scales with the array size. The subscript “one-way” is to emphasize that the FWHM is for focusing of only the transmit aperture (or only the receive aperture due to acoustic reciprocity). An RCA array can only perform one-way focusing in each lateral dimension, if conventional DAS beamforming is used, and its detail resolution is therefore defined by (4). As opposed to this, a 2-D matrix array can focus in each lateral dimension both in transmit and receive. The resulting pulse-echo field is proportional to the Fourier transform of the convolved transmit and receive apertures [26]. If the same aperture is used for transmitting and receiving, the pulse-echo field along one dimension becomes:

$$p_{x,\text{two-way}} = \tilde{g}[A] \ast \tilde{g}[A] = |\tilde{g}[A]||\tilde{g}[A]|^2.$$  (5)

The FWHM of two-way focusing is:

$$\text{FWHM}_{\text{two-way}} = \frac{0.886\lambda_z}{L_x} = 0.886\lambda_z f_s.$$  (6)

The ratio between the resolution of one-way focusing and the resolution of two-way focusing is therefore

$$\frac{\text{FWHM}_{\text{one-way}}}{\text{FWHM}_{\text{two-way}}} = \frac{1.208\lambda_z f_s}{0.886\lambda_z f_s} \simeq 1.36.$$  (7)

Thus, for the same aperture size, the theoretically expected FWHM of an RCA array is 36% larger than the FWHM of a two-way focused 2-D array. Based on the FWHM, the detail resolution for the two types of arrays will consequently be equal, if the side-length of the RCA array is increased by 36% relative to the fully addressed 2-D array. In Fig. 1, the resulting fields from the one-way focused array (black), the two-way focused array with 36% larger aperture side-length (orange), and a two-way focused array (red). The two former are plotted using a normalized (3), while the latter uses (5).

To have the same lateral resolution for both fully addressed and RCA 2-D arrays, the number of row or column elements on an RCA array has to get increased only by a factor of $1.208/0.886 = 1.36$, i.e., by a factor of $2 \times 1.36 = 2.72$ for the total number of elements. For instance for a 2-D array with $256 \times 256$ elements, row–column addressing corresponds to a reduction in the total number of channels by 99.6%, i.e., from 65,536 channels to 512 channels. Any $N + N$ channel RCA array with $N \geq 3$ will, thus, achieve a better detail resolution than a fully addressed 2-D array with the same total number of channels. However, changing the aperture size will only affect the argument in the sinc function in (3), not the shape of the function. Hence, the normalized amplitudes remain unchanged, and so do the side-lobe levels relative to the main lobe level. This is seen in Fig. 1, where the two one-way focused arrays have side-lobe levels of equal magnitude. As a consequence of the squaring of the Fourier transform of the apertures given in (5), the amplitudes of the side-lobes are halved by a factor of two in dB, when two-way focusing is performed. One measure of contrast is the side-lobe level [25]. Therefore, a fully addressed 2-D array will have superior contrast performance relative to an RCA 2-D array.

Note that the above calculations are only strictly valid for a CW emission at a single frequency [25], and as such they should only be seen as estimates of the imaging performance. Also, the estimates are made using an un-apodized aperture, and does therefore not take into account the effects of applying different apodization functions.
IV. METHODS

A. Imaging Quality Assessment Measures

The imaging performance is computed using four measures:

1) SNR: The SNR is calculated as the ratio of the signal energy to the electronic noise. The SNR is calculated by:

\[
\text{SNR}(\vec{r}) = \sqrt{\frac{1}{N} \sum_{n=1}^{N} s_n(\vec{r})^2 - \frac{1}{N} \sum_{n=1}^{N} s_{\text{bck}}(\vec{r})^2},
\]

where \(\vec{r} = (x,y,z)\) is the voxel coordinate, and \(s_n\) a single IQ-beamformed image frame with index \(n\). The point where SNR falls below 0 dB is the penetration depth.

2) Spatial resolution: The spatial resolution is calculated as the FWHM of the imaging system’s point spread function (PSF).

3) Cystic resolution: The cystic resolution (CR) is the ability to detect an anechoic cyst in a uniform scattering medium [27]–[29]. The relative intensity (RI) of the anechoic cyst was shown by Ranganathan and Walker [28], to be quantified as the clutter energy to total energy ratio,

\[
\text{RI}(R) = \sqrt{\frac{E_{\text{out}}(R)}{E_{\text{tot}}}} = \sqrt{1 - \frac{E_{\text{bck}}(R)}{E_{\text{tot}}}},
\]

where \(E_{\text{in}}\) is the signal energy inside a circular region with radius, \(R\), centred at the peak point spread function, \(E_{\text{tot}}\) is the total PSF energy, and \(E_{\text{bck}}\) is the PSF energy outside the circular region. The \(R(R)\) curve can be compressed to a single number by sampling the curve at e.g. 20 dB. The result is the required cyst radius at which the intensity at the cyst’s center is 20 dB lower than its surroundings, written as \(R_{\text{20dB}}\).

4) Contrast resolution: The contrast resolution in B-mode images, i.e., contrast-to-noise ratio (CNR), is defined as

\[
\text{CNR} = \frac{\mu_{\text{cyst}} - \mu_{\text{bck}}}{\sqrt{\sigma_{\text{cyst}}^2 + \sigma_{\text{bck}}^2}},
\]

where \(\sigma_{\text{cyst}}^2\) and \(\sigma_{\text{bck}}^2\) are variances, and \(\mu_{\text{cyst}}\) and \(\mu_{\text{bck}}\) are mean values of gray levels within the background and lesion, respectively.

B. Equipment and Measurement Setup

The dimensional parameters of both arrays are found in the companion paper [1]. The probes are plugged into the experimental ultrasound scanner, SARUS [4]. The measured IQ-modulated RF signals are beamformed using MATLAB (MathWorks Inc., Massachusetts, USA) implementations of the DAS and SMP beamformers described in Section II.

To evaluate the imaging performance of both probes, several ultrasound phantoms are used. An iron needle with diameter of 300 µm facing towards the transducer along its central axis, was used as a point target in a water bath for characterizing the 3-D PSF. To evaluate the FWHM and the CR as a function of depth, a geometrical copper wire phantom was used as line targets, where wires were located at different depths with 1 cm spacing. The wire grid phantom has three columns separated by 1 cm and each has 13 rows of wires.

A tissue mimicking phantom with cylindrical anechoic targets, model 571 from Danish Phantom Design (Frederikssund, Denmark) with attenuation of 0.5 dB/(cm MHz) was used for SNR and contrast measurements.

The transmit pressure intensity and temperature measurements of the probes were carried out using the AIMS III intensity measurement system (Onda Corporation, Sunnyvale, California, USA) connected to the experimental research scanner SARUS [30], [31]. The mechanical index (MI) and the spatial-peak-temporal-average intensity (I_{STAI}) are both measured for each of the transducers and for both imaging sequences. The current prototyped probes do not have the required permissions to be used on humans, therefore no in vivo data have been acquired.
C. Defocused SAI Sequence Choice of Parameters

To accomplish the best performance, the location and number of virtual sources have to be optimized in a trade-off between spatial resolution, field-of-view, and SNR. Fig. 2 illustrates the position of three virtual line sources behind the array. In the figure, D is the active aperture, α is the maximum steering angle, and F denotes the distance to the active aperture centre, i.e., \( f_b = F/D \). The rest of the virtual line sources are interspaced equally between the two virtual line sources placed at the edges. A simulation parameter study is carried out over the maximum steering angle for placing the 62 virtual line sources and the transmit \( f_b \) of the defocused SAI sequence to image a point scatterer at a depth of 20 mm in front of the array. The lateral FWHM and CR values of the beamformed PSFs for maximum steering angles in range of ±10° to ±60° and transmit \( f_b \) from −3 to −0.5 are shown in Fig. 3. The criteria to choose the best parameters is to have the best contrast and spatial resolution for the lowest steering angle and transmit \( f_b \). As a trade-off between contrast and spatial resolutions, the maximum steering angle of ±30° and transmit \( f_b = -1 \) are chosen for the defocused SAI sequence. The parameters of both SAI sequences are listed in Table I.

V. IMAGING PERFORMANCE ASSESSMENT

Fig. 4 illustrates three cross-planes (azimuth, elevation, and C-plane) of the volumetric pulse-echo beam patterns measured with both probes in comparison with Field II simulations using the DAS beamformation method. The iron needle faces towards the transducer, and it is imaged with the single element transmissions SAI sequence. A Hanning apodization is applied over the receive and synthesized transmit apertures. The measured pulse-echo impulse responses of both probes [1] as well as the diameter of the needle are taken into account for the simulations by imaging a disk consisting of 500 point targets to represent the tip of the needle.

Note the secondary lobes after the main lobe along the axial direction for both simulated PSFs of the probes at the range of 20 mm to 22 mm of depth, which are due to the internal reflections from the RF shielding foils covering the arrays. These secondary reflections are visible in the impulse responses of the probes [1]. Among these secondary echoes are the edge echoes, which originate from the either ends of the line elements. They were not suppressed below 40 dB [2]. To choose the best parameters is to have the best contrast and spatial resolution, consequently leading to a larger speckle size in the axial direction.

<table>
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<th>Simulation</th>
<th>Measurement</th>
<th>CMIUT</th>
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<td>mm</td>
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Table II: FWHM and CR of simulated and measured 3-D PSFs

Fully addressed (FA)
The more pronounced secondary lobes in the measurements compared with the simulations in azimuth and elevation planes of Fig. 4 are due to the reflections coming towards the transducer from the shaft of the needle. Thereby, the transmitted waves have no deconstructive effect after the tip of the needle, which is not the case in simulations for an ideal point target. This had a more dramatic effect on the SMF beamforming method, which resulted in a larger axial FWHM reported in Table II.

To study the focusing abilities of both probes as a function of depth, a wire grid phantom is used to quantify the line spread function (LSF) characteristics of both probes, since the echoed signals from the needle are too low at higher depths. The diameter of the wires is 200 µm, which is smaller than a wavelength in water, and therefore it is used as line targets. The phantom has three columns of wire, which are separated by 10 mm in the axial and lateral directions. Both of the SAI sequences are used to image the wire phantom with the PZT and CMUT probes by placing them centered around the middle column. A volume region of 26 × 10 × 5 mm³ centered around each beamformed wire in the middle column is used for the LSF characteristics evaluation as a function of depth. Fig. 5a and Fig. 5b illustrate the calculated FWHM in the lateral and axial directions for both of the SAI sequences with the PZT and CMUT probes at different depths. Using the SAI sequence with transmit $f_e = -1$ has increased the axial FWHM values of the PZT probe compared with the SAI sequence with single element at a time. This is because of the phase delay differences between the PZT elements, which have been reported in the companion paper [1]. The phase delay differences of the elements affect the defocused transmit SAI sequence with $f_e = -1$, however in receive stage those delay difference can be compensated. Likewise what was observed previously with the 3-D PSFs, here for the LSFs, the CMUT probe has smaller FWHM values in the axial direction comparing with the PZT probe. Although the lateral FWHM values increase linearly with depth similar to (4), the SMF beamforming has lowered the FWHM values for the SAI sequence with transmit $f_e = -1$. As it is illustrated in Fig. 6a and Fig. 6b, CR values increase almost linearly with depth for both probes.

A volume region of a tissue mimicking phantom with no cysts was imaged 20 times for calculating the SNR. The measured SNRs for both probes are illustrated in Fig. 7. The PZT probe has a penetration depth of around 14 cm for single element transmissions, whereas the CMUT probe can only penetrate down to 10 cm. However by using all the elements in transmit in the SAI sequence with transmit $f_e = -1$, the CMUT probe has a penetration depth of around 14 cm, whereas the PZT can penetrate down to almost 25 cm. Using a SAI sequence with transmit $f_e = 1$ by placing the virtual sources in front of the array has increased the penetration depth for the CMUT probe down to 15 cm, whereas the PZT can penetrate down to almost 30 cm.
(a) PZT probe
(b) CMUT probe

Fig. 5. Axial and lateral FWHM of the PZT (a) and CMUT (b) probes as a function of depth for DAS beamformed images of the wire phantom. The solid lines correspond to lateral FWHM (left axis) and the dashed lines correspond to axial FWHM (right axis).

\( \lambda \) was calculated for soft tissue and is 0.5 mm.

The red dotted line shows the estimated lateral FWHM based on the Fresnel approximation in (4).

Due to the perpendicular orientation of the transmit and the receive directions, the field-of-view of RCA arrays is limited to the forward looking region in front of the array, e.g., 26×26 mm² for these probes. Two cross-planes (azimuth and C-plane) are shown in Fig. 8 and Fig. 9 at a dynamic range of 40 dB from a volume of 26×26×80 mm³ acquired with both probes using a cyst phantom and beamformed with the DAS and SMF beamforming methods. Only down to 80 mm are shown, due to the limited space in the paper. The origin corresponds to the center of the transducer surface. Data were acquired with both probes using the single element emission SAI sequence as well as the SAI sequence with a transmit \( f_b = -1 \). The hollow cysts are located along a 10° tilted plane and therefore, the lower hollow cysts are not completely visible at regions farther from the array. See the videos in the supplementary materials for the full sized DAS beamformed volumes of the cyst phantom using the SAI sequence with transmit \( f_b = -1 \).

The CNR measure is calculated in a cylindrical region centered at each of the large hollow cysts with a diameter of 8 mm. The calculated CNR values for each imaging sequence with each probe are shown in the Fig. 10. Due to the higher generated pressure with the PZT probe compared with the CMUT probe, the CNR of the PZT probe is almost 2 times larger than the CMUT probe.

VI. INTENSITY AND TEMPERATURE MEASUREMENT

\( \text{MI and } I_{\text{pta}} \) are measured for the SAI sequence with a transmit \( f_b = -1 \) at a pulse repetition frequency of 5 kHz, since it uses all row elements in transmit and thereby has the largest emitted energy. For the PZT probe they are \( \text{MI} = 0.67 \) and \( I_{\text{pta}} = 0.53 \text{mW/cm}^2 \). They are \( \text{MI} = 0.06 \) and \( I_{\text{pta}} = 0.18 \text{mW/cm}^2 \) of the CMUT probe. All are within the FDA safety limits of \( \text{MI} \leq 1.9 \) and \( I_{\text{pta}} \leq 720 \text{mW/cm}^2 \) for
abdominal imaging [30]–[32]. For the SAI sequence with a transmit \( f_t = 1 \) at a pulse repetition frequency of 5 kHz, the measured MI = 0.88 and \( I_{\text{pnt}} \) = 5.5 mW/cm\(^2\) for the PZT probe, and MI = 0.13 and \( I_{\text{pnt}} \) = 0.55 mW/cm\(^2\) for the CMUT probe are also both within the FDA safety limits. All sequences are MI limited and the transmit voltage could be scaled by a factor between 2.16 and 31.7 giving an increased SNR of 6.7 to 30 dB. This could result in a penetration increase of 1.1 cm to 5 cm.

Another criteria that has to be addressed is the heating of the probe, which has to be within the FDA safety limits [32]. The linear voltage regulators used for the amplifiers in the prototype probes are dissipating power and generating waste heat. Therefore, the temperature rise in the probe is a combination of heating produced by the linear regulators as well as the transducer arrays. To separate the heating caused by the amplifiers in the handle from the transducers themselves, two spots on the probes were measured for temperature changes in a still air environment. One sensor located at the sole of the probe and the other sensor was located on the body of the probe, where the amplifiers are located. These two sensor measurements are shown in Fig. 11 for situations when only the amplifiers are turned on and no voltage is applied to the transducers. However, when the transducers are transmitting the excitation pulses with amplifiers turned on, Fig. 11 illustrates the temperature rise for both probes using all of the imaging sequences firing sequentially after each other without any delay between each sequence using ±75V transmit voltage in still air situation. The temperature rise of both probes is dominated by the dissipated heat of the linear voltage regulators. Both probes satisfy the FDA safety limits on the absolute temperature in still air, which requires the absolute temperature to be less than human body temperature. However, on the temperature rise, due to the heat generated by the linear voltage regulators, both probes did not satisfy the requirement of less than 10°C temperature rise in 60 minutes and therefore for future clinical use, modifications to the power supplies of these prototype probes need to be made.

