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Bera, Deep; Van Den Adel, Franc; Radeljic-Jakic, Nikola; Lippe, Boris; Soozande, Mehdi; Pertijs, Michiel A.P.; Verweij, Martin D.; Kruizinga, Pieter; Daeichin, Verya; Vos, Hendrik J.

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Fast volumetric imaging using a matrix TEE probe with partitioned transmit-receive array

D. Bera[1], F. van den Adel[2], N. Radeljic-Jakic[2], B. Lippe[2], M. Soozande[1], M.A.P. Pertijs[3], M.D. Verweij[3,1], P. Kruizinga[1], V. Daeichin[3], H.J. Vos[1,3], A.F.W. van der Steen[1,3], J.G. Bosch[1], N. de Jong[1,3]

1 Department of Biomedical Engineering, Thoraxcenter, Erasmus MC, Rotterdam, The Netherlands

2 Oldelft Ultrasound, Delft, The Netherlands

3 Delft University of Technology, Delft, The Netherlands

Corresponding author: Deep Bera

P.O. Box 2040, 3000 CA Rotterdam, The Netherlands

kamer EE-2302, Wytemaweg 80, 3015 CN Rotterdam,

Tel: +31 10 704 34 74

Email: d.bera@erasmusmc.nl

Abstract

We present a 3D multiline parallel beamforming scheme for real-time volumetric ultrasound imaging using a prototype matrix TEE probe with diagonally diced elements and separated transmit and receive arrays. The elements in the smaller rectangular transmit array are directly wired to the ultrasound system. The elements of the larger square receive aperture are grouped in 4×4-elements sub-arrays by micro-beamforming in an ASIC. We propose a beamforming sequence with 85 transmit-receive events that shows good performance for a volume sector of 60°×60°. The beamforming is validated using Field II simulations, phantom measurements and in-vivo imaging. The proposed parallel beamforming achieves up to 59 Hz volume rate and produces good image quality by angle-weighted combination of overlapping sub-volumes. Point Spread Function, Contrast Ratio and Contrast-to-Noise Ratio in the phantom experiment closely match with simulation. *In-vivo* 3D imaging at 22 Hz volume rate in a healthy adult pig clearly shows the cardiac structures, including valve motion.

Keywords: Transesophageal echocardiography, matrix transducer, sub-array beamforming, parallel beamforming, volumetric ultrasound imaging

INTRODUCTION

Echocardiography is an indispensable diagnostic modality to assess the anatomy and function of the heart. In general, there are two types of echocardiography routinely performed in the clinic: transthoracic echocardiography (TTE), and transesophageal echocardiography (TEE). In TTE, an ultrasound transducer is placed on the chest, and the imaging is performed through an acoustic window in between the ribs. The ultrasound images produced by TTE may suffer from poor image quality because of the limited acoustic window and attenuation, aberration, shadowing and reflections due to the skin, fat, and ribs. In TEE, a transducer is mounted on the tip of a gastroscopic tube and inserted via the mouth into the patient's esophagus to image the heart. Therefore, unlike TTE, TEE images are not deteriorated by the skin, fat, or ribs. Moreover, as the esophagus is located only millimeters away from the heart, the received ultrasound signals in TEE are less attenuated than in TTE. TEE, therefore, produces a superior image quality to TTE, especially for the cardiac structures such as the aorta, pulmonary artery, valves, atria, atrial septum, atrial appendages and even the coronary arteries.

At present, TEE is routinely performed in several heart conditions. It is most commonly used to evaluate valvular disease, prosthetic heart valve dysfunction, cardiac sources of embolism, aortic dissections or aneurysms, and endocarditis (Khandheria et al. 1994). In addition, TEE is performed both to verify the preoperative diagnosis and to monitor the progress in many cardiac surgical procedures such as congenital heart disease corrections or valve repair (Cheitlin et al. 2003).

Several studies have shown that real-time 3D imaging is preferred over 2D imaging in most cardiac diagnostics (Kapoor et al. 2016; Montealegre-Gallegos et al. 2014) because of its superior visualization of 3D structures in the heart. Unlike 2D imaging where the acquisitions are performed only at fixed imaging planes corresponding to standard views, in 3D TEE all the important information is captured in a single dataset. In a comprehensive study, an acquired volume can be rotated and cropped at any desired plane to view different cardiac structures. Moreover, 3D TEE provides more consistent measurements of clinically relevant parameters such as volumes of the left ventricle (LV) and the right ventricle (RV), ejection fraction, and cardiac output (Montealegre-Gallegos et al. 2014), compared to 2D TEE. Furthermore, 3D TEE allows better morphological and dynamic evaluation of 3D cardiac structures such as the tricuspid valve, the aortic valve, and the mitral valve. Consequently, 3D TEE has become an essential diagnostic modality for a comprehensive examination of cardiac anatomy and function as well as for guiding and monitoring operative and catheter-based interventions (Frank et al. 2014; Sugeng et al. 2008).

For 3D TEE in adults, there are a number of commercially available matrix array TEE probes : X7-2t from Philips Ultrasound, Bothell, WA (Salgo 2007); V5M TEE from Siemens Healthineers GmbH, Erlangen, Germany (Siemens Healthineers 2012); and 6VT-D from General Electric Healthcare, Amersham, U.K.(GE Healthcare 2013). These matrix TEE probes are capable of real-time acquisition and live 3D display. The Philips X7-2t probe has an active aperture of 10×10 mm^2 with 2500 elements of frequency range 2-7 MHz (Salgo 2007). The V5M TEE probe has an aperture size of 14.5×11.5 mm^2 (operating frequency 3-7 MHz). The 6VT-D TEE probe has an effective aperture size of 14.3×12.7 mm^2, and its operating frequency range is 3-8 MHz. These probes comprise of complicated interconnect circuitry to have integrated transmit and receive elements. In these probes, the transmit beamforming is limited by the capabilities of the on-chip high voltage pulsers. Additionally, using these probes, volume imaging at a high volume rate is achievable only with limited viewing angle and compromised image resolution. To produce volumes of larger viewing angle with high resolution, volume stitching using ECG is used, which reduces the achievable volume rate and may introduce image artifacts with irregular heartbeats. Hence, a 3D TEE probe avoiding these challenges will certainly be very helpful for high frame rate volumetric imaging.

