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Synthetic Aperture High Quality B-mode Imaging with a Row-Column Array Compared to Linear Array Imaging

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Abstract—Row-column (RC) matrix probes can yield 3-D volumetric imaging using a number of receiving elements similar to traditional linear arrays for commercial scanners. Some doubts have, however, been raised on the B-mode image quality of RC probes. It is hypothesized that synthetic aperture (SA) RC imaging can yield a better volumetric resolution than commercial spatially translated linear arrays, and at the same time attain volume rates similar to frame rates for single slice linear array scanning. A commercial 6 MHz 128 × 128 elements RC array with λ pitch was used on a Verasonics Vantage 256 scanner. A SA sequence with 2 × 96 emissions on the rows and columns and reception on all 128 orthogonal elements were employed giving a 62.5 Hz volume rate for a pulse repetition frequency of 12 kHz. The resolution and contrast were compared to an optimized SA volume rate for a pulse repetition frequency of 12 kHz. Reception on all 128 orthogonal elements were employed giving a 62.5 Hz volume rate comparable to normal linear array imaging. RC arrays can yield higher quality B-mode images than linear arrays, and therefore the four rows of point scatterers could not be differentiated in the elevation direction due to the fixed elevation focus of the array, and therefore the four rows of point scatterers could not be differentiated in the elevation direction due to the fixed elevation focus of the array. Imaging was conducted on a 3-D printed point spread function (PSF) phantom with scattering voids in a 6 × 4 × 4 grid with a 2.05 mm spacing in all three directions. The exposed kidney of a Sprague-Dawley rat was also scanned in vivo with the RC probe. B-mode images of the 3-D printed PSF phantom were shown with a 40 dB dynamic range along with the linear array scans. An isotropic resolution of (1.05λ, 1.10λ, 0.62λ) = (x, y, z) was obtained for the row-column probe. The linear array probe had an elevation resolution determined by the geometric elevation focus of the array, and therefore the four rows of point scatterers could not be differentiated in the elevation direction due to the fixed elevation focus. The data rates were identical for the two arrays, but the RC array yielded an isotropic PSF with an improved contrast, and a 62.5 Hz volume rate comparable to normal linear array imaging. In vivo kidney images for the three orthogonal planes were shown with a 60 dB dynamic range demonstrating the isotropic speckle pattern in all three directions for all depths. The SA imaging RC sequence thus yielded a PSF independent of orientation and depth. Any slice plane in the volume therefore had a uniform speckle pattern, contrast, and resolution, demonstrating that RC arrays can yield higher quality B-mode images than linear arrays. The penetration depth of the probe and sequence was also measured to be 550λ corresponding to 141 mm.

I. INTRODUCTION

Three-dimensional ultrasound imaging has been devised and investigated since the pioneering work by von Ramm et al. [1, 2] from 1991. Matrix arrays have been employed, where the number of elements on each side is N for a total number of elements of N². The number N should be determined from the focusing demands on the array, where Full-Width-Half-Maximum resolution is FWHM = λF# = λD/W = λD/(NPp), where F# is F-number, λ wavelength, D imaging depth, W transducer width, and p_e probe element pitch. The point spread function (PSF) width is, thus, inversely proportional to the number of elements, which makes fabricating good resolution matrix arrays difficult due to the many elements. Further, a doubling of resolution necessitates quadrupling the total number of elements. Many approaches for making sparse arrays have been suggested [3–7], but they suffer from reduced contrast and reduced penetration depth due to the reduction in active transducer surface area.

A remedy for breaking the N² dependence on resolution is to employ row-column (RC) arrays, where only the rows or the columns are accessed at a time [8–20]. Doubling resolution for these arrays also doubles the number of elements, but quadruples the probe surface. Therefore very large arrays can be made, which can maintain a low F-number for large depths and further yield an excellent penetration depth [21]. The remaining question is then whether a sufficient contrast can be attained for the arrays. The hypothesis of this paper is that row-column arrays can yield better volumetric images than translated linear array probes.

A commercial 128 × 128 RC array from Vermon, (Tours, France) is compared to a state-of-the-art GE linear array probe through Field II simulations, and results similar to the simulation are shown for the RC array. In-vivo data from scanning a rat kidney is also presented. The synthetic aperture imaging scheme and the setup is presented in the next Section and the results in Section III. A discussion of the results and method along with conclusions are presented in the last Section.

II. DATA ACQUISITION AND BEAMFORMING

A commercial 6 MHz 128 × 128 elements RC probe from Vermon with a λ pitch is used for the imaging with a Verasonics Vantage 256 system. Synthetic aperture (SA) imaging
is used for acquiring the volumetric data. A number of virtual sources are evenly spread out over the aperture to ensure the minimum possible F-number, and both rows and columns are used alternatively in transmit to ensure the best possible contrast. Signals are received on all elements orthogonal to the transmit elements, so rows are used in receive when transmitting on columns. The virtual sources employ 32 elements in transmit and 96 + 96 transmissions (rows + columns) are made for a complete data set. Edge effects are avoided by having the virtual source aperture start at the edge of the probe - hence 128-32=96 transmissions.

The SA sequence was made using an F-number of -0.7 with a Hanning apodization in transmit. Beamforming was conducted with an F-number of 0.7 in receive, and a Hanning apodization for both transmit and receive using the GPU beamformer presented in [22].

