Harmonic source wavefront aberration correction for ultrasound imaging

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A method is proposed which uses a lower-frequency transmit to create a known harmonic acoustical source in tissue suitable for wavefront correction without *a priori* assumptions of the target or requiring a transponder. The measurement and imaging steps of this method were implemented on the Duke phased array system with a two-dimensional (2-D) array. The method was tested with multiple electronic aberrators [0.39π to 1.16π radians root-mean-square (rms) at 4.17 MHz] and with a physical aberrator 0.17π radians rms at 4.17 MHz) in a variety of imaging situations. Corrections were quantified in terms of peak beam amplitude compared to the unaberrated case, with restoration between 0.6 and 36.6 dB of peak amplitude with a single correction. Standard phantom images before and after correction were obtained and showed both visible improvement and 14 dB contrast improvement after correction. This method, when combined with previous phase correction methods, may be an important step that leads to improved clinical images.

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I. INTRODUCTION

Pulse-echo ultrasound is in widespread use as a diagnostic tool. However, there are a significant number of patients that do not make high-quality ultrasound images. Often these patients have a thick layer of fat that results in a degraded image due to increased scattering, wavefront distortion, and increased attenuation. The distortions, or phase aberrations, include not just timing or phase delays, but also amplitude and waveshape distortions.¹ Imaging systems generally operate on the assumption that all tissues in the body have the same sound speed. Variations in sound speed for different tissues degrade the imaging beam, leading to a loss in resolution, contrast, and image dynamic range.² As imaging frequency increases, the subsequent decrease in wavelength causes arrival-time variations to become more significant, and these more significant arrival-time variations result in increased focusing errors. It is known that by applying wavefront correction that the imaging beam can be improved. The problem has been to determine the necessary correction functions for the imaging beam.³ Many wavefront correction methods developed^{4–9} benefit greatly from the presence of a specific acoustical target, such as a point reflector or an active ultrasound source acting as a transponder. Point reflectors are very rarely encountered during a diagnostic scan, however, and more recent wavefront correction methods have been extended to better utilize more general acoustical targets such as speckle regions.^{10–12} A new approach which seeks to create a known acoustical source and broaden the applicability of previous methods is described here.

In media allowing non-linear acoustic propagation, such as water and biological tissue, harmonics of the transmit frequency develop along with the fundamental beam.^{13,14} These

harmonics develop as the fundamental propagates, with the lower-frequency fundamental acting as a three-dimensional (3-D) source for the higher frequency harmonics. For diagnostically relevant acoustical pressures, harmonic levels in the focal plane are proportional to the fundamental beam according to

$$D_n(\theta) = [D_1(\theta)]^n; \quad \text{Generally for } \sigma < 1, \quad (1)$$

where σ indicates the shock number, n indicates the harmonic number, and $D(\theta)$ indicates the amplitude-normalized beam directivity pattern.¹⁵ Hence, the second harmonic beam is proportional to the square of the fundamental beam. Because it is proportional to the square of the fundamental, the unaberrated second harmonic beam is always narrower than the fundamental beam. On the other hand, an unaberrated second harmonic beam will be wider than an unaberrated beam transmitted directly at the frequency of the second harmonic. As an example, consider the measured beam plots of Fig. 3(a): The main lobe of the beam transmitted at 4 MHz is narrower than the 4 MHz harmonic beam generated from a 2 MHz transmit. In the measured beam plots of Fig. 3(c) we see an example of how aberration can affect the beams. Simulations have shown that when aberration is present, the beam transmitted directly at 4 MHz is more susceptible to focusing errors than the second harmonic 4 MHz beam, which is generated from the lower-frequency 2 MHz fundamental beam.¹⁶ Simulations by Varslot *et al.*¹⁷ have shown that aberration of the harmonic beam is closely tied to aberration of the fundamental beam. Based on this previous simulation work, we hypothesize that the second harmonic beam produced by a less aberrated fundamental can be used as an operator positionable source point for wavefront correction and in conjunction with two-dimensional (2-D) arrays permits improved B-mode imaging at twice the fundamental frequency.

