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Ultrasonic Reverberation Clutter Suppression Using Multi-Phase Apodization With Cross-Correlation

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Abstract

Despite numerous recent advances in medical ultrasound imaging, reverberation clutter from near-field anatomical structures, such as the abdominal wall, ribs, and tissue layers, is one of the major sources of ultrasound image quality degradation. Reverberation clutter signals are undesirable echoes, which arise as a result of multiple reflections of acoustic waves between the boundaries of these structures, and cause fill-in to lower image contrast. In order to mitigate the undesirable reverberation clutter effects, we present in this paper a new beamforming technique called multi-phase apodization with cross-correlation (MPAX), which is an improved version of our previous technique, DAX. While DAX uses a single pair of complementary amplitude apodizations, MPAX utilizes multiple pairs of complementary sinusoidal phase apodizations to intentionally introduce grating lobes from which an improved weighting matrix can be produced to effectively suppress reverberation clutter. Our experimental sponge phantom and preliminary *in vivo* results from human subjects presented in this work suggest that MPAX is a highly effective technique in suppressing reverberation clutter and has great potential for producing high contrast ultrasound images for more accurate diagnosis in clinics.

Index Terms

Beamforming; clutter suppression; contrast enhancement; reverberation clutter

I. Introduction

A. Background

Medical ultrasound imaging is one of the most commonly used imaging modalities in clinics today. While it may not provide as much anatomical detail as other modalities like CT or MRI, ultrasound imaging has long been regarded as a preferred imaging modality for numerous clinical applications because of its high portability, high temporal resolution, inexpensive price, and no or minimal hazardous effects due to ionizing radiation [1]. However, medical ultrasound imaging often suffers from poor image contrast caused by at least 3 distinct mechanisms: 1) phase aberration effects caused by sound speed inhomogeneities in soft tissues, 2) clutter due to off-axis scatterers, and 3) reverberation clutter caused by near-field structures [2, 3]. Typically, phase aberration is unavoidable because the conventional delay-and-sum (DAS) beamforming utilized in clinical ultrasound imaging systems assumes a constant sound speed of 1540 m/s when the actual sound speed in the body may deviate from this value depending on the tissue compositions in the patient [4]. This would result in reduced ultrasound beam focusing quality and thereby, lowering image contrast and spatial resolution. Off-axis clutter is also an inherent feature of medical ultrasound imaging because human anatomy and hardware limitations often limit the dimension of the array transducer and make it impractical to tightly focus the beam to every point throughout the entire image. Hence, leakage of ultrasound beam energy is unavoidable and low-energy reverberations originating from off-axis scatterers would show up as clutter. Reverberation clutter is present if multipath scattering of ultrasound occurs due to the presence of near-field anatomic structures, such as the abdominal wall, ribs, and tissue layers [5]. In fact, a previous simulation study has demonstrated that along with phase aberration, reverberation clutter is the dominant mechanism of image quality degradation for fundamental frequency B-mode imaging [6].

B. Review of Existing Techniques

Various techniques have been proposed in the past in order to suppress clutter and obtain high contrast ultrasound images. One of the most widely studied approaches among these techniques is one that employs developing a weighting matrix based on the coherence of the received RF data. The coherence factor (CF), first introduced in 1999 [7], uses the ratio of the coherent energy to the total incoherent energy of the received RF data to weight each image point at every depth. The Generalized Coherence Factor (GCF) was later developed by modifying CF to account for the energy spread by speckle-generating targets that CF does not take into consideration [8]. Received RF signals from the mainlobe region correspond to low frequency components of the spectrum of the aperture domain data as they are coherent while those from sidelobes and clutter correspond to high-frequency components as they are incoherent. By taking advantage of this, GCF is computed as a ratio of the spectral energy within a low frequency region to the total spectral energy. A matrix of GCF values is then used to weight each pixel within the field-of-view (FOV). Phase coherence factor (PCF) and sign coherence factor (SCF) employ a sidelobe reduction approach similar to GCF, but the matrix for pixel-by-pixel weighting is based on phase distributions of the delayed channel RF signals across the aperture rather than coherence [9].

More recently, a new technique known as short-lag spatial coherence (SLSC) imaging has been introduced [10, 11]. Unlike conventional B-mode imaging that forms images based on echo brightness, SLSC forms images similar to conventional B-mode images using lateral spatial coherence as the basis. SLSC images are generated by first calculating the normalized spatial correlation of the delayed backscattered echoes at every axial position. The short-lag spatial coherence value for every axial depth and image line in the FOV is then obtained by summing the resulting normalized spatial correlation of the delayed backscattered echoes over the predetermined short-lag region. Other methods relevant to contrast enhancement in medical ultrasound that have been proposed in the past include a Fourier transform-based technique [12], the parallel adaptive receive compensation algorithm (PARCA) [13], and most recently a chirp model-based approach [14].

