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3-D Spatial Compounding Using a Row-Column Array

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Abstract

2-D spatial compounding has long been investigated to reduce speckle in ultrasound images. To further reduce speckle, several 3-D spatial compounding studies using 1-D and 1.5 D arrays with mechanical translation and position tracking have been reported. However, the fixed elevational focus and mechanical translation can degrade image quality in elevation. Using 2-D arrays, a better elevational resolution can be achieved with electronic focusing. Furthermore, 2-D arrays can generate greater number of independent images than 1-D arrays, and the need for mechanical scanning is eliminated.

In this paper, we present our 3-D spatial compounding images of two gel-based contrast phantoms and one resolution phantom. These images were acquired using a prototype 4 cm × 4 cm ultrasonic row-column prototype 2-D array operating at 5 MHz. Compounding nine decorrelated volumes showed a speckle signal-to-noise ratio (SNR) improvement of 2.68. The average improvement of the lesion contrast-to-noise ratio (CNR) was 2.45. However, using a smaller aperture to generate these volumes worsened the lateral resolution as predicted by theory.

Keywords

2-D array; 3-D imaging; spatial compounding; ultrasound

I. Introduction

Speckle is a common artifact in coherent imaging modalities (including ultrasound) that degrades target contrast against the surrounding background. It rises from constructive and destructive interference of echoes backscattered from scatterers, giving ultrasound images a granular appearance. Several methods have been proposed to reduce speckle noise including filtering of B-mode images, frequency compounding, and spatial compounding.^{1–6}

Compounding involves averaging images of the same target that have uncorrelated speckle patterns. In spatial compounding, this is achieved by viewing the target from different angles using steering or aperture translation techniques.^{3,5–7} Besides speckle noise reduction, spatial compounding offers additional advantages such as reduced acoustic shadowing and suppressed refraction artifacts.⁶ Clinically, real-time spatial compounding has shown usefulness in the evaluation of breast lesions, peripheral blood vessels, and musculoskeletal injuries.^{8–10}

A number of studies investigated the decorrelation of speckle with lateral translation of the transducer to find the optimum displacement for 2-D compounding.^{2,3,11} The degree of statistical independence between the averaged images governs the success of spatial compounding. Speckle SNR, defined as the ratio of the mean to standard deviation of the detected data, increases by the square root of the number of independent compounded images. If the compounded images are partially correlated, the number of independent images N can be estimated using the following equation:^{3,7}

$$N = \frac{n^2}{n+2 \sum_{X=1}^n \sum_{Y=X+1}^n \rho_{X,Y}} \quad \text{Eq.(1)}$$

where n is the number of compounded images, $\rho_{X,Y}$ is the correlation coefficient between images X and Y . The correlation coefficient of two $m \times n$ pixel image regions, X and Y can be calculated using (Trahey and Smith et al. 1986):

$$\rho_{X,Y} = \frac{\sum_{o=1}^m \sum_{p=1}^n (X_{o,p} - \bar{X})(Y_{o,p} - \bar{Y})}{\sqrt{\sum_{o=1}^m \sum_{p=1}^n (X_{o,p} - \bar{X})^2 \sum_{o=1}^m \sum_{p=1}^n (Y_{o,p} - \bar{Y})^2}} \quad \text{Eq.(2)}$$

where $X_{o,p}$ is the mean envelope-detected echo magnitude value of image X at location o,p , and $\bar{X}$ and $\bar{Y}$ are the mean pixel values of image region X and Y , respectively.

3-D spatial compounding has the advantages of providing a greater number of independent images and reducing out-of-plane refraction artifacts.¹² To date, most 3-D spatial compounding research reported in the literature has used a 1-D linear or phased array with mechanical translation and position tracking.^{13–17} 1-D arrays lack the capability of electronic focusing in elevation which degrades the image quality away from the fixed focus. Additionally, mechanical translation with position tracking can introduce registration errors.¹² Krucker et al. used a 1.5-D transducer array and compounded 5 partially correlated volumes. They used mechanical steering in elevation and applied nonrigid registration to align the steered volumes. The reported CNR improvement for –12dB spherical cysts was 1.97.¹²