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Aberrating layers in the body are generally distributed in two-dimensions; however, typical medical ultrasound transducers used to correct wavefront distortions are one-dimensional (1-D) arrays of elements, with spacing between elements typically less than one wavelength. While the elements are small in the direction of the array (azimuthal direction), they are over 50 times larger in the elevation dimension. For any waveform distortion correction method to function well, the wave's distortion should be constant across the surface of the transducer element.³ As a result of spatial variation of the aberrating layer, distortions may be constant over only a few elements of the array in the azimuthal direction.^{18,19} However, because most elements are many times taller than they are wide, the distortion may not be constant across a single element's elevation dimension, severely limiting the ability of 1-D arrays to correct *in vivo* wavefront distortions regardless of the correction method used.²⁰ Even 1.5- and 1.75-D arrays, which split the large elevation aspect of 1-D elements into three to eight separate elements, are still 5–15 times larger in the elevation than azimuth dimension,²¹ again potentially hampering correction performance. For adequate wavefront correction, the single correction applied to each element must adequately compensate for the wavefront distortion measured across the entire element, which means that *in vivo* wavefront distortion correction requires a true 2-D array with element size less than the spatial variations of the aberrator.

In this work we evaluate the performance of the harmonic source wavefront aberration correction method paired with a simple cross-correlation delay estimator. Correction of electronic aberrators is performed in a water tank using point, specular, and speckle correction targets. Correction of a physical aberrator is accomplished using multiple off-axis correction targets. Finally, correction based on the received second harmonic signal is compared to correction based on the received fundamental.

II. METHODOLOGY

A. Ultrasound system

All measurements were conducted using T5, the Duke University phased array scanner, and a custom built 2-D array. The Duke University ultrasound scanner T5 was entirely constructed at Duke University as a test bed system for advanced ultrasound imaging experiments, and is capable of real-time 3-D scanning. The system features 512 independent transmit channels with programmable transmit waveforms, and 1024 independent receive channels with 32-to-1 parallel processing. System bandwidth is in excess of 9 MHz with radio frequency (RF) sampling at 50 MHz. The system is capable of synthetic aperture imaging and features a highly capable and flexible display system. Multiple images can be acquired and displayed simultaneously, real-time volume-rendered images can be manipulated interactively and various cut-planes of volumes can be displayed. Echo data from each of the receive channels can be accessed easily to permit measurements necessary to implement wavefront corrections. All beam steering and focusing data is under software control and wavefront corrections are determined and calculated

FIG. 1. (a) Transducer bandwidth, with response at 2.08 and 4.16 MHz marked with circles. (b) A six-cycle transmitted pulse at the focal point (left) has developed significant, spectrally separated harmonics as seen in the spectrum (right).

using a separate computer system and then are downloaded into the T5 beam steering computer.

The 2-D matrix array was sampled by 512 active elements in a checkerboard pattern to form a roughly circular aperture of diameter 15.3 mm. All elements operated in both transmit and receive. Elements were square with dimension 0.3 mm; active elements were spaced 0.6 mm in azimuth and elevation. Measurements of the main lobe and its vicinity for the transmit–receive beam were made in a water tank with a hydrophone serving as a point reflector. The array's bandwidth was such that transmit and receive was possible at both 2.08 and 4.16 MHz with less than 5 dB loss relative to the array's center frequency of 3.57 MHz [Fig. 1(a)]. A six-cycle pulse was transmitted by T5 at either 2.08 or 4.17 MHz; a 4.16 MHz transmit was not possible due to limitations in hardware. A six-cycle pulse was chosen because it was narrowband enough to permit separation of the 2.08 MHz fundamental and 4.16 MHz harmonic by bandpass filtering [Fig. 1(b)], and to improve the signal-to-noise ratio (SNR). Bandpass filtering was performed digitally by the separate computer data acquisition system using a 183-tap filter centered at 4.16 MHz with a Hamming window to suppress the 2.08 MHz signal by > 50 dB. Table I summarizes

TABLE I. Peak amplitude of the 4.16 MHz harmonic beam relative to the 2.08 MHz fundamental for increasing focal distance.

Focal distance (mm)	20	40	60	80
Harmonic level (dB)	−25.00	−19.06	−16.56	−14.69

the measured level of the harmonic generated by the fundamental for the T5 ultrasound system. It can be seen that as focal distance increases, the harmonic has more distance over which to develop, resulting in increased amplitude relative to the fundamental. A focal distance of 60 mm was chosen in transmit and receive as a typical midrange distance in medical images.