Imaging was conducted on a 3-D printed point spread function (PSF) phantom [23] with scattering cavities in a $6 \times 4 \times 4$ grid with a 2.05 mm spacing in all three directions. The scatterers are 205 $\mu$m wide along the x–y axes, but only 80 $\mu$m in the z direction.

The RC volumetric images were compared to Field II [24, 25] simulated images obtained from a 256-elements GE L3-12D 6 MHz linear array probe with a geometric elevation focus at 22 mm giving an F-number of 4.4. A 12 emissions SA sequence was used with 32 active elements and an F-number of -0.7. The probe was translated in the elevation direction over the phantom in steps of 0.2 mm obtaining 100 images for the volume.

Finally, the exposed kidney of an anesthetized Sprague-Dawley rat was also scanned in vivo with the RC probe.

### III. RESULTS

An example of the PSF and image quality obtainable from a 6 MHz Vermon 128 $\times$ 128 elements RC array is shown in Fig. 1. Top row displays images from measurements on the phantom, and the corresponding Field II simulation is shown in the second row. An isotropic resolution of $(0.5\lambda, 1.10\lambda, 0.62\lambda) = (x, y, z)$ is attained for the measured data, and a similar performance is seen for the simulated data. The data are also compared to the linear array volume scan in the third row. The linear array probe has an elevation resolution determined by the geometric elevation focus F-number of 4.4, and the four rows of point scatterers can therefore not be differentiated due to the fixed elevation focus. The acquisition of the linear array data set necessitated $12 \times 100 = 1200$ emissions and mechanical translation, whereas the RC data set used 192 emissions corresponding to a normal focused linear array image. A volume rate of 62.5 Hz can therefore be attained down to a depth of 6 cm (pulse repetition frequency $f_{rep} = 12$ kHz).

Finally the bottom row in Fig. 1 shows in vivo images of a Sprague-Dawley rat kidney. The dynamic range is 60 dB and an isotropic speckle pattern is seen in all three imaging planes due to SA imaging, a constant F-number throughout the image, and the large size of the RC array.

Fig. 2 shows the attained penetration depth for the probe and sequence when measured on a tissue mimicking phantom with an attenuation of 0.5 dB/[MHz cm]. Twenty different images have been measured and the mean of these subtracted to yield the noise. The penetration depth is defined as the depth when the Signal-to-Noise Ratio (SNR) reaches zero, which is at $550\lambda$ at 6 MHz corresponding to 141 mm.

### IV. DISCUSSION AND CONCLUSION

Three-dimensional ultrasound is important for capturing all information in a scan. It offers the possibility of acquiring a full volume and then retrospectively inspect the volume as is normally done in radiology for CT, MR, and PET images. Such an imaging system could be devised for the abdomen for scanning the full liver or kidneys. The current resolution offered by matrix arrays is limited due to the small size of the probes because of restrictions in the number of elements. Resolution essentially scales with the squared number of elements, and trying to restrict the large number of elements by employing sparse arrays leads to a limited penetration depth and reduced contrast. A possible solution is to use a translating aperture for 3-D imaging, but it has been shown here that the elevation resolution is quite poor due to the fixed lens of the array yielding a non-isotropic point spread function. Dual stage focusing could be employed in the elevation direction, but this demands a very accurate registration of the probe position down to $\lambda/100$ to get a good resolution and contrast, which is difficult in vivo.

A viable alternative is to use RC arrays, where resolution scales linearly with the number of elements and the penetration depth is quite large. The large probe size makes it possible to maintain a constant F-number over a large range, as was demonstrated for the rat kidney example shown in Fig. 1. The speckle pattern has a constant size in all three orthogonal planes and for all depths, and any slice in the volume will therefore have the same speckle appearance. A matrix array with the same number of active elements would only have $16 \times 16$ elements, and the resolution would be $9.7\lambda$ at 20 mm for an elements pitch of $\lambda/2$, or 9 times worse than the RC array. Such a small size array would also have a very low penetration depth, whereas the RC array can penetrate down to $550\lambda$, when using spherical emissions with 32 elements for a very low MI and $I_{spta}$.

Currently, the SA sequence employed for the RC arrays contains 96 + 96 emissions for a high quality result in terms of resolution and contrast. The sequence length should be the subject of further optimization to increase the volume rate from 62 Hz to higher levels. A second challenge is the availability of RC arrays. Only a few arrays exist, and the pitch of the arrays is $\lambda$, which is not optimal for SA imaging as it restricts the acceptance angle for the transmitted and received signals. A third limitation is that RC cannot be steered outside the rectangular region below the active aperture. This restricts the available volume, and convex or lensed RC arrays should be developed [26].
Fig. 1. Point spread functions obtained from a 3-D printed phantom with isolated point targets using a 6 MHz Vermon 128 × 128 elements row-column array with \( \lambda \) pitch. The top row shows the measured images in the \( x - z \), \( y - z \) and \( y - x \) planes (left to right). The corresponding simulated data from the phantoms is shown in the next row. Simulated data for a linear array probe translated across the phantom is shown in the next row for a GE 6 MHz linear array using a SA sequence. The bottom row shows images of a rat kidney in all three planes obtained using the RC array.
A complete imaging system would also necessitate the estimation of blood velocity, tissue motion and should be applicable for non-linear and super resolution imaging. This has lately been demonstrated in [27], which showed examples of 3-D super resolution volumetric images [28] and full tensor velocity imaging in a volume [29]. RC arrays can, thus, acquire high quality data for both anatomic and functional imaging, and the volumetric data can be studied retrospectively in full 3-D.

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REFERENCES