In this paper, we describe our 3-D ultrasound imaging system using a prototype 2-D array with row column addressing used for 3-D spatial compounding. In this work, we use subaperture translation to compound nine uncorrelated volumes. Using this technique eliminates the need for registration. We present images of gelatin phantoms with a cylindrical –11 dB cyst and a cylindrical +11 dB tumor and quantify the improvement in CNR and speckle SNR for the compounded images over standard B-mode images. We were able to achieve CNR improvements of 2.38 and 2.51 for the cyst and tumor 3-D images respectively. The speckle SNR improvements were 2.65 for the cyst image and 2.70 for the tumor image. Using five pairs of nylon wires embedded in a clear gelatin phantom, the decrease in lateral resolution caused by our spatial compounding technique was found to be a factor of 2.04.

II. Methods

A. System description

A prototype 2-D array with row-column addressing was used in our 3-D imaging experiments. This array has a 5.3 MHz center frequency, 53% –6 dB fractional bandwidth, $256 \times 256 = 65,536$ elements with $\lambda/2 = 150 \mu\text{m}$ pitch and row-column addressing.²¹ The design of this 2-D array utilizes a two-layer electrode pattern where the bottom layer consists of a series of vertical electrodes (Fig. 1A) and the top layer consists of a series of horizontal electrodes (Fig. 1B). In transmit, the vertical electrodes serve as the “ground” and the top electrodes serve as the “transmitters”.

In this example, we excite transmit channel D as indicated by the arrow to the right of channel D in Fig. 1B. This row of elements, shown in the gray shading in Fig. 1C, then emits a cylindrical wavefront into the field. In elevation, the wavefront appears omnidirectional since the aperture behaves like a single small element. In the azimuth direction, the emitted beam is a planar wavefront because all elements fire simultaneously, and the aperture behaves as a single long element. For receive mode, receive channels A–H are active and the desired receive column is selected (Fig. 1D). In receive mode, the individual elements along one column (gray shading) is used to record the echoes (Fig. 1F). With this design, transmit beamforming can be done in the vertical or elevational direction while receive beamforming can be done horizontally or azimuthally. A schematic illustrating this beamforming method is shown in Fig 2. Multiple rows can be used for elevational beamforming in transmit and multiple columns can be used for azimuth beamforming in receive. By stepping transmit subapertures across the array with multiple receive beams within the transmit beam, a 3-D rectilinear volume can be acquired.¹⁸

The 2-D array was interfaced with the Ultrasonix (Richmond, BC, Canada) Sonix RP ultrasound system using a custom printed circuit board. This system has a 40 MHz sampling frequency, 128 channels, and 32 analog to digital converters. It allows the user to acquire raw radio frequency (rf) data and gives the user control over transmit aperture size, transmitted power, transmitted frequency, receive aperture size, filtering, and time-gain compensation.

Using a synthetic aperture approach, two rows were excited at a time and signals from all 256 columns were acquired. Exciting two $\lambda/2$ pitch elements simultaneously makes the effective transmit pitch equal to λ . Excitation was done in this manner because transmit elements were accessed manually, which makes exciting the 256 transmit elements individually a labor intensive process. A two-cycle, 5 MHz transmit pulse was used. Two channels from the Sonix machine were used for transmission and manually multiplexed to access all 128 pairs of transmitters. Sixty-four system channels were used for receiving. A different set of 64 receive elements was connected to the ultrasound system until rf data from all 256 columns were collected. Data was sampled at 40 MHz, collected 100 times, and averaged to suppress random noise. Averaged elements rf data were filtered using a 64-tap FIR bandpass filter with frequency range of 3.75 – 6.25 MHz.

Off-line 3-D beamforming was applied to the rf element data using Matlab (Mathworks Inc., Natick, MA). Three 32-element adjacent transmit apertures with 1λ pitch and three 64-element adjacent receive apertures with $\lambda/2$ pitch were used to generate the spatially compounded 3-D image (Fig. 3). Using delay-and-sum beamforming, the nine combinations of these apertures were used to create nine decorrelated volumes. For the contrast phantoms images, dynamic focusing every 1 mm in transmit and receive was used along with an expanding aperture to keep the F number at 2. The resulting B-mode images were $38.4 \times 38.4 \times 41.4$ mm with $129 \times 129 = 16,641$ scan lines and 0.3 mm line spacing. The wires phantom image was beamformed using dynamic focusing every 0.5 mm in transmit and receive with 257×257 scan lines and 0.15 mm line spacing. This was done in order to get a more accurate estimation of the system resolution with and without spatial compounding.