B. Aberrators

Two types of aberrators were utilized: A physical aberrator placed in the near-field of the transducer and electronic aberrators implemented as focusing errors applied to the beam steering data in transmit and receive. The physical aberrator consisted of a slab of room temperature vulcanizing (RTV) rubber ($c = 940$ m/s, $\alpha = 10.7$ dB/cm at 4.16 MHz) 3 mm thick on average with 2-D random thickness variations of about 0.5 mm randomly distributed across it. The RTV aberrator was placed parallel to and contacting the face of the array. The electronic aberrators acted as single phase screens, and were calculated by assigning focusing delay errors to each element from a uniform random distribution. Fifteen aberrators each were generated from one of five uniform random distributions ranging from 0 to $2\pi/3$, π , $4\pi/3$, $5\pi/3$, and 2π radians at 4.17 MHz. Respectively, the aberrators resulted in root-mean-squared (rms) phase errors of 0.39π , 0.58π , 0.78π , 0.97π , and 1.16π radians at 4.17 MHz. A total of 75 electronic aberrators were created: 15 aberrators $\times$ 5 aberration delay ranges. The same focusing errors were applied in both transmit and receive.

C. Beam plots

Beam plots were measured in a water tank at 22°C ($c = 1480$ m/s) using the beamformed receive signal from T5. The receive RF signal was displayed on an Agilent DSO-6034-A digital oscilloscope. Measurement of the 2.08, 4.16, and 4.17 MHz beam levels was made using the displayed fast Fourier transform (FFT) of the receive signal. Lateral beam plots were measured by sweeping a hydrophone, acting as a point reflector, laterally in the azimuthal plane using a translation stage. A fixed focus of 60 mm was used in transmit and receive while measurements were made in lateral increments of 0.25 mm. Beam plots were measured for the unaberrated beam, beams with each of the electronic aberrators applied, and for beams with correction applied.

D. Reflectors

Ultrasound medical images result from reflections from a mixture of targets. These include point reflectors, specular reflectors, and regions of pure speckle. In order to successfully correct medical images, harmonic correction must be able to correct signals received from each type of target. Point reflectors provide the most straightforward source for correction because the signal received by each transducer element is the same except for a time-delay corresponding to the focusing error. Specular reflectors are more challenging than point reflectors, since the beam expands from the reflecting surface non-spherically. For tightly focused transmit beams, reflections originate from a small point-like

region at the focal point and can be approximated by a spherical receive surface. This is not the case for more distant foci where the focal spot becomes larger. Also, aberrations will cause beam spreading making the spherical approximation even less valid. Clearly, using a lower-frequency and the second harmonic work toward mitigating the aberration-induced beam spread. Speckle consists of a large number of sub-resolution scatters in an insonified volume and reflections from these regions are the most challenging source for correction. Reflections from speckle can produce both wave-shape distortions and time-delays in the signal received by transducer elements, even from tightly focused transmit beams. However, the tighter the transmit focus, the smaller the region of speckle that will reflect the transmitted signal and thus improve correction effectiveness.⁵

The reflections of the transmitted beam from various targets is the received signal used to determine the delays necessary to compensate for focusing errors introduced by the electronic or physical aberrators. The point reflector consisted of the tip of a hydrophone (radius = 1 mm) placed in the water tank at 60 mm from the transducer. The specular reflector consisted of a sheet of Lucite ($c = 2650$ m/s) approximately 24 mm thick placed in the water tank so that the first surface was parallel to the transducer face and 60 mm away. A custom phantom constructed by Professor Ernie Madsen, University of Wisconsin–Madison, with a large number of sub-resolution scatterers ($c = 1540$ m/s, $\alpha = 0.5$ dB/cm/MHz) was used as a speckle target. Data were recorded from a speckle region 60 mm away from the transducer.

E. Wavefront correction

The wavefront correction process consisted of two phases: measurement and imaging. During the first phase a six-cycle measurement pulse was transmitted at 2.08 MHz. This relatively low-frequency pulse was minimally affected by the aberrator and through non-linear propagation produced a minimally aberrated 4.16 MHz harmonic component.²² The fundamental and harmonic pulses reflected from a target or speckle region [Fig. 2(a)], and the 4.16 MHz harmonic pulse was isolated and used to determine the aberrating delays present. The second transmit event consisted of a two-cycle imaging pulse transmitted and received at 4.17 MHz with corrective time-delays, determined from the received harmonic pulse, compensating for the aberrating delays [Fig. 2(b)].

