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Phase modulation beamforming for ultrafast plane wave imaging

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Abstract

In order to improve the spatial resolution for high frame rate imaging, a new image formation approach is proposed based on introducing very weak aberration into received data, then combining multiple results by taking the pixel-wise standard deviation of multiple aberrated images and subtracting the result from the delay-and-sum image. This approach is demonstrated in simulations, tissue-mimicking phantom experiments, and *in vivo* imaging. Simulations indicate the lateral full-width half-max (FWHM) of targets decreases by $38.24 \pm 6.38\%$. In imaging wire targets in a tissue-mimicking phantom at 7.8 MHz, wire target FWHM decreases by $35.91 \pm 5.39\%$. However, contrast was observed to decrease by 1.23 dB and CNR by 18.5% in phantom studies due to the subtraction of similar images, which increases the number of dark pixels in the image. Finally, the proposed technique is tested *in vivo*, with images showing improvements similar to those in tissue-mimicking phantoms, including increased separation between closely-spaced targets.

Keywords

ultrafast; plane wave compounding; aberration; beamforming

I. Introduction

DURING the past decade, ultrafast plane wave imaging has emerged as a useful new approach for ultrasound imaging due to its high acquisition rates, with losses in spatial and contrast resolution due to the use of unfocused transmit events restored by compounding multiple steered transmit events [1]. High frame rate plane wave imaging has also been extended to many functional imaging modes, including Doppler [2-4], contrast-enhanced imaging [5, 6], and elasticity imaging [1].

Recently, several authors have presented techniques for improving image quality in plane wave imaging. Jensen *et al.* investigated the number of transmit events and maximum steering angle resulting in maximal image quality for a given array [7]. Alternatively, Garcia *et al.* [8] demonstrated improved plane wave imaging based on the on the f - k migration methods of Stolt [9] and Lu [10, 11]. Chen *et al.* recently updated this approach to include

angular weighting and coherent compounding in the frequency domain, which improves contrast without loss of resolution [12]. Mozumi *et al.* have also applied coherence-based weighting and adaptive beamforming to unfocused plane wave imaging [13]. Other researchers have developed techniques for improving image quality by accounting for element directivity [14], applying an angular coherence factor [15], and suppressing side lobes [16]. Furthermore, Agarwal *et al.* have recently presented an approach for high resolution imaging based on application of multiple apodization functions, improving spatial resolution by as much as 35 $\times$, though with reduced contrast and contrast-to-noise ratio (CNR) [17].

While novel beamforming techniques can improve image quality for B-mode imaging, conversely, aberration can degrade the point spread function (PSF) and corresponding image quality. Aberration, or focusing errors resulting from spatial variations in the speed of sound in tissue, can be estimated and corrected using acquired radiofrequency (RF) data [18-23]. Even small focal errors resulting from weak aberration cause main lobe broadening and increased clutter. By modifying acquired RF data sets prior to beamforming based on time domain estimates of shifts between channel data [18, 22], or by selecting the phase relationship between adjacent elements that maximizes speckle brightness [21], the PSF can be partially restored to a state similar to the PSF in the absence of aberration.

Because adaptive, coherence-based beamforming as well as correction of aberration can both provide higher contrast and/or spatial resolution [18-28], the phase relationship between RF channel data during image formation clearly affects image quality. Through modifying the phase relationship between RF channel data during the beamforming process, it may be possible to produce a PSF with a specific advantageous characteristic, such as narrower main lobe width or reduced off-axis clutter.

In this work, we present a technique for improving lateral resolution in high frame rate ultrasound imaging with multiple unfocused transmit events by introducing small focal errors during receive beamforming, then subtracting the variation resulting from this aberration from non-aberrated data sets to reduce main lobe width (Figure 1). The ability of the proposed technique to improve spatial resolution for a given number of transmit events is investigated in simulations, phantom studies, and *in vivo* imaging. The lateral resolution is quantified using the full-width half-max (FWHM). Contrast, CNR, and the distribution (histogram) of pixel values for this processing technique are also quantified.