The resulting nine volumes were then envelope detected and averaged on a linear scale. To quantify the improvements of 3-D spatial compounding, standard beamforming was applied to all data sets using a 64-element transmit aperture with 1λ pitch and 128-element receive aperture with $\lambda/2$ pitch.

B. Phantom fabrication

Three $70 \times 70 \times 70$ mm gel-based phantoms were used to test our 3-D spatial compounding technique. Two contrast phantoms were made to quantify the contrast enhancement in the

compounded images. One phantom included a 10 mm cylindrical -11 dB cyst and the other phantom included a +11 dB tumor. Both inclusions were located approximately at the center. The recipes for the inclusions were determined empirically where we used the background material recipe while varying the amount of graphite. To quantify the loss in resolution, a clear gel phantom with embedded wires was made. Since the 2-D array is symmetric in azimuth and elevation, the wires short axis can be used to reasonably evaluate the loss in resolution. The wires phantom included pairs of nylon wire targets with axial separations of 0.5, 1, 2, 3 and 4 mm located at the center of the phantom. The bottom wire in each pair was shifted laterally by 1 mm with respect to the top wire. The diameter of the nylon wire was 400 μm . Table 1 show the recipes used.¹⁹

Figure 4 shows images of the phantoms acquired using the Ultrasonix L14-5/38 linear array with synthetic transmit and receive focusing every 1mm. The contrast ratio calculated using these images are -10.50 dB for the cyst phantom and +10.94 dB for the tumor phantom. The cyst is not perfectly at the center in figure 4 because of variable mechanical compression of the phantom during the data acquisition process.

III. Results

Two sets of 3-D rf data of the cylindrical -11 dB cyst and +11 dB tumor phantoms were acquired using the 2-D array. 3-D CNR values were calculated by choosing equal sized volumes of the interior of the lesion and the background located at the same depth in the linear scale B-mode image. The following equation was used to evaluate CNR:⁵

$$CNR = \sqrt{\frac{(\bar{s}_t - \bar{s}_b)^2}{\text{var } s_t + \text{var } s_b}} \quad \text{Eq.(3)}$$

where s and $\text{var } s$ denote mean and variance, respectively. The subscripts t and b represent the target and background, respectively. A CNR kernel size of $110 \times 13 \times 250$ voxels (elevation $\times$ azimuth $\times$ axial) was used. The CNR of the uncompounded image (standard beamforming) and the image compounded from nine volumes for the cyst phantom are 1.16 and 2.76 respectively. For the tumor phantom, the CNR of the uncompounded image is 1.03 and the CNR of the image compounded from nine volumes is 2.59. This gives a CNR improvement of 2.38 for the cyst image and 2.51 for the tumor image.

Figure 5 shows the monotonic increase of mean CNR values as the number of compounded volumes increases. These values were obtained experimentally using Equation 3 and theoretically using Equation 1. For number of compounded volumes $N=9$, CNR was calculated for the only one possible combination. For number of compounded volumes $N=1$ and 8, CNR values were calculated for the only nine combinations possible. For number of compounded volumes $N=2$ to 7, CNR values were calculated for nine randomly selected combinations.

To evaluate speckle reduction, 25 independent 3-D blocks with a size of 100 resolution cells were chosen from the background of the nine volumes at a depth of around 35 mm. The following equation was used to evaluate speckle SNR on a linear scale:^{3,11}

$$SNR = \frac{\mu}{\sigma} \quad \text{Eq.(4)}$$

where μ is the mean and σ is the standard deviation of the detected data for a speckle region that contains no resolvable structures.

Table 2 shows the mean cross-correlation and mean speckle SNR values along with the theoretical and experimental speckle SNR improvement values for the cyst and tumor 3-D images.