To determine the effectiveness of wavefront correction, lateral beam plots were measured for the 2.08, 4.16, and 4.17 MHz signals without aberration and with each of the electronic aberrators applied. A point reflector, specular reflector, or speckle region was then placed at the peak of the unaberrated beam. Each electronic aberrator was then applied and 512 six-cycle transmits at 2.08 MHz using the full transmit aperture were made. For each transmit pulse, 1 of the 512 receive channels was digitized at 50 MHz for 20 μ s; the 20 μ s recording window included the focal point. After 512 transmits, the entire receive aperture had been measured and digitized. This measurement process was

FIG. 2. Diagram of experimental setup. Wavefront correction consisted of two phases, (1) measurement and (2) imaging.

repeated until a total of 16 measurements for each receive element were obtained. These measurements were averaged together to reduce noise. The averaged signals were bandpass-filtered at 4.16 MHz to isolate the harmonic signal from the 2.08 MHz fundamental. The aberrating delays were estimated using a multi-lag least-means-squares cross-correlation algorithm

$$\mathbf{T} = (\mathbf{M}^T \mathbf{M})^{-1} \mathbf{M}^T \mathbf{D}, \quad (2)$$

where $\mathbf{T}$ is the aberrating delay estimate vector, $\mathbf{D}$ is the differential time shift vector between pairs of elements, and $\mathbf{M}$ is the model matrix relating arrival times to time shift estimates between element pairs.^{23,24} The matrix $\mathbf{M}$ is determined from array geometry and specifies which elements are spatially near each other. The vector $\mathbf{D}$ can be determined by any of the many arrival-time estimation techniques in existence; cross-correlation of signals from neighboring elements was implemented as it is both straightforward and effective.²⁵

Correction of aberrating time-delays was implemented by subtracting the resultant delays, $\mathbf{T}$, from the current aberration function. Proper correction results in a decrease in magnitude of the focusing error. A 4.17 MHz beam was then transmitted and received in the water tank with correction applied. Lateral beam plots of the corrected 4.17 MHz beam were made in the water tank using the corrective delays from each electronic aberrator and correction target combination. An additional correction iteration was accomplished for the 1.16π radian aberrator-point correction target combination by applying corrective delays to the 2.08 MHz transmit and repeating the delay estimation process. The delays calculated from the second correction iteration were subtracted from the combined electronic aberrator-first iteration correction, and beam plots measured. Further correction iterations were not investigated, as they would lead to a significant decrease in frame rate for a clinical implementation of aberration correction.

F. Off-axis correction

The majority of harmonic source wavefront aberration corrections were performed with the correction target located on the imaging beam axis (0° elevation, 0° azimuth). The region of the scan area over which the focusing delay errors are relatively constant is termed a target isoplanatic patch, and can be corrected with a single set of delays. Aberrators encountered in a clinical setting generally have multiple target isoplanatic patches,²⁶ thus harmonic source wavefront aberration correction must be able to correct using targets off the imaging beam axis. The ability of harmonic source wavefront aberration correction to correct using off-axis correction targets was evaluated using a finite thickness physical aberrator composed of RTV paired with a point correction target. Correction was performed at seven locations: point target at -18° , -12° , -6° , 0° , 6° , 12° , 18° azimuth, all at 0° elevation. No modification of the harmonic source wavefront aberration correction method was necessary for off-axis correction; the insonification beam was simply steered to the desired correction region. To evaluate the improvement in image quality from correction, B-mode images of a Gammex-RMI phantom (Middleton, WI) were captured with the RTV aberrator present and correction from each of the seven locations applied. A corrected image was created by taking the scan lines around the correction point from each of the seven off-axis corrected B-modes and combining those scan lines to form a composite image.

G. Harmonic versus fundamental correction

SNR is an important factor to be considered in any correction scheme. It is measured by calculating the average correlation coefficient (ρ) for the array using the relationship²⁷

$$\text{SNR} = \sqrt{\frac{\rho}{1 - \rho}}. \quad (3)$$

The correlation coefficient for each element is determined for the signal received from successive measurements; the average correlation coefficient for the array is the average of these element correlation coefficients.