In our previous studies, we proposed a novel beamforming technique called dual apodization with cross-correlation (DAX), which utilizes phase differences in grating lobe signals from two complementary receive (RX) apodizations for clutter suppression [15, 16]. The novelty in DAX is that it employs the counter-intuitive idea that grating lobes, which are often thought to have detrimental effects on the image quality, can in fact provide information that may be useful for improving the image contrast by suppressing unwanted clutter signals. The findings of our previous studies showed that DAX is robust with low to medium level phase aberrations [16–18].

In our subsequent studies, we explored a number of different approaches to overcome the limitations of DAX and improve ultrasound image contrast in a more robust manner when a high level of phase aberration is present. First, we proposed a hybrid technique, which integrates DAX with phase aberration correction (PAC) [17, 18] or harmonic imaging [19, 20]. PAC aims to restore coherence by estimating and correcting for focusing errors that cause image quality degradation whereas harmonic imaging takes advantage of the reduced aberration effects at the second harmonic content of the received echo signals. Integration of DAX with such imaging techniques seeks to achieve synergistic enhancements of ultrasound image contrast from their independent contrast enhancement mechanisms. Second, we developed a modified version of DAX called phase apodization with cross-correlation (PAX), which introduces sinusoidal time or phase delays in the received channel RF data to generate grating lobes [21]. Our initial results showed that the % CNR improvement for PAX with a strong aberrator was 125 % higher in simulation and 218 % higher in experiment when compared with DAX.

Although they were shown to be promising in suppressing clutter caused by off-axis scatterers and phase aberrations effects, the techniques described above are often limited in suppressing high levels of reverberation clutter originating from near-field structures. Integration with harmonic imaging could be an attractive solution as it has been shown to suppress reverberation clutter effectively without additional computational burden in data processing [22, 23]. However, rather than relying on a different imaging technique to overcome its limitations, it is desirable to improve the robustness of the clutter suppression technique itself in the presence of high levels of reverberation clutter. Therefore, in this paper, we propose a new technique called multi-phase apodization with cross-correlation (MPAX), an extension of PAX that is designed to be more robust particularly in an *in vivo*

environment where high reverberation clutter level is expected. This work is also a development of our earlier work in which multiple amplitude apodization pairs are utilized for a similar purpose [24]. We demonstrate in this paper with simulation, experimental phantom, and *in vivo* imaging results that by utilizing multiple RX phase apodization pairs, MPAX achieves greater suppression of reverberation clutter and thus, greater contrast enhancement when compared with DAX. The rest of this paper is organized as follows: Section II first gives an overview of DAX and a formal theoretical description of MPAX. Section III describes the detailed methods employed in this study. Finally, Section IV presents our results from simulation, phantom experiments, and initial *in vivo* scanning of two human subjects along with our analysis and discussion. Section V summarizes our findings and concludes the paper.

II. Theory

In this section, we begin with a brief review of DAX and then employ Rayleigh-Sommerfeld diffraction theory to present a full description of the newly proposed technique, MPAX, and explain its underlying principles for sidelobe/clutter suppression.

A. Dual Apodization with Cross-correlation

In ultrasound imaging with array systems, the transmit (TX) aperture and its normalized far-field complex beam pattern form a Fourier transform relationship [25]. Therefore, a TX aperture having a rectangular amplitude apodization results in a sinc function for the far-field beam pattern. In conventional pulse-echo ultrasound imaging, the far-field complex beam pattern becomes the product of the TX and RX far-field beam patterns. Hence, when a rectangular amplitude apodization is used for both TX and RX apertures, the far-field pulse-echo complex beam pattern becomes a square of the sinc function. In DAX, the TX aperture remains unchanged, but two complementary square-wave apodizations exhibiting a group of alternating elements are applied on RX. An example of such apodizations having a 4-4 alternating pattern is shown in Fig. 1. In this case, the 2nd RX apodization (RX2), is identical to the 1st RX apodization (RX1) but spatially shifted to the right by 4 elements.

The magnitudes of the far-field pulse-echo beam patterns from the two RX apodizations are identical. However, their phases are not because of the different linear phase tilts in the beamspace domain, which are induced by the spatial shifting of the two complementary RX apodizations. In addition, the inter-element spacing or the pitch of the array d , no longer holds when such square-wave apodizations are applied. The pitch effectively becomes $2Md$, where M is the number of alternating elements and as a result, places q^{th} grating lobes at:

$$\theta_{g,q} = \pm \sin^{-1}\left(\frac{q\lambda}{2Md}\right). \quad (1)$$

It is also worthwhile to note that the amplitude of the grating lobes relative to the mainlobe is determined by the envelope and the effective element width Md , associated with DAX places the i^{th} zeros of the envelope at:

$$\theta_{e,i} = \pm \sin^{-1} \left(\frac{i\lambda}{Md} \right). \quad (2)$$

Since the fill-factor, defined as the ratio between the element width and the pitch, for square-wave apodizations is 1/2, the even-order grating lobes vanish as their positions coincide with those of the zeros in the envelope [26], leaving only the mainlobe ($q = 0$) and the odd-order grating lobes that exhibit a relative phase difference of π .