II. Methods

A. Processing techniques

The phase modulation processing technique, which relies on beamforming data multiple times with different phase realizations (applied aberration profiles), is illustrated in Figure 1. Received RF channel data are processed multiple times for each acquisition: 1) using delay-and-sum (i.e. without aberration, blue line in Figure 1A), and 2) one or more times with an aberration profile (time delay error) introduced across the aperture (red and yellow lines in Figure 1A) according to the equation:

$$\tau_{error}(n) = A \sin\left(\frac{n\pi}{k} - d\phi\right), \quad (1)$$

where A is the amplitude of the introduced error, n is element number from 1 to 128, k is the depth-dependent parameter controlling the spatial frequency of the aberration profile, and $d\phi$ indicates an incremental phase shift applied to each different processing $d = 1 \dots m$ of the acquired RF data. In this work, $A = 3.2$ ns and $\phi = \pi/10$, thus there is a shift of $\pi/10$ between different aberration profiles applied to the same data (i.e. $d = 1$ for the red line and $d = 2$ for the yellow line in Figure 1A). The frequency of the applied aberration profile is varied with depth by adjusting parameter k in Equation 1. As shown in Figure 2A, increasing k (left axis) increases the spatial frequency of the applied aberration profile, as quantified by the FWHM correlation length (right axis). In Figure 2B, the spatial response of the array transducer is shown, and in Figure 2C, the FWHM width of the beam in Figure 2B is plotted as a function of depth. By comparing Figure 2A and Figure 2C, it can be seen that the spatial frequency parameter k is varied in depth to be inversely proportional to the beam width. This variation in spatial frequency content of the applied aberration is necessary because high frequency aberration (high value of k) produces variation across the entire lateral extent of a given target, which is required to produce the modulation resulting in higher resolution. However, when the beam is narrower (around 20 mm in depth as in Fig. 2B-C), then the target is more laterally compact in the acquired RF data and decorrelates rapidly with lateral distance. Thus when the PSF is narrow, introduction of only a small focusing error across the elements of the array produces substantial modulation, resulting in improvement when the standard deviation is taken and subtracted from the conventional delay-and-sum image.

For conventional delay-and-sum beamforming, receive beamforming delays for a given element n are computed according to:

$$\tau_{DAS}(r_p, n) = \frac{\|r_n - r_p\|}{c}, \quad (2)$$

where r_p indicates the coordinates of the point in the field, r_n indicates the coordinates of the receiving element n , and c is the speed of sound. For the cases with small focal errors, receive beamforming delays for each element n are computed according to:

$$\tau_{Aberrated}(r_p, n) = \tau_{DAS}(r_p, n) - \tau_{error}(n). \quad (3)$$

Delayed signals both without (τ_{DAS} , Equation 2) and with introduced focusing errors ($\tau_{Aberrated}$, Equation 3) are stored in matrices $\mathbf{X}_{DAS}$ and $\mathbf{X}_{PM_{d=1 \dots m}}$, respectively, for summation, where there are m versions of $\mathbf{X}_{PM_d}$ depending on the total number of phase realizations m processed (where $d = 1 \dots m$ in Equation 1). The resulting beamformed RF signal for a given pixel is produced according to:

$$\mathbf{RF}_{DAS}(r_p) = \mathbf{X}_{DAS}^T \mathbf{w} \text{ and } \mathbf{RF}_{PM_d}(r_p) = \mathbf{X}_{PM_d}^T \mathbf{w}, \quad (4)$$

where $\mathbf{X}_{DAS}$ and $\mathbf{X}_{PM_d}$ contain delayed signals received by an n -element array from a point at location r_p with delays for $\mathbf{X}_{DAS}$ and $\mathbf{X}_{PM_d}$ computed according to Equations 2 and 3, respectively, and $\mathbf{w}$ is an $n \times 1$ vector with all values equal to one for all depths. Applying Equation 4 for all angles α produces $\mathbf{RF}_{DAS}(r_p, \alpha)$ as well as $\mathbf{RF}_{PM_d}(r_p, \alpha)$ for each value of d , where α is the steering angle of the transmitted wave. Phase modulation images are formed by taking the pixel-wise standard deviation across the multi-angle coherently compounded, envelope-detected versions of $\mathbf{RF}_{DAS}(r_p, \alpha)$ and $\mathbf{RF}_{PM_d}(r_p, \alpha)$, then subtracting the normalized result from the conventional image:

where $Env[\cdot]$ denotes the envelope of beamformed RF signal, $\max\{\cdot\}$ denotes the maximum pixel value and $std(v_1, v_2, \dots)$ denotes the standard deviation of pixel values $v_1, v_2, \dots$ at a given location across multiple images. As described, Equation 5 denotes subtracting the normalized envelope of the pixel-wise standard deviation of the DAS image and some number of aberrated images from the normalized DAS image. While the example in Figure 1 shows *ImagePhaseModulation* incorporating 2 aberrated data $\mathbf{RF}_{PM_d}$ (i.e. $d = 1$ and $d = 2$) in addition to $\mathbf{RF}_{DAS}$, for the remainder of the work in this paper, 4 aberrated data $\mathbf{RF}_{PM_d}$ (i.e. $d = 1$ to 4 with $\phi = \pi/10$) are utilized in addition to $\mathbf{RF}_{DAS}$.

In phase modulation beamforming, taking the pixel-wise standard deviation across $\mathbf{RF}_{DAS}$ and $\mathbf{RF}_{PM_{d=1\dots m}}$ (second term in Equation 5) emphasizes small differences in two similar PSFs; these variations primarily occur along the slope of the main lobe and in the clutter pedestal (Figure 1C-D). When the standard deviation image is subtracted from the conventional delay-and-sum image after compounding and envelope detection (Equation 5), the main lobe width is decreased.

B. Simulations

Simulations were performed using Field II [29] to mimic the acquisition conditions in the tissue-mimicking phantom (Section C). Briefly, a 128-element linear array transducer operating at 7.8 MHz and 50% bandwidth was simulated. Point targets were positioned at depths of 6.5, 7.5, 8.5, 9.5, 22, and 32 mm as in the physical tissue-mimicking phantom. Weaker scatterers were positioned surrounding these targets (>10 per resolution cell). The scattering amplitude of these weaker background scatterers relative to point targets was varied from -29 dB to -11 dB in 6 dB increments to assess the effect of acoustic scattering ratio on target FWHM. Electronic noise was added to each channel with an electronic SNR of 30 dB. Radiofrequency (RF) data resulting from these simulations were beamformed to produce both delay-and-sum (DAS) and phase modulation (PM) images with the number of angles varying from 1 to 31 angles over an angular extent of $\pm 14^\circ$. The lateral FWHM of targets were evaluated in the resulting images under conditions of: 1) variation in acoustic scattering

$$ImagePhaseModulation = \frac{Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{DAS}(r_p, \alpha)\right]}{\max\left\{Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{DAS}(r_p, \alpha)\right]\right\}} - \frac{\text{std}\left(\begin{matrix} Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{DAS}(r_p, \alpha)\right], Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{PMd=1}(r_p, \alpha)\right], \\ Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{PMd=2}(r_p, \alpha)\right], \dots \end{matrix}\right)}{\max\left\{\text{std}\left(\begin{matrix} Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{DAS}(r_p, \alpha)\right], Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{PMd=1}(r_p, \alpha)\right], \\ Env\left[\sum_{\alpha=-\alpha_{max}}^{\alpha_{max}} \mathbf{RF}_{PMd=2}(r_p, \alpha)\right], \dots \end{matrix}\right)\right\}}, \quad (5)$$

ratio, 2) variation in number of angles, and 3) variation in target depth.

Additional set simulations were performed with 11 angles and -11 dB background scattering amplitude with two closely-spaced point targets centered on the transducer separated by 0.4 mm. The ability to separate two closely-spaced targets was quantified by determining the minimum intensity level reached between the two targets.

C. Phantom studies

In order to assess spatial resolution in the lateral direction, wire targets were imaged in a tissue-mimicking phantom (ATS Labs Model 539) using a Verasonics Vantage 256 (Kirkland, WA, USA) and an L11-5 transducer operating at 7.8 MHz. This array has 128 elements with inter-element separation of 260 μm . A field of view of ± 15 mm laterally and up to 35 mm in depth was imaged. Acquired RF data were beamformed offline to produce DAS and PM images with the number of angles varying from 1 to 31 angles over an angular extent of $\pm 14^\circ$. Three data sets were acquired by translating the transducer by 1 mm in the lateral direction between acquisitions, resulting in targets at different lateral positions in each acquisition. The lateral width (FWHM) of each target was evaluated in each acquired data set, allowing experimental evaluation of the FWHM due to variation in target depth and number of angles. All images were displayed with matched histograms using *imhistmatch* in Matlab.