As an alternative to the commercially available 3D TEE probes, Oldelft Ultrasound (Delft, the Netherlands) has recently developed a prototype matrix probe for 3D TEE to facilitate full volume imaging with good resolution at a sufficient frame rate (>20 Hz) for visualizing the motion of the 3D structures of the heart. To reduce the complexity and power dissipation of the Application-Specific

Integrated Circuit (ASIC) design, the prototype matrix probe is divided into separate transmit and receive arrays based on a split-array architecture (Blaak et al. 2011; Yu 2012). This split-array concept offers several advantages. The transmit elements are directly wired out to an external ultrasound system, thereby enabling the use of a compact low-voltage (1.8 V) 180 nm CMOS process for the ASIC, which is only connected to the receive elements. Moreover, the absence of in-probe transmit electronics reduces power dissipation and provides full flexibility in defining the transmit pulse shapes; on-chip high voltage pulsers for transmission mostly can only provide very simple pulse shapes. Additionally, for non-fundamental imaging techniques such as (sub)harmonic imaging, in the split-array architecture the transmit and receive arrays could be optimized separately; however, this was not realized in this prototype. To reduce the receive channel count, micro-beamforming (or sub-array beamforming) is performed by applying small analog delays before summing the received RF signals from the individual elements of each sub-array. These micro-beamformed RF receive signals are then transferred to the external ultrasound system (Blaak et al. 2011). The prototype transducer comprises a small rectangular transmit array at the distal end of the gastroscopic tube and a larger square receive array proximal to the transmit array (Figure 1). The PZT material is diced at an angle of 45° to the azimuth and elevation plane, which produces elements and sub-arrays rotated diagonally. This diagonal dicing reduces the overlap between transmit sidelobes and grating lobes and receive grating lobes for the separated transmitter-receiver layout, as will be shown in this study.

The specific transmitter-receiver layout will affect the image characteristics in several ways. First of all, the misalignment between the transmit and receive beams produced by the separated transmit and receive array will cause slightly tilted speckles and PSFs. Secondly, the combination of rectangular transmit aperture and square receive aperture will produce asymmetric PSFs (narrower in azimuth direction than in elevation direction). Finally, the diagonal dicing will produce transmit and receive grating lobes that are most prominent in the diagonal directions. Because of these effects, the appearance of the volume image might change based on the orientation of the probe with respect to the imaging object. For any 3D beamforming technique using the prototype probe, these effects on the image characteristics are expected, as they are caused by the intrinsic properties of the prototype. In this paper, we examine these effects and discuss the implications.

To solve the challenge of producing volume images at a sufficiently high frame rate (>20 Hz), parallel beamforming (Shattuck et al. 1984) can be used. The relatively wide transmit beams of the small transmit array illuminate a sector of 3D space, and by processing the received channel signals in parallel, several image lines can be reconstructed simultaneously. The PSF of the reconstructed beams is wider, because of the limited transmit focusing. Artifacts could also appear in the resulting images, in the form of sharp intensity changes between scanlines from neighboring transmissions (Hasegawa and Kanai 2011; Hergum et al. 2007; Tong et al. 2014). We refer to these artifacts as *crossover artifacts*. One solution to produce good quality volume images at high frame rate using parallel beamforming with a small transmit aperture has already been proposed by us in (Bera et al. 2016). In that study, we minimized the crossover artifacts by combining the overlapping sub-volumes corresponding to each transmission. We refer to this scheme as *angle-weighted sub-volume combination*. In the current study, the prototype matrix probe is very different from the transducer in (Bera et al. 2016). It has a different layout and aperture size of the transmit and receive arrays. Accordingly, the transmit and receive opening angles are also different. Additionally, this probe has a more flexible pre-steering capability. Hence, the 3D beamforming technique as proposed in (Bera et al. 2016) is adapted for the prototype matrix probe based on these characteristics.

In this paper, we describe a 3D beamforming scheme that can produce good quality volume images at an adequate frame rate (>20 Hz) with the prototype adult matrix probe. The scheme uses parallel beamforming. Due to the rectangular transmit aperture, the transmit beam has an elliptical cross-section and more receive scanlines are reconstructed in elevation direction than in the azimuth direction. Hence, to reconstruct a volume (of equal size in elevation and azimuth direction), less transmit beams are required in the elevation direction than in the azimuth direction. The receive sub-arrays are pre-steered to the same direction as the transmit direction, creating one transmit-receive event (tx-rx event). To produce a volume image at an adequate frame rate, a 3D beamforming scheme should use a minimal number of tx-rx events and also should be able to minimize the image artifacts introduced by the parallel beamforming. Therefore, to judiciously choose the tx-rx events scheme for high frame rate volume imaging, we first show the results from the acoustic characterization of the transmit array and the receive array. Based on these results, we propose a parallel beamforming scheme exploiting the capabilities of the prototype probe. The proposed 3D beamforming scheme requires 85 tx-rx events (17 directions in azimuth × 5 directions in elevation) to reconstruct a volume of 60°×60° field of view. To avoid the crossover artifacts in the final image, we use angle-weighted sub-volume combination in the proposed 3D beamforming scheme. In this paper, we validate and compare the performance of the 3D beamforming scheme in both simulation and experiment. Finally, we did an *in vivo* acquisition on a porcine heart to show the real-time 3D imaging capability of the prototype matrix probe.