A first-order model can approximately describe the thermal dynamics of the transducers inside the probes, i.e.,

\[
T(t) = (T_0 - T_s)(1 - e^{-t/\tau}) + T_s,
\]

where \( T(t) \) is the temperature at time \( t \), \( T_0 \) is the initial temperature, \( T_s \) is the steady state temperature, and \( \tau \) is the time constant, which the temperature is at 63% of total temperature change. By comparing the time constants in Fig. 11, it can be seen that both probes heat up with the same rate, when only the electronics on the handle are turned on and no voltage is applied to the transducers. However, by applying the excitation voltage across the transducers inside the probes, the CMUT probe reaches the equilibrium state relatively quicker than the PZT probe. For both probes the temperature measured with the sensor located on the handle and above the electronics, increases quicker than the sensor located on the sole of the probe, however the maximum measured temperature on the handle is below the temperature measured at the sole of the probe. That indicates the dissipated heat on the sole of the probe is a combination of the heat generated by the transducer array and electronics inside the probe.

### VII. Discussion

Based on the comparison presented in Section III, to have the same lateral resolution for both fully addressed and RCA 2-D arrays with the same aperture size, the number of row or column elements on an RCA array has to get increased only by a factor of 1.36. However, suppressing the secondary lateral lobes of the PSF of an RCA array as shown in Fig 1, which are artefacts for the contrast resolution, is a challenging problem. Using SAI technique by placing the virtual sources behind the transducer, which improves the SNR in combination with SMF beamforming method, was a novel idea that improved the imaging sensitivity. Both of the SAI technique and the SMF beamforming method are computationally very demanding.

It is important to note that the roll-off apodization will not affect the lateral resolution within the rectilinear imaging field-of-view of the array but it only removes the secondary range lobes (ghost echoes), and therefore the FWHM measurements should be comparable to (3), since no electronic apodization has been applied. For the 3-D PSF measurement at the depth of 19.8 mm as shown in Fig. 4, but without using electronic apodization, the simulated lateral FWHM values listed in Table II were slightly larger than 0.71 mm that is estimated from (3). As it was expected from the SMF beamforming method, using SMF increased the axial FWHM values for both CMUT and PZT probes, instead the CR and CNR have improved compared to DAS. The effect of using SMF was more pronounced with the CMUT probe compared to the PZT probe, which can be attributed to the higher bandwidth of the CMUT probe for the spatio-temporal convolution with the spatial impulse response in the SMF beamformation method.

In Fig. 5a the larger axial FWHM values for the defocused SAI sequence compared to the single element transmissions can be attributed to the phase delay difference of the elements in the PZT probe discussed in Part I [1], which leads to imperfect transmit and receive focusing. Although the receive delay can be compensated after the measurements, the transmit...
delays can only be compensated during the measurements. The measured lateral FWHM values are close to the estimated lateral FWHM values from (3), as indicated by a red dotted line in Fig. 5a and Fig. 5b. The best lateral resolution is achieved with the single element transmissions SAI sequence using DAS. In Fig. 6a and Fig. 6b by using SMF, the CR values have been improved for both CMUT and PZT probes for the single element transmission SAI sequence. However, for the CMUT probe, using the defocused SAI sequence, no significant difference between DAS and SMF has been observed. For the PZT probe, on the other hand, using SMF for the defocused SAI sequence has worsened the CR compared to DAS. For both probes, by using the SMF method, the seventh wire at the depth of 1543 had a lower main-lobe to side-lobe ratio compared to DAS, thus increased the CR value at that depth.

Theoretically, transmitting with row element and receiving with column elements should image exactly the same rectilinear volume as transmitting with column element and receiving with row elements. However, because of the manufacturing process, the sensitivity of row and column elements might be slightly different. For the PZT probe the difference is small, since transmitting with row elements and receiving the echoes with column elements or vice versa, are similar as shown in Fig. 12. For the CMUT probe, because of a capacitive substrate coupling of the bottom electrodes, as discussed in the companion paper, part I [1], the receive sensitivity of the bottom elements is...
lower and therefore in our imaging set-up, the elements with higher receive sensitivity, i.e., the top elements, are chosen for receiving, while the bottom elements are used for transmitting.

As a result, for the CMUT probe by coherently compounding the two volumes, assuming no movement, the spatial resolution and contrast will degrade.

Using single element for transmission, the CMUT probe has a lower penetration depth on a tissue mimicking phantom compared to the PZT probe. On the contrary by using all the elements in transmit, both probes penetrate down to 15 cm, with PZT probe even down to 30 cm. In the same way, the CNR values increased, when using all the elements in transmit. Placing the virtual focus lines in front of the transducer would increase the penetration depth further, but larger transmit \( f_t \) values will degrade the spatial resolution. For SAI sequences with the transmit focus in front of the array, the transmitted acoustic energy is focused along a line in contrast to a point by using fully addressed 2-D arrays and therefore for the same size of the 2-D arrays, the MI and the I_{psta} are lower for RCA 2-D arrays.

VIII. CONCLUSION

In this paper, the imaging performance of two prototyped 62×62 RCA 2-D array probes fabricated in CMUT and PZT transducer technologies were demonstrated quantitatively and comparatively. Using SAI technique both probes were able to image down to 14 cm at a volume rate of 88 Hz. DAS and SMF beamforming methods were both able to
Fig. 10. CNR values measured with both PZT and CMUT probes as a function of depth. Calculated over the beamformed volumes of the cyst phantom beamformed with DAS and SMF methods. λ for soft tissue is 0.5 mm.

Fig. 11. Temperature values of the PZT and the CMUT probes as a function of time for both cases, when only the amplifiers in the handle are turned on, and when both the amplifiers in the handle and the ±75 volts amplifiers are turned on. The whole imaging sequences has been running in still air while the temperature sensors were mounted on the centre of the probes soles and also on the probes handles.

perform dynamic transmit-receive focusing throughout the rectilinear field-of-view. The performance of both probes was evaluated through simulation and experiments. Results show that both probes can image a rectilinear volume in front of the transducer successfully. Integrated hardware apodization along each line-element effectively removed the ghost echoes without altering the main echo’s beam width. It was demonstrated that volumetric imaging with equipment in the price range of conventional 2-D imaging is possible. Both probes were prototypes and not optimized, which limited the imaging performance. Future work will focus on configuring the probes for better performance through adjusting the DC bias voltage for the CMUT probe for achieving higher penetration and using a better shielding method for both probes to eliminate the reflections within the probes.

The results of this study have demonstrated the promising potentials of RCA 2-D arrays compared with fully addressed 2-D arrays, which are their low channel count, low MI and $I_{peak}$ values, and high penetration depth. It was shown that, due to one-way focusing of RCA 2-D arrays in each lateral dimension, the spatial resolution is lower than fully addressed arrays, however that can be compensated by increasing the size of the array by 36% in each lateral dimension. Moreover, the contrast resolution was improved by using SMF beamforming method. Our future work involves applying contrast enhancing algorithms in combination with SMF beamforming, such as adaptive imaging by dual apodization with cross-correlation (DAX) [33] or short-lag spatial coherence (SLSC) imaging [34] to improve the image quality.

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3-D Imaging using Row–Column Addressed 2-D Arrays with a Diverging Lens

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3-D Imaging using Row–Column-Addressed 2-D Arrays with a Diverging Lens

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Abstract—it has been shown that row–column-addressed (RCA) 2-D arrays can be an inexpensive alternative to fully addressed 2-D arrays. Generally imaging with an RCA 2-D array is limited to its forward-looking volume region. Constructing a double-curved RCA 2-D array or applying a diverging lens over the flat RCA 2-D array, can extend the imaging field-of-view (FOV) to a curvilinear volume without increasing the aperture size, which is necessary for applications such as abdominal and cardiac imaging. Extended FOV and low channel count of double-curved RCA 2-D arrays make it possible to have 3-D imaging with equipment in the price range of conventional 2-D imaging. This study proposes a delay-and-sum (DAS) beamformation scheme specific to double-curved RCA 2-D arrays and validates its focusing ability based on simulations. A synthetic aperture imaging (SAI) sequence with single element transmissions at a time, is designed for imaging down to 14 cm at a volume rate of 88 Hz. The curvilinear imaging performance of a \( \lambda / 2 \)-pitch 3 MHz 62+62 RCA 2-D array is investigated as a function of depth, using a diverging lens with f-number of -1. The results of this study demonstrate that the proposed beamforming approach is accurate for achieving correct time-of-flight calculations, and hence avoids geometrical distortions.

I. INTRODUCTION

An \( N \times N \) element 2-D array can be operated utilizing only 2N connections, when a row–column or cross-electrode addressing scheme is used [1]–[3]. This is contrary to the \( N^2 \) connections needed, when conventionally addressing the elements. In general, a row–column-addressed (RCA) array is a 2-D matrix array, which is addressed via its row- and column indices. Effectively, this makes two 1-D arrays arranged orthogonal to each other. As an example, a 256×256 RCA array will have 512 elements. A 2-D matrix array of equivalent size would have 65,536 elements, over a factor of 7 more than the current state-of-the-art X6-1 PureWave xMATRIX probe from Phillips (Eindhoven, Netherlands) that has 9212 elements [4]. This exhibits the potential of having very large RCA 2-D arrays with low channel count and real-time capabilities.

It has been demonstrated in several studies [1]–[3] that row–column technology is a realistic alternative to the state-of-the-art matrix probes, especially as a low-cost alternative. However, one major issue with the RCA arrays is that they can only emit acoustic energy directly below the array and in a cross-shape to the sides. For applications such as abdominal imaging, it is relevant to have a probe with a large aperture capable of phased array imaging. True volumetric phased array imaging is possible with RCA arrays, provided that the array is double curved to spread the energy during transmit [1]. However, manufacturing curved transducer elements is challenging for both capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technologies. Another approach to spread the acoustic energy is by using a double curved diverging acoustic lens on top of the RCA array [5]. Using a lens makes it easier to fabricate curved transducers, as it is not needed to manufacture curved elements, and also making a lens is a well-tested technology. An in-depth study of the possibilities of this approach is therefore the main goal of this study.

It is investigated whether curvilinear volumetric imaging is possible with an RCA array equipped with a diverging lens. A dedicated beamformer is developed and the performance is evaluated using Field II [6], [7] simulations. The quality assessments of the B-mode images, i.e., spatial resolution and contrast resolution, are carried out on the simulated data using SAI technique. The remainder of the paper is organized as follows: The current limitations with flat RCA arrays and different approaches to disperse the acoustic energy are described in Section II. Section III presents a DAS beamformer for a double curved RCA array. In section IV, a detailed overview of the simulation setup is presented. Section V explains the simulation results and the final section concludes the paper.

II. CURVED RCA 2-D ARRAYS

To spread the acoustic energy of a line-element, the element has to be curved to generate a diverging wave. However, manufacturing double curved transducers is very challenging. Alternatively, the defocusing of the waves can be made by using a fixed electronic delay profile along each flat line-element to generate a diverging wavefront. This can be seen as using micro-beamforming with a fixed first stage. Another simpler approach to spread the acoustic energy is by using a double curved diverging acoustic lens on top of the RCA array [5]. A concave diverging lens can be designed with a material having a lower speed of sound compared to the human tissue, so that it has higher thickness around the corners and the sides of the array, and less thickness close to center of the array.
Alternatively, a convex diverging lens can be made from a material with a higher speed of sound compared to the human tissue, which is preferred for a better contact surface. A flat diverging lens can also be made by using a combination of two different materials, one with higher and other one with lower speed of sound compared to the human tissue, which results in a flat surface for good contact.

Note that, for the same aperture size, lower lens $f_s$ values for the lens corresponds to larger thicknesses of the lens, and therefore the attenuation becomes higher through the lens material. Thus, there is a trade-off between field-of-view (FOV) and attenuation. The function $p$ from 1 to the number of receive line elements $N_r$ be written as:

$$ p = \frac{\text{shortest distance from the arc source}}{\text{contact points on the transmit and receive elements}}. $$

Fig. 1. The time-of-flight of a wavefront is given by the shortest distance to the point being focused $p$ and back to the receiving element $r_o$ divided by the speed of sound. The points $q_l$ and $q_r$ are the closest contact points on the transmit and receive elements $s_m$ and $r_o$ to the point $p$.

Fig. 2. Distance between a point $p$ and an arc $\hat{ab}$ is calculated using (4).

The arc segment from point $a$ to point $b$ with center $c$ is termed $ab$ assuming the center at origin. This is illustrated in Fig. 2. The projection of point $p$ onto the plane passing through the arc $\hat{ab}$ and its center $c$ is termed $p'$ and is determined by the usual equation for projection. To determine if the vector $cp'$ is in between vector $ca$ and vector $cb$, we define the normalized cross products $\hat{l}_a$ and $\hat{l}_b$ as

$$ \hat{l}_a = \frac{cp' \times ca}{|cp'||ca|}, \quad \hat{l}_b = \frac{cp' \times cb}{|cp'||cb|}. $$

Depending on the location of the point $p$, vectors $\hat{l}_a$ and $\hat{l}_b$ can be either $\hat{j}$ or $-\hat{j}$, where $\hat{j}$ is the unit vector of the $z$-axis. $\hat{l}_a$ and $\hat{l}_b$ have different signs, when $\alpha_1 \leq \phi \leq \alpha_2$, they have the same sign, when $\alpha_2 \leq \phi \leq \alpha_1$. Here $\alpha_1$, $\alpha_2$, and $\phi$ are the angles between the $x$-axis and the vectors $ca$, $cb$, and $cp$, respectively, as shown in Fig. 2.

When $\hat{l}_a$ and $\hat{l}_b$ have different signs, i.e. $\hat{l}_a = \hat{j}$ and $\hat{l}_b = -\hat{j}$, or $\hat{l}_a = -\hat{j}$ and $\hat{l}_b = \hat{j}$, the standard formula for the distance between an arc and a point can be used:

$$ d = \sqrt{|pp'|^2 + (|cp'| - R)^2}, $$

where $R$ is the curvature of the arc and equal to $|ca|$ or $|cb|$. When $\hat{l}_a$ and $\hat{l}_b$ have the same signs, i.e. $\hat{l}_a = \hat{j}$ and $\hat{l}_b = \hat{j}$, or $\hat{l}_a = -\hat{j}$ and $\hat{l}_b = -\hat{j}$, the shortest distance from the arc segment to the point is the distance from the closest end of the arc segment $(a$ or $b)$ to the point $p$. The following therefore determines the minimum distance between the point $p$ and the arc segment $ab$:

$$ d(\overset{\rightarrow}{ab}, p) = \begin{cases} \sqrt{|pp'|^2 + (|cp'| - R)^2} & \text{if } \begin{cases} \hat{l}_b = -\hat{j} \text{ and } \hat{l}_a = \hat{j} \\ \hat{l}_b = \hat{j} \text{ and } \hat{l}_a = -\hat{j} \end{cases} \\ |ap| & \text{if } \hat{l}_b = -\hat{j} \text{ and } \hat{l}_a = -\hat{j} \\ |bp| & \text{if } \hat{l}_b = \hat{j} \text{ and } \hat{l}_a = \hat{j} \end{cases}. $$

**III. DAS BEAMFORMING WITH CURVED RCA 2-D ARRAYS**

The time-of-flight (ToF) of a wavefront is given by the shortest distance from the arc source $s_m$ to the point being focused $p$, and back to the receiving element $r_e$, divided by the speed of sound. Using the notations from Fig. 1 this can be written as:

$$ \text{ToF}_{ab}(n,p) = \frac{d(s_m,p) + d(r_e,p)}{c}, $$

where $c$ is the speed of sound in the medium, $n$ is an index from 1 to the number of receive line elements $N_r$ and $m$ is the emission index. The function $d(\ldots)$ calculates the shortest distance between an arc and a point in space, which will be defined in the remainder of this section.
Using (4), the distances $d(\mathbf{s}_m, \mathbf{p})$ and $d(\mathbf{r}_x, \mathbf{p})$ can now be determined. The focused signal at point $\mathbf{p}$ is calculated by summing all receive signals at the time instances given by (1):

$$z_m(\mathbf{p}) = \sum_{n=1}^{N} a_{elec}(n, \mathbf{p}) y_{m,t}(\text{ToF}_m(n, \mathbf{p})).$$

where $N$ is the number of receive elements, $a_{elec}$ is the electronic receive apodization, and $y_{m,t}(t)$ is the measured signal from emission $m$ on the receive element $n$ at time $t$.