The 4.16 MHz harmonic beam is more point-like than the 2.08 MHz fundamental beam, but suffers from decreased SNR relative to the fundamental. To determine the relative importance of the opposing effects of beam width and SNR, the correction performance of the 2.08 MHz fundamental was compared to that of the 4.16 MHz harmonic beam. To determine whether fundamental or harmonic correction performed better when using the T5 scanner with the 2-D transducer, correction of the electronic aberrators was performed using a region of the speckle phantom as a correction target. A modification to the harmonic wavefront correction method, i.e., the substitution of the fundamental for the harmonic beam, was used to perform the correction. The 2.08 MHz fundamental signal was bandpass-filtered from the received signal and fed to the delay estimation algorithm instead of the 4.16 MHz signal. Peak amplitude of the corrected 4.17 MHz imaging beam was measured for both fundamental and harmonic correction.

To further investigate the impact SNR had on correction performance, correction was performed using the 2.08 MHz fundamental and the 4.16 MHz harmonic signals of equal SNR. Decreasing the SNR of the 2.08 MHz fundamental to the level of the 4.16 MHz harmonic simulates the case where overall system SNR is high enough that the increased SNR of the 2.08 MHz fundamental over the 4.16 MHz harmonic is not significant. Comparing correction performance between the 2.08 MHz fundamental and 4.16 MHz harmonic before and after SNR equalization will indicate the impact SNR has on correction.

Fifteen spatially uncorrelated electronic aberrators with 1.16π radians rms focusing error were corrected using a specular reflector as a correction target. Correction was first performed using the 4.16 MHz harmonic, and the SNR of the 4.16 MHz harmonic was measured. Transmit amplitude was then decreased until the SNR of the 2.08 MHz fundamental beam was equal to the value measured for the 4.16 MHz harmonic beam. Correction using this decreased transmit amplitude was then performed using the 2.08 MHz fundamental.

III. RESULTS

A. Aberrated beam plots

As the focusing error magnitude of the electronic aberrator increased, the quality of the 4.17 MHz imaging beam decreased much more rapidly than the 2.08 MHz fundamental beam or 4.16 MHz harmonic beam (Fig. 3). The unaberrated beam plots in Fig. 3(a) show the 4.17 MHz imaging beam (measured and simulated with Field II²⁸), the measured 2.08 MHz fundamental beam, and the measured 4.16 MHz generated harmonic beam. There is a close match between the measured and simulated unaberrated imaging beams; the mismatch of the right side lobe is likely due to a misalignment of the transducer and beam plot axes. The beam plots in Figs. 3(b) and 3(c) show measurements from a single example of

FIG. 3. (Color online) Plot of 4.17 MHz imaging beam measured (dashed circle) and simulated (solid), 2.08 MHz fundamental (solid diamond), and 4.16 MHz harmonic (dashed triangle). Beams are normalized to their peak value and plotted (a) with no aberration, (b) with a 0.58π radian rms (at 4.17 MHz) electronic aberrator applied, and (c) with a 1.16π radian rms electronic aberrator applied.

aberrators of 0.58π radians rms and 1.16π radians rms (at 4.17 MHz), respectively. An increase in relative side-lobe level accompanied by a decrease in beam amplitude and dynamic range as focusing error magnitude increases is seen. These effects are more pronounced for the 4.17 MHz imaging beam than for the 4.16 MHz harmonic beam, especially when stronger aberration is applied. Comparing the effect of focusing errors on peak amplitude for the 4.17 MHz imaging beam, the 2.08 MHz fundamental beam, and the 4.16 MHz harmonic beam (Fig. 4), it can be seen that increasing focusing errors cause a decrease in amplitude for the fundamental, harmonic, and imaging beams. However, the harmonic beam is less

FIG. 4. Plot of average aberrated peak beam level for the 4.17 MHz imaging beam (dashed circle), the 4.16 MHz harmonic beam (dashed triangle), and the 2.08 MHz fundamental beam (solid diamond). Beams are normalized to their unaberrated value. Values from fifteen beams were averaged for each point; error bars represent a range of 2 standard deviations (SDs).

affected by focusing errors than the imaging beam because it is generated from the lower-frequency 2.08 MHz fundamental.

