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THREE-DIMENSIONAL REAL-TIME SYNTHETIC APERTURE IMAGING USING A ROTATING PHASED ARRAY TRANSDUCER

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Abstract - Current 3D real-time imaging is done either with sparse 2D arrays, or with mechanically moved phased arrays. The former results in a poor resolution and contrast due to a limited amount of elements. The latter has the disadvantage of low frame rates due to the sequential acquisition of the volume line-by-line and plane-by-plane.

This paper describes an approach which combines mechanically moved phased array with synthetic transmit aperture imaging, resulting in high volume acquisition rates without a trade-off in image quality.

The scan method uses a conventional fully populated 64 element phased array, which is rotated over the volume of interest. The data is acquired using coded signals and synthetic transmit aperture imaging. Only one group of elements transmits at a time. The delays are set such as to form a cylindrical wave. The back-scattered signal carries information not only from the plane located directly below the transducer, but also from neighboring planes.

A complete dataset for all elements for the whole rotation is acquired and stored. The volume is then focused from this complete data set in order to obtain dynamic transmit and receive focusing in all directions.

I INTRODUCTION

Currently 3D volumes are scanned sequentially - line-by-line and plane-by-plane. This technique uses conventional linear/phased arrays for scanning each of the planes. After a plane is scanned, the transducer is moved to a new position to scan a new plane. The motion can be either a linear translation or rotation. This is very slow, and can be used only to acquire still volumetric images. The typical scan depth for cardiac imaging is 15 cm. Assuming a speed of sound $c = 1540$ m/s, this gives a pulse repetition frequency of $f_{prf} = 5000$ Hz. Having 64 planes and 64 lines per plane results in 1 volume per second. Wide beam transmission and parallel beamforming in receive, known also as "exploso-scan" increases the speed of acquisition [1]. The exploso-scan has been used with 2D phased arrays [2]. The parallel processing of 16:1 (16 receive beams within one transmitted) resulted in more than 12 B-mode volumes per second. Due to the a wide beam in transmit, the spatial and contrast resolution of the images is worse than the one achieved by modern ultrasound scanners.

Lockwood et al. [3] suggested the use of a linear array transducer combined with synthetic *transmit* aperture (STA) focusing. The planes in the volume are scanned one-by-one. Within each plane the image is created by sparse STA focusing. The idea is that a plane can be reconstructed by using only 3 to 5 emissions, giving roughly 10 volumes per second. The drawback of the approach is that due to the low number of transmissions the images have high-sidelobes. The resolution in the elevation direction of the transducer is determined by the physical size of the elements, and is typically 5 to 6 times worse than for the elevation plane [4].

In this paper we suggest to use a rotating phased array to scan the volume. The rotation is continuous (no two emissions are done in the same plane). The motion of the transducer is used to synthesize both the transmitting and receiving apertures.

The scanning technique, the beamformation procedure, and the implementation details are described in Sections II and III. The results of scanning a phantom are presented in Section IV. Finally, conclusions are drawn in Section V.

II SCANNING TECHNIQUE

Figure 1 illustrates the scanning process. The scanned volume is conical. The top of the cone is at the center of the rotating array. The array rotates at a constant speed. The transmissions are done using a group of elements, whose delays are set in such a way as to emulate the radiation pattern of a single element. The reception is performed using all of the transducer elements. It is assumed that the rotation of the transducer is negligible for the time of one pulse-echo event.

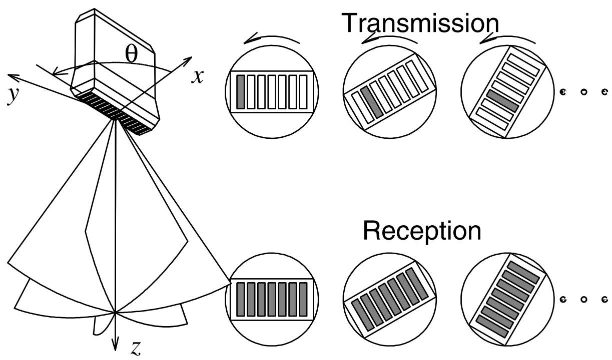

Figure 1: The scanned volume has a conical shape. The transducer rotates at a constant speed. The transmissions are done with a single element. The reception is done with all transducer element. The active elements are drawn in gray.