IV. SIMULATION AND MEASUREMENT SETUP

In this work, Field II [6], [7] is used for all simulations. The simulated receive signals are beamformed using a MATLAB (MathWorks Inc., Massachusetts, USA) implemented simulation parameters of a RCA 62+62 element 2-D array. The parameters are shown in Table I. The receive array is rotated 90° with respect to the transmit array. Field II is set up to use lines to describe the apertures and each line-element is divided into square mathematical sub-elements with a side length of $\lambda/4$. To remove the otherwise apparent secondary echoes originating from the either ends of arc shaped elements, two roll-off apodization regions are placed at both ends of each element [9]. The length of each apodization region was equal to 15 times the pitch of the array. Each mathematical sub-element in both transmit and receive arrays is delayed according to the lens delay profile.

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</table>

V. RESULTS AND DISCUSSION

In Fig. 3, the pulse-echo energy as a function of lateral position for different lens $f_a$ is illustrated. For the flat array, the pulse-echo energy drops by moving away from the forward looking region of the array. This drop of the energy is due to the diffraction of the sound waves. At around 20 mm lateral position the pulse-echo energy drops by 40 dB, when no lens is used. However, by using a diverging acoustic lens on top of an RCA 2-D array a larger FOV can be attained. The FOV can be adjusted by using different $f_a$ values for the lens. The main advantage of using a diverging lens or designing a curved 2-D array, is to disperse the transmit and receive fields, so that they overlap in a larger area. By using a diverging lens with $f_a = -1$, the overlapped transmit and receive region increases to $\pm 26.5^\circ$ in both directions and the energy is maintained within the FOV (yellow solid line in Fig. 3). By using all the elements in the transmit and placing the transmit focus in front of the array, this drop of the energy might be compensated partially.

Fig. 4 is illustrating three cross-planes (azimuth, elevation and C-plane) of a phantom with point targets simulated and beamformed with the proposed DAS beamforming method, with and without a diverging lens. The point targets are located along elevation dimension from –30 mm to 30 mm in an axial range of 5 mm to 95 mm. It can be seen from the figures that by using a diverging lens the FOV is extended compared to the flat RCA array.

To study the PSF characteristics as a function of lateral angle and radial distance, a point scatterer is imaged by sweeping it from 0° to 40° in lateral plane with steps of 10° at radial distances from 10 mm to 60 mm from the center of the array. At each radial distance and lateral angle the full-width at half-maximum (FWHM) and cystic resolution (CR) values are calculated over a volume of 10 mm × 10 mm × 10 mm surrounding the point target. Fig. 5 is illustrating the measured FWHM and CR values as a function of depth and lateral angle. Moving away from the center of the elements towards the edges, the transmit wavefronts contact each other at a sharper point compared with the contact point at the center. This can be observed in Fig. 5 that, by moving away from the center towards the higher angular position in lateral plane, the elevation FWHM values become smaller while the CR values become larger. On the other hand, the lateral FWHM values stay constant, this is due to the intersection of the wavefronts in receive direction which is at the center of the receive elements and therefore the elevation FWHM values stay constant for all lateral angles.

VI. CONCLUSION

In this paper, the imaging performance of a curved 62+62 RCA 2-D array with a diverging lens, is quantitatively demon-
strated. A SAI sequence with single element transmissions at a time, was designed for imaging down to 14 cm at a volume rate of 88 Hz. The capabilities of a curved RCA 2-D array to effectively focus in both transmit and receive are investigated. A suitable DAS beamformer was introduced and implemented. Simulated results confirm that using a diverging lens can increase the imaging FOV and also that it is possible to perform dynamic transmit-receive focusing throughout the curvilinear FOV. Thereby, the inherent imaging limitation with flat RCA 2-D arrays, i.e., its forward looking rectilinear FOV, is overcome by using a diverging lens. Overall, having a low channel count and a large FOV, offers the potential to fabricate arrays with large aperture sizes, which is important for abdominal scans. Thus by using a curved RCA 2-D array, 3-D imaging with equipment in the price range of conventional 2-D imaging is possible.

ACKNOWLEDGMENT

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Curvilinear 3-D Imaging Using Row–Column Addressed 2-D Arrays with a Diverging Lens: Feasibility Study

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Curvilinear 3-D Imaging Using Row–Column Addressed 2-D Arrays with a Diverging Lens: Feasibility Study

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Abstract—Constructing a double-curved row–column-addressed (RCA) 2-D array or applying a diverging lens over the flat RCA 2-D array can extend the imaging field-of-view (FOV) to a curvilinear volume without increasing the aperture size, which is necessary for applications such as abdominal and cardiac imaging. Extended FOV and low channel count of double-curved RCA 2-D arrays make 3-D imaging possible with equipment in the price range of conventional 2-D imaging. This study proposes a delay-and-sum (DAS) beamforming scheme specific to double-curved RCA 2-D arrays and validates its focusing ability based on simulations. A synthetic aperture imaging sequence with single element transmissions is designed for imaging down to 14 cm at a volume rate of 88 Hz. Using a diverging lens with f-number of -1 circumscribing the underlying RCA array, the imaging quality of a double-curved λ/2-pitch 3 MHz 62+62 RCA 2-D array is investigated as a function of depth within a curvilinear FOV of 60°×60°. The simulated double-curved 2-D array exhibits the same full-width-at-half-maximum values for a point scatterer within its curvilinear FOV at a fixed radial distance compared with a flat 2-D array within its rectilinear FOV. The results of this study demonstrate that the proposed beamforming approach is accurate for achieving correct time-of-flight calculations, and hence avoids geometrical distortions.

I. INTRODUCTION

An \(N \times N\) element 2-D array can be operated utilizing only \(2N\) connections, when a row–column or cross-electrode addressing scheme is used [1]–[7]. This is contrary to the \(N^2\) connections needed, when conventionally addressing the elements. In general, a row–column-addressed (RCA) array is a 2-D matrix array, which is addressed via its row- and column indices. Effectively, it consists of two 1-D arrays arranged orthogonal to each other as shown in Fig. 1. As an example, for a 256×256 RCA array, a 2-D matrix array of equivalent size would have 65,536 elements, over a factor of 7 more than the current state-of-the-art X6-1 PureWave xMATRIX probe from Philips (Eindhoven, Netherlands) having 9212 elements [8]. This exhibits the potential of having very large RCA 2-D arrays with low channel count and real-time capabilities.

It has been demonstrated in several studies that row–column technology is a realistic alternative to the state-of-the-art matrix probes, especially as a low-cost alternative [1]–[7], [9]. However, one major issue with the RCA arrays is that they can only emit acoustic energy directly below the array and in a cross-shape to the sides. Therefore, imaging can only be done in a rectilinear region in front of the array. For applications such as cardiac imaging, it is relevant to have a probe with a small foot-print capable of phased array
imaging, such that the heart can be visualized through the ribs. True volumetric phased array imaging is possible with RCA arrays, provided that the array is double curved to spread the energy during transmit [2]. However, manufacturing curved transducer elements is challenging for both capacitive micromachined ultrasonic transducer (CMUT) and piezoelectric transducer (PZT) technologies. Another approach to spread the acoustic energy is by using a double curved diverging acoustic lens on top of the RCA array [10]. Using a lens makes it easier to fabricate curved arrays, as it is not needed to manufacture curved elements, and also making a lens is a well-tested technology. An in-depth study of the possibilities in this approach is therefore the main goal of this study.

In this paper, the curvilinear volumetric imaging performance of an RCA array equipped with a diverging lens is investigated based on Field II [11], [12] simulations. The quality assessments of the B-mode images, i.e., spatial resolution and contrast resolution, are carried out based on the simulations using synthetic aperture imaging (SAI) technique. The SAI sequence is designed for imaging down to 14 cm of depth.

The paper is organized as follows: The current limitations with flat RCA arrays are discussed in Section II-A, and different approaches to disperse the acoustic energy are introduced in Section II-B. Section II-C presents an overview of the DAS beamformation with a double curved RCA, and the utilized SAI sequence is explained in Section III-A. In Section III-B, the imaging quality assessment measures are explained. In Section III-C, a detailed overview of the simulation setup is presented. Section IV explains and discusses the simulation results with an RCA 2-D array equipped with a diverging lens. The final section concludes the paper with suggestions for future work.

![Figure 2](image1.png)

**Figure 2.** The static roll-off apodization layout is applied to either ends of the line-elements of the array, e.g., here a 16+16 RCA array with roll-off apodization regions is shown. The central region, shown in black, has an apodization value of one. Each roll-off region is connected to each line-element.

![Figure 3](image2.png)

**Figure 3.** Relative transmit and receive pressure fields at radial distance of 80 mm for azimuth and elevation steering angles from $-45^\circ$ to $45^\circ$. The imaging area is the intersection of these two fields, which is (a) the rectilinear forward-looking box, and (b) the curvilinear forward-looking region in front of the transducer using a lens with $f_x = -1$.

## II. RCA 2-D Arrays

### A. Flat RCA 2-D Arrays

In 3-D ultrasound imaging with flat RCA 2-D arrays, the two orthogonal 1-D transmit and receive arrays are both used for focusing in the lateral and elevation directions separately. Each of the two 1-D arrays can electronically focus in one lateral dimension, when delays are applied to the elements in the array. One of the 1-D arrays is used to transmit ultrasound into the object of interest. For example, the transmit array is able to focus the beam in the $x$- and $z$-directions, whilst no electronic transmit focusing can be performed in the $y$-direction. As a result, the emitted ultrasound is focused along a line parallel to the $y$-direction. By adjusting the delays on the transmit elements, this focal line may be translated to any position...
in the $xz$-plane. The orthogonal 1-D array then receives the echoes scattered from the illuminated region of the volume. By applying delays, the received signals can also be focused in a line normal to any position in the $yz$-plane. The combination of the two orthogonal line-foci of the transmit and receive array produces a point focus in the volume. By translating this focus throughout the volume, a 3-D rectilinear image may be formed.

DAS beamformers usually assume the geometry of the sound sources and receivers to be points. However, by row–column addressing the elements on a 2-D matrix array, each row and column is acoustically equivalent to a line-element. Furthermore, the emitted wavefront of a line-element has the shape of a cylinder, i.e., it is a plane wave in the plane aligned along the line-element and a circle are in the plane orthogonal to the line-element. Assuming the geometry of the line-elements to be points is therefore a poor approximation. A more accurate approximation assumes the line-elements to be line segments instead of points, and the beamformer should calculate the distances between line-elements and the point [5]. However, the long length of the line-elements results in prominent edge effects [2], [5]. These edge effects are due to the limited size of the aperture and originates from both ends of the line-element. It was shown that using a static roll-off apodization as shown in Fig. 2 along each row and column element reduces those edge effects without altering the main echo response [5], [9].

The pulse-echo field for the flat RCA 2-D array is limited to the forward looking rectilinear region in front of the transducer as shown in Fig. 3a, due to the perpendicular orientation of the transmit and receive fields. The relative orthogonal transmit and receive pressure fields at the depth of 80 mm are shown when steering the beam to the sides. Both transmit and receive beams were steered by $\pm 45^\circ$. When the horizontal array is used as a transmit array, it can steer the transmit beam in the $z-x$ plane, and at the same time the vertical array is receiving in the $z-y$ plane. Only the region indicated by white dashed lines, which is the intersection of transmit and receive pressure fields, can be imaged at any depth with an acceptable dynamic range.

## B. Curved RCA 2-D Arrays

Using a double-curved RCA 2-D array can extend the volumetric imaging field-of-view (FOV) to a curvilinear region. To spread the acoustic energy of a line-element curvilinearly along its larger dimension, it has to be curved to generate a diverging wave. The defocusing of the waves can be made by using a fixed electronic delay profile along each flat line-element, similar to a fixed first stage in micro-beamforming with 2-D arrays [13]. Another approach is to use a double curved diverging acoustic lens on top of the flat RCA array [10].

A concave diverging lens could be designed with a material, which has lower speed of sound compared to the human tissue. It will have a higher thickness around the corners and the sides of the array, and less thickness close to center of the array. Alternatively, a convex diverging lens can be made from a material with a higher speed of sound compared to the human tissue, which is preferred for a better contact surface. A flat diverging lens also can be made by using a combination of two different materials, one with higher and other one with lower speed of sound compared to the human tissue.

In Fig. 4 the pulse-echo energy as a function of lateral position for different lens f-numbers ($f_\theta$) is illustrated on an RCA array. The $f_\theta$ is defined as a ratio between the focal distance to the lens diameter. The pulse-echo energy drops by moving away from the forward looking region of the array. Almost at around $8^\circ$ steering angle the pulse-echo energy drops by 40 dB, when no lens is used. However, by using a diverging acoustic lens on top of an RCA 2-D array, a larger FOV can be illuminated. The FOV can be adjusted by using different $f_\theta$ values for the lens. By using a diverging lens with $f_\theta = -1$, the overlapped transmit and receive region increases to about $\pm 30^\circ$ in both directions as shown in Fig. 3b compared to Fig. 3a.

Note that, for the same aperture size, lower $f_\theta$ values for the lens corresponds to larger thicknesses of the lens and therefore the attenuation becomes higher through the lens material. Thus, there is a trade-off between FOV and attenuation. For example the delay profile can be in a range of 0 µs to 3.5 µs for a lens with $f_\theta = -0.7$ and a speed of sound of 1400 m/s, which corresponds to a thickness range of 0 mm to 5 mm. A suitable material for a lens could be Sylgard 160 (PDMS) with a density of 1580 kg/m$^3$ and a speed of sound of 950 m/s and attenuation of $0.4 f^{1.4}$ dB/cm MHz, where $f$ is the operating frequency in MHz. Therefore, for an operating frequency of 3 MHz the maximum attenuation is 6.14 dB at the largest thickness [14]. This might be compensated by doubling the amplitude of the excitation pulse.

## C. DAS Beamforming with Curved RCA 2-D Arrays

The time-of-flight (ToF) of a wavefront is given by the shortest distance from the arc source $s_n$ to the point being focused $p$, and back to the receiving element $r_n$, divided by the speed of sound. Using the notations from Fig. 5 this can be written as:

Figure 4. Reduction in the pulse-echo energy using a diverging lens relative to a flat transducer. The points are located on a line at 80 mm away from the surface of the transducer.
where the cross products \( \hat{c}_a \) between vector \( c_p \) equation for projection. To determine if the vector through the arc

\[ \hat{c}_a = \frac{c_p \times c_b}{\|c_p\| \|c_b\|}, \]

\[ \hat{i}_n = \frac{c_p' \times c_b}{\|c_p'\| \|c_b\|}, \] (3)

Depending on the location of the point \( p \), vectors \( \hat{i}_n \) and \( \hat{i}_b \) can be either \( j \) or \( -j \), where \( j \) is the unit vector of the z-axis. \( \hat{i}_n \) and \( \hat{i}_b \) have different signs, when \( \alpha_1 \leq \phi \leq \alpha_2 \) and same sign, when \( \alpha_2 \leq \phi \leq \alpha_1 \), where \( \alpha_1, \alpha_2 \), and \( \phi \) are the angles between the x-axis and vectors \( c_a, c_b \), and \( c_p \), respectively, as shown in Fig. 7.

When \( \hat{i}_n \) and \( \hat{i}_b \) have different signs, i.e. \( \hat{i}_a = j \) and \( \hat{i}_b = -j \), or \( \hat{i}_n = -j \) and \( \hat{i}_b = j \), the standard formula for the distance between an arc and a point can be used:

\[ d = \sqrt{\|pp'\|^2 + (\|cp\| - R)^2}, \] (4)

where \( R \) is the curvature of the arc and equals to \( \|ca\| \) or \( \|cb\| \).