MATERIALS AND METHODS

Description of the prototype matrix transducer

Figure 1 shows the probe and the layout of transmit and receive elements of the prototype matrix probe. The probe has an outer dimension of 15×11×35 mm^3 which is similar to other TEE probes. The acoustic aperture measures about 10 × 9 mm^2 overall and consists of 2176 individual PZT elements that operate at an ultrasound frequency of 5 MHz. The transducer array is fabricated by dicing a bulk piezo-electrical material (3265HD, CTS Corporation, Albuquerque, MN, USA) with a pitch of 181 µm and a dicing kerf width of 30 µm at an angle of 45°, which produces elements rotated by 45°. The PZT stack is mounted directly on top of a front-end ASIC using a PZT-on-CMOS integration scheme.

The rectangular transmit aperture area (5.76 × 0.90 mm^2) consists of 128 elements distributed in 6 rows at the distal side of the tube. The transmit elements are wired out directly to the external ultrasound scanner. The receive aperture (8.7 × 8.7 mm^2) comprises 2048 elements that are connected to the front-end ASIC for micro-beamforming. The receive elements are divided into 128 sub-arrays of 4×4 elements. Each sub-array has a micro-beamformer that applies analog delays to the signals from the individual elements before summing. Thus, the micro-beamformers reduce the number of required receive cables by a factor of 16, to 128. The analog delays for the elements in a sub-array can be programmed from 0 ns up to 550 ns in steps of 10 ns. Thus, an individual sub-array can be steered to any angle in the 3D hemisphere. A time gain compensation is configured as an attenuation of the signal from individual receive sub-arrays. The attenuation ranges from 16 dB to 0 dB. The attenuation is applied with time steps of 640 ns up to a maximum time of ~80 μs. The prototype is realized as a fully functional device, mounted in a standard gastroscopic tube with manipulation handle.

Setup for the acoustic characterization of the transmit and receive array

The prototype matrix probe was mounted on the rotating arm of a fully automated UMS3 Scanning Tank (Precision Acoustics, Dorchester, U.K.) as shown in Figure 2. To characterize the transmit array, a 0.2 mm needle hydrophone with integrated amplifier (Precision Acoustics, Dorchester, U.K.) was mounted on the XYZ stage of the water tank. The amplified hydrophone signals were captured using an oscilloscope (DSOX4054A, Keysight Technologies, Santa Clara, CA, USA). For the receive characterization, the needle hydrophone was replaced with an unfocused 0.5-inch transducer (V309, 5.8 MHz center frequency with 80% bandwidth, Olympus Scientific Solutions, Waltham, MA, USA). This external transducer was excited with an arbitrary waveform generator (AWG) (33250A, Keysight Technologies, Santa Clara, CA, USA). The transmit array and the receive array were evaluated separately. The transmit elements were connected to 128 channels of a Verasonics V1 ultrasound system (Verasonics Inc. Kirkland, Washington, USA). The ultrasound system was configured to generate pulses with a peak-to-peak of 15 V for all the transmit elements with appropriate time delays to steer the transmit beams. The micro-beamformed RF signals from the 128 receive sub-arrays were recorded simultaneously using the ultrasound system. In this paper, the set of output RF signals from all the sub-arrays from a single tx-rx event is referred to as a micro-beamformed RF (μBRF) dataset. The host PC was connected to the controller FPGA in the handle via an interface box. This interface box converts the PC's USB interface to the Serial Peripheral Interface (SPI) used by the FPGA and ASIC for control signals and pre-steering settings. The interface box was also connected to a power supply (± 5 V) for the controller FPGA and the ASIC.

Characterization of the transmit array

To characterize the transmit array, we measured the transmit beam profiles (both non-steered and steered), the transmit efficiency and the frequency response. Each of the 128 transmit elements was excited by a 5 MHz single cycle sinusoidal pulse from the ultrasound system, and the acoustic signal was recorded on-axis using the hydrophone placed at a distance of 20 mm from the surface of the transmit array. The 2D beam profiles of the transmit array were measured on a C-plane at 20 mm by the hydrophone mounted on the XYZ stage of the water tank. To measure the transmit beam profiles, the transmit array was steered to (0°,0°), (30°,0°) and (30°,30°) [expressed as (θ, φ), where θ is the angle in azimuth direction and φ is the angle in elevation direction]. For every steering angle, the peak-to-peak pressure was recorded by the hydrophone at each scanning position. The transmit efficiency and the frequency response of the transmit array were computed based on the highest peak-to-peak acoustic pressure signal received by the hydrophone for (0°,0°) steering.

Characterization of the receive array

To characterize the receive array, we measured the receive sensitivity variation among the sub-arrays, the receive beam profiles of the sub-arrays (for several pre-steering angles) and the receive efficiency. For the receive characterization, the prototype was mounted on the rotational arm of the water tank (as shown in Figure 2), in front of the unfocused single-element ultrasound transducer used as a source. The source transducer was excited with a single-cycle 5 MHz sinusoidal pulse, generated from the AWG. To assure a plane-wave excitation, the source transducer was placed at a sufficient distance from the receive array. In order to mimic the echoes from different angles, the prototype was mechanically rotated in azimuth direction from -60° to +60° in steps of 2°. The received acoustic signals from the 128 sub-arrays were captured at every angle using the ultrasound system, and the beam profiles of the individual sub-arrays were computed. For each pre-steering angle, the beam profile of the entire receive array was calculated by delaying the signals from all the sub-arrays before summing. To measure the receive beam profiles, the receive sub-arrays were pre-steered to 5 angles (-40° to +40° with steps of 20°) in the azimuth direction. To measure the sensitivity of the receive sub-groups and receive efficiency of a sub-group we used the data acquired for (0°,0°) pre-steering. The beam profiles in elevation direction are considered to be the same because of the symmetry of the receive array.