B. Reflector correction

Harmonic correction was very successful when dealing with reflections from a point reflector. Figure 5 shows that even with a severely aberrated beam, significant correction is achieved in just the first correction iteration (33.4 dB). Looking down the plot of the aberrated RF [Fig. 6(a)], the received signal is nearly identical for each channel except for a shift in time, making the correlation-based correction algorithm an ideal choice. After correction [Fig. 6(b)], a

FIG. 5. (Color online) Plot of the unaberrated (square), and one example of the electronically aberrated (circle) and corrected (diamond and asterisk) 4.17 MHz imaging beams using a point reflector as a correction target.

FIG. 6. (a) Aberrated and (b) corrected RF data measured from the point reflector after bandpass filtering to extract the 4.16 MHz harmonic signal. The aberrated RF signal in (a) corresponds to the aberrated curve in Fig. 5, while the corrected RF signal in (b) corresponds to the first correction iteration curve in Fig. 5.

well-focused signal appears as alternating dark and light vertical bands between samples 25 and 175. The slight horizontal shifts present in the corrected signal are due to residual focusing errors.

A clear relationship can be seen in Fig. 7 between the rms focusing error and the amplitude of the corrected beam; when smaller focusing errors are present in the measurement beam, the corrected imaging beam more closely approaches the amplitude of the unaberrated beam. An additional correction iteration was investigated to determine if the reduced focusing errors of a corrected beam could be used as a measurement beam to further improve correction. The most aberrated beam (1.16 π radian rms error) was corrected using a point reflector, and then retransmitted as a measurement beam. This less aberrated measurement beam was again corrected (additional 3.9 dB), resulting in a beam closely approaching the unaberrated beam (Fig. 5). The results for a second correction iteration with a point reflector agree with the iterative correction study by Måsøy *et al.*,²⁹ and suggest that the imaging beam will rapidly approach the unaberrated beam after only a few correction iterations.

Harmonic correction on the specular reflector was slightly less effective than the point reflector. The relationship between aberrated peak beam level and the resulting corrected peak beam level (Fig. 7) is similar to that for point reflector correction except for a downward shift indicating less accurate focusing error estimation and thus correction. This decrease is expected for the specular reflector because, unlike the point reflector, the signal received by each transducer element originates from a distributed region due to

FIG. 7. (Color online) Plot of average peak beam level relative to the unaberrated beam for the 4.17 MHz imaging beam when a point target (square), specular reflector (diamond), and speckle region (x) were used as targets to correct the aberrated beam (circle). Values from fifteen beams were averaged for each point; error bars represent a range of 2 SDs.

finite transmit focal spot size. The compact focus of the harmonic transmit beam insonifies a small region of the specular reflector in the focal plane, leading to a received signal that is not entirely spherical.

The speckle target presented the most challenging setup for the corrective delay estimation algorithm. The cross-correlation delay estimator used is not ideally suited to process RF data from speckle regions.⁵ In addition to problems associated with beam size as in the case of the specular reflector, the speckle target had lower reflectivity than either the point or specular reflectors. A lower reflectivity for the speckle reflector means that significantly less signal is received by the transducer than for the point or specular reflector cases, thus decreasing the SNR available for correction measurement. The reduction in SNR from reduced reflectivity leads to less accurate corrective delay estimation by the correlation algorithm.³⁰

C. Imaging correction

B-mode images were used to illustrate the effect of aberration and correction on image quality. However, the 2-D array used in these experiments can also be employed to acquire 3-D images, and it is anticipated that the described wavefront correction technique will also improve 3-D images. The effect of aberration and correction on B-mode images is shown in Fig. 8. All images were formed using a two-cycle 4.17 MHz pulse focused at 60 mm in transmit and receive. It should be noted that 4.17 MHz falls outside the transducer's typical imaging bandwidth. The aberrated image [Fig. 8(c)] was formed by applying an electronic aberrator with 1.16π rms focusing errors to transmit and receive. Because a single set of focusing errors was used, a single correction could be applied in transmit and receive regardless of scan angle or depth.