By introducing grating lobes having a relative phase difference of π , DAX attempts to generate two different point spread functions having similar mainlobe signals but very different clutter patterns manifested as phase differences in the two beamformed RF data sets. In DAX, any signals coming from the on-axis mainlobe (i.e. $x = 0$) are expected to be highly correlated as they show no phase differences while those coming from the off-axis grating lobes located away from the mainlobe are expected to be negatively correlated with large phase differences, which can be detected and quantified by means of normalized cross-correlation. Although 1D axial normalized cross-correlation followed by 2D median filtering was used in our earlier works [15–19], 2D normalized cross-correlation is adopted in this work to maximize its robustness [27]. The final nonlinear DAX weighting matrix $\rho_{DAX}(k, h)$ is computed by:

$$\rho_{DAX}(k, h) = \max \left(\frac{\sum_{h=m-B}^{m+B} \sum_{k=l-A}^{l+A} RX1(k, h) RX2(k, h)}{\sqrt{\sum_{h=m-B}^{m+B} \sum_{k=l-A}^{l+A} RX1(k, h)^2} \sqrt{\sum_{h=m-B}^{m+B} \sum_{k=l-A}^{l+A} RX2(k, h)^2}}, \varepsilon \right), \quad (3)$$

where RX1 and RX2 are the beamformed RF data sets from the complementary apodization functions, k is the k_{th} sample on image line h with a kernel size of $2A+1$ by $2B+1$ samples, and ε is the minimum threshold value, typically chosen to be 0.001 such that a $20\log_{10}(0.001) = 60$ dB amplitude reduction is applied to clutter signals manifesting a relative phase difference of $\pi/2$ or greater while preserving the mainlobe components having no or negligible phase difference. 2D median filtering, which was applied to the weighting matrix in our earlier works, is omitted as computation of 2D normalized cross-correlation is effective in removing artifacts in the weighting matrix caused by the random nature of speckle. Finally, the DAS-beamformed RF data is multiplied by the weighting matrix such that clutter-dominated signals are suppressed while the mainlobe-dominated signals remain intact.

B. Multi-Phase Apodization with Cross-correlation

Despite the promising results from simulation studies and phantom experiments demonstrated in our earlier work [15, 16], one of the limitations of DAX, particularly in an *in vivo* environment with a high level of reverberation clutter, is that it could lead to a noisy

weighting matrix as the contributions from reverberation clutter may often dominate over the grating lobe signals introduced by the complementary square-wave RX apodizations. This is, in part, due to the fact that the design of DAX algorithm allows for only one pair of apodizations and hence, relies on a single measurement (*i.e.* normalized cross-correlation coefficient) in quantifying the clutter contributions for each pixel. In this section, we describe a new technique called multi-phase apodization with cross-correlation (MPAX), which extends the concepts exhibited in our earlier work on PAX [21], for increased robustness particularly in the presence of high levels of near-field reverberation clutter.

Motivated by the concept of thin sinusoidal phase gratings and its mathematical formulations from Fourier optics [28], a pair of complementary sinusoidal phase apodizations, such as those shown in Fig. 2, is introduced. Their effects on the pulse-echo field can be approximated using the Rayleigh-Sommerfeld diffraction theory and the Fresnel approximation. At the TX focal depth, the complex pulse-echo field, $\psi_{\omega,PA}$ as a result of DAS beamforming with a sinusoidal phase apodization applied to the RX aperture at frequency ω at field point (x, z) can be expressed as [28, 29]:

$$\psi_{\omega,PA}(x, z) = \left(\frac{e^{jkz}}{j\lambda z} \right)^2 A_T(u_x) \int_{-\infty}^{\infty} a_R(x_0) \times e^{-\frac{jkx_0}{z}} e^{j\frac{m}{2} \sin(2\pi f_0 x_0)} dx_0. \quad (4)$$

where λ is the wavelength, $k = \frac{2\pi}{\lambda}$ is the wave number, m is the peak-to-peak phase delay in radians, and f_0 is the spatial frequency of the phase apodization in cycles per millimeters. The term $A_T(u_x)$ represents the spatial Fourier transform of the transmit aperture $a_T(x_0)$ at spatial frequency $u_x = \frac{x}{\lambda z}$ (*i.e.* $\mathcal{F}[a_T(x_0)]_{u_x = \frac{x}{\lambda z}}$) where x_0 is the azimuth coordinate on the aperture surface. The term $e^{j\frac{m}{2} \sin(2\pi f_0 x_0)}$ describes the sinusoidal phase apodization introduced to the RX aperture $a_R(x_0)$. The analysis can be simplified by use of the identity:

$$e^{j\frac{m}{2} \sin(2\pi f_0 x_0)} = \sum_{q=-\infty}^{\infty} J_q\left(\frac{m}{2}\right) e^{j2\pi q f_0 x_0}, \quad (5)$$

where J_q is the Bessel function of the first kind and of order q [28]. Hence, the equation becomes:

$$\begin{aligned} \psi_{\omega,PA}(x, z) &= \left(\frac{e^{jkz}}{j\lambda z} \right)^2 A_T(u_x) \sum_{q=-\infty}^{\infty} J_q\left(\frac{m}{2}\right) \times \int_{-\infty}^{\infty} a_R(x_0) e^{-\frac{jkx_0}{z}} e^{j2\pi q f_0 x_0} dx_0 \\ &= \left(\frac{e^{jkz}}{j\lambda z} \right)^2 A_T(u_x) \sum_{q=-\infty}^{\infty} J_q\left(\frac{m}{2}\right) A_R(u_x - q f_0). \end{aligned} \quad (6)$$