Parallel beamforming with micro-beamformed datasets

We selected the proposed beamforming scheme based on three arguments. First, the transmit beams should have sufficient overlap to insonify the entire 60°×60° viewing angle with sufficient amplitude and allow the angle-weighted combination of several sub-volumes to avoid the crossover artifacts. The transmit aperture produces a beam with a theoretical -6 dB opening angle of ~4° in azimuth direction and ~27° in elevation direction. The pre-steered sub-array receive beams have a theoretical -6 dB opening angle of ~20° in both azimuth and elevation directions. Hence, to cover the entire viewing angle, at least fifteen transmissions in azimuth direction and at least three transmissions in elevation direction are required, but a larger number is beneficial for sufficient transmit/receive amplitude. This should be faced with a trade-off to the second argument, which is that less transmissions means higher volumetric frame rate. With a final aim of roughly 50 volumes per second and a maximum of 5000 tx-rx events per second, this would lead to a maximum of 100 transmissions per volume. The third argument is that the extremities of the volume may be allowed to have a little lower quality in terms of SNR/signal amplitude, contrast, and resolution, as the natural emphasis will, in general, be on the central part of the imaged volume. Based on these three arguments, we propose a 3D parallel beamforming scheme which is steered to 17×5 angles, distributed over a combination of azimuth directions of -24°:3°:24° and elevation directions of -20°:10°:20°. The angular weights for the combination of the sub-volume receive scanlines were chosen as coefficients from a Hanning window, based on the angular distance of the scanlines from the tx-rx direction.

To summarize, the following steps were performed to produce the final volume with 121×121 scanlines:

Step 1: A wide transmit beam was steered to one angle in 3D space from the set of 17×5 predefined angles.

Step 2: The receive sub-arrays were pre-steered to the same transmit angle and the 128-channel µBRF dataset was acquired.

Step 3: To produce a sub-volume with 3D beamformed scanlines (~13×41 lines near the transmit/receive angle), Delay-and-Sum parallel beamforming using dynamic receive focusing was applied to the µBRF dataset of the tx-rx event.

Repeat step 1-3 for all 85 angles (85 tx-rx events).

Step 4: The scanlines from the overlapping sub-volumes of neighboring tx-rx events were linearly combined to produce the final scanlines by angle-weighting.

Ideal delay-and-sum (DAS) beamforming

To compare the image quality of the volumes produced by the proposed beamforming scheme with ideal volume images, in simulation we produced volume images with a slow but ideal DAS 3D beamforming. To achieve the best image quality with the given transmit and receive aperture, for each of the 61×61 scanlines in the volume one tx-rx event was used, where the transmit steering and receive pre-steering were set to the exact direction of the scanlines. Comparison with the acquired volumes generated with the proposed beamforming scheme was done after scan conversion.

Imaging simulations and experimental setup

Simulations

The transducer parameters used in simulations using FieldII (Jensen 1996) are shown in Table 1. The impulse response of the transducer was modelled as a 4 cycle sinusoid (46% fractional bandwidth). The parameters used for the acoustic characterization and the 3D imaging in both simulation and experiment are shown in Table 2 and Table 3, respectively. In Table 3, the parameters used for the simulation were chosen to produce volume images using ideal DAS beamforming.

Experimental setup for 3D imaging

To investigate the performance of the proposed 3D beamforming scheme, the micro-beamformed datasets were acquired with the prototype probe connected to the Verasonics ultrasound system. The experiment was done with a commercial tissue phantom (multi-purpose ultrasound phantom 040-GSE, CIRS, Virginia, USA) (CIRS) with 0.5 dB/MHz/cm attenuation and containing wire targets and hyperechoic and anechoic cysts. We performed imaging experiments for three orientations of the prototype matrix probe on the phantom. Figure 3 shows one of the three setups, where the central azimuth plane of the prototype matrix probe was positioned perpendicular to the wires and to the axis of the hyperechoic cylinder in the phantom. To show the effects of the rectangular transmit aperture and the diagonal dicing, the other two imaging orientations were achieved by rotating the probe to ~45° and ~90° with respect to the position shown in Figure 3. The beam was produced by placing a virtual source behind the transducer plane at a distance of -100 mm, effectively producing a plane steered wave. The pulse repetition time was 0.200 ms (85×0.2 ms = 17 ms per volume recording) and the wait time between frame recordings was 28.5 ms, needed for reliable synchronization in the current setup. This synchronization issue was caused by a slow data connection for control signals, not by the prototype itself. The effective volume rate was therefore not 59 Hz but 22 Hz. The µBRF dataset received by the sub-arrays was acquired with fixed gain settings in the ultrasound system, digitized at 20 MHz sampling rate, and stored for post-processing. Before 3D beamforming, the μBRF datasets were filtered with a band-pass filter of 100% bandwidth with 5 MHz center frequency.

In-vivo porcine experiment

For the *in vivo* experiment, imaging was performed in an experimental intervention in an adult pig of 50 kg (Erasmus MC Animal Experiments Committee protocol #109-14-12). The anesthetized pig was laid down in the supine position and ventilated. Since the pig esophagus is oriented differently than in humans with respect to the heart, direct transesophageal imaging in pigs is cumbersome. Therefore, the imaging was performed through a hole in the chest wall and the diaphragm. The probe head was positioned directly next to the heart to get a view similar to the standard short axis (SAX) view. Afterwards, imaging was done in an open-chest setting, with the probe head on the anterior heart wall. The acquisitions were carried out by an experienced cardiologist. The imaging settings were kept the same as in the phantom imaging. A 2.5-s acquisition was recorded up to ~90 mm depth in each view using the same tx-rx scheme as the phantom imaging, but using a transmit excitation pulse with a peak-to-peak of 40 V. We oriented the probe to capture the following 3D structures of the heart: the aortic valve, the mitral valve, the tricuspid valve and the interventricular septum.