When \( \hat{i}_n \) and \( \hat{i}_b \) have the same signs, i.e. \( \hat{i}_a = j \) and \( \hat{i}_b = j \), or \( \hat{i}_n = -j \) and \( \hat{i}_b = -j \), the shortest distance from the arc segment to the point is the distance from the closest end of the arc segment (a or b) to the point \( p \). The following therefore determines the minimum distance between the point \( p \) and the arc segment \( ab \):

\[ d(ab, p) = \begin{cases} \sqrt{\|pp'\|^2 + (\|cp\| - R)^2} & \text{if } \hat{i}_a = j \text{ and } \hat{i}_b = j, \hat{i}_a = -j \text{ and } \hat{i}_b = -j \end{cases} \]

\[ \sqrt{\|ap\|^2} & \text{if } \hat{i}_a = j \text{ and } \hat{i}_b = j, \hat{i}_a = -j \text{ and } \hat{i}_b = -j \] (5)

Using (5), the distances \( d(s_m, p) \) and \( d(r_n, p) \) can now be determined. The focused signal at point \( p \) is calculated by summing all receive signals at the time instances given by (1):

\[ z_m(p) = \sum_{n=1}^{N} a_{elec}(m, p) y_{m,n}(ToF_m(n, p)), \] (6)

where \( N \) is the number of receive elements, \( a_{elec} \) is the electronic receive apodization, and \( y_{m,n}(t) \) is the measured signal from emission \( m \) on the receive element \( n \) at time \( t \).

The synthetic transmit aperture (STA) focused signal at point \( p \) is calculated by summing the focused signals from all emissions:

\[ l_{STA}(p) = \sum_{m=1}^{M} b_{elec}(m, p) z_m(p), \] (7)

where \( M \) is the number of transmissions, \( b_{elec} \) is the electronic transmit apodization, and \( z_m(p) \) is the focused received signal from emission \( m \) at point \( p \). In general, both \( a_{elec} \) and \( b_{elec} \) are dependent on the imaging point \( p \) so that, a dynamic apodization can be achieved in transmit and receive. In this study however, they are fixed to an apodization window, e.g., Hanning, for all imaging points.

Figure 5. The time-of-flight of a wavefront is given by the shortest distance from the source \( s_m \) to the point being focused \( p \) and back to the receiving element \( r_n \), divided by the speed of sound.

Figure 6. Distance between a point \( p \) and an arc \( ab \) is calculated using (5).
III. METHODS

A. Synthetic Aperture Imaging Technique

In conventional ultrasound imaging it will be a tedious method to transmit for each steering angle so many times to cover the whole volume. However, this will not be a problem, if a SAI technique is used [15]. Thereby, all the transmit delay calculations can be done after the acquisition. A SAI sequence is designed for imaging down to 14 cm of depth. It utilizes single element transmissions on the row elements and the echoes are collected with all the column elements. For a speed of sound of 1540 m/s, 182 µs is required to acquire a single image line to a depth of 14 cm. For 62 emissions this is equivalent to a volume rate of 88 Hz. IQ-modulated RF data are used for beamforming a low-resolution volume for every emission and finally, by summing all the low-resolution volumes, a high-resolution volume is generated.

B. Imaging Quality Assessment Measures

The imaging performances of a double-curved RCA 2-D array is computed using the two measures described below:

1) Spatial Resolution: The spatial resolution is calculated as the full-width at half-maximum (FWHM) of the imaging system’s point spread function (PSF).

2) Contrast resolution: The spatial resolution is calculated as the cystic resolution (CR), which is the ability to detect an anechoic cyst in a uniform scattering medium [16]–[18].

<table>
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<th>Parameter name</th>
<th>Notation</th>
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<tr>
<td>Lens focal ratio</td>
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C. Simulation Setup

In this study, Field II [11], [12] is used for all simulations. A MATLAB (MathWorks Inc., Massachusetts, USA) beamformer that implements (7) was programmed to beamform data from curved RCA arrays and produce the PSFs included in this paper. The simulation parameters of a RCA 62×62 element 2-D array are shown in Table I. The receive array is rotated 90° with respect to the transmit array. Field II is set up to use lines to describe the apertures and each line-element is divided into square mathematical sub-elements with a side length of $\lambda/4$. To remove the otherwise apparent secondary echoes originating from the either ends of line-elements, two roll-off apodization regions are placed at both ends of each element [5], [9]. The length of each apodization region was equal to 15 times the pitch of the array. Each mathematical sub-element in both transmit and receive arrays is delayed according to the lens delay profile and no attenuation is assumed for the lens. Theoretically, transmitting with row elements and receiving with column elements should image exactly the same curvilinear volume as transmitting with column elements and receiving with row elements. Thus, no preference is considered in transmitting with row elements and receiving the echoes with column elements, or vice versa.

Fig. 8 illustrates two different ways to integrate a diverging lens over the array. The lens shown in Fig. 8a circumscribes...
the whole underlying array. On the other hand, the lens shown in Fig. 8b does not cover the whole array, instead the lens is inscribed in the array. In this configuration, essentially there is no diverging focusing applied to the end-most elements, and all elements between the end and the middle have compromised divergence. Thus, the defocusing is applied inconsistently across the array.

Both inscribed and circumscribed cases can provide apodization from lens attenuation as the lens becomes thicker toward the edges. The circumscribed case actually provides more apodization because in this case the lens gets thicker in the corners than the inscribed lens. The inscribed lens is advantageous because it has a smaller lens arc height and shorter chord length than the circumscribed lens. This reduced arc height improves patient contact possibilities, but, more importantly, the shorter chord length enables lower \( f_g \) defocusing. Fresnel lens could be another configuration as a diverging lens, however in this study the configuration shown in Fig. 8a has been chosen for the simulations.

IV. RESULTS AND DISCUSSION

The beamformer can IQ-beamform 250,000 voxels from a complex data set of 1.5 MiB from 62 receive line elements in approximately 14.1 s on a PC with a 3.4-GHz Intel Core i7-4770 CPU (Intel Corp., Santa Clara, CA, USA) and 32 GiB of RAM. The proof-of-concept Matlab implementation of the beamformer can therefore not achieve a frame rate useful for real-time applications, but the frame rate is adequate for research purposes.

Fig. 9 shows the received echoes that are generated from a single scatterer located at \((x, y, z) = (0, 0, 30)\) mm. The secondary echoes after the main echo are suppressed below \(-40\) dB by using the static roll-off apodization regions. The overlaid blue line is the predicted time-of-flight using (6).

Fig. 10 shows five 3-D PSFs simulated with Field II [11], [12] using SAI technique and beamformed for both flat and curved RCA 2-D array. The point targets are located at \((x, y, z) = (0, 0, 30)\) mm and \((x, y, z) = (0.5, -15, 25.9)\) mm and Hanning electronic apodization is applied over the received RF data. A Hanning apodization is applied over the low-resolution volumes before summing in the SAI technique. The PSFs are normalized to their maximum value and shown in a dynamic range of \(40\) dB. For the PSF 1 and PSF 2, the roll-off apodization is disabled. The secondary lobes located slightly after 30 mm depth in PSF 1 and PSF 2 are the apparent edge echoes and cannot be suppressed by using the electronic apodization. On the other hand, for the PSF 3 to PSF 5, the roll-off apodization is activated. It can be noticed that the apparent

Figure 10. Three cross-planes (azimuth, elevation, and C-plane) of 3-D PSFs are shown at a dynamic range of \(40\) dB. The origin corresponds to the center of the transducer surface aligned with a point target positioned at \((x, y, z) = (0, 0, 30)\) mm for PSF 1 to PSF 4, and a point target positioned at \((x, y, z) = (0, -15, 25.9)\) mm for PSF 5. The C-planes are at depth of 30 mm.
secondary echoes after each main echo in both PSF 3 and PSF 4 are suppressed by using the static roll-off apodization regions. Using the roll-off apodization does not change the lateral resolution of the main echo, this can be seen by comparing PSF 1 and PSF 2 with PSF 3 and PSF 4. The effect of roll-off apodization is mostly outside of the curvilinear imaging FOV of the array, and therefore will not affect the lateral resolution within the imaging FOV. Table II lists the FWHM and the CR of the simulated 3-D PSF 3 ~ PSF 5 shown in the Fig. 10.

Table II
FWHM and CR of simulated 3-D PSF 3 ~ PSF 5 shown in Fig. 10

<table>
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<th>PSF 4</th>
<th>PSF 5</th>
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<td><strong>Elevation FWHM</strong></td>
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<td>1.9</td>
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To study the PSF characteristics as a function of lateral angular position and radial distance, a point scatterer is imaged by sweeping it from 0° to 40° in the lateral plane with steps of 10° at radial distances from 10 mm to 60 mm from the center of the array. At each radial distance and angular position the FWHM and CR values are calculated over a volume of 10 mm × 10 mm × 10 mm surrounding the point target. Fig. 11 is illustrating the measured FWHM and CR values as a function of depth and angular position in lateral plane.

Using multiple elements in the transmit or receive and by adjusting their delays, the flat RCA 2-D array generates a focal line, however, with a curved RCA 2-D array, focusing in transmit or receive generates two intersection points instead of a focal line as shown in Fig. 12. Similar to flat RCA 2-D array, either of these two intersection points can be focused in receive and in this way suppressing the secondary intersection points in both transmit and receive. Looking at Fig. 12, it can be noticed that the characteristics of the focused intersections can be different at different angles. Moving away from the center towards the higher angular position in lateral plane, the elevation FWHM values become smaller while the CR values become larger. On the other hand, the lateral FWHM values stay constant, this is due to the intersection of the wavefronts in the receive direction which is at the center of the receive elements and therefore the elevation FWHM values stay constant for all angular positions in the lateral plane.

Fig. 13 illustrates three cross-planes (azimuth, elevation, and C-plane) of a phantom with point targets in water simulated with and without a diverging lens. The point targets are distributed in both lateral and elevation dimensions from –60 mm to 60 mm with step size of 10 mm, as well as in an axial range of 5 mm to 95 mm with step size of 4.5 mm. It can be seen from Fig. 13 that, by using a diverging lens the FOV is extended compared to the flat RCA array.

Fig. 14 illustrates three cross-planes (azimuth, elevation, and C-plane) of an anechoic cyst vessel phantom simulated with and without a diverging lens. The phantom with volume
size of $40 \times 40 \times 20 \text{ mm}^3$ contains an anechoic cyst vessel with radius of 5 mm along the azimuth dimension located at a 20 mm depth. The phantom is simulated in water with average scatterer density of 8 per mm$^3$. The FOV is extended compared to the flat RCA array. In Fig. 15, the cyst phantom is located deeper at a 60 mm depth and beamformed with the proposed DAS beamforming method, with and without a diverging lens. Similar to Fig. 14 here also using a diverging lens extends the FOV compared to the flat RCA array.

Diverging the wavefronts has the negative effect of lowering the pulse-echo energy as it is shown in Fig. 4 compared with the conventional row–column imaging using flat arrays. This loss of the energy can be compensated for by using all the elements in transmit and placing the transmit focus in front of the array.

Using a diverging lens, the elements at the middle of the array can be represented as an arc, but the 3-D focusing characteristics of the off-center elements should make their representations more complicated. It requires to formulate the trigonometric functions that are used for delay calculations in a spherical geometry. It is also possible to have a different curvature in transmit and receive, however that requires to formulate the delay calculations in bispherical coordinates, which was beyond the scope of this study.

V. Conclusion

In this paper the qualitative imaging performance of a curved 62×62 RCA 2-D array was evaluated. The capabilities of a curved RCA 2-D array to effectively focus in both transmit and receive were investigated, and a suitable DAS beamformer introduced and implemented. Using SAI technique it was possible to image down to 14 cm at a volume rate of 88 Hz. To validate the performance simulations of the imaging performance with a curved RCA 2-D array at several different situations was evaluated. Results confirm that using a diverging lens with $f_\theta = -1$ can increase the imaging FOV to $60^\circ \times 60^\circ$, and it is also possible to perform dynamic transmit-receive focusing throughout the curvilinear FOV. Thereby, the inherent imaging limitation with flat RCA 2-D arrays, i.e., its forward looking rectilinear FOV, is overcome using a diverging lens. Overall, having a low channel count and a large FOV offers the potential to fabricate arrays with large aperture sizes, which is important for abdominal scans. Thus, by using a curved RCA 2-D array, 3-D imaging is possible with equipment in the price range of conventional 2-D imaging. These advantages might contribute to an increased use of real-time 3-D ultrasound imaging in medical diagnostics, and to the development of new clinical applications.

If each line-element can be divided into two equal sub-
elements as shown in Fig. 16, by activating each sub-element of the row elements it is possible to eliminate either of those intersection points. The advantage of doing so is that, if the echoed signals are collected with the same transmitting elements, a two-way focusing profile can be produced, which is not possible with traditional row-column imaging since the transmit and receive apertures are perpendicular to each other. Therefore, by dividing the curvilinear FOV into 2 sub-volumes, each sub-volume can be beamformed with only row or column elements. Although this was not the main focus of this study, it could be interesting to investigate the focusing abilities using only the curved row or column elements.

Figure 15. Three cross-planes (azimuth, elevation, and C-plane) of a hollow tube with a diameter of 10 mm inside a rectangular box imaged with an RCA 2-D array with a diverging lens ($f_{D} = -1$) are shown in 40 dB dynamic range. The cyst box dimensions are 40 x 40 x 20 mm$^3$. The C-planes are at depth of 60 mm.

Figure 16. If each arc-shaped elements $s_1$ and $s_2$ is divided into two sub-elements, by activating each sub-element, only one intersection can be produced, $t_1$ or $t_2$ depending on which side has been activated. Thus, it is possible to accurately calculate the time-of-flight using only either row or column elements independently. Thereby, a two-way focusing profile can be achieved.

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REFERENCES


Patent Applications
Row-column Addressed 2-D Array with a Double Curved Surface

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A.1 Technical Field

The following generally relates to ultrasound imaging and more particularly to a row-column addressed 2-D with a double curved surface and/or 3-D imaging with the row-column addressed 2-D with the double curved surface.

A.2 Background

An ultrasound imaging system includes a transducer array, which includes a one-dimensional (1-D) or a two-dimensional (2-D) array of transducing elements. For three dimensional (3-D) imaging with a 2-D array, the elements can be individually addressed or group-wise addressed using row-column addressing where the elements are accessed by their row or column index, and each row and column in the array thereby acts as one large element. With individual addressing, an \(N \times N\) array would require \(N^2\) electrical connections and channels to fully address the array. As a result, 2-D arrays in the medical environment have been small with small fields of view, which are not well-suited for applications such as abdominal, breast, vascular, etc. examinations. With row-column addressing, the \(N \times N\) array would require only 2N electrical connections and channels to fully address the array.

Figure 1 schematically illustrates an example 6x6 flat array 102 configured for row-column addressing. Each column 106 includes an electrically conductive trace or path 108 in electrical communication with each element 104 of the column 106. The column 106 also includes an electrode 110, in electrical communication with the electrically conductive trace or path 108, which is used to excite the column 106. Each row 112 includes an electrically conductive trace or path 114 in electrical communication with the elements 104 of the row 112. The row 112 also includes an electrode 116, in electrical
communication with the electrically conductive trace or path 114, which is used to excite the row 112. The row-column addressing of the array 102 effectively transforms the 36-element 2-D array into a 6-element 1-D flat column array 118 and a 6-element 1-D flat row array 120.

For 3-D imaging, one of the 1-D arrays transmits waves into an object under evaluation and the other 1-D arrays receives echoes scattered from the insonified region. Both of the 1-D arrays can be focused in the lateral and elevation directions separately, and each of the 1-D arrays can electronically focus in one lateral dimension when delays are applied to the elements in the array. For example, the 1-D array 118 is able to focus the beam in x- and z-directions, but not in a y-direction. As a result, the emitted ultrasound is focused along a line segment or arc parallel to the y-direction. By adjusting the delays on the transmit elements, this focal line may be translated to any position in the xz-plane. The 1-D array 120 receives the echoes, and by applying delays, the received signals can be focused in a line segment or arc normal to any position in the yz-plane.