If both the TX and RX apertures are 1-D arrays of finite length $2a$, the above equation becomes:

$$\begin{aligned}\psi_{\omega,PA}(x, z) &= \left(\frac{e^{jkz}}{j\lambda z}\right)^2 (2a)^2 \text{sinc}[2\pi a(u_x)] \sum_{q=-\infty}^{\infty} \times J_q\left(\frac{m}{2}\right) \text{sinc}[2\pi a(u_x - qf_0)] \\ &= \left(\frac{e^{jkz}}{j\lambda z}\right)^2 (2a)^2 \text{sinc}\left[2\pi a\left(\frac{x}{\lambda z}\right)\right] \sum_{q=-\infty}^{\infty} \times J_q\left(\frac{m}{2}\right) \text{sinc}\left[\frac{2\pi a}{\lambda z}(x - qf_0\lambda z)\right].\end{aligned}\quad (7)$$

The equation (7) obtained from Rayleigh-Sommerfeld diffraction theory predicts that the sinusoidal phase apodization on the RX aperture deflects the main beam energy into multiple grating lobes characterized by the Bessel function J_q of the first kind. The theoretical diffraction efficiencies for mainlobe ($q=0$), 1st grating lobe ($q=1$) and 2nd grating lobe ($q=2$) are depicted in Fig. 3. It describes how much of the mainlobe energy is deflected into the 1st and 2nd grating lobes as the peak-to-peak excursion of the phase apodizations is varied. Hence, it may be useful in selecting the proper m values. The 0th order mainlobe vanishes completely whenever $m/2$ is a root of J_0 and the largest possible diffraction efficiency into one of the 1st order grating lobes is equal to the maximum value of J_1^2 , which is far greater than that of the amplitude apodizations [28].

The equation (7) predicts that the grating lobes are located at a distance of $\pm q f_0 \lambda z$ from the mainlobe. Carefully selecting the peak-to-peak excursion, m and the spatial frequency, f_0 of the sinusoidal phase apodization allows for more flexibility in manipulating the locations and the magnitude of the grating lobes in a controlled manner. Multiple pairs of grating lobes over a range of locations are generated by varying the spatial frequency f_0 in each phase apodization pair. This allows for computing multiple normalized cross-correlation coefficients for each pixel. Since each of these coefficients is computed using a unique pair of grating lobes, they contain different information, which can be averaged to yield a robust weighting matrix even in the presence of high levels of reverberation clutter.

In a manner similar to DAX, 2D normalized cross-correlation is used in MPAX instead of 1D axial normalized cross-correlation in order to maximize its robustness [27]. Therefore, the final nonlinear MPAX weighting matrix, $\rho_{MPAX}(k, h)$ is computed by:

$$\rho_{MPAX}(k, h) = \max \left(\frac{1}{N} \sum_{i=1}^N \frac{\sum_{h=m-B}^{m+B} \sum_{k=l-A}^{l+A} \text{RX1}_i(k, h) \text{RX2}_i(k, h)}{\sqrt{\sum_{h=m-B}^{m+B} \sum_{k=l-A}^{l+A} \text{RX1}_i(k, h)^2} \sqrt{\sum_{h=m-B}^{m+B} \sum_{k=l-A}^{l+A} \text{RX2}_i(k, h)^2}}, \varepsilon \right), \quad (8)$$

where RX1_i and RX2_i are the beamformed RF data sets with the i^{th} pair of complementary phase apodization functions, k is the k^{th} sample on image line h with a kernel size of $2A+1$ by $2B+1$ samples, ε is the minimum threshold value, and N is the number of grating lobe pairs. Similar to DAX with 2D normalized cross-correlation, 2D median filtering is no longer utilized. While the resulting N coefficients corresponding to mainlobe signals tend to have a mean value close to 1 and a relatively small variance, those corresponding to signals dominated by reverberation clutter tend to have a much smaller mean value and a larger

variance. Hence, the resulting coefficients associated with each pixel are averaged to yield values near 1 in a speckle region and near 0 in a clutter region.

1. Channel RF signals time-aligned in receive are summed after a uniform apodization is applied to obtain standard DAS-beamformed RF data.
2. The same time-aligned channel RF signals are fed into the MPAX algorithm in which a pair of complementary sinusoidal phase apodizations (i.e. $RX1_j$ and $RX2_j$) with predetermined parameters m and f_0 is applied to obtain two different beamformed RF data after summing.
3. 2D normalized cross-correlation is then performed using the two resulting beamformed RF data to yield a matrix filled with 2D normalized cross-correlation coefficients.
4. Steps 2–3 are repeated for N different pairs of complementary sinusoidal phase apodizations. Each pair of sinusoidal phase apodizations would have a unique combination of the parameters m and f_0 .
5. The N different 2D normalized cross-correlation coefficient matrices as a result of steps 2–4 are averaged and thresholded with a minimum threshold value, ϵ , to yield the final weighting matrix.
6. The final weighting matrix is then multiplied to the standard DAS-beamformed RF data.
7. Additional steps including bandpass filtering, envelope detection, and log compression are performed to obtain the MPAX-applied image.