In order to assess contrast, additional experimental data sets were acquired using a different region of the same phantom containing hypoechoic lesions. Again, three data sets were acquired by repositioning the transducer before each acquisition. Contrast ratio

$$(20 \log_{10}(\bar{S}_{in} / \bar{S}_{out})), \text{CNR} \left(\frac{\bar{S}_{in} - \bar{S}_{out}}{\sqrt{\sigma_{in}^2 + \sigma_{out}^2}} \right), \text{speckle variance } (\sigma_{out}^2), \text{ and speckle SNR } (\bar{S}_{out} / \sigma_{out})$$

were characterized for DAS and PM images. $\bar{S}_{in}$ and $\bar{S}_{out}$ denote the mean of the signal inside and outside of the hypoechoic lesions in the image, respectively, while σ_{in} and σ_{out} denote the standard deviations in the same regions. In addition, the generalized CNR was computed for all images. Generalized CNR is a recently-proposed metric [30], which allows comparison of lesion contrast for different image formation techniques by comparing the overlap between the histograms for the lesion and surrounding regions, thus removing the influence of any change in dynamic range due to the image formation technique.

D. In vivo imaging

The same array transducer and system were used to acquire images of the carotid artery and surrounding anatomy in a healthy volunteer with 31 angles over an angular extent of $\pm 14^\circ$. Imaging data were processed using both DAS and PM techniques. Contrast ratio, CNR, and generalized CNR (gCNR) were evaluated for these images for the hypoechoic artery lumen relative to the surrounding hyperechoic tissue.

III. Results

A. Simulations

Simulated imaging results are shown in Figure 3 for acquisition with 11 angles with background scattering amplitude at -29 dB for both DAS (A) and PM processing (B). The cross-sectional views through the point target at a depth of 22 mm (C) shows a decrease in target width for PM processing. Decrease in FWHM with varying acoustic background level relative to the point target scattering amplitude is shown in Fig. 4. The resolution improvement is greater at lower target-to-background ratios because these conditions have greater likelihood of high amplitude scatterers close to the target, thus increasing the ability to separate targets as in the simulations for closely-spaced point targets in Fig. 5. This percent improvement decreases gradually with increasing target-to-background ratio because for many scatterers per resolution cell, the likelihood of one strong “background” scatterer being located within the main lobe of the PSF is higher as the background scattering amplitude is increased. Due to multiple scatterers and averaging across multiple simulations for each set of conditions, the variation in performance response appears to be gradual rather than discrete. In Fig. 5, results for 0.4 mm-spaced point targets are shown, with two illustrative cross sections at depths of 8.5 mm (A) and 22 mm (B) indicated improved target separation for PM relative to DAS, as the minimum intensity value between the two targets is lower for PM processing. These results are summarized in Fig. 5C.

Following phantom imaging results in Fig. 6, the percent decrease in FWHM as a function of varying depth are shown for both experimental and simulation cases in Fig. 7A (dashed blue line); percent decrease in FWHM with varying degree of angular compounding is shown in Fig. 7B (dashed blue line).

These results, their relationship with experimental results, and the effect of the target to background scattering ratio (Fig. 4) are discussed in detail in the Discussion section. For the simulated data in Figure 7A, the FWHM decreased by $38.24 \pm 6.38\%$ across all depths.

B. Phantom studies

An illustrative result from phantom imaging is shown in Fig. 6, with lateral widths of wire targets decreasing in the lateral direction. In Fig. 7A, the decrease in FWHM in the lateral direction is shown for the wire targets in the tissue-mimicking phantom as a function of depth (black line). The percent change is similar with increasing depth, with only the most shallow wire target showing a smaller improvement as in the simulations. In Fig. 7B, the percent decrease in the lateral FWHM of the wire targets is shown with varying degree of angular compounding. The trend is similar to the simulations but overall the experimental

data shows greater improvement in lateral resolution, which is discussed in detail in the Discussion section. For the experimental data in Figure 7A, the FWHM decreased by $35.91 \pm 5.39\%$ across all depths.