The 2-D array 102 can be used for phased array imaging. However, due to the 1-D arrays being flat, when transmitting plane waves, only a rectilinear forward-looking volume region can be imaged. Although it is possible to focus the ultrasound wavefronts curvilinearly, the pulse-echo field is limited only to a forward-looking volume region. This can be seen in Figure 2, which shows the transmit and receive pressure fields 202 and 204 when steering the transmit and receive beams by ±45° with a same radial distance. The imaging area is the intersection of these two fields, which, unfortunately, is limited to a rectilinear forward-looking boxed-shaped region 206 in front of the transducer 102. Furthermore, it is only possible to use each 1-D flat array either in transmit or receive focusing to jointly target a single point in 3-D space.

A.3 Summay

Aspects of the application address the above matters, and others. In one aspect, a transducer array for an ultrasound imaging system includes a row-column addressed 2-D array of transducer elements. The row-column addressed 2-D includes a first array of 1-D arrays of elements, a second array of 1-D arrays of elements, which is orthogonal to the first array, and a double-curved surface. In another aspect, an apparatus includes a transducer array with an array-wise addressable 2-D array with a curved surface. The 2-D array includes a set of 1-D column array elements and a set of 1-D row array elements. The apparatus further includes transmit circuitry that conveys an excitation pulse to the transducer array, receive circuitry that receives a signal indicative of an ultrasound echo from the transducer array, and a beamformer that processes the received signal, generating ultrasound image data. In another aspect, a method includes transmitting an ultrasound signal with 2-D row-column addressed transducer array with a curved surface, transmitting an echo signal with the 2-D row-column addressed transducer array with a curved surface, beamforming the echo signal to create an image, and displaying the image. Those skilled
in the art will recognize still other aspects of the present application upon reading and understanding the attached description.

A.4 Brief Description of the Drawings

The application is illustrated by way of example and not limited by the figures of the accompanying drawings, in which like references indicate similar elements and in which:

Figure 1 schematically illustrates a 2-D array configured for row-column addressing;

Figure 2 schematically illustrates a rectilinear forward-looking imaging region of the prior art 2-D array of Figure 1;

Figure 3 schematically illustrates an example ultrasound imaging system with a row-column addressed transducer array with a curved surface;

Figure 4 schematically illustrates an example row-column addressed transducer array for the configuration of Figure 3;

Figure 5 schematically illustrates another example row-column addressed transducer array for the configuration of Figure 3;

Figure 6 schematically illustrates an example of the curved surface of the row-column addressed transducer array;

Figure 7 schematically illustrates a field of view for the row-column addressed transducer array with the curved surface;

Figure 8 shows the intersection of transmitted wavefronts with the row-column addressed transducer array with the curved surface is a point;

Figure 9 shows wavefronts that create two focus points using the row-column addressed transducer array with the curved surface;

Figure 10 provides a diagram showing how to extend beamforming to calculate a time-of-flight in dynamic receive focusing with the curved surface for single element transmissions;

Figure 11 shows a beamformed B-mode image of a point scatterer imaged with the curved surface as the system’s point spread function;

Figure 12 shows a beamformed B-mode image of a point scatterer imaged without the curved surface as the system’s point spread function;

Figure 13 shows a beamformed B-mode image of point scatterers imaged with the curved surface;

Figure 14 shows a beamformed B-mode image of point scatterers imaged without the curved surface;

Figure 15 graphically shows pulse-echo energy as a function of steering angles for different f#s and without the curved surface; and

Figure 16 illustrates an example method in accordance with an embodiment herein.
Figure 3 schematically illustrates an example ultrasound imaging system 300. The ultrasound imaging system 300 includes a 2-D transducer array 302 configured for row-column addressing. The 2-D array 302 include a plurality of detector elements arranged in a N×M matrix of N rows and M columns, where N and M are positive integers and \(N=\text{M}\) or \(N\neq\text{M}\) (e.g., \(N>\text{M}\) or \(N<\text{M}\)). Examples of square arrays include 64x64, 192x192, 256x256, 512x512 and/or other arrays, including larger and/or smaller arrays. Examples also include non-square arrays such as rectangular, circular, irregular and/or other shaped arrays. The elements can be piezoelectric (PZT), capacitive micromachined ultrasonic transducer (CMUT) elements, and/or other transducing elements.

Figure 4 schematically illustrates a non-limiting example of the row-column addressed array 304. Each column 402 includes an electrically conductive trace or path 404 in electrical communication with the elements 404 of the column 402. The column 402 also includes an electrode 408, in electrical communication with the electrically conductive trace or path 404, which is used to excite the column 104. Each row 410 includes an electrically conductive trace or path 412 in electrical communication with the elements 406 of the row 410. The row 410 also includes an electrode 414, in electrical communication with the electrically conductive trace or path 412, which is used to excite the row 110.

Either the rows 410 or the columns 402 transmit while the other receives. In this example, the elements 406 of the column 402 are arranged in a first or “y” direction of an “x-y” plane, and the elements 406 of the row 410 are arranged in a second or “x” direction of the “x-y” plane. The elements 406 of the column 402 are configured to transmit, in response to being excited with an electrical pulse, a pressure wave in a “z-x” plane, and the elements 406 of the rows 410 are configured to receive echoes, produced in response to the transmitted pressure wave interacting with matter, in a “z-y” plane. In this example, “x,” “y,” and “z” are axes of the Cartesian coordinate system 416.

A geometry of the elements 406 in Figure 4 are square and/or rectangular. Figure 5 schematically illustrates a variation of Figure 4 in which a geometry of outer or perimeter elements 502 are not square and/or rectangular, and elements at corners 504 are omitted. The triangular elements 502 provide integrated apodization that linearly scales (rather than discretely scales, as shown in Figure 4) the output pressure transmitted by (during transmit) and the electrical signal generated by (during receive) of the elements of the periphery. The illustrated apodization decreases in a direction towards the periphery and away from the center region. Other geometries (e.g., hexagon, circle, etc.) are also contemplated herein.

The examples discussed in connection with Figures 4 and 5 and/or other examples of row-column addressed 2-D arrays are described in international application serial number PCT/IB2013/002838, filed on December 20, 2013, and entitled “Ultrasound Imaging Transducer Array with Integrated Apodization,” the entirety of which is incorporated herein by reference. An example in international application PCT/IB2013/002838 includes PZT transducer elements. Another example in international application PC-
T/IB2013/002838 includes CMUT elements. Other configurations of the row-column addressed 2-D array 102 are also contemplated herein.

Returning to Figure 3, the transducer array 302 further includes a curved surface 306. In one instance, the curved surface 306 is a diverging lens disposed in front of an active transducing surface of the 2-D array 304. In another instance, the curved surface 306 is the active transducing surface of the 2-D array 304. In yet another instance, the curved surface 306 is a combination of the diverging lens and the active transducing surface. As described in greater detail below, the curved surface 306 disperses the transmit and receive fields so that they overlap in an area that is larger than the forward-looking volume region of the 2-D array 304. Also described in greater detail below, the curved surface 306 allows for two-way focusing by focusing at any point in the 3-D space both in transmit and in receive separately. As a result, at least two elements are enough to image a whole curvilinear volume.

Transmit circuitry 308 generates pulses that excite a predetermined set of the addressed columns (or rows) to emit one or more ultrasound beams or waves. Receive circuitry 310 receives signals indicative echoes or reflected waves, which are generated in response to the transmitted ultrasound beam or wave interacting with (stationary and/or flowing), from a predetermined set of addressed rows (or columns). The receive circuitry 310 may also pre-process and/or condition the received signals, e.g., by amplifying, digitizing, etc. the signals. A switch 312 switches between the transmit circuitry 308 and the receive circuitry 310, depending on whether the transducer array 302 is in transmit or receive mode.

A beamformer 312 processes the received echoes, for example, by applying time delays and weights, summing, and/or otherwise processing the received echoes. Alternatively, the beamformer 312 can process the received echoes by applying spatial matched filtering to focus the RF-data at any time and location in space. An example of this is described in Jensen, & Gori. (2001), “Spatial filters for focusing ultrasound images,” 2, 1507–1511 vol.2. doi:10.1109/ULTSYM.

A display 316 is configured to visually display images and/or other information. A scan converter 318 scan converts the beamformed data, converting the beamformed data (e.g., images or volumes) into the coordinate system of the display 316, which visually displays the images. In one instance, the data is visually displayed in an interactive graphical user interface (GUI), which allows the user to selectively rotate, scale, and/or manipulate the displayed data through a mouse, a keyboard, touch-screen controls, etc.

A controller 320 controls one or more of the components of the system 300 such as at least one of the transmit circuitry 308 or receive circuitry 310, the switch 312, and the beamformer 314. Such control can be based on the mode of operation (e.g., B-mode, etc.) of the system 300 and/or otherwise. A user interface 322 includes include an input device (e.g., a physical control, a touch-sensitive surface, etc.) and/or an output device (e.g., a display screen, etc.). A mode, scanning, and/or other function can be activated by a signal indicative of input from the user interface 322.

In one instance, the transducer array 302 is part of a probe and the transmit circuitry
308, the receive circuitry 310, the switch 312, the beamformer 314, the scan converter 318, the controller 320, the user interface 322, and the display 316 are part of a console. Communication there between can be through a wired (e.g., a cable and electro-mechanical interfaces) and/or wireless communication channel. In this instance, console can be a portable computer such as a laptop, a notebook, etc., with additional hardware and/or software for ultrasound imaging. The console can be docked to a docking station and used.

Alternatively, the console can be part (fixed or removable) of a mobile or portable cart system with wheels, casters, rollers, or the like, which can be moved around. In this instance, the display 316 may be separate from the console and connected thereto through a wired and/or wireless communication channel. Where the cart includes a docking interface, the console can be interfaced with the cart and used. An example of such a system is described in US publication 2011/0118562 A1, entitled “Portable ultrasound scanner,” and filed on November 17, 2009, which is incorporated herein in its entirety by reference.

Alternatively, the transducer array 302, the transmit circuitry 308, the receive circuitry 310, the switch 312, the beamformer 314, the scan converter 318, the controller 320, the user interface 322, and the display 316 are all housed by and enclosed within a hand-held ultrasound apparatus, with a housing that mechanically supports and/or shields the components within. In this instance, the 2-D array 304 is structurally integrated as part of the housing. An example of a hand-held device is described in US patent 7,699,776, entitled “Intuitive Ultrasonic Imaging System and Related Method Thereof,” and filed on March 6, 2003, which is incorporated herein in its entirety by reference.

As briefly discussed above, the transducer array 302 includes the curved surface 306. Figure 6 illustrates a representation of the curved surface 306 in connection with the row-column addressed 2-D array 304 of Figure 5. However, it is to be understood that the curved surface 306 can be employed with the row-column addressed 2-D array 304 of Figure 4 and/or other row-column addressed 2-D array. As discussed herein, the curved surface 306 can be a diverging lens disposed adjacent to the transmitting/receiving side of the row-column addressed 2-D array. Examples of suitable lenses include a spherical, cylindrical, Fresnel, and/or other lens. In a variation, the curved surface 306 is the active transducing surface of the 2-D array 304. The curved surface 306 may or may not have integrated apodization.

The illustrated curved surface 306 is a double-curved (i.e., curved in the x-z and y-z planes) convex surface. In one instance, the curvature is the same in the x-z and x-y planes. In another instance, the curvature is different in the x-z and x-y planes. The curved surface 306 has a first thickness at peripheral regions 602 and a second thickness at a center region 604, where the first thickness is greater than the second thickness. The thickness of a region controls the delay provided by that region and hence the divergence.

A non-limiting example of suitable thicknesses includes a thickness in a range of 0 to 5 mm for a lens with f#=-0.7 and a speed of sound of 1400 m/s, which corresponds to a delay range of 0 to 3.5 µs. The f# is defined as a ratio between a focal distance to a
A.5. Detailed Description

diameter of the lens. A non-limiting example of a suitable material of the curved surface 306 is Sylgard® 170 (PDMS) with a density of 1000 and a speed of sound of 1400 m/s and attenuation of 3.7. Sylgard® 170 is a product of Dow Corning Corporation, MI, USA. The curved surface 306 can be disposed centered over the 2-D array 304 or disposed off-center, e.g., at a corner region.

The curved surface 306 disperses the transmit and receive fields so that they overlap in a larger area relative to the forward-looking region of the 2-D array 304. Generally, for the flat 2-D array 102 of Figure 1, each line element produces a cylindrical wave in one direction and a plane wave in the other direction, which restricts the width in the other direction to a line segment having the width of the element. In contrast, the curved surface 306 produces a spherical wave, which originates from a virtual point source, located behind the array, which propagates in a larger field of view. This can be seen by comparing Figures 2 and 7. In Figure 2, the transmit and receive pressure fields 202 and 204 intersect and provide the rectilinear forward-looking region 206. In Figure 7, diverging transmit and receive pressure fields 702 and 704 intersect and provide a larger intersecting and thus larger imaging region 706, which is larger than the rectilinear forward-looking region 206 of Figure 2. In the example of Figure 7, the overlapped transmit and receive region 706, compared to Figure 2, increases to ±26.5° in both directions.

With the 2-D flat array 102 of Figure 1, the intersection of transmitted wavefronts has a shape of a line segment. With the configuration shown in Figures 3 and 6, with the curved surface 306, the intersection of the transmitted wavefronts is not a line segment but a point. This can be seen in Figure 8, which shows how the intersection of multiple wavefronts creates a curved line instead of a straight line. Each curved line 802, 804 and 806 is an intersection of two spherical wavefronts. Depending on a distance of each element to a focusing point, each wavefront has a different curvature and therefore all the wavefronts contact at only one point 808. With the 2-D flat array 102 of Figure 1, focusing is not possible in the orthogonal plane to the transmit steering direction.

With the curved surface 306, focusing in transmit direction can be achieved by delaying the wavefronts so that they pass the first point of contact and generate two focus points. This can be seen in Figure 9, which shows curved lines 902, 904 and 906 and two focus points 908 and 910. This allows for not only focusing in the transmit plane but also in the orthogonal plane. In conventional ultrasound imaging it will be a tedious process to transmit for each steering angle so many times to cover the whole volume. However, this can be done with the approach described herein, e.g., by employing a synthetic aperture imaging (SAI) algorithm, which allows all the delay calculations to be done after data acquisition. Furthermore, since it is possible to focus at any point in the 3-D space both in transmit and in receive separately, two-way focusing can be achieved.

An example beamforming algorithm is described next. The example first explains flat row-column beamforming and then extends this to row-column beamforming for the curved surface 306.

Delay-and-sum (DAS) beamformers usually assume the geometry of the sound sources and receivers to be points. However, by row-column addressing the elements on a
2-D matrix array, each row and column is acoustically equivalent to a line-element. Furthermore, the emitted wavefront of a line-element has the shape of a cylinder, i.e. it is a plane wave in the plane aligned along the line-element and a circle arc in the plane orthogonal to the line-element. Assuming the geometry of the line-elements to be points is therefore a poor approximation. A better approximation assumes the line-elements to be line segments instead of points. At the focal zone where an array of line-elements is focused, the geometry is also a line segment.