Nearly all the features of the RMI phantom such as (1) the anechoic cyst, (2) the resolution targets, (3) the high-contrast cysts, and (4) the range targets become obscured by the presence of aberration. Application of corrective delays from each of the reflectors significantly improves the image. The point reflector provides the most effective correction, making it possible to easily view the anechoic cyst, resolution targets, and high-contrast cyst, while the range targets can be seen with some difficulty [Fig. 8(d)]. The corrections from the specular reflector and speckle regions result in significant image improvement over the aberrated image, but are not quite as effective as the point reflector, with the anechoic cyst and resolution targets easily seen while the high-contrast cysts can be seen with some difficulty [Figs. 8(e) and 8(f)]. The overall improvement in image quality for the correction targets follows the same trends seen in Fig. 7 of improvement in peak beam amplitude, which can be considered a proxy for SNR.

Analysis of the interior of the anechoic cyst and neighboring speckle region measured the contrast that is provided by the corrected imaging beams. The results in Table II clearly show that correction increases contrast, making the interior of the anechoic cyst darker while brightening the

FIG. 8. (a) B-mode images of an RMI phantom showing unaberrated image and (b) position of the phantom targets. (c) Electronically aberrated images as well as corrections obtained from a (d) point reflector, (e) specular reflector, and (f) speckle reflector are shown. Each range marker represents 1 cm. Brightness of unaberrated, aberrated, and corrected images was normalized.

TABLE II. Contrast between anechoic cyst and surrounding speckle before and after corrective delays from different correction targets are applied.

Image		Contrast (dB)
Unaberrated	[Fig. 8(a)]	23.15
Aberrated	[Fig. 8(c)]	1.26
Point target corrected	[Fig. 8(d)]	15.49
Specular reflector corrected	[Fig. 8(e)]	12.92
Speckle region corrected	[Fig. 8(f)]	9.96

neighboring speckle. The greatest contrast improvement occurred when a point target was used as a correction target, followed by improvement seen with use of a specular reflector as a correction target, and finally with use of a speckle region as a correction target. It should be noted that the values in Table II were calculated on the detected data, not the brightness-compensated data displayed in Fig. 8.

D. Off-axis correction

The aberrated and corrected images of the RMI phantom with the RTV aberrator present in Fig. 9 clearly show that off-axis correction was effective at restoring image quality. (1) The resolution targets, (2) anechoic cyst, and (3) range targets are all visibly improved when off-axis correction is applied to the RTV aberrated scan. Comparing the corrected B-mode image with the unaberrated B-mode image, the most conspicuous difference is the greater brightness of the unaberrated B-mode. This brightness difference is due largely to attenuation of the imaging beam by the RTV aberrator and is not a byproduct of correction.

Off-axis correction was successful at compensating for the focusing errors that varied with scan angle caused by the finite thickness RTV aberrator. While harmonic source wavefront aberration correction can be applied at any scan angle, correction performance is expected to degrade as the angle between the correction target and the imaging beam axis increases. This is due to the angular response of the transducer elements,

$$V_i = \varepsilon P_i \cos \theta_i, \quad (4)$$

where P_i is the acoustic pressure wave incident at element i , V_i is the output voltage produced by element i , ε is piezoelectric conversion factor between acoustic pressure and voltage, and θ_i is the angle between element i and the source of acoustic pressure P_i . The angular response described by Eq. (4) is ideal, the angular response of most transducer elements degrades more rapidly than $\cos \theta$. Therefore, correction performed off-axis will suffer from reduced transmit and receive amplitude due to the element angular response. These reduced amplitudes lead in turn to reduced SNR, degrading correction performance. It should be noted, however, that grating lobes, a common concern when steering a beam off-axis, are greatly suppressed by the non-linear generation of the second harmonic beam.

E. Harmonic versus fundamental correction

With the current experimental setup, correction using the 2.08 MHz fundamental was actually more effective than when using the 4.16 MHz harmonic, despite the larger, less point-like region insonified by the fundamental beam. The improved performance of the 2.08 MHz fundamental beam was due to its increased SNR over the 4.16 MHz harmonic, which is characteristic of non-linear acoustics. The Cramer–Rao lower bound is the limit on the smallest variance an estimator can obtain and for a cross-correlation estimator is given by

$$\sigma(\tau) \leq \sqrt{\frac{3}{2f_0^3 \pi^2 T (B^3 + 12B)} \left(\frac{1}{\rho^2} \left(1 + \frac{1}{\text{SNR}^2} \right)^2 - 1 \right)} \quad (5)$$

where τ is the estimated delay, f_0 is the imaging center frequency, T is the observation time, B is the -6 dB fractional bandwidth, ρ is the correlation coefficient between two signals, and SNR is the signal-to-noise ratio of the data.³⁰ According to Eq. (5), increased SNR reduces the theoretical lower bound of error estimates. However, Table III shows that increases in SNR led to diminishing improvements in correction, both in terms of remaining residual error and peak beam amplitude. In particular, Table III shows that in