Figure 6 shows the monotonic increase of mean speckle SNR values as the number of compounded volumes increases. These values were obtained experimentally using Equation 3 and theoretically using Equation 1. For number of compounded volumes $N=9$, CNR was calculated for the only one possible combination. For number of compounded volumes $N=1$ and 8, SNR values were calculated for the only nine combinations possible. For number of compounded volumes $N=2$ to 7, SNR values were calculated for nine randomly selected combinations.

Figure 7A–C show the azimuth B-scan, elevation B-scan, and C-scan respectively of the cyst phantom generated using standard beamforming. Figure 7D–F show the image compounded from nine volumes for the cyst phantom. The C-scans and the elevation B-scans are taken approximately at the widest cross section of the cyst and the azimuth B-scans are taken approximately at the center of the phantom. All images are log-compressed and shown on a 30 dB dynamic range.

Figure 8 shows isosurface renderings of the cyst standard beamforming 3-D image and the 3-D image compounded from nine volumes. In each case, the isosurface level was adjusted for optimal display of the cyst.

Figure 9A–C show the azimuth B-scan, elevation B-scan, and C-scan respectively of the tumor phantom generated using standard beamforming. Figure 9D–F show the compounded image of the tumor phantom. The C-scans and the elevation B-scans are taken approximately at the widest cross section of the tumor and the azimuth B-scans are taken approximately at the center of the phantom. All images are log-compressed and shown on a 30 dB dynamic range.

Figure 10 shows isosurface renderings of the tumor standard beamforming 3-D image and the 3-D image compounded from nine volumes. In each case, the isosurface level was adjusted for optimal display of the tumor.

Figure 11 shows the azimuth B-scan of the wires phantom 3-D image with the short axis of the wires in the azimuth direction. Figure 11a was generated using standard beamforming, and figure 11b was generated by compounding nine volumes. Using detected data and the wire nearest the transducer, the -6 dB beamwidths of the uncompounded and compounded images were 0.70 mm and 1.43 mm respectively. This gives a reduction in the lateral resolution by a factor of 2.04.

IV. Discussion and Conclusions

In this paper, we presented initial 3-D spatial compounding results using a prototype 256×256 2-D array with row-column addressing, 40×40 mm aperture and 5 MHz center frequency. Using our experimental setup, we acquired several 3-D images to identify the benefits and the drawbacks of 3-D spatial compounding with aperture translation.

Using different subsets of uncorrelated volumes for compounding shows the expected monotonic increase of mean CNR and speckle SNR values as the number of compounded volumes increases. Our results show good agreement with values predicted by theory (Fig. 5,6). Compounding nine volumes shows a significant improvement in CNR and speckle SNR

for the contrast phantoms 3-D images. Qualitatively, it is easier to identify the inclusions in the compounded images than in the uncompounded images (Fig. 7,9). Also, the isosurface renderings of the -11 dB cyst and the $+11$ dB tumor show significant clutter reduction in the compounded images compared to the uncompounded ones (Fig. 8,10). The loss in lateral resolution was found to closely match the expected factor of two. This is caused by the fact that the aperture size in the compounded image is half of that in the uncompounded image.

In figure 9E, an inclined “ghost” artifact at the lateral edges of the image above the tumor can be seen. This is mainly caused by grating lobes that show up in the transmit direction of the 2-D array (elevational direction in figure 9E). Since the transmit side of the 2-D array has a pitch of 1λ , grating lobes for this side are theoretically located at 90° . Aperture number 3 (Fig. 3) creates the far left scan lines at large steering angles (maximum of 45° for a depth of 35 mm). Hence, a “ghost” of the tumor appears at the left edge of the elevational slices. This also applies to the far right scan lines created using aperture number 1 (Fig. 3). This effect was confirmed by finding the presence of this artifact in the individual compounded volumes. Using $\lambda/2$ pitch on the transmit side would minimize this problem. However, this requires manual excitation of the 256 transmit elements individually which would make data acquisition a lengthy process.