Measures of image quality

A. Point Spread Function (PSF): To estimate the performance of the proposed 3D beamforming scheme, the widths of PSFs in the azimuth and elevation directions were measured at -6 dB.

B. Contrast to Noise Ratio (CNR) and Contrast Ratio (CR): The performance of the proposed 3D beamforming scheme on the cysts was estimated using the CNR and CR as defined in (Van Wijk and Thijssen 2002), which are given by

$$CNR = \frac{\mu_s - \mu_c}{\sqrt{(\sigma_s^2 + \sigma_c^2)/2}} \tag{2}$$

$$CR = \frac{\mu_s - \mu_c}{(\mu_s + \mu_c)/2} \tag{3}$$

where μ_s and μ_c are the mean amplitudes of a speckle region and a cyst region, respectively, and σ_s^2 and σ_c^2 represent the variances of the speckle and cyst region.

RESULTS

Transmit and receive beam characteristics

Transmit characteristics

Figure 4 shows the measured beam profiles of a single transmit element in azimuth and elevation direction at 20 mm depth. From this figure, it can be observed that the full width half maximum (FWHM) beamwidth of a single element is ~50°. Figure 5 shows the time trace at 20 mm for the position of maximum peak-to-peak pressure signal received by the hydrophone for the same transmit element when excited using a pulse with a peak-to-peak of 15 V. For this transmit element, the transmit efficiency (calculated using the peak-to-peak pressure of 10 kPa) was 0.67 kPa/V.

Figure 6 shows the measured and simulated beam profiles of the entire transmit array when steered to (0°,0°). As expected, due to the rectangular transmit aperture the FWHM in azimuth direction was narrower than the elevation direction. The FWHM in azimuth direction was ~3.6° in the measurement compared with ~4° in the simulation. In elevation direction, the FWHM was ~20° in the measurement compared to ~28° in the simulation. The measured beam profile in elevation direction was narrower than in the simulation. One of the reasons could be a misalignment of the measurement plane with respect to the transmit beam; as will be visible in Figure **7**(a) below, a small misalignment in the azimuth plane will result in a large mismatch in the elevation beam profile. This effect is much less prominent for the azimuth beam profile.

Figure 7 shows the measured and simulated 2D beam profiles on a C-plane at 20 mm for three steering angles [(0°,0°), (30°,0°) and (30°,30°)]. From this figure, it can be observed that the measured beam profiles were very similar to the simulated beam profiles, although in the measured beam profiles shown in Figure 7(b) and (c) the clutter level was 10 dB higher than in simulation. In Figure 7(c), for steering to (30°,30°), the transmit grating lobes appeared in the diagonal direction at (-35°,-35°). This is due to the element pitch of 181μm being larger than λ/2 for the transmit array in the diagonal direction.

Figure 8 shows the time trace of the transmit array at 20 mm for the position of the maximum peak-to-peak pressure signal received by the hydrophone and the corresponding frequency response of the transmit array. From this figure, it can be observed that for an excitation using a single sinusoid with a peak-to-peak of 15 V, the maximum peak-to-peak pressure generated by the transmit array was 1.2 MPa. The frequency response shows that the transmit array has a center frequency of 4.8 MHz with 50% bandwidth.

Receive characteristics

The receive beam profiles of two sub-arrays (chosen randomly) for 5 pre-steering angles (-40°:20°:40° in azimuth direction) are shown in Figure 9. It can be observed that the directivity patterns of the sub-arrays are nearly uniform for all the pre-steering directions. The received intensity drops only by ~3 dB from pre-steering (0°,0°) to pre-steering (0°,40°). This suggests that even with micro-beamforming it is possible to achieve a wide directivity pattern similar to a single element. Additionally, there is no significant difference in the directivity patterns of individual sub-arrays. The FWHMs of the received beams for the sub-arrays are ~30° compared to the theoretical value of ~20°. The theoretical FWHM was computed considering a circular transducer with a diameter (1.024 μm) same as the width of a sub-array in elevation/azimuth direction.

The receive sensitivity variation among the sub-arrays for pre-steering to (0°,0°) is shown in Figure 10. From this figure, it can be observed that the sub-arrays have almost a uniform receive sensitivity (variation of ~2 dB).

To measure the beam profiles of the complete receive array, for each pre-steering angle, delays were computed based on the center position of each sub-array for that steering angle and applied to the RF signals. Figure 11 depicts the beam profiles of the receive array for 5 pre-steering angles (-40°:20°:40°). From this figure, it can be observed that by applying different pre-steering to the sub-arrays, it is possible to achieve good sensitivity over a wide sector in 3D space (opening angle at least 60 degrees). The mean FWHM of the received beam for the receive array was ~2.5° which is similar to the theoretical value of ~2.3°. The peaks of the beam profiles for all the 5 pre-steerings were shifted by -2°. This may have been caused by a misalignment in the measurement setup. Based on the symmetry of the receive array, the beam profiles in the elevation direction are expected to be the same as in the azimuth direction.