For all experimental results in this study, performance is evaluated in terms of contrast-to-noise ratio (CNR) as defined by:

$$CNR = \frac{|\bar{S}_t - \bar{S}_b|}{\sigma_b}, \quad (9)$$

where $\bar{S}_t$ is the mean of the target, $\bar{S}_b$ is the mean of the background and σ_b is the standard deviation of the background in dB [30].

III. Methods

A. Field II Simulation

Computer simulations for a point target were performed using Field II [31] to compare the beamplots for DAS, DAX, and MPAX. Imaging parameters were chosen to model a 64-element ATL P4-2 phased array having parameters summarized in Table I. For DAX, a 4-4 alternating pattern shown in Fig. 1 was used. These are equivalent to two complementary square-wave apodizations having an effective pitch of $2Md = 2 \times 4 \times 0.32 \text{ mm} = 2.56 \text{ mm}$. For MPAX, 8 different f_0 values (i.e. $N = 8$) ranging from 0.342 cycles/mm to 0.635 cycles/mm at an increment of 0.042 cycles/mm were selected. The minimum and the maximum f_0 values were empirically selected such that the grating lobe level is always

higher than that from 4-4 DAX and roughly varies from -30 dB to -40 dB. The N value was also selected empirically based on our experimental phantom results which will be presented and discussed in the next section. With the selected f_0 values, the grating lobes move from lateral positions of ± 1.47 cm to ± 2.74 cm at an increment of 0.181 cm. Among the 8 phase apodization pairs for MPAX used in this study, the first and last f_0 values, namely 0.342 cycles/mm (solid lines) and 0.635 cycles/mm (dash-dot lines), are shown for $m = 3.6$ rad in Fig. 2.

B. Experimental Data from Custom Sponge Phantom

The performance was evaluated for DAS, DAX, and MPAX with a custom sponge phantom (Grease Monkey Pro Cleaning Hydrophilic Sponge, Big Time Products LLC, Rome, GA) with a 4 cm-diameter circular hole. A previous study reported that a highly-reflective copper wire mesh produces clutter having similar characteristics to that of *in vivo* data [3]. Hence, in order to mimic near-field reverberation effects, a wiry copper household scouring pad (Practical Matter Copper Mesh Scourers, IMS Trading LLC, Los Angeles, CA) was cut to roughly 1 cm in thickness and placed at the face of the transducer. Individual channel RF signals were acquired from the custom sponge phantom immersed in a container filled with de-gassed water. Data acquisition was performed using a Verasonics data acquisition system (Verasonics, Redmond, WA) with a 64-element ATL P4-2 phased array transducer at a rate of 15 frames/s. A total of 18 frames were acquired from a custom sponge phantom with and without a copper wire mesh at the transducer face. A 1-cycle pulse with a center frequency of 2.5 MHz was used and a total of 128 transmit beams with a transmit focus at an axial depth of 7 cm were used over a 72° field-of-view at an angular beam spacing of 0.57° . All RF data sets were sampled at a sampling frequency of 10 MHz. A 2D-kernel size of $2\lambda \times 1.3\lambda$ was empirically chosen for 2D normalized cross-correlation in both DAX and MPAX.

C. Initial In vivo Evaluation

For *in vivo* performance evaluation, RF data sets were collected from two human subjects using the same transmit and acquisition settings as those described for sponge phantom imaging after obtaining an Institutional Review Board (IRB) approval. One of them was a healthy volunteer and the other was a patient recruited at the Keck School of Medicine at the University of Southern California, Los Angeles, CA. In order to ensure safety of the human subjects, transmit power for the pulse sequences was determined prior to scanning based on acoustic output measurements obtained with a HNP-0150 needle hydrophone (Onda, Sunnyvale, CA) and AH-1100 amplifier, such that the mechanical index and spatial-peak pulse average intensity do not exceed the limits established by the Food and Drug Administration. Cardiac data sets were acquired from the apical four-chamber and subxiphoid views from the healthy volunteer while the long axis view of the inferior vena cava was obtained from the recruited patient. Performance evaluation in terms of image contrast was performed for DAS, DAX, and MPAX. Improvement in image contrast was again quantified by the CNR as defined by equation (9).

IV. Results and Discussion

A. Field II Point Target Simulations

Fig. 5 depicts simulated beamplots for DAX with 4-4 alternating pattern and those for MPAX. In Fig. 5a, a simulated 4-4 DAX beam (dotted) from a complementary RX amplitude apodization pair as shown in Fig. 1 is compared with the beam generated from standard DAS with uniform apodization (solid). It is shown that introduction of the RX amplitude apodizations resulted in 1st grating lobes at $u_{g,1} = \pm \sin^{-1} \left(\frac{\lambda}{2Md} \right) = \pm 13.9^\circ$, which is equivalent to ± 1.7 cm from the mainlobe at an axial depth of 7 cm. The final beamplot (dashed) after DAX weighting has been applied is also shown.