In imaging the region of the ATs 539 phantom containing hypoechoic lesions, the contrast is reduced slightly and there is a visible increase in the number of dark pixel values (Table 1, Fig. 8). This increase in the number of dark pixels can be seen when comparing corresponding cross sections of DAs and PM images (Fig. 8c) and in the histogram of image values (Fig. 8D). Potential reasons underlying these results are described in greater detail in the Discussion section.

C. In vivo imaging

An illustrative result from *in vivo* imaging is shown in Figure 9. A large, bright region within the arterial lumen in the DAs image (bisected by the yellow dashed line in Figure 9A) can be seen to be two separate targets in the PM image, with the surrounding bright artifact also decreased in the PM image. The cross sections indicated by the yellow dashed lines in the two images can be compared quantitatively in Figure 9C. *In vivo* values of contrast ratio, CNR, and gCNR were: -7.14 ± 1.83 dB (DAS) and -5.63 ± 2.02 dB (PM), 2.63 ± 0.37 (DAS) and 1.95 ± 0.51 (PM), and 0.919 ± 0.019 (DAS) and 0.881 ± 0.026 (PM), respectively. Further study is required to evaluate the effect on image contrast and the effect of number of compounding angles on spatial resolution for *in vivo* imaging.

IV. Discussion

A. Lateral resolution in simulations and phantom experiments

As summarized in Fig. 7, the spatial resolution in the lateral direction is improved by applying the proposed processing technique, which consists of beamforming the received channel data with intentionally introduced phase errors, then forming an image of the difference between the standard deviation image and the conventional delay-and-sum image. As described in Figure 1, phase modulation processing yields a narrower main lobe relative to DAs. This occurs because the small aberration introduced creates a PSF that is slightly different from the PSF in the absence of aberration (Figure 1C-D). When the standard deviation is taken, these differences—which primarily occur along the slope of the main lobe and in the clutter pedestal—are emphasized. By subtracting the standard deviation image from the image without aberration (Equation 5), the main lobe width is decreased.

In Fig. 7A, lateral resolution increases for targets at varying depths, however the improvement is smaller at shallower depths. As can be seen in both the experimental and simulated images, there appears to be a visible modulation in the images at these shallow depths, with variation in speckle brightness visible at the left side of the image at depths < 10 mm in Fig. 3B and 6B. This is likely due to the sinusoidal aberration profiles applied, which at shallow depths produces a highly spatially-varying PSF, whereas at greater depths, the effects of weak aberration tend to average out. As seen in Fig. 7B, the lateral resolution improves for all compounding angles tested, though improvement is slightly greater for fewer angles, likely because SNR is lower in these images.

The results of simulations (blue lines in Fig. 7) demonstrate similar trends to experimental data, however the improvement in lateral resolution as quantified by percent decrease in FWHM of the lateral target is smaller for the simulations. There are two potential explanations for this: 1) targets in Field II are comprised of true point targets rather than the finite width wire targets in experimental data, and 2) although every effort was made, conditions in simulations may not be exactly matched to those encountered in the phantom, i.e. acoustic scattering characteristics between the target and background or electronic noise may differ slightly. An initial attempt to investigate this was made by varying the intrinsic contrast or acoustic scattering ratio between the simulated targets and the background (Figure 4), with results indicating the measured FWHM decreases (and standard deviation increases) with decreasing acoustic scattering ratio. This is likely because for high levels of background scattering, the measured FWHM is more likely to experience a high amplitude scatterer close to the point target, and the measured FWHM is also more highly variable due to the presence of stronger scatterers under the side lobes of the PSF.

B. Contrast in tissue-mimicking phantom studies

In imaging the hypoechoic lesions in the ATS 539 tissue-mimicking phantom, slight decreases in contrast and speckle variance were observed (Table I, Fig. 8). This decrease in contrast can also be seen in the cross section in the simulated image in Figure 3C. CNR and gCNR also decrease slightly due to the decrease in contrast and increase in speckle variance (Table I). The decrease in CNR is likely due to decrease in contrast, visible in the cross section in Fig. 8C, and also due to increase in speckle variance, which is likely more significant than the variation in contrast. Concerning contrast alone, In examining the histogram in Fig. 8D, it can be seen that PM processing results in an increased number of lower pixel values, i.e. the curve moves to the left and the peak increases. As can be seen in the images and cross sections in Fig. 8, this is the result of the subtraction operation, which produces very low values within a speckle-producing medium when the DAS pixel value and standard deviation value (two terms in Equation 5) are nearly equivalent.