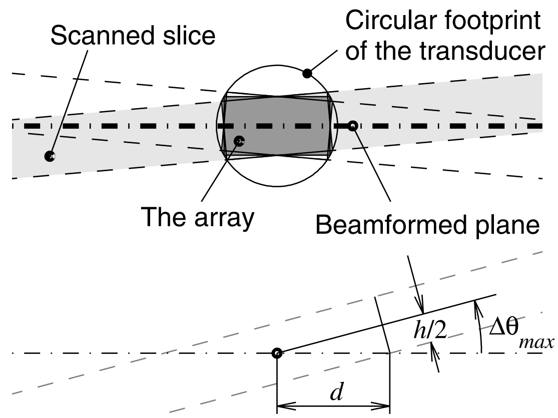

Figure 2: The received by the array echo carries information from a thick slice. A plane from the volume is covered by several such slices.

The volume is described in spherical coordinates r, θ, ϕ, where r is the distance to the origin of the coordinate system, θ is the angle of rotation in the $(x-y)$ plane, and ϕ is the scan angle measured from the plane $(x-y)$ (see Fig. 1). The relation between the spherical and Cartesian coordinates are given by:

$$\begin{aligned} x &= r\cos\phi\cos\theta \\ y &= r\cos\phi\sin\theta \\ z &= r\sin\phi. \end{aligned} \tag{1}$$

Each of the planes in the volume is characterized by a fixed angle θ_k ($0 \le \theta_k \le \pi$).

It is assumed that the echo received by the transducer carries information from a slice with a thickness equal to the height h of the transducer elements as shown in Fig. 2. The figure shows three positions of the transducer, the regions (slices) from which the information was gathered, and the plane to be reconstructed. The plane is covered by several slices, scanned at different rotation angles θ of the transducer. The maximum and minimum deviation of the angles θ from θ_k is dependent on the distance $d = \sqrt{x^2+y^2} = r\cos\phi$ from the z axis in the beamformed plane (see Fig. 2). For the conical scan format the relation between the scan depth z and the lateral distance d are given by:

$$\frac{z}{d} = \tan\phi. \tag{2}$$

The maximum angle $\Delta\theta_{max} = \theta - \theta_k$ is given by:

$$\Delta\theta_{max} = \arcsin\frac{2d}{h}, \tag{3}$$

where h is the height of the transducer elements, and d is the lateral distance in the beamformed plane. In other words, the line which coincides with the z axis is covered by all of the transducer positions. The further from the z axis a point lies, the smaller is the usable range of rotation angles θ.

Assuming that the transducer is rotating with a constant angular velocity ω, the angle of rotation of the transducer at transmission n is $\theta_n = n \cdot \omega / f_{prf}$. The beamformation of the points lying in the plane k defined by θ_k is expressed as:

$$\begin{aligned} H(r,\theta_k,\phi) &= \sum_{n\in(|\Delta\theta_n|<\Delta\theta_{max})} \sum_{j=0}^{N_{xdc}} s_{ij}(t_{ij}(r,\Delta\theta_n,\phi)), \\ \Delta\theta_n &= \theta_n - \theta_k, \\ t_{ij}(r,\Delta\theta_n,\phi) &= \frac{1}{c}|\vec{r}_f - \vec{r}_i| + |\vec{r}_j - \vec{r}_f|, \end{aligned} \tag{4}$$

$$\begin{aligned} \vec{r}_f &= (x_f, y_f, z_f), \\ x_f &= r_f\cos\phi\cos\theta_k, \\ y_f &= r_f\cos\phi\sin\theta_k, \\ z_f &= r_f\sin\phi, \end{aligned} \tag{5}$$

$$\begin{aligned} \vec{r}_i &= (x_i,y_i,z_i), \vec{r}_j = (x_j,y_j,z_j) \\ x_i &= p\left(i - \frac{N_{xdc}-1}{2}\right)\cos\Delta\theta_n \\ y_i &= p\left(i - \frac{N_{xdc}-1}{2}\right)\sin\Delta\theta_n \\ x_j &= p\left(j - \frac{N_{xdc}-1}{2}\right)\cos\Delta\theta_n \\ y_j &= p\left(j - \frac{N_{xdc}-1}{2}\right)\sin\Delta\theta_n \\ z_i &= z_j = 0; \\ i,j &\in [1, N_{xdc}], \end{aligned} \tag{6}$$