In Fig. 5b, simulated PAX beams (dotted and dashed-dot) used in MPAX are compared with standard DAS with uniform apodization (solid). Note that for illustration purposes, PAX beams from only 2 of the 8 phase apodization pairs corresponding to those depicted in Fig. 2 are shown. Each phase apodization pair generates grating lobes having a unique magnitude and location. As described in our analysis in the previous section, the grating lobes from the phase apodization pair with a spatial frequency of f_0 are located at a distance of $\pm q f_0 \lambda z$ from the mainlobe. Hence, the 1st grating lobe location corresponds to ± 1.47 cm and ± 2.74 cm for $f_0 = 0.342$ cycles/mm and $f_0 = 0.635$ cycles/mm, respectively at $z = 7$ cm. All the other grating lobes associated with f_0 values between 0.342 cycles/mm and 0.635 cycles/mm appear at lateral locations equally spaced between ± 1.47 cm and ± 2.74 cm. The final beamplot after MPAX weighting (dashed) is also shown. Since the grating lobes from MPAX RX apodizations are located at different lateral positions, the resulting cross-correlation coefficient matrices would exhibit different patterns of artifacts caused by off-axis clutter as well as near-field reverberation clutter. In this way, we can obtain information about the clutter signals from multiple directions. Reverberation clutter artifacts that appear highly correlated from one phase apodization pair could appear less correlated if a different phase apodization pair is employed.

With additional information obtained from employing multiple pairs of phase apodizations in MPAX, it is possible to avoid relying on a single pair of DAX grating lobe signals whose expected phase difference of π may be lost and become highly correlated in the presence of high levels of reverberation clutter. By averaging cross-correlation coefficients from multiple pairs of phase apodizations in MPAX, undesirable effects from any erroneous estimation due to contributions of high levels of reverberation clutter are reduced. Furthermore, PAX beams can have greater grating lobe magnitudes when compared with DAX beams. Hence, each estimation is expected to be more reliable in high levels of reverberation clutter. The improved robustness associated with MPAX in the presence of high levels of reverberation clutter will become more evident with experimental and *in vivo* patient data presented in the next section.

B. Experimental Data from Custom Sponge Phantom

Fig. 6 shows DAS, DAX, and MPAX images of a 4 cm-diameter circular hole in a sponge without (top row) and with (bottom row) a copper wire mesh at the transducer face. The CNR values were calculated from the pixels within the anechoic target (solid-line box in Fig.

6) and the pixels within the speckle background (dashed-line box in Fig. 6) using equation (9). The results are summarized in Table II. Fig. 7 shows DAX and MPAX weighting matrices which were used to obtain the corresponding DAX- and MPAX-weighted images shown in Fig. 6. The weighting matrices were obtained from a custom sponge phantom (top) without and (bottom) with a copper wire mesh. The MPAX weighting matrices were obtained by averaging 2D normalized cross-correlation matrices from 8 different sinusoidal phase apodization pairs (*i.e.* $N=8$) evenly spaced between 0.342 cycles/mm and 0.635 cycles/mm. Fig. 8 shows the performance, in terms of CNR, of MPAX as a function of N on the custom sponge phantom a) without and b) with a copper wire mesh. The spatial frequency spacing for each N value is varied such that the sinusoidal phase apodizations are always evenly spaced between 0.342 cycles/mm and 0.635 cycles/mm. In both cases, the CNR increases with increasing N until it reaches its maximum approximately at $N=8$. Hence, the N value was chosen to be the value beyond which no further increase in CNR is observed.

Using no copper wire mesh to serve as a control, the CNR value for DAS was 6.0. In this case, both DAX and MPAX were highly effective in suppressing acoustic clutter artifacts from off-axis scatterers and achieved CNR values of 16.0 and 16.4, respectively. These values correspond to CNR improvements of 167 % and 173 %, respectively. When near-field reverberation clutter effects were induced by placing a copper wire mesh at the transducer face, the CNR value for DAS was reduced to 2.7 and the circular anechoic target was severely obscured. Since much of the DAX grating lobe signals become highly correlated because of the strong reverberation clutter signals, image contrast improvement associated with DAX is limited with a CNR value of 6.6. Although this corresponds to 144 % CNR improvement, it is clear from the results shown in Fig. 6 that much of the reverberation clutter effects still remain inside the circular hole in the sponge phantom. However, MPAX takes advantage of the increased grating lobe magnitude when compared with DAX and the additional cross-correlation coefficients computed from grating lobes from multiple locations, to suppress reverberation clutter effects in a more robust manner, yielding a CNR of 10.5, which corresponds to a 289 % CNR improvement. The visibility of the anechoic target is greatly improved and the image contrast is enhanced. A larger grating lobe magnitude can be beneficial as it better preserves the phase differences in the grating lobe signals particularly in the presence of high levels of reverberation clutter. However, a larger grating lobe magnitude may reduce signal correlation for main lobe signals. Therefore, a more effective suppression of reverberation clutter is obtained at a cost of reduced main lobe signal preservation. Such a trade-off is inevitable and the optimization of MPAX performance entails finding the optimal grating lobe level that maximizes the CNR improvement. However, the optimal grating lobe magnitude may not be fixed but depend on the reverberation clutter level. Furthermore, additional cross-correlation coefficients computed from grating lobes from multiple locations also help improve the accuracy of the final weighting matrix. The suppression of the speckle signals in the near field region of DAX and MPAX images is expected as a single transmit focus at 7 cm was used.