C. In vivo imaging

In the *in vivo* image of the short axis of the carotid artery in Figure 9, phase modulation processing increases separation between adjacent point-like targets. Furthermore, due to the effect of subtraction, which causes an increase in the number of dark pixels in the image as previously described for the images of the hypoechoic lesions in the tissue-mimicking phantom, the image appears to be darker, especially in the lumen of the artery. Closely spaced bright targets such as those surrounding the lumen also appear to have increased separation, though further investigation is required to draw any conclusions regarding performance for in vivo imaging.

D. Future work

Substantial additional work remains to characterize phase and frequency relationships between aberrated data sets and speckle statistics. There are also many different parameter variations within the presented technique that could be explored, including the effect of varying the number of phase realizations (i.e. applied aberration profiles, as $m = 4$ profiles were used in this work), the nature of the applied aberration profiles (i.e. spatial frequency,

magnitude, relative phase shift between profiles $d\phi$, and the shape of the profile, which need not be sinusoidal), and the spatial variation of the aberrator spatial frequency.

The current variation in spatial frequency (Fig. 2) is intended to maintain higher spatial frequency aberration close to the transducer, then initially decrease spatial frequency with depth as the beam narrows, before finally increasing the spatial frequency of the aberration profile as the beam broadens with increasing depth. This piecewise approach also ensures smooth variation in spatial frequency with depth, avoiding any discontinuities in the image. The advantage of high frequency aberration profiles is the production of variation across the entire lateral extent of a target in acquired RF data, which allows the target to become narrower when the standard deviation envelope data is subtracted from the DAS envelope data (Fig. 1). However, in locations where the PSF is naturally narrower, the target is more laterally compact within the RF data and decorrelates more rapidly. In these cases, only a small variation is needed from one element to the next in order to produce improvement when the standard deviation is taken and subtracted. A high spatial frequency variation over a target with a small lateral extent can result in a lateral shift of the target. This is likely the reason for small lateral shifts seen in some results (i.e. Fig. 6C): in cases where the applied aberration profile is small relative to the lateral extent of the target, only a partial segment (< 1 cycle) of a sine wave aberration profile is applied to the data. Presently, the exact nature of these tradeoffs is unknown and thus further investigation is required.

Furthermore, it is unclear whether these techniques might be applied to other types of transmit events such as focused beam or diverging wave, where the spatial variation in the PSF is different from that in this work. The spatial frequency of the applied aberration profile(s) (Fig. 2) would likely need to be different from that used in this work. In general, the applied aberration profile(s) may need to be modified to correspond to the spatially-varying PSF or spatial frequency content, thus depending on the combination of aperture, frequency, depth, and transmit type, though further investigation is required.

V. Conclusion

Improved spatial resolution can be realized by multiple processing of single received data sets. This processing operates by introducing aberration into received RF data, then subtracting increased variation along the slope of the main lobe in these aberrated images from the non-aberrated image. The FWHM of wire targets decreased by $35.91 \pm 5.39\%$ as compared with $38.24 \pm 6.38\%$ in simulations. Despite small decreases in contrast and CNR due to the subtraction operation, images acquired *in vivo* demonstrate sufficient contrast to distinguish targets in addition to increased separation of point targets (Fig. 9).

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Biography

Bowen Jing was born in Gansu, China, in 1989. He received Ph.D. degree in biomedical engineering from Xi'an Jiaotong University, Xi'an, China, in 2017. In 2018, he joined the Wallace H. Coulter Department of Biomedical Engineering at Georgia Institute of Technology and Emory University, where he currently works as a postdoctoral researcher. His current research interests include ultrafast ultrasound imaging, ultrasound beamforming, contrast-enhanced ultrasound imaging and transcranial ultrasound imaging.