Name	Notation	Value	Unit
No. of channels	N_{xdc}	64	NA
No bits / sample	b	12	NA
Sampling freq.	f_s	40	MHz
Pulse repetition freq.	f_{prf}	5000	Hz
Transducer pitch	p	220	μm
Element height	h	3.8	mm
Acoustic elevation focus	F	85	mm
Center frequency	f_0	3.2	MHz
Fractional bandwidth	BW	> 60	%
Rotation amplitude	$\pm\max(\theta_k)$	± 135	deg.

Table 1: Parameters of the ultrasound system and the transducer array.

where θ_n is the rotation angle of the transducer at transmission n; i and j are the indexes of the transmitting and receiving elements, respectively; $t_{ij}(r, \Delta\theta_n, \phi)$ is the propagation time from the transmitting element i to a point located at $\vec{r}_f$ and back to the receiving element j; c is the speed of sound; p is the distance between the centers of two neighboring transducer elements, and N_{xdc} is the number of transducer elements.

III Implementation

The scanning technique was implemented on the experimental system RASMUS developed at the Center for Fast Ultrasound Imaging using a 64 elements phased array PA3.5/3D from Vermon [5, 6]. The system is capable of gathering raw sampled data from 64 channels. The 3D beam formation is carried out off-line. The off-line 3D visualization is done using software from GE Medical Systems Kretz Ultrasound.

The parameters of the system and the transducer are given in Table 1.

The transducer is alternately rotated clock- and counter clock-wise with an amplitude of $\pm 90°$. The targeted scan rate is 10 volumes per second, resulting in an average angular speed of $\omega = 10\pi$ rad/s.

The pulse repetition frequency is $f_{prf} = 5000$ Hz, giving 500 emissions per volume. Assuming that imaging all of the transducer elements are used in receive in STA, then the resolution is determined by the distance between the outermost transmitting elements. The number of transmissions used to beamform one point, on the other hand, determines the side- and grating-lobe levels. The angular speed ω is constant. From Fig. 2 it can be seen that the maximum angle of rotation θ_{max} which can be used for beamformation is smaller for points which are farther from the axis of rotation z. Fewer transmit/receive events are available for their beamformation. In order to achieve good resolution, a larger part of the transmitting aperture must be covered in short time, meaning that we have to transmit sparsely, say, with elements $i \in [1, 9, \cdots, N_{xdc}]$. The points lying closer to z can be beamformed using a larger number of transmissions (eventually all of the 500 emissions per volume). This requirements lead to the following sequence of the indexes of the transmit elements:

$$[1, 9, 17 \ldots 57,\ 2, 10, \ldots 58,\ \ \ldots,\ \ 8, 16, \ldots 64].$$

The points which lie in the outskirts of the planes are beamformed using only 8 emissions (the transducer moves of 2.4 degrees). The points which lie close to the z axis are beamformed using all of the 64 emissions. The image has higher contrast close to the center, but the resolution is maintained throughout the whole volume.

In order to achieve a good penetration depth, 13 elements were used in transmit, creating a cylindrical wave. The excitation was a linear frequency modulated pulse with duration of 20 μs and frequency range of $f \in [1, 6]$ MHz.

IV Results

The imaging abilities of a system can be described by the point spread function (PSF). The resolution of the system is determined by the lateral dimensions of the PSF, and the contrast by the level of the side- and grating-lobes.

To obtain the PSF of the imaging system a tissue mimicking phantom with point scatterers was scanned. The point scatterers are metal spheres, 0.3 mm in diameter, placed randomly in a low-scattering medium. The average backscattering coefficient of the surrounding medium is -30 dB, which prohibits the examination of the grating and side lobe energy.