Effect of the diagonal dicing and the transmit-receive layout on the PSF

To estimate the effect of the diagonal dicing we have simulated the pulse-echo beam profile for a (30°,30°) steering angle. In Figure 12 we show the 2D beam profiles of the transmit array, receive array and the pulse-echo on a C-plane at 20 mm for a steering angle of (30°,30°). From this figure, it can be concluded that the grating lobes of the transmit and receive array are non-overlapping. Due to the diagonal dicing and the eccentric transmit array, the receive grating lobes do not overlap with the wide transmit sidelobes in the elevation direction. Thus, in the pulse-echo beam profile, the off-axis energies are greatly suppressed. The FWHMs in azimuth and elevation directions were 2.3° and 2.6°, respectively. Due to the wider transmit beam profile in elevation direction, the FWHM in elevation direction was slightly wider compared to the azimuth direction.

To further estimate the effect of the separated transmit and receive array, we simulated imaging of 18 point scatterers placed at different depths (starting from 2.5 mm to 50 mm with an interval of 2.5 mm) in the central region. Figure 13 shows the central elevation and central azimuth planes of the volume with the point scatterers. The dynamic range of the displayed images is 40 dB. From this figure, it can be observed that the PSFs at depths <20 mm are tilted. This is caused by the non-aligned transmit and receive beams close to the transducer. However, this effect was not at all visible at larger depths.

3D beamforming

Volume images of phantom

Figure 14 shows the three volume rendered images of the phantom produced using the proposed 3D beamforming scheme for the three orientations of the prototype probe: azimuth plane perpendicular to the wires and hyperechoic cyst (shown in Figure 3) and by rotating the probe to ~45° and ~90° with respect to the first position. In this figure, the structures (wires at different depths and hyperechoic cylinder) of the CIRS phantom are clearly visible in all the three volumes. The color in the rendered volume changes from white to blue as we move from the front to the back. Because of the specular reflections from the wires, only the reflections perpendicular to the wires are visible. For the three probe orientations, the rendered volumes were rotated to get the same view of the hyperechoic cylinder and the wires. From this figure, it can be observed that the reflection patterns of these three images are slightly different. As expected, the orientation of the tilted PSFs at the shallower depths changes depending on the orientation of the probe. However, the hyperechoic cylinder and the wires at a larger depth appear very similar in all three images. This suggests that the separated transmit-receive array and diagonal dicing have no severe effect on the volume images.

Figure 15 shows the azimuth plane at $y = 4$ mm and the elevation planes at $x = 12$ mm and $x = 3$ mm of the volume produced for the imaging setup as depicted in Figure 3. The azimuth plane is almost perpendicular to the wires and to the axis of the hyperechoic cylinder. As a result, in Figure 15(a) the wires appear as points, and the hyperechoic cylinder as a hyperechoic circular region (cross-sectional area) at $z = 30$ mm. The hyperechoic cylinder is clearly visible in the elevation plane shown in Figure 15(b). Although the wires are parallel to the elevation plane shown in Figure 15(c), they do not appear as elongated intensity lines. In Figure 15(c), the wires at 10 mm and 20 mm appear as tilted similar to the simulation results in Figure 13. This is caused by the eccentric transmit array.

Width of PSF in azimuth and elevation directions

Figure 16 shows the width of the PSF at different depths in azimuth and elevation direction for the simulation and experiment. In both elevation and azimuth directions, the width of the PSF at -6 dB averaged over the different depths is a factor 1.4 of the simulation with the slow but ideal DAS beamforming. Both in experiment and simulation, the widths of PSFs in elevation direction were slightly (average ~8%) broader than in the azimuth direction. This was expected because of the smaller aperture size in the elevation direction of the rectangular transmit array (as also visible in Figure 12).

3D imaging of an anechoic stepped cylinder in the CIRS phantom

To estimate the imaging performance of the prototype probe on cysts, the anechoic stepped of the CIRS phantom was imaged. Figure 17 shows the 2D azimuth and elevation planes of the volume produced by the proposed 3D beamforming scheme, where the azimuth plane was perpendicular to the anechoic cylinder axis. In Figure 17(a), the anechoic cylinder appears at 20 mm as a circular cystic region and in Figure 17(b) the stepped radius of the cylinder is visible. To quantify the imaging performance of the proposed beamforming scheme on the cyst, we computed CNR and CR as defined in Eq. (2) and (3) for both simulation and experiment, using a circular window of radius 1.5 mm [shown in Figure 17(a)]. The CNR values for the simulation and experiment were 2.8 and 2.5, respectively. The CR value for the simulation was 1.6 compared with 1.18 in the experiment. From these CNR and CR values, we can conclude that the image quality of the volume produced in the experiment is similar to the simulation with ideal DAS beamforming.

In vivo imaging of the porcine heart

Figure 18 shows three 2D slices and a volume rendered 2D image from the volume reconstructed using the *in vivo* acquisition of the healthy adult pig focused on mitral valve (with the SAX-like view). The reconstructed volume was sliced at specific oblique planes for better visualization of the mitral valve. In this figure, several structures of the heart such as the left atria (LA) and the right atria (RA), the left ventricle (LV) and the right ventricle (RV) are visible along with the interventricular septum (IVS). The tricuspid valve (TV) and the two leaflets of the mitral valve are prominently visible in Figure 18B. For better visualization of the 3D image sequences, two videos are available as multimedia attachments (Video 1, Video 2). Both the videos were generated from the 3D acquisition through the diaphragm at a volume rate of 22 Hz during 2-3 heart cycles (2.5 secs). The first video is focused on the mitral valve motion and the second video is focused on the aortic valve motion. These results confirm that with the prototype matrix probe using the proposed 3D beamforming scheme we can produce 3D images with good image quality at a volume rate of >20 Hz. The acquisition scheme was using 85 tx-rx events per volume, resulting in a minimal volume acquisition time of 85×200 μs for a depth of up to 15 cm, which corresponds to a theoretical volume rate of 59 Hz. The achieved acquisition rate of 22 Hz was purely limited by the USB data communication issues of the used data acquisition setup, not by the hardware of the prototype probe.