Calculating the distances between the line-elements and a given point should therefore be calculated as the distance between a line segment and a point. For beamforming with line-segment virtual sources, the time-of-flight for the sound propagating through the media can be calculated as shown in Equation 1:

\[ t_{ToF} = \frac{|r_{fp} - r_{xmt}| \pm d(AB_{fp}, P) + d(CD_{rcv}, P)}{c}, \]

where \( r_{xmt} \) and \( r_{fp} \) are vectors from a center of the 2-D array to a center of each transmit line-element and also to a center of a focal line-segment, \( P \) is a position vector of any beamforming point, \( AB_{fp} \) is a vector from one end to another end of a focal line-segment, \( CD_{rcv} \) is a vector from one end to another end of each receive line element, \( d(\cdot; \cdot) \) calculates a shortest distance between the point \( P \) and each of the transmit or receive line-elements which finds a minimum distance between a point and a line-segment, and \( \pm \) refers to whether the point \( P \) is in between a focal line-segment and a surface of the transducer, i.e. -, or the point \( P \) is located after the focal line-segment, i.e. +. The minimum distance between the point \( P \) and the line segment \( AB \) can be calculated as shown in Equation 2:

\[ d(AB, P) = \begin{cases} \frac{\|AB \times AP\|}{\|AB\|} & \text{if } 0 \leq \frac{AB \cdot AP}{\|AB\|^2} \leq 1 \\ \|AP\| & \text{if } \frac{AB \cdot AP}{\|AB\|^2} < 0 \\ \|BP\| & \text{if } \frac{AB \cdot AP}{\|AB\|^2} > 1 \end{cases}, \]

Figure 10 is used to explain how to extend the above to calculate the time-of-flight in dynamic receive focusing with the curved surface 306 for single element transmissions. The curved surface 306 is not shown for sake of clarity, and is located above or on top of the 2-D array 304. A shortest path 1002 from a source line-element 1004 to an imaging point \( P \) 1006 is in a plane 1008 that goes through a curved surface center 1012 and the point \( P \) 1006 and that is orthogonal to a transmit plane B 1010, which goes through the transmit element 1004 and the curved surface center 1012.

To calculate the distance from the transmit element 1004 to the imaging point \( P \) 1006, \((BP)\), a perpendicular projection \( P_{TXproj} \) 1014 of the point \( P \) 1006 is located on the plane
1010. An intersection 1016 of the plane B 1008 (which goes through the point P 1006, \( P_{TX_{proj}} \) 1014 and the center 1012), with the transmitting element 1004, is identified at a point B 1018. \( BP \) is then calculated as shown in Equation 3:

\[
BP = \sqrt{BP_{TX_{proj}}^2 + P_{TX_{proj}}^2},
\]

where \( P_{TX_{proj}} \) is a line segment from \( P_{TX_{proj}} \) 1014 to \( P \) 1006, and \( BP_{TX_{proj}} \) is a line segment, which is computed by subtracting the vector which is pointing from the center 1012 to the \( P_{TX_{proj}} \) 1014 by the radius of the curved surface 306, which is fixed. This approach is also used to calculate \( AP \) 1024, using a perpendicular projection \( P_{RX_{proj}} \) 1026 and a point A 1028 on a receive segment 1030, as shown in Equation 4:

\[
AP = \sqrt{AP_{RX_{proj}}^2 + P_{RX_{proj}}^2},
\]

where \( P_{RX_{proj}} \) is a line segment from \( P_{RX_{proj}} \) 1014 to the \( P \) 1106, and \( AP_{RX_{proj}} \) is a line segment, which is computed by subtracting the line segment from the center 1012 to the \( P_{RX_{proj}} \) 1026 by the radius of the curved surface 306.

A total distance is computed as shown in Equation 5:

\[
BP + PA.
\]

This is repeated for the other columns of the 2-D array 304 for the point \( P \) 1006. The computed shortest distances for all of the columns to the point \( P \) 1006 provide the data to beamform the point \( P \) 1006. This can be achieved using a synthetic aperture imaging and/or other algorithm. In synthetic transmit aperture imaging, by taking advantage of the superposition theorem, the transmit focus may be synthesized in every location by delaying and summing a plurality of datasets (before or after conventional beamforming) acquired from successive transmissions.

In synthetic transmit aperture imaging with a linear array with \( N \) elements, for each image point \((r, \theta)\), the A-scan signal is as shown in Equation 6:

\[
A_{STA}(r, \theta) = \sum_{n=0}^{N-1} \sum_{m=0}^{N-1} S_{m,n} \left( \frac{2r}{c} - \tau_n - \tau_m \right),
\]

where \( S_{m,n} \) is the echo signal. The first and second summations correspond to transmit and receive beamforming. \( \tau_n \) and \( \tau_m \) are beamforming delays for transmit \( m \) and receive \( n \) element combination shown in Equations 7 and 8:

\[
\tau_m = \frac{1}{c} \left( r - \sqrt{x_m^2 - r^2 - 2x_m r \sin \theta} \right)
\]

\[
\tau_n = \frac{1}{c} \left( r - \sqrt{x_n^2 - r^2 - 2x_n r \sin \theta} \right).
\]
An example of this is described in Jensen, J. A., Nikolov, S., Gammelmark, K. L., & Pedersen, M. H. (2006). “Synthetic Aperture Ultrasound Imaging. Ultrasonics,” 44(SUPPL.), e5–e15, e5–e15. doi:10.1016/j.ultras.2006.07.017. The above shortest distance calculation is repeated for all the points of interest in the field of view or region 706. The points can be inside the planes, outside of the planes and/or on a plane(s).

Figures 11 and 12 respectively show two beamformed B-mode images of a point scatterer imaged with the curved surface 306 (Figure 11) and without the curved surface 306 (Figure 12). The point scatterer is positioned at eighty (80) millimeters (mm) distance from the 2-D array 304. The origin corresponds to the center of the transducer surface. Data was generated using synthetic aperture imaging with 62 single-element transmissions, emitting a 2-cycle sinusoidal excitation pulse with every row of elements at a time and receiving the echoes with all column elements. These figures show how energy falls off with angle with and without the curved surface 306.

Figures 13 and 14 respectively show two beamformed B-mode images for multiple point scatterers imaged with the curved surface 306 (Figure 13) and without the curved surface 306 (Figure 14). For these images, seven scatterers 1301, 1302, 1303, 1304, 1305, 1306 and 1307 are positioned at forty (40) mm distance from the 2-D array 304 within ±45° in the elevation plane. One 1-D array includes sixty-two columns, and the other 1-D array includes sixty-two rows. The receive array is rotated 90° with respect to the transmit array. The parameters as shown in Table 1.

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</tbody>
</table>

Figure 13 shows all of the scatterers 1301, 1302, 1303, 1304, 1305, 1306 and 1307, including those (1301, 1302, 1306 and 1307) beyond the forward-looking region 206, which are in the larger area 706. Figure 14 shows the scatterers within the forward-looking region 206 of the 2-D array 304, which include scatterers 1303, 1304 and 1305. Using a diverging lens can thus enable imaging a large field of view.

Figure 15 graphically shows a comparison of the normalized pulse-echo energy as a function of steering angle from center to one side of the array with and without the curved
The points are located on an arc with radii of eighty (80) millimeters (mm) away from the center of the 2-D array 304 and spanned from 0° to 80°. A curve 1502 shows the pulse-echo energy without the curved surface 306. The curves 1504, 1506, 1508 and 1510 respectively show the pulse-echo energy with the curved surface 306 for a decreasing f#, which shows an increasing imaging region. For example, at -40 dB a lens increases the pulse-echo field-of-view up to 70 degrees for a lens with f# = -0.5.

Figure 16 illustrates an example method. At 1600, the transducer array 302 is used to scan a subject. As described herein the transducer array 302 includes the row-addressed 2-D array 304 with the curved surface 306. Two-way focusing may be employed. The curved surface 306 may be configured with integrated apodization. At 1602, the acquired data is beamformed as described herein. At 1604, the beamformed data is processed to generate an image. At 1606, the image is displayed.

The functions described herein may be implemented via one or more hardware and/or software computer processors (e.g., a micro-processor, a central processing unit (CPU), etc.) executing one or more computer readable instructions encoded or embodied on computer readable storage medium (which excludes transitory medium) such as physical memory which causes the one or more processors to carry out the various acts and/or other functions and/or acts. Additionally, or alternatively, the one or more processors can execute instructions carried by transitory medium such as a signal or carrier wave.

The application has been described with reference to various embodiments. Modifications and alterations will occur to others upon reading the application. It is intended that the invention be construed as including all such modifications and alterations, including insofar as they come within the scope of the appended claims and the equivalents thereof.

A.6 Claims

What is claimed is:

1. A transducer array (302) for an ultrasound imaging system (300), the transducer array comprising: a row-column addressed 2-D array of transducer elements (304), wherein the row-column addressed 2-D array comprises: a first array of 1-D arrays of elements; a second array of 1-D arrays of elements, which is orthogonal to the first array; and a double-curved surface (306).

2. The transducer array of claim 1, wherein the double-curved surface is a lens.

3. The transducer array of claim 2, wherein the lens is a convex lens.

4. The transducer array of claim 3, wherein the convex lens is a spherical lens.

5. The transducer array of any of claims 2 to 4, wherein the lens has a same curvature in two orthogonal directions.

6. The transducer array of any of claims 2 to 4, wherein the lens has a first curvature in a direction of the first array and a second curvature in a direction of the second array, wherein the first and second curvatures are different.
7. The transducer array of any of claims 2 to 6, wherein the lens has a first thickness at a periphery and a second thickness at a center region, and the first thickness is greater than the second thickness.

8. The transducer array of any of claims 2 to 6, wherein the lens is centered over the 2-D array.

9. The transducer array of any of claims 2 to 6, wherein the lens is disposed off-center with respect to the 2-D array.

10. The transducer array of any of claims 2 to 6, wherein the 2-D array has a first imaging region (206) and the combination of the 2-D array and the lens has a second imaging region (706), and the second imaging region is larger than the first imaging region.

11. The transducer array of any of claims 2 to 10, wherein the lens has integrated apodization.

12. The transducer array of any of claims 2 to 11, wherein the lens is a Fresnel lens.

13. The transducer array of claim 1, wherein the curved surface is an active transducing surface of the 2-D array.

14. The transducer array of any of claims 1 to 13, wherein the 2-D array includes piezoelectric or capacitive micromachined ultrasonic transducer transducing elements.

15. An apparatus (300), comprising: a transducer array (302) with an array-wise addressable 2-D array with a curved surface, wherein the 2-D array includes a set of 1-D column array elements and a set of 1-D row array elements; transmit circuitry (308) that conveys an excitation pulse to the transducer array; receive circuitry (310) that receives a signal indicative of an ultrasound echo from the transducer array; and a beamformer (314) that processes the received signal, generating ultrasound image data.

16. The apparatus of claim 15, further comprising: a controller (320) configured to control the transmit circuitry for two-way focusing to focus the set of 1-D column array elements or the set of 1-D row array elements.

17. The apparatus of claim 16, wherein the beamformer is configured to determine a shortest distance from a source to a point to a drain for each column and each row for a plurality of points in an imaging field of view.

18. The apparatus of claim 17, wherein the beamformer sums coherently low resolution beamformed images for a point for all transmissions.

19. The apparatus of any of claims 17 to 18, wherein the point is The points can be inside the planes, outside of the planes and/or on a plane(s).

20. A method, comprising: transmitting an ultrasound signal with 2-D row-column addressed transducer array with a curved surface; transmitting an echo signal with the 2-D row-column addressed transducer array with a curved surface; beamforming the echo signal to create an image; and displaying the image.
A transducer array (302) for an ultrasound imaging system (300) includes a row-column addressed 2-D array of transducer elements (304). The row-column addressed 2-D includes a first array of 1-D arrays of elements, a second array of 1-D arrays of elements, which is orthogonal to the first array, and a double-curved surface (306). In another aspect, an apparatus includes a transducer array with an array-wise addressable 2-D array with a curved surface. The 2-D array includes a set of 1-D column array elements and a set of 1-D row array elements. The apparatus further includes transmit circuitry (308) that conveys an excitation pulse to the transducer array, receive circuitry (308) that receives a signal indicative of an ultrasound echo from the transducer array, and a beamformer (314) that processes the received signal, generating ultrasound image data.
FIGURE 1
Fig. 3

- TRANSDUCER ARRAY 302
- RCA 2-D ARRAY 304
- CURVED SURFACE 306
- SWITCH 312
- TRANSMIT CIRCUITRY 308
- CONTROLLER 320
- USER INTERFACE 322
- RECEIVE CIRCUITRY 310
- BEAMFORMER 314
- SCAN CONVERTER 318
- DISPLAY 316

FIGURE 3
FIGURE 4
FIGURE 6

FIGURE 7
FIGURE 8
FIGURE 10
FIGURE 15

FIGURE 16
3-D Imaging and/or Flow Estimation with a Row-Column Addressed 2-D Transducer Array

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B.1 Technical Field

The following generally relates to ultrasound imaging and more particularly to three-dimensional (3-D) imaging with a row-column addressed 2-D transducer array and/or flow estimation with a row-column addressed 2-D transducer array.

B.2 Background

For 3-D imaging with a two-dimensional (2-D) array of transducing elements, the elements can be individually addressed or group-wise addressed, e.g., using row-column addressing, where groups of elements are accessed either by a row index or a column index such that each row and column is utilized as a single larger element. With traditional row-column addressing, the row and column arrays each steer the transmit and receive beams in one direction. However, the transmit and receive directions are orthogonal to each other. For example, when the row array is used as a transmit array, it can steer the transmit angle in the $z-x$ plane while at the same time the column array receives in the $z-y$ plane. When the sequence is complete, the two arrays switch function, and now the column array is used as a transmit array and the row array as a receive array. This leads to two identical volumes; however, at each point only one-way focusing in transmit and receive is achievable. Three-dimensional vector flow has been implemented with a row and column array in a single plane as described in Christiansen et al., serial number 14/599,857, filed January 2015, and entitled “3-D flow estimation using row-column addressed transducer
arrays,” which is incorporated herein by references in its entirety. Unfortunately, the lack of two-way focusing and limitations with only 3-D vector flow in a plane render traditional row-column addressing not well-suited for real-time volumetric 3-D vector flow imaging. In view of at least the above, there is an unresolved need for another approach for 3-D imaging with a row-column addressed transducer array and/or flow estimation with a row-column addressed transducer array.

B.3 Summay

Aspects of the application address the above matters, and others.

In one aspect, an ultrasound imaging system includes a 2-D transducer array with a first 1-D array of one or more rows of transducing elements configured to produce first ultrasound data and a second 1-D array of one or more columns of transducing elements configured to produce second ultrasound data. The first and second 1-D arrays are configured for row-column addressing. The ultrasound imaging system further includes a controller configured to control transmission and reception of the first and second 1-D arrays, and a beamformer configured to beamform the received first and second echoes to produce ultrasound data, and an image processor configured to process the ultrasound data to generate an image, which is displayed via a display.

In another aspect, a method includes controlling transmission and reception of first and second 1-D arrays of a 2-D transducer array, wherein the first 1-D array includes one or more rows of transducing elements configured to produce first ultrasound data, and the second 1-D array includes one or more columns of transducing elements configured to produce second ultrasound data, wherein the first and second 1-D arrays are configured for row-column addressing, beamforming the received first and second echoes to produce ultrasound data, and processing the ultrasound data to generate an image, which is displayed via a display.

In another aspect, a computer readable medium is encoded with non-transitory computer executable instructions which when executed by a processor causes the processor to: control transmission and reception of first and second 1-D arrays of a 2-D transducer array, wherein the first 1-D array includes one or more rows of transducing elements configured to produce first ultrasound data, and the second 1-D array includes one or more columns of transducing elements configured to produce second ultrasound data, wherein the first and second 1-D arrays are configured for row-column addressing, and at least one of beamform the received first and second echoes to produce ultrasound data with two-way focusing in elevation or process the received first and second echoes to estimate volumetric 3-D vector flow information.

Those skilled in the art will recognize still other aspects of the present application upon reading and understanding the attached description.
B.4  Brief Description of the Drawings

The application is illustrated by way of example and not limited by the figures of the accompanying drawings, in which like references indicate similar elements and in which:

FIGURE 1 schematically illustrates an example imaging system with a 2-D row-column addressed array;
FIGURE 2 schematically illustrates an example of the 2-D row-column addressed array;
FIGURE 3 schematically illustrates an example of an effective 1-D column array resulting from the row-column addressing with the 2-D row-column addressed array;
FIGURE 4 schematically illustrates an example of an effective 1-D row array resulting from the row-column addressing with the 2-D row-column addressed array;
FIGURE 5 depicts a point spread function for the azimuth direction in accordance with an embodiment herein;
FIGURE 6 depicts a point spread function for the elevation direction in accordance with an embodiment herein;
FIGURE 7 depicts a point spread function for elevation versus azimuth in accordance with an embodiment herein;
FIGURE 8 depicts a point spread function for the azimuth direction for traditional row-column addressing;
FIGURE 9 depicts a point spread functions for the elevation direction for traditional row-column addressing;
FIGURE 10 depicts a point spread functions for elevation versus azimuth for traditional row-column addressing;
FIGURE 11 schematically illustrates single element transmission synthetic aperture imaging;
FIGURE 12 schematically illustrates processing of the output of the single element transmission synthetic aperture imaging of FIGURE 11;
FIGURE 13 schematically illustrates row-column steered sequence for 3-D vector flow obtained in a cross sectional plane with TO; and
FIGURE 14 schematically illustrates row-column steered sequence for volumetric 3-D vector flow with TO.