C. Initial In Vivo Cardiac Imaging Results

Fig. 9 shows DAS, DAX, and MPAX images of an apical 4-chamber view (top row) and a subxiphoid view (bottom row) of the heart from a healthy volunteer. It is assumed that reverberation clutter effects from near-field structures like the ribs and the tissue layers are inherent in these *in vivo* data sets and as a result, the visibility of the cardiac chambers is compromised. The CNR values were calculated from the left ventricle (LV) as well as the right ventricle (RV) in both data sets using equation (9). The solid- and dashed-line boxes in Fig. 9 indicate the target and background pixels, respectively, that are used. The calculated CNR values are summarized in Table III.

For the apical 4-chamber view, the CNR value was 1.6 in the LV and 1.1 in the RV for DAS. DAX enhanced image contrast in both ventricles, yielding a CNR value of 3.5 in the LV and 1.6 in the RV. These correspond to CNR improvements of 119 % and 45 %, respectively. MPAX achieved more favorable results in both ventricles with a CNR value of 4.8 in the LV and 1.7 in the RV, corresponding to improvements of 200 % and 55 %, respectively. In addition to the superior performance of MPAX over DAX in terms of CNR in the LV, the enhancement of the visibility of the left ventricular chamber is clearly seen in the MPAX image. However, suppression of the reverberation clutter effects in the RV, which appeared at least 10 dB stronger than those in the LV, was more challenging. This is possibly due to a suboptimal probe positioning when collecting data. As a result, the CNR improvements in the RV were much smaller than those in the LV.

For the subxiphoid view, the CNR value was 3.8 in the LV and 3.1 in the RV for DAS. Image contrast was enhanced in both ventricles with DAX, yielding a CNR value of 5.3 in the LV and 4.9 in the RV. These correspond to CNR improvements of 39 % and 58 %, respectively. Again, MPAX achieved more favorable results in both ventricles with a CNR value of 5.9 in the LV and 7.3 in the RV, corresponding to improvements of 55 % and 134 %, respectively. It is worthwhile to point out that the reverberation clutter suppression associated with DAX was not quite uniform in the chambers because the reverberation clutter often dominates over the grating lobes and DAX relies on a single cross-correlation coefficient for each pixel. Hence, the DAX-processed image lacks smoothness and shows an increased amount of sharp changes in the pixel values particularly in the chambers. However, much of these undesirable features are suppressed with MPAX as it can create a weighting matrix that is artifact-free as a result of averaging multiple cross-correlation coefficients computed from different phase apodization pairs. The MPAX-processed image looks much smoother and the suppression of the reverberation clutter effects is achieved throughout the whole image in a more uniform manner. Therefore, in addition to the larger CNR improvements, the MPAX-processed images are qualitatively more desirable as they further enhance the visibility of the chambers and delineation of the endocardial borders.

D. Initial In Vivo Abdominal Imaging Results

Fig. 10 shows DAS, DAX, and MPAX images of an inferior vena cava from a patient recruited at the Keck School of Medicine at the University of Southern California. As in the case of *in vivo* cardiac data sets, it is assumed that reverberation clutter effects from near-field structures are inherent in the *in vivo* abdominal data. The effects of the reverberation

clutter is clearly seen from the DAS-beamformed image as the visibility of the inferior vena cava, which is located between axial depths of 6 cm and 9 cm, is low. The CNR values were calculated for DAS, DAX, and MPAX using equation (9). The solid- and dashed-line boxes in Fig. 10 indicate the target and background pixels, respectively, that are used in CNR calculation. The calculated CNR values are summarized in Table III.

The CNR value of the inferior vena cava was 2.8 for DAS. DAX yielded a CNR value of 5.0, which corresponds to a CNR improvement of 79 %. Despite the contrast enhancement associated with DAX, some of the reverberation clutter effects are still visible within the inferior vena cava. MPAX further enhanced image contrast and yielded a CNR value of 5.7, corresponding to an improvement of 104 %. Similar to the trend seen with *in vivo* cardiac data, MPAX not only showed higher CNR when compared with DAX, but it also achieved a qualitatively more favorable result with a greater and more uniform suppression of the reverberation clutter effects in the anechoic region. Furthermore, MPAX was more effective than DAX in improving the visibility of blood vessels and other anatomical structures in the near-field (*i.e.* <7cm).