DISCUSSION

In this paper, we presented 3D beamforming using a prototype TEE matrix probe featuring a separated transmit-receive aperture and diagonal element dicing. To obtain acceptable image quality at a high volume rate (>20 Hz), a previously introduced 3D parallel beamforming method (Bera et al. 2016) was extended in this paper to utilize the capabilities of the prototype matrix probe. The proposed 3D beamforming scheme, based on the acoustic characteristics of the probe, was validated using Field II simulations, phantom experiments and an *in vivo* experiment; the results established the capabilities of the prototype matrix transducer and the performance of the proposed 3D beamforming.

The unique architecture of the separated transmit-receive arrays of the prototype matrix probe allows good image quality and high volume rate in combination with a low electric power budget. By separating high-voltage transmit circuitry from the low-voltage receive electronics, it is possible to use low-power, higher-density CMOS technology for the implementation of the ASIC. The split-array architecture also eliminates the need for transmit/receive switches to separate the high voltage transmit signals from the low voltage receive electronics. Moreover, directly wiring out all transmit elements eliminates the need for any transmit electronics in the probe head, which further reduces power dissipation and provides freedom in choosing any complex arbitrary transmit waveforms for transmission. The possible disadvantages of this split-array architecture were proven to be very limited. The limited transmit aperture was able to produce sufficient pressure for adequate imaging. The asymmetric aperture resulted in slightly wider PSFs in elevation direction, but in the *in vivo* tests, this was not noticeable. The eccentric transmit aperture causes a tilt in the elevational PSF and speckle pattern, only observed at shallow depth (< 20 mm). This is not important for diagnostics, and at larger depth, this effect was not at all visible. Thus, it can be concluded that the separated transmitter-receiver design can be advantageous to the transducer characteristics without significant degradation in image quality and speckle pattern beyond depths of a few centimeters.

One aspect of the separated architecture that could be explored further is the possibility of optimizing the transmit array and receive arrays separately for a specific imaging technique. For example, for second harmonic or subharmonic imaging, the center frequency and pitch of the transmit array could be chosen differently from the receive array. This possibility has not yet been explored in this prototype.

The diagonal dicing proves to be very beneficial for the performance. The small rectangular transmit aperture produces wide transmit beams with higher sidelobes, especially in elevation direction. A full volume that would be created from only a single pre-steering setting would result in high grating lobe levels. This can be understood by considering the pre-beamformed RF signal of each sub-array as the output of a large, tilted element. Thus, the sub-arrays represent a large effective pitch (> 2λ) resulting in strong receive grating lobes. In a classical matrix transducer diced parallel to azimuth and elevation directions, these receive grating lobes would be oriented in azimuth and elevation direction. Consequently, the PSFs would severely degrade because of the overlap of these grating lobes with the wide transmit main beams and associated sidelobes. But by dicing the transducer at 45°, the receive grating lobes are positioned at 45° degrees, whereas the transmit sidelobes are still parallel to azimuth and elevation directions (see Figure 12). Thus, diagonal dicing for the prototype probe helps greatly in achieving a good PSF.

The results from the transmit acoustic characterization measurements were in good agreement with the simulated results. However, the measured transmit beam profiles in both azimuth and elevation directions were narrower than the simulation. One possible reason could be a misalignment of the scan plane with respect to the transmitted beam.

Based on the results in the *in vivo* imaging, the transmit efficiency of the prototype probe in terms of producing acoustic pressure is suitable for sufficient depth (~90 mm) of penetration. The central frequency of 4.8 MHz with 50% bandwidth is well suited for fundamental cardiac imaging.

To achieve sufficient overlap between the transmit beams, the number of tx-rx events (85, 17×5) for the proposed beamforming scheme was determined using the FWHMs of the transmit beam in azimuth and elevation directions. The FWHM of the measured receive beam profiles in the experiment matched closely with the simulated values. The sub-arrays showed a very uniform directivity pattern. The drop in intensity for pre-steering of the sub-arrays from 0° to 40° was only 3 dB. Thus, the combined directivity pattern of a single sub-array for different pre-steering angles is comparable with the directivity pattern of a single element. The sensitivity variation among the sub-arrays was very low, showing an excellent yield of the transducer manufacturing process.

The proposed 3D beamforming scheme with 85 tx-rx events has shown nice performance in terms of image quality and volume rate for both phantom and *in vivo* experiment. The widths of the PSFs in the phantom experiment were slightly wider than in the simulations. The CNR and CR values in the phantom experiment were lower than in the simulation. The image quality of the simulated volumes differs from the experimental volumes, as in simulation, we use ideal DAS beamforming to produce the best quality volume images. The results confirm the performance of the transducer and the 3D beamforming scheme in terms of the image quality. Due to the rectangular transmit array, the PSFs are asymmetric, they are narrow in azimuth direction and wide in elevation direction. However, it has not introduced any noticeable artifacts in the 3D images. In the real-time 3D volumes of the *in vivo* experiment, the 3D cardiac structures (chambers and valves) and their motion were clearly visible. These results show the capability of the prototype probe to produce image quality at an adequate volume rate of 22 Hz, making it suitable for real-time 3D imaging.