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FIG. 1. S. Awad, Row-column addressing scheme

Row-column addressing scheme. The transmit side consists of 8 horizontal electrodes. The receive side consists of 8 vertical electrodes which serve as ground in transmit. Transmit beamforming is done in the vertical or elevational direction while receive beamforming is done horizontally or azimuthally.¹⁸

FIG. 2. S. Awad, Transmit and Receive Beamforming for 3-D rectilinear imaging
Transmit and Receive Beamforming for 3-D rectilinear imaging.¹⁸

FIG. 3. S. Awad, The transmit and receive apertures used

The three apertures used to create the first scan line for (a) the transmit side and (b) the receive side of the row-column 2-D array. The apertures were walked to create on-axis scan lines whenever it is possible. All apertures are indicated by braces and shading.

FIG. 4. S. Awad, B-mode images of contrast phantoms used

B-mode images of (a) the -11 dB cyst phantom and (b) the $+11$ dB tumor phantom using Ultrasonix L14-5/38 linear array. Both images are log-compressed and shown with 30 dB dynamic range.

FIG. 5. S. Awad, CNR improvement curves

Mean of CNR values for 1 to 9 compounded volumes for: (a) -11 dB cyst phantom (b) +11 dB tumor phantom. Solid lines show experimental CNR values and dashed lines show theoretical CNR values.

FIG. 6. S. Awad, Speckle SNR improvement curves

Mean of speckle SNR values for 1 to 9 compounded volumes for: (a) -11 dB cyst phantom (b) +11 dB tumor phantom. Solid lines show experimental speckle SNR values and dashed lines show theoretical speckle SNR values.

FIG. 7. S. Awad, Slices of the 3-D B-mode images of the -11 dB cyst with and without compounding
 Experimental cyst 3-D images with the -11 dB cyst short-axis in azimuth using: (A–C) Standard 3-D beamforming. (D–F) Compounding of nine volumes. All images are log-compressed and shown with 30 dB dynamic range.

FIG. 8. S. Awad, Isosurface renderings of the -11 dB cyst 3-D images with and without compounding

Isosurface renderings of the -11 dB cyst 3-D image generated using: (a) standard beamforming and (b) compounding of nine volumes. In each case, the isosurface level was adjusted for optimal display of the cyst.

FIG. 9. S. Awad, Slices of the 3-D B-mode images of the +11 dB tumor with and without compounding

Experimental tumor 3-D images with the +11 dB tumor short-axis in azimuth using: (A–C) standard 3-D beamforming and (D–F) compounding of nine volumes. All images are log-compressed and shown with 30 dB dynamic range.

FIG. 10. S. Awad, Isosurface renderings of the +11 dB tumor 3-D images with and without compounding

Isosurface rendering of the +11 dB tumor 3-D image using: (a) Standard 3-D beamforming. (b) Compounding of nine volumes. In each case, the isosurface level was adjusted for optimal display of the tumor.

FIG. 11. S. Awad, A slice of the 3-D B-mode images of the wires with and without compounding
A short axis slice of the 3-D B-mode images of the wires phantom generated using: (a) standard beamforming and (b) Compounding of nine volumes. Both images are log-compressed and shown with 20 dB dynamic range.

Table 1

S. Awad, Phantom recipes

Recipes for background and inclusion materials in the gel phantoms used.

Material	Background	Inclusion
DI water	400 g	50.00 g
N-propanol	36.70 g	4.60 g
Formaldehyde	0.24 g	0.030 g
Gelatin 275 Bloom	24.00 g	3.00 g
Graphite	3.89 g (contrast phantoms)	0.10 g (−11 dB cyst)
	0 g (wires phantom)	3.43 g (+11 dB tumor)

Table 2

S. Awad, Speckle SNR comparison

Mean cross-correlation, mean speckle SNR values and the theoretical and experimental speckle SNR improvement values for the cyst and tumor 3-D images.

	-11 dB cyst	+11 dB tumor
Mean cross-correlation values for the nine volumes	0.024	0.022
Mean SNR value for the compounded image	4.99	5.03
Theoretical SNR improvement	2.92	2.81
SNR improvement over the nine volumes	2.67	2.69
SNR improvement over the uncompounded image	2.65	2.70