V. Conclusions and Future Work

The primary emphasis of this study was to develop and evaluate a beamforming technique that can suppress near-field reverberation clutter effects in a highly robust manner and achieve great enhancement in ultrasound image contrast. Although our previously developed technique, DAX has been proven quite effective in suppressing off-axis clutter and low-to-medium level phase aberrations [15, 18], a limitation of this method is its reduced efficacy in the presence of high levels of near-field reverberation clutter. Therefore, we presented in this work a new contrast enhancement technique called multi-phase apodization with cross-correlation (MPAX), which attempts to overcome the limitations of DAX and achieve greater improvement in image contrast in the presence of reverberation clutter. To our knowledge, this work is the first attempt to adapt the concept of sinusoidal phase apodization in Fourier optics for image contrast enhancement in medical ultrasound imaging. The common underlying principle behind DAX and MPAX is that grating lobes are intentionally introduced as a means to detect, quantify, and suppress contributions from clutter signals. The key difference between them is the methods utilized in generating and controlling the grating lobes. While DAX employs a pair of complementary square-wave amplitude apodizations, MPAX employs multiple pairs of complementary sinusoidal phase apodizations to create grating lobes with a more precise control of their magnitudes and locations.

In this study, we formally introduced MPAX and compared the performance of conventional DAS beamforming, DAX, and MPAX in terms of CNR improvement from our custom-made sponge phantom as well as our preliminary *in vivo* cardiac and abdominal imaging data sets from two human subjects. Our experimental and preliminary *in vivo* results in this study show that MPAX utilizing only 8 pairs of phase apodizations is highly robust in the presence of high levels of reverberation clutter and effectively enhances ultrasound image contrast without introducing any artifacts or undesirable features in the image. Regardless of the presence of the reverberation clutter, MPAX always resulted in higher CNR values when

compared with DAX. In addition to the higher CNR improvements, MPAX also produces images that are qualitatively better as it suppresses unwanted clutter in a much more uniform manner.

In practice, it is expected that coupling MPAX with harmonic imaging would yield additional improvements in image quality as the contrast enhancement mechanism of harmonic imaging is distinct from that of MPAX. Since beamforming needs to be performed $2N$ times in MPAX, a $2N$ -fold increase in computational burden with additional computation for 2D normalized cross-correlation is expected when compared with DAS. *In vivo* evaluation of this technique with additional human subjects is underway in both cardiology and radiology.

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Fig. 1.
Receive amplitude apodizations for 4-4 DAX.

Fig. 2.

Examples of complementary sinusoidal phase apodizations used in MPAX. Sinusoidal time delays with 2 different f_0 values are shown for $m = 3.6$ rad: left) $f_0 = 0.342$ cycles/mm, right) $f_0 = 0.635$ cycles/mm.

Fig. 3.

Diffraction efficiency $J_q^2(m/2)$ vs. $m/2$ for grating orders $q = 0, 1$, and 2 (reproduced from Goodman [28]). Fig. 4. MPAX System Diagram.

Fig. 4. illustrates a system diagram that describes the data processing steps in MPAX. The main steps in MPAX are summarized as follows:

Fig. 5.

Simulated lateral beamplots for (a) 4-4 DAX and (b) MPAX. A beamplot for conventional DAS beamforming is shown with solid lines as control in both cases. Beamplots shown with dotted and dashed-dot lines are generated by RX apodizations while those shown with dashed lines are the final beamplots after clutter suppression.

Fig. 6. Experimental results from a custom sponge phantom (top) without and (bottom) with a copper wire mesh for DAS, DAX, and MPAX. All images are displayed on a 60 dB dynamic range.

Fig. 7. DAX (left column) and MPAX (right column) weighting matrices from a custom sponge phantom (top) without and (bottom) with a copper wire mesh.

Fig. 8.
Effect of number of phase apodization pairs on the performance of MPAX in terms of contrast-to-noise ratio.

Fig. 9.

In vivo images of (top) the apical 4-chamber view and (bottom) the subxiphoid view at end-diastole for DAS, DAX, and MPAX. All images are displayed on a 60 dB dynamic range.

Fig. 10.

In vivo images of the inferior vena cava for DAS, DAX, and MPAX. All images are displayed on a 60 dB dynamic range.

Table I

Field II Simulation Parameters

Parameters	Value
Total Number of Elements	64
Center Frequency	2.5 MHz
Bandwidth	50 %
Azimuthal Element Pitch	0.32 mm
Elevation Element Height	13 mm
Sound Speed	1540 m/s
Transmit Focus	7 cm
Angular Beam Spacing	0.46°

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Table II

Summary of CNR values for Experimental Custom Sponge Phantom

		DAS	DAX	MPAX
Sponge Phantom	Without copper wire mesh	6.0	16.0	16.4
	With copper wire mesh	2.7	6.6	10.5

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Table III

Summary of CNR values for *in vivo* imaging

	Imaging View	Target	DAS	DAX	MPAX
Cardiac Imaging	Apical	LV	1.6	3.5	4.8
		RV	1.1	1.6	1.7
	Four-chamber	LV	3.8	5.3	5.9
		RV	3.1	4.9	7.3
	Subxiphoid	LV	3.8	5.3	5.9
		RV	3.1	4.9	7.3
Abdominal Imaging	Long Axis	IVC	2.8	5.0	5.7