B.5  Detailed Description

The following describes an approach to achieve two-way focusing in elevation with data acquired with a 2-D row-column addressed array and/or estimate vector flow information with data acquired with the 2-D row-column addressed array.

Figure 1 schematically illustrates an example ultrasound imaging system 100. The ultrasound imaging system 100 includes a 2-D transducer array 102 with at least two one-dimensional (1-D) arrays 104 of transducing elements 106 where the 1-D arrays
104 are arranged orthogonal with respect to each other. The 2-D transducer array 102 includes N rows (or columns) and M columns (or rows) of the transducing elements 106, where N and M are positive integers and \( N = M \) or \( N \neq M \). The 2-D transducer array 102 may include a 16x16, 32x32, 64x64, 128x128, 512x512 larger or smaller array, a non-square/rectangular array, a circular array, and/or another 2-D transducer array. Figure 2 illustrates an example of the 2-D transducer array 102.

In Figure 2, the 2-D transducer array 102 is 6x6 array (N=M=6). The 2-D array 102 includes a plurality of rows 2041, 2042, 2043, 2044, 2045, and 2046, collectively referred to herein as rows 204. The 2-D array 102 also includes a plurality of columns 2061, 2062, 2063, 2064, 2065, and 2066, collectively referred to herein as columns 206. The rows 204 and columns 206 include individual elements 2081,1, . . . , 2081,6, . . . 2086,1, . . . , 2086,6, collectively referred to herein as elements 208. The individual rows 204 and columns 206 are addressable (individually or in groups) respectively through contacts 2101, 2102, 2103, 2104, 2105, and 2106, and 2121, 2122, 2123, 2124, 2125, and 2126, collectively referred to as row contacts 210 and column contacts 212. Row-column addressing effectively transforms the 36-element 2-D array 102 into a six-element, 1-D column array 302 (FIGURE 3) and a six-element, 1-D row array 402 (FIGURE 4). The axial direction is along the beam direction, the azimuth direction is orthogonal to the axial direction and along the transmitting elements, and the elevation direction is orthogonal to the azimuth and axial directions.

Returning to FIGURE 1, the transducing elements 106 may include piezoelectric, capacitive micromachined ultrasonic transducer (CMUT), and/or other elements. Furthermore, the transducing elements 106 may include integrated apodization, which may be identical or different for the individual elements. An example is described in patent application PCT/IB2013/002838, filed December 19, 2013, and entitled “Ultrasound Imaging Transducer Array with Integrated Apodization,” the entirety of which is incorporated herein by reference. Furthermore, the 2-D array 102 may have flat 1-D arrays, one curved 1-D array, two curved 1-D arrays, a single curved lens in front of or behind one of the 1-D arrays, a double curved lens in front of or behind the 1-D arrays, a combination of at least one curved 1-D array and at least one curved lens, etc. An example is described in patent application PCT/IB2016/053367, filed June 8, 2016, and entitled “Row-Column Addressed 2-D array with a Double Curved Surface,” the entirety of which is incorporated herein by reference.

Transmit circuitry 108 generates pulses that excite a predetermined set of addressed 1-D arrays of the 2-D array 102 to emit one or more ultrasound beams or waves, e.g., into a scan field of view. Receive circuitry 110 receives echoes or reflected waves, which are generated in response to the transmitted ultrasound beam or wave interacting with (stationary and/or flowing) structure in the scan field of view, from a predetermined set of addressed arrays of the 2-D array 102. A controller 112 controls the transmit circuitry 108 and/or the receive circuitry 108. Examples of control include: 1) transmitting and receiving with row elements, 2) transmitting and receiving with column elements, 3) transmitting with row elements and receiving with column elements, 4) transmitting with
column elements and receiving with row elements, 5) transmitting with row elements and receiving with row and column elements, 6) transmitting with column elements and receiving with row and column elements, 7) transmitting with row elements and receiving with row and column elements and transmitting with column elements and receiving with row and column elements, etc.

As described in greater detail below, the controller 110 can control the transmit and receive circuitries 106 and 108 to acquire data to create a two-way focusing profile in elevation in the transmit direction. This can be achieved, for example, by controlling the transmit and receive circuitries 106 and 108 to transmit and receive with both rows and columns (example 7 above). This approach improves spatial resolution relative to traditional row-column addressing. As a result, relative to traditional row-column addressing, the size of the array can be maintained to yield the full spatial resolution improvement, the size of the array can be reduced while still yielding improved spatial resolution, and/or the size of the array can be reduced to maintain a particular resolution. For example, to maintain a particular resolution, the size of the array in each dimension can be reduced by a factor of 2 relative to traditional row-column addressing.

A beamformer 114 processes the echoes, for example, by applying time delays, weighting on the channels, summing, and/or otherwise beamforming received echoes, producing data for generating images in A-mode, B-mode, Doppler, and/or other ultrasound imaging modes. An image processor 116 processes the beamformed data. For B-mode, this may include generating a sequence of focused, coherent echo samples along focused scanlines of a scanplane. The image processor 116 may also be configured to process the scanlines to lower speckle and/or improve specular reflector delineation via spatial compounding, apply filtering such as FIR and/or IIR, etc. A scan converter 118 scan converts the output of the image processor 118 and generates data for display, for example, by converting the data to the coordinate system of the display. The scan converter 118 can be configured to employ analog and/or digital scan converting techniques.

The illustrated embodiment further includes a velocity processor 120. In a variation, the velocity processor 120 is omitted and/or is located remote from the imaging system 100, such as in a computing device such as a computer or the like, which is remote from and not part of the imaging system 100. The illustrated velocity processor 120 is configured to process the beamformed row-column addressed data to determine 3-D velocity components. As described in greater detail below, this may include estimating 3-D velocity components from unfocussed diverging waves in combination with synthetic aperture (SA) and directional transverse oscillation (DTO), which yields higher volume rates, estimating 3-D velocity components from focused emissions and TO, and/or estimating 2-D and/or 3-D velocity components using DTO, which yields higher spatial resolution. Other methods could be transmission of plane waves and using a velocity estimator based on speckle tracking, e.g., Bohs et al., “Speckle tracking for multi-dimensional flow estimation,” 2000, vol. 38, or vector Doppler techniques.

A rendering engine 122 visually presents one or more of the images and/or the velocity information via a display monitor 124. In one instance, the data is visually displayed in
an interactive graphical user interface (GUI), which allows the user to selectively rotate, scale, and/or manipulate the displayed data through a mouse, a keyboard, touch-screen controls, etc. A user interface 126 includes one or more input devices (e.g., a button, a knob, a slider, a touch pad, etc.) and/or one or more output devices (e.g., a display screen, lights, a speaker, etc.). The user interface 126 can be used to select an imaging mode such as row-column addressing with two-way focusing in elevation and/or 3-D velocity component estimation, e.g., using one or more of the 3-D velocity component estimation approaches described herein.

In one instance, the transducer array 102 is part of a probe and the transmit circuitry 108, the receive circuitry 110, the controller 112, the beamformer 114, the image processor 116, the scan converter 118, the velocity processor 120, the rendering engine 122, the display 124, and the user interface 126 are part of a separate console such as a computing system. Communication there between can be through a wired (e.g., a cable and electro-mechanical interfaces) and/or wireless communication channel. In this instance, the console can be similar to a portable computer such as a laptop, a notebook, etc., with additional hardware and/or software for ultrasound imaging. The console can be docked to a docking station and used.

Alternatively, the console can be part (fixed or removable) of a mobile or portable cart system with wheels, casters, rollers, or the like, which can be moved around. In this instance, the display 124 may be separate from the console and connected thereto through a wired and/or wireless communication channel. Where the cart includes a docking interface, the laptop or notebook computer type console can be interfaced with the cart and used. An example of a cart system where the console can be selectively installed and removed is described in US publication 2011/0118562 A1, entitled “Portable ultrasound scanner,” and filed on November 17, 2009, which is incorporated herein in its entirety by reference.

Alternatively, the transducer array 102, the transmit circuitry 108, the receive circuitry 110, the controller 112, the beamformer 114, the image processor 116, the scan converter 118, the velocity processor 120, the rendering engine 122, the display 124, and the user interface 126 are housed within a hand-held ultrasound apparatus, where the housing mechanically supports and/or encloses the components therein. In this instance, the transducer array 102 and/or the display 124 can be part of the housing, being structurally integrated or part of a surface or end of the hand-held ultrasound apparatus. An example of a hand-held device is in US 7,699,776, entitled “Intuitive Ultrasonic Imaging System and Related Method Thereof,” and filed on March 6, 2003, which is incorporated herein in its entirety by reference.

As briefly discussed above, in one non-limiting instance, the controller 112 controls the transmit circuitry 108 and the receive circuitry 110 to acquire data to create a two-way focusing profile in elevation in the transmit direction. For this, the controller 112 controls the transmit circuitry 108 and the receive circuitry 110 to transmit with row elements and receive with both row and column elements and then transmit with column elements and receive with both row and column elements, or vice versa, i.e. transmit with column
elements and receive with both row and column elements and then transmit with row elements and receive with both row and column elements.

Where the transmit and receive elements are the same (i.e. both rows, or both columns), the acquired data is used for two-way focusing in elevation, e.g., at least because the transmit and receive focus lines are both in the same plane. Where the transmit and receive elements are perpendicular to each other (i.e. rows and columns, or columns and rows), the acquired data is used to focus along each transmit focus line with only one-way focusing in elevation. The beamformer 114 beamforms the received echo signals, producing two volumes, a one for transmitting with row elements and receiving the echoes with both row and column elements, and another for transmitting with column elements and receiving the echoes with both row and column elements, both with a two-way focusing profile in elevation in transmit and a one-way profile in receive.

These two volumes are combined to produce a volume with a two-way focusing profile in elevation in the transmit direction. In one instance, the two volumes are combined by multiplying them and taking the square root. In general, this approach is well suited for static or moving tissue, e.g. at least because it is not very sensitive to movement. In another instance, the two volumes are combined by summing phase coherent signals. This approach is also well suited for static or moving tissue, although it may be more sensitive to movement. In another instance, the two volumes are combined by taking a minimum value of an absolute value of the two volumes at each point in space. These approaches create a two-way focusing profile in elevation in the transmit direction, which increases spatial resolution in both dimensions, relative to traditional row-column addressing where orthogonal arrays (row and column, or column and row) are used to transmit and receive. In general, the spatial resolution in the perpendicular dimension is improved by using the two-way focusing profile for each point along the transmit focus-line instead of the one-way beam profile.

FIGURES 5, 6, and 7 show point spread functions (PSF’s) respectively for the azimuth direction, the elevation direction, and elevation versus azimuth. In FIGURE 5, a first or y-axis represents range in units of millimeters (mm) and a second or x-axis represents azimuth in the same units. In FIGURE 6, a first or y-axis represent the range similar to FIGURE 5, and a second or x-axis represents elevation in the same units. In FIGURE 7, a first or y-axis represents elevation range and a second or x-axis represents azimuth, both in the units of millimeters. For comparative purposes, FIGURES 8, 9, and 10 show PSF’s for traditional row-column addressing. In FIGURE 8, a first or y-axis represents range in units of millimeters and a second or x-axis represents azimuth in the same units. In FIGURE 9, a first or y-axis represent the range similar to FIGURE 8, and a second or x-axis represents elevation in the same units. In FIGURE 10, a first or y-axis represents elevation range and a second or x-axis represents azimuth, both in the units of millimeters. FIGURES 7 and 10 show improved spatial resolution with the approached described herein (FIGURE 7) relative to traditional row-column addressing (FIGURE 10).

Again, as a result of the improved resolution, the number of transmissions can be maintained to yield the full spatial resolution improvement, the number of transmissions
can be reduced while still yielding improved spatial resolution, and/or the number of transmissions can be reduced to maintain a particular resolution, relative to traditional row-column addressing. To have a same lateral resolution for both fully addressed and row-column addressed 2-D arrays, the number of row or column elements on a row-column addressed array is increased. Changing the aperture size will not change the normalized amplitudes, and the side-lobe levels relative to the main lobe level. By squaring the Fourier transform of the apertures, the amplitudes of the side-lobes are halved by a factor of two in decibels (dB) when two-way focusing is performed. A measure of contrast is the side-lobe level. Therefore, the approached described herein will have superior contrast performance relative to the traditional row-column addressed 2-D array one-way focusing.

Super resolution technique using ultrasound can overcome the diffraction limit and provide enhanced visibility of vascular features. It is possible to study the micro-vasculature and thereby directly the perfusion, of tissues and tumors. The resolution of standard clinical ultrasound systems typically ranges between 50-500 µm, and even high frequency setups struggle to resolve micro-vessels with a diameter around 100 µm or less. However, it is possible to go beyond the diffraction limit when applying contrast agents consisting of gas filled microbubbles, which is disclosed in Errico et al., “Ultrafast ultrasound localization microscopy for deep superresolution vascular imaging”, Nature, vol. 527, pp. 499-502, November 2015. Microbubbles are enhanced in ultrasound images due to their non-linear properties and strong back-scattering ability. Based on the received RF data, it is possible to locate and track individual microbubbles in 2-D when a 1-D transducer is used or in a full volume when a 2-D transducer is applied. The precision of the estimated microbubble position highly depends on the focusing performance. With two way RC focusing, as described herein, the location of the microbubble is therefore expected to be improved as well as the overall performance of mapping micro-vasculatures in a volume or a plane.

As briefly discussed above, in one non-limiting instance, the velocity processor 120 processes the beamformed row-column addressed echoes to estimate 3-D velocity components from unfocussed diverging waves in combination with synthetic aperture (SA) and directional transverse oscillation (DTO). The technique is not limited to this combination, such that focused or plane waves can be utilized in transmit and can be combined with e.g. speckle tracking, vector Doppler techniques etc. An example of this described next in connection with FIGURES 11 and 12.

In traditional synthetic aperture imaging with a 1-D array, the transmit sequence consists of several unfocused emissions, which can be either single element transmissions or multiple element transmission using virtual sources. After each transmit event, a low resolution image is beamformed by using all elements in receive. When all transmit events have been executed, the low resolution images are added together to form a high resolution image. The high resolution image is equally focused everywhere in the plane. The high resolution image can be processed to render a B-mode image, but can also be used for vector flow estimation. Patent application PCT/IB2015/051526, filed March 2, 2015, and entitled “Vector velocity estimation with directional transverse oscillation,” which is
incorporated herein by reference in its entirety, describes an approach in which a high resolution image is obtained with synthetic aperture (SA) techniques and used to obtain the lateral velocity component in the entire plane, when directional transverse-oscillation (DTO) is applied.

The approach described herein expands this to 3-D vector flow for the 2-D row-column addressed transducer array 102, which results in high resolution volumes (HRV’s). FIGURE 11 shows an example data acquisition sequence for obtaining high resolution volumes with the row-column addressed array 102, which are processed by the velocity processor 120 to produce 3-D vector flow estimation in a volume. Due to the large area of each element in a row-column addressed array, enough energy from a single element emission is generated to beamform a low resolution volume. When an emission is made with a column element, all row elements are used in receive to beamform a low resolution volume, and when a row emission is made, all column elements are used in receive to beamform yet another low resolution volume. The interleaved transmit sequence consists of N emissions distributed between N/2 row emissions and N/2 column emissions. Adding all the N/2 low resolution images beamformed with the aperture containing the row elements yields the high resolution volume HRVCR, and adding all the N/2 low resolution images beamformed with the aperture containing the column elements yields the high resolution volume HRVCR.