Fig. 3 shows the image of the point scatterer located at $(x, y, z) = (-3.5, -4.5, 44.05)$. The top graph shows a C-scan, which is taken along a spherical surface whose center is placed at the geometric center of the transducer. The bottom graph shows a standard B-mode phased array image. The contours are drawn at levels of -3, -6, -12, and -18 dB. The dashed lines show the cross-sections between the two plots. It can be seen that the PSF is anisotropic - the 6 dB width of the PSF in the azimuth plane (along d) is 0.94 mm, and the 6 dB beam width in the elevation plane is 3.27 mm. The 6 dB beamwidth in the axial direction is 0.59 mm, which is 1.2λ.

Fig. 4 shows a screen-shot from the visualization program on the left, and a 55 dB B-mode image on the right. The scanned object is a tissue mimicking phantom with embedded cysts. The attenuation is 0.5 dB/(MHz·cm). The cysts are small cylinders 8, 4, 2 and 2 mm in diameter. It can be seen that the penetration depth of the system is up to a 120 mm. The dynamic range is limited by the range sidelobes of the FM pulse, which in this case are at -45 dB from the peak.

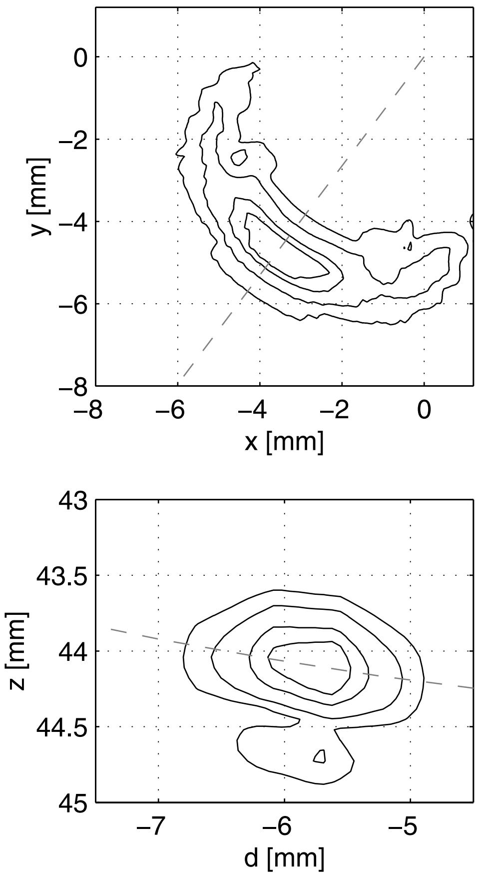

Figure 3: The point spread function. The top plot shows a C-scan (parallel to the surface of the transducer) and the bottom plot shows a B-mode scan.

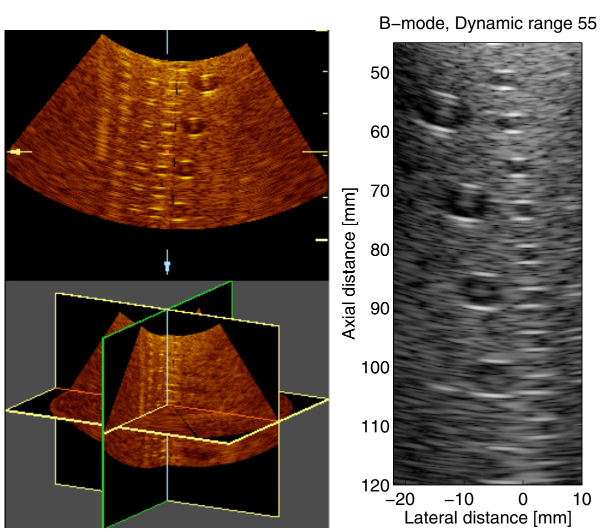

Figure 4: Scan of a cysts phantom. A screen-shot from the visualization program is given on the left.

V CONCLUSIONS

A method for real-time 3D synthetic aperture imaging using a rotating transducer is described in this paper. Ten volumes are scanned per second. Unlike other methods the motion of the transducer is uninterrupted. The beamformation of the points in the volume which are below the transducer are done using 64 emissions, and the outermost points only 8 emissions.

The images have spatial resolution comparable to that of a phased array. The contrast resolution is currently limited by the range side lobes of the non-optimized FM pulse.

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