The 3D images produced by the proposed beamforming scheme have good image quality; the 85-tx-rx events scheme was chosen for good image quality at a sufficiently high frame rate. However, the volume acquisition rate of 22 Hz is lower than what is expected for the 85-tx-rx events scheme (59 Hz). This low volume acquisition rate is only determined by practicalities in the triggering and the communication between the host computer and the probe, which would have been overcome if the probe would have been addressed in its native SPI communication protocol. The probe itself poses no limitations to achieve a 59 Hz full volume acquisition rate for 85 tx-rx events (or higher if depth is limited). Much higher volume rates can be easily achieved by employing broader and lesser transmissions per volume, presumably at the cost of some image quality. In the future, we will explore such high frame rate volumetric imaging for advanced applications such as 3D particle image velocimetry or speckle tracking. Considering the

FWHMs (~30°) and the intensity profiles of the sub-arrays for different pre-steering, a volume of 90°×90° can be covered with only 9 tx-rx events. This will improve the volume rate at least by a factor of 9.

CONCLUSIONS

This paper presents 3D beamforming with a prototype volumetric TEE probe. The prototype probe has two key design features. First, it has separated, adjacent transmit and receive arrays with rectangular transmit aperture. Second, it consists of oblique square elements produced by diagonal dicing. The separated arrays, while offering important implementational advantages, did not lead to any noticeable artifacts at larger depth (> 20 mm). The diagonal dicing helped in improving the image quality by separating the diagonal receive grating lobes from the transmit sidelobes in elevation direction. The *in vivo* experiment showed that the prototype matrix probe can produce good quality 3D images with 60°×60° field of view at a volume rate of 22 Hz using the proposed 3D beamforming scheme with 85 transmit-receive events. The 22 Hz currently is limited by communication overhead, and can, in principle, be increased to 59 Hz with the proposed scheme. Adaptations to this scheme will allow much higher framerates. Hence, this matrix TEE probe with adapted 3D beamforming schemes will open up possibilities for new applications like 3D speckle tracking and particle image velocimetry in the heart.

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List of figures:

Figure 1: The prototype matrix TEE probe. (a) The probe with the rectangular tip, the probe handle and the gastroscopic tube, (b) a zoom-in of the tip, and (c) the effective aperture, with element-wise layout of the transmit and receive arrays. The black square indicates a 4×4 receive sub-array.

Figure 2: Measurement setup for acoustic characterization. SPI = Serial Peripheral Interface; USB = Universal Serial Bus.

Figure 3: Experimental imaging setup for the position of the probe having the central azimuth plane perpendicular to the structures in the phantom

Figure 4: Beam profiles of a single transmit element in (a) azimuth direction and (b) elevation direction.

Figure 5: The time trace at 20 mm for the position of the maximum peak-to-peak pressure signal received by the hydrophone for a single transmit element

Figure 6: Beam profiles of the transmit array in (a) azimuth direction and (b) elevation direction

Figure 7: Transmit beam profiles on a C-plane at 20 mm. Simulated (left side) and measured (right side) for steering to (a) (0°,0°), (b) (30°,0°), (c) (30°,30°)

Figure 8: (a) The time trace of the transmit array at 20 mm for the position of maximum peak-to-peak pressure signal received by the hydrophone, (b) the frequency response of the transmit array

Figure 9: Receive beam profiles of two sub-arrays for 5 pre-steering angles

Figure 10: Receive sensitivity variation among the sub-arrays for pre-steering to (0°,0°)

Figure 11: Beam profiles of the receive array for 5 pre-steering angles

Figure 12: 2D Beam profiles on the C-plane at z = 20 mm. (a) Transmit, (b) Receive and (c) Pulse-echo

Figure 13: 2D slices of the simulated point scatterers. (a) central elevation plane and (b) central azimuth plane.

Figure 14: Volume rendered 2D images of the CIRS phantom for the three orientations of the probe. The angles between the azimuth plane and the structures in the phantom were (a) ~90°, (b) ~45° and (c) ~0°. The volumes were reconstructed until 60 mm depth with 60°×60° field of view

Figure 15: 2D images in (a) Azimuth plane at y = 4 mm, Elevation plane at (a) x = 12 mm and (b) x = 3 mm

Figure 16: Width of PSF for simulation and measurement in azimuth and elevation directions.

Figure 17: 2D images of the anechoic stepped cylinder, (a) central azimuth plane and (b) central elevation plane

Figure 18: 3D in vivo imaging of the porcine heart. A. azimuth plane, B. elevation plane, C. C-plane sliced to focus on the mitral valve and D. volume rendered image in 2D

List of videos:

Video 1: Real-time volumetric image of the porcine heart focused on the mitral valve

Video 2: Real-time volumetric image of the porcine heart focused on the aortic valve

Table 1: Transducer simulation parameters

	Parameters	Value
Transducer	Center frequency	5 MHz
	Number of transmit elements	128
	Number of receive elements	2048
	Diagonal pitch	181 μm
	Number of receive sub-arrays	128
	Receive sub-array diagonal pitch	724 μm
Simulation parameters	Excitation pulse	Single sinusoid with a Hamming window
	Apodization	Box-car
	Sampling frequency	100 MHz

Table 2: Parameters for acoustic characterization

Transmit Characterization	Transmit focus	20 mm
	Transmit steering angles	(0°,0°), (30°,0°) and (30°,30°)
Receive Characterization	Receive pre-steering angles	[-40°: 20°: +40°] (-40° to +40° with steps of 20°) in azimuth direction

1

Table 3: Imaging parameters

	Parameters	Value
For simulation	Transmit steering angles	[-30°: 1°: +30°] in both azimuth and elevation direction
	Receive pre-steering angles	61×61= 3721 angles co-aligned with the transmit directions
	Field of view	60°×60°
	Number of receive scanlines	61×61
	Transmit focus	-100 mm
	Receive focusing	Dynamic receive focus
For experiment	Transmit steering angles	[-24°: 3°: +24°] in azimuth direction in combination with [-20°: 10°: +20°] in elevation direction
	Receive pre-steering angles	85 angles co-aligned with the transmit directions
	Field of view	60°×60°
	Number of receive scanlines	121×121
	Transmit focus	-100 mm