As shown in FIGURE 12, each of the HRV’s is separated into multiple high resolution planes (HRP’s), and processed by a transverse oscillation (TO) estimator to yield the lateral velocity component. An example of a suitable TO estimator is described in US 6,148,224 A, filed December 30, 2016, and entitled “Apparatus and method for determining movements and velocities of moving objects,” which is incorporated herein by reference in its entirety. The TO estimator requires two TO signals as input, which need to be phase shifted by a quarter of the lateral wavelength. The TO signals can be created in the receive beamforming by changing the apodization function to contain two separated peaks. However, a lateral oscillation can also be generated in the Fourier domain, known as k-space, to provide better control over the lateral oscillation function to contain a desired lateral oscillation frequency. Example approaches are described in Jensen et al., “High frame rate vector velocity estimation using plane waves and transverse oscillation,” in Proc. IEEE Ultrason. Symp., 2015, pp. 1–4, and Salles et al., “2-D arterial wall motion imaging using ultrafast ultrasound and transverse oscillations,” IEEE Trans. Ultrason., Ferroelec., Freq. Contr., vol. 62, no. 6, pp. 1047–1058, 2015.

FIGURE 12 shows multiplication of the filter and the Fourier transformed plane yields a TO HRP. The plane is filtered in the lateral dimension, while the axial dimension is not filtered, or untouched. The directional information of the flow is preserved by applying a Hilbert transform on the filtered plane (spatial domain) for each of the lines in the lateral direction. The directional information of the flow could also be obtained in the Fourier domain, by setting all negative frequencies equal to 0. These two signals (the Hilbert
transformed and non-Hilbert transformed signal) are now used as input to the velocity processor 120. The output of the velocity processor 120 is the 2-D vector flow information for the axial and the lateral velocity components in the entire plane. This routine is then performed on all the planes that makes up for the HRV to yield 2-D vector flow in a volume. The HRVCR is used to estimate the direction and the magnitude of the velocity component in the direction parallel to the column elements, and the HRVRC is used to estimate the axial velocity and the azimuth velocity components.


Combining the estimated axial velocity component with the lateral velocity component found from HRVCR and with the respective lateral velocity component found from HRVRC yields the 3-D vector flow information for the entire volume. Additional combinations can also be used to estimate the velocities, such that the high resolution volume can be constructed from the addition or any multiplication of HRVCR, HRVRR, HRVCC, or HRVRC.

As briefly discussed above, in one non-limiting instance, the velocity processor 120 processes the beamformed row-column addressed echoes to estimate 3-D velocity components from focused emissions in a plane and TO. An example of this described next in connection with FIGURE 13.

application serial number 20160106391 A, publication number 2016/06391 A1, filed on May 5, 2013, and entitled “Three dimensional (3D) transverse oscillation vector velocity ultrasound imaging,” which is incorporated herein by reference. The result is one component perpendicular to the element orientation in addition to an axial component. Three beamformed lines are needed, including one center line for the axial estimator and two steered lines for the transverse estimate. The center line rcenter is beamformed along the direction (0, 0, z), using delay-and-sum and a traditional apodization profile. For the two steered lines, a traditional TO apodization profile with two separated peaks is applied and beamforming is performed along the lines (x, y, z) = (λx(z)/8, 0, z) to create the λx/4 spatial separation.

This approach can be expanded to estimate 3-D velocities with the 2-D row-column addressed transducer array 102. The third velocity component can be obtained by applying the same procedure as for the transverse component, but this time by beamforming the two steered lines at ±λy/8 in the orthogonal direction. All five lines are beamformed from two transmit events and combined afterwards. The five beamformed signals are subsequently used as input to the TO velocity estimator. From each transmit event three lines are beamformed at multiple direction. Two of the lines, rleft and rright, are used to estimate the velocity component perpendicular to the tallest dimension of the receiving elements using the TO method, and the third line, rcenter, is used to estimate the axial velocity with an autocorrelation approach, such as that describe in Kasai et al., “Real-Time Two-Dimensional Blood Flow Imaging using an Autocorrelation Technique,” IEEE Trans. Son. Ultrason., vol. 32, pp. 458–463, 1985 or Loupas et al., “An Axial Velocity Estimator for Ultrasound Blood Flow Imaging, Based on a Full Evaluation of the Doppler Equation by Means of a Two-dimensional autocorrelation approach,” UFFC, 1995, vol 42, pp. 672-688. By combining the estimated transverse velocity components, one from each transmit event, with one of the two independent axial estimates, a 3-D velocity vector along the direction of the respective beamformed centerline is obtained.

The transmit sequence can either be designed to yield M-mode data, where 3-D vector flow is estimated in points along the axial (0,0,z) direction, or it can be expanded to contain 3-D vector flow in a plane, when several steered emissions in one plane are added to the sequence, and finally, if steered emissions are made in two planes, 3-D volumetric flow can be obtained. The steered transmit sequence is used to estimate 3-D vector flow in the cross-sectional plane in a vessel. This sequence consists of one focused emission C1 using column elements and N focused emissions Ri using row elements, where i = 1...N. 3-D vector flow is estimated in points along the N steered directions in the zy-plane. The column emission generated a plane wave within the cross sectional zy-scan plane, whereas plane waves perpendicular to the scan plane were steered in the zy-plane when using the row elements.

From the row transmit event Ri, the vxi and vzi velocity components are estimated in points along the direction of the respective beamformed centerline. However, the C1 column transmit event provides the data for beamforming the lines needed for estimating all vyi and vzi velocity components, as this transmit event sonifies the zy scan plane.
The steered transmit sequence used is schematically written as: C1 → R1 → R2 → R3 → ...RN, and C1 → R1 → R2 → R3 → ...RN. The sequence can be modified to yield volumetric 3-D vector flow, if several column emissions are added. A sequence to yield volumetric 3-D vector flow could be written as: C1 → R1 → C2 → R2 → ...CN → RN and C1 → R1 → C2 → R2 → ...CN → RN. This is shown in FIGURE 13.

Compared to the M-mode sequence, the steered sequence differs in two ways. First, after each column emission C1, multiple steered row emissions RN are emitted. From each row emission three lines are beamformed according to the steering directions and vx and vz can be estimated along each direction. Second, from a single column emission C1, three lines are beamformed along each steering direction yielding vy and vz velocity estimates along the N directions. 3-D vector flow is estimated in points along directions originating from the center of the aperture and through the intersection between the focal lines. The estimation plane is obtained when interpolating the combined 3-D vector flow estimates.

To achieve volumetric 3-D flow, TO beamforming is performed in multiple directions. TO beamforming is performed at all sites where the focal line from a row emission and a column emission are intersecting. Both sequences yield continuous data, which means that the distance between each identical emission type is equally distributed in time for all time. An advantage of continuous data is that very high frames rate can be obtained, and that dynamic ensemble lengths and any echo canceling filters can be applied. The higher obtainable frame rate with continuous data occurs, since a sliding window can be applied on the beamformed data to generate one velocity estimate. The velocity estimate can be updated from each new similar emission, since the new data can replace the oldest data in the estimator.

As briefly discussed above, in one non-limiting instance, the velocity processor 120 processes the beamformed row-column addressed echoes to estimate 2-D and/or 3-D velocity components in a volume using DTO. An example of this described next in connection with FIGURE 14.

Compared to the single plane sequence of FIGURE 13, the volumetric sequence differs in at least two ways. First, after multiple column emission CN are emitted, and second, TO beamforming is performed where ever the focal line from a row emission or a column emission are intersecting. 3-D vector flow is estimated in all points along directions originating from the center of the aperture and through the intersection between the focal lines. The estimation volume is obtained when interpolating the combined 3-D vector flow estimates. An alternative sequence could be: C1→C1→R1→R1→...CN→CN→CN→RN→RN. This gives a high velocity range and a continuous sequence. Although the sequence becomes longer however the time difference between the two sequences for every direction becomes smaller compared to the previous sequences. This increases the maximum detectable velocity as this is given by \( v_{\text{max}} = \frac{\lambda_x}{2T_{\text{prf}}} \), where \( \lambda_x \) is the lateral or azimuth wavelength and Tprf is the time between measurements. Keeping Tprf low, this, ensures a high maximum detectable velocity.

The application has been described with reference to various embodiments. Modifica-
ations and alterations will occur to others upon reading the application. It is intended that the invention be construed as including all such modifications and alterations, including insofar as they come within the scope of the appended claims and the equivalents thereof.

B.6 Claims

What is claimed is:

1. An ultrasound imaging system (100), comprising:
   a 2-D transducer array (102), including:
   a first 1-D array (104, 204) of one or more rows of transducing elements (106, 204, . . . 204₆) configured to produce first ultrasound data; and
   a second 1-D array (104, 206) of one or more columns of transducing elements (106, 206₁, . . . 206₆) configured to produce second ultrasound data, wherein the first and second 1-D arrays are configured for row-column addressing;
   a controller (112) configured to control transmission and reception of the first and second 1-D arrays;
   a beamformer (114) configured to beamform the received first and second echoes to produce ultrasound data; and
   an image processor (116) configured to process the ultrasound data to generate an image, which is displayed via a display (224).

2. The ultrasound imaging system of claim 1, wherein the controller is configured to control the first and second 1-D arrays to transmit a first ultrasound signal with the first 1-D array and receive first echoes with the first and second 1-D arrays, and subsequently transmit a second ultrasound signal with the second 1-D array and receive second echoes with the first and second 1-D arrays, and the beamformer is configured to combine the beamformed first and second echoes to produce ultrasound data with two-way focusing in an elevation direction in transmit.

3. The ultrasound imaging system of claim 2, wherein the beamformer combines the first and second echoes by multiplying the beamformed first and second echoes.

4. The ultrasound imaging system of claim 2, wherein the beamformer combines the beamformed first and second echoes by summing the first and second echoes.

5. The ultrasound imaging system of claim 2, wherein the beamformer combines the beamformed first and second echoes by taking a minimum value of an absolute value of the first and second echoes at each point in space.

6. The ultrasound imaging system of claim 1, further comprising: a velocity processor (120) configured to processes the beamform data to produce 3-D vector flow volumetric imaging data.

7. The ultrasound imaging system of claim 6, wherein the controller is configured to control transmission of the first and second 1-D arrays to produce single element transmission.
8. The ultrasound imaging system of claim 7, wherein the velocity processor is configured to process the beamformed data using a synthetic aperture and a directional transverse oscillation estimator.

9. The ultrasound imaging system of claim 6, wherein the controller is configured to control transmission of the first and second 1-D arrays to produce focused steered emission sequence.

10. The ultrasound imaging system of claim 9, wherein the velocity processor is configured to process the beamformed data using a transverse oscillation estimator to estimate 3-D vector flow at least one of a plane or a volume.

11. The ultrasound imaging system of claim 9, wherein the velocity processor is configured to process the beamformed data using directional transverse oscillation to compute at least one of a 2-D in-plane or a 3-D vector flow estimate.

12. The ultrasound imaging system of claim 9, wherein the beamformer employs directional beamforming in the flow direction estimated by directional transverse oscillation.

13. The ultrasound imaging system of any of claims 1 to 12, further comprising: a diverging lens coupled to a transducing side of the 2-D transducer array.

14. The ultrasound imaging system of any of claims 1 to 13, wherein at least one of the first 1-D array or the second 1-D array includes a curved array.

15. A method, comprising: controlling transmission and reception of first and second 1-D arrays of a 2-D transducer array, wherein the first 1-D array includes one or more rows of transducing elements configured to produce first ultrasound data, and the second 1-D array includes one or more columns of transducing elements configured to produce second ultrasound data, wherein the first and second 1-D arrays are configured for row-column addressing; beamforming the received first and second echoes to produce ultrasound data; and processing the ultrasound data to generate an image, which is displayed via a display.

16. The method of claim 15, wherein the controlling includes controlling the first and second 1-D arrays to transmit a first ultrasound signal with the first 1-D array and receive first echoes with the first and second 1-D arrays, and subsequently transmit a second ultrasound signal with the second 1-D array and receive second echoes with the first and second 1-D arrays, and wherein the beamforming combines the first and second echoes to produce the ultrasound data with two-way focusing in an elevation direction.

17. The method of claim 16, further comprising: processing the two-way focused data to estimate and correct for motion in at least one of 1-D, 2-D or 3-D ultrasound data.

18. The method of claim 16, further comprising: controlling transmission to produce single element transmission or constructing a virtual source transmit; and processing the received echoes using a synthetic aperture algorithm.

19. The method of claim 18, further comprising: processing the data to produce super resolution imaging using micro bubbles in at least one of 1-D, 2-D or 3-D ultrasound data.

20. The method of claim 15, further comprising: controlling the transmission to produce single element transmission or constructing a virtual source transmit; and processing the received echoes using a synthetic aperture to produce high resolution volumes.
21. The method of claim 20, further comprising: estimating flow by adding the high resolution volumes.

22. The method of claim 20, further comprising: estimating flow by multiplying the high resolution volumes.

23. The method of claim 20, further comprising: estimating flow in a row direction by processing a high resolution volume of the volumes for the flow direction; and estimating flow in a column direction by processing a high resolution volume of the volumes for the flow column.

24. The method of claim 20, further comprising: employing directional beamforming in a flow direction estimated by transverse oscillation to refine a flow estimate.

25. The method of any of claims 15 to 24, further comprising: displaying only one line in M-mode.

26. A computer readable medium encoded with non-transitory computer executable instructions which when executed by a processor causes the processor to: control transmission and reception of first and second 1-D arrays of a 2-D transducer array, wherein the first 1-D array includes one or more rows of transducing elements configured to produce first ultrasound data, and the second 1-D array includes one or more columns of transducing elements configured to produce second ultrasound data, wherein the first and second 1-D arrays are configured for row-column addressing; beamform the received first and second echoes to produce ultrasound data; and process the ultrasound data to generate an image.

**B.7 Abstract**

An ultrasound imaging system (100) includes a 2-D transducer array (102) with a first 1-D array (104, 204) of one or more rows of transducing elements (106, 204, . . ., 204$_6$) configured to produce first ultrasound data and a second 1-D array (104, 206) of one or more columns of transducing elements (106, 206, . . ., 206$_6$) configured to produce second ultrasound data. The first and second 1-D arrays are configured for row-column addressing. The ultrasound imaging system further includes a controller (112) configured to control transmission and reception of the first and second 1-D arrays, and a beamformer (114) configured to beamform the received first and second echoes to produce ultrasound data, and an image processor (120) configured to process the ultrasound data to generate an image, which is displayed via a display (224).
2. D Transducer Array

TRANSMIT CIRCUITRY
CONTROLLER
USER INTERFACE
DISPLAY
RECEIVE CIRCUITRY
BEAMFORMER
SCAN CONVERTER

VELOCITY PROCESSOR
IMAGE PROCESSOR

2-D ARRAYS
1-D ARRAYS
TRANSDUCING ELEMENTS

FIGURE 1

ANA1315-WO
(BKM-10-8040-PCT)
FIGURE 5
FIGURE 6
FIGURE 7
FIGURE 8
FIGURE 9
FIGURE 10
FIGURE 11

Transmit aperture

Receive aperture

Low resolution volume

High resolution volume

FIGURE 12

Filtering in 2D Fourier domain

HRV

HRV

HRV

HRV
Volumetric Sequence

TO beamforming on columns  2-D velocity estimates

Combined 3-D vector flow

TO beamforming on rows

Focal line  Centerline  Velocity lines

Active elements  +λ/8

FIGURE 14
Combined Bibliography

References from Chapter 1


Griffith, J. M. and W. L. Henry (1974). “A Sector Scanner for Real Time Two-Dimensional Echocardiography”. In: **Circulation** 49.6, pp. 1147–1152. DOI: 10.1161/01.CIR.49.6.1147. URL: [http://circ.ahajournals.org/content/49/6/1147.abstract](http://circ.ahajournals.org/content/49/6/1147.abstract) (cit. on p. 1).


**References from Chapter 2**


References from Chapter 3


References from Chapter 4


References from Chapter 5


References from Chapter 6