Brooks D. Lindsey (S'10, M'13) received the B.S. degree in electrical engineering from the University of Illinois at Urbana-Champaign and the Ph.D. degree in biomedical engineering from Duke University, Durham, NC, for his work in real-time 3D transcranial ultrasound imaging. While at Duke, he was the recipient of a pre-doctoral fellowship from the National Institutes of Health. In 2013, he joined the Joint Department of Biomedical Engineering at the University of North Carolina-Chapel Hill and North Carolina State University as a postdoctoral research associate. While a postdoc, he received the Ruth L. Kirschstein National Research Service Award from the National Institutes of Health. In 2017, Dr. Lindsey joined the Wallace H. Coulter Department of Biomedical Engineering at Georgia Institute of Technology and Emory University as Assistant Professor, where he directs the Ultrasonic Imaging and Instrumentation Lab. His lab's research interests include interventional and functional imaging, transducer and system development, and ultrasound signal processing.

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Fig. 1.

Phase modulation processing for 21 angles with 3.2 ns aberration amplitude. (A) Weak aberration profiles are applied to received RF data (blue line is without aberration, red and yellow lines show weak aberration profiles), (B) resulting in the formation of 3 similar images. (C) Lateral cross sections through the wire targets in the 3 images in B are similar but not identical, thus when the standard deviation is taken across similar images at each pixel (direction indicated by green arrow in B and D), a new standard deviation image results (E). (F) The result of subtracting the standard deviation image (E) from the non-aberrated image (first image in B).

Fig. 2.

(A) Variation of spatial frequency of applied aberration profiles as a function of depth. k is the parameter used to adjust the frequency of the applied aberration profile (Equation 1). The corresponding FWHM correlation length of the profile is shown on the right axis. (B) Field II simulated pulseecho response for the array transducer with unfocused transmit. (C) Full-width half-max beam width for the simulated data in B as a function of depth.

Fig. 3.

(A) Simulated image with DAS processing and 11 angles for a background acoustic scattering level of -29 dB, (B) the same data with PM processing, (C) cross-sectional cut through the point target at $z=9.5$ mm. Images are displayed with matched histograms.

Fig. 4. Percent decrease in target full-width half-maximum (FWHM) as a function of target to background scattering amplitude (acoustic SNR) in simulated images.

Fig. 5.

Simulated images (A) and cross-sections (B-C) through two targets separated by 0.4 mm from simulated image with 11 angles at a depth of (B) 8.5 mm, and (C) 22 mm. (D) Summary of the minimum measured intensity between the two targets as function of varying depth of the two targets for both delay-and-sum (DAS) and phase modulation (PM) processing with 3 simulations per point. Images are displayed with 50 dB dynamic range and histogram matched.

Fig. 6.

Images of wire targets in a tissue-mimicking phantom with 3 angles with (A) delay-and-sum, and (B) phase modulation beamforming for a single unfocused transmit event. The cross section for the wire target located at $z=11$ mm is shown in (C). Images are displayed with matched histograms.

Fig. 7.

(A) Percent decrease in target full-width half-maximum (FWHM) as a function of target depth phantom and simulation data acquired with 3 angles. (B) Percent decrease in target full-width half-maximum (FWHM) with varying number of angles at a depth of 11 mm. For simulations (blue lines), the background acoustic scattering level is -29 dB.

Fig. 8.

Images of ATS 539 tissue-mimicking phantom using (A) DAS and (B) PM image formation displayed with matched histograms. Blue and red lines indicate the cross-sectional cuts through two hypoechoic lesions shown in (C). (D) Histogram plots of envelope amplitudes for the data in (A) and (B).

Fig. 9.

(A) In vivo DAS image with 31 angles, (B) PM image of the same data, (C) lateral cross section through the location indicated by the dashed lines in A and B shows improved separation of adjacent targets and reduction of the large blurred object visible in (A). Images are displayed with matched histograms.

TABLE I

Image quality statistics in imaging cysts in ATS 539 tissue-mimicking phantom

	DAS	PM	Variation
<i>Contrast ratio</i>	-25.50 ± 0.05 dB	-24.27 ± 0.28 dB	-1.23 dB
<i>Speckle variance</i>	$8.28 \times 10^{-5} \pm 1.0 \times 10^{-7}$	$3.72 \times 10^{-5} \pm 5.0 \times 10^{-7}$	-55.1%
<i>CNR</i>	1.79 ± 0.04	1.46 ± 0.15	-18.5%
<i>gCNR</i>	0.966 ± 0.018	0.941 ± 0.038	-2.66%
<i>Speckle SNR</i>	1.86 ± 0.04	1.61 ± 0.12	-13.4%