FIG. 9. B-mode images of an RMI phantom showing (a) an unaberrated image, (b) an image with the RTV aberrator applied, and (c) an image with off-axis correction applied. Off-axis correction was performed at seven locations between -18° and 18° . Each range marker represents 1 cm. Brightness of unaberrated, aberrated, and corrected images was normalized.

TABLE III. Comparison of SNR, measured rms error, and peak beam amplitude for correction using the 2.08 MHz fundamental and 4.16 MHz harmonic. A speckle region was used as a correction target.

Aberration (radians at 4.16 MHz)	SNR (dB)		Residual error (ns)		Corrected amplitude (dB)	
	Fundamental	Harmonic	Fundamental	Harmonic	Fundamental	Harmonic
1.16π	21.48	6.32	44.0	64.9	-9.3	-22.6
0.97π	23.22	7.10	37.3	58.6	-6.7	-16.6
0.77π	23.54	7.88	31.5	51.7	-4.7	-11.4
0.58π	24.32	8.53	26.0	39.2	-3.2	-6.0
0.39π	25.56	9.17	20.9	31.5	-2.2	-3.6

the presence of weak aberrators, where the SNR of both the fundamental and harmonic beams is relatively high, the difference between correction using the 2.08 MHz fundamental and correction using the 4.16 MHz harmonic becomes quite small. Thus, for a clinical implementation of harmonic source wavefront aberration correction that provides sufficiently high values of SNR, the increased SNR of the 2.08 MHz fundamental over the 4.16 MHz harmonic should have negligible impact on correction performance.

Indeed, when SNR of the 2.08 MHz fundamental and 4.16 MHz harmonic signals is equalized, Table IV shows correction using the 4.16 MHz harmonic is more effective than correction using the 2.08 MHz fundamental. The correction target used, a specular reflector, provided a good tool with which to evaluate quality of the fundamental and harmonic beams. As the focus of a beam narrows, the area of the specular reflector that is insonified shrinks, becoming more and more like a point target. As the area of the specular reflector that is insonified becomes more like a point target, the received signal increasingly assumes the shape of a spherical wave, improving performance of the delay estimation algorithm. Thus, the improved correction performance of the 4.16 MHz harmonic proves that the harmonic beam creates a more point-like source than the 2.08 MHz fundamental.

IV. DISCUSSION

Harmonic correction was shown to improve imaging beam amplitude for reflections arising from point targets, specular reflectors, and speckle regions. Improvements in imaging beam amplitude translated to improvements in overall image quality, as seen in Fig. 8. Harmonic correction was also shown to be effective when off-axis correction targets were utilized. As expected, point targets proved to be the most effective in permitting accurate delay measurements. Specular targets were slightly less effective than point targets but still resulted in correction amplitudes very close to the unaberrated beam. Reflections from regions of pure speckle were the most challenging. The reduction in SNR

from the decreased reflectivity for the speckle target reduced the accuracy of the correction estimation algorithm.

The results of the speckle reflector highlight the importance of signal level in determining corrective delays. In general, the second harmonic is at least 12 dB lower in amplitude than the fundamental, with additional signal losses present due to frequency dependent attenuation, transducer frequency response, and aberration. When dealing with speckle regions, where there is little to reflect a stronger transmit beam, it may be necessary in 2-D arrays to group elements together in 2×2 or 3×3 grids to increase received signal, though groups will need to be small enough that distortion is constant across all the elements.^{31,32} Multiple correction iterations can be used to compensate for aberration by increasing transmit beam amplitude thus increasing the harmonic signal received by the array elements. However, additional correction iterations decrease imaging frame rate. A clinical implementation of aberration correction will need to be near real-time, making use of more than a single aberration measurement and correction iteration unlikely. Thus, improving the quality of the first aberration measurement and correction is essential.

With the current ultrasound system, correction utilizing signal from the 2.08 MHz fundamental was more effective than correction utilizing signal from the 4.16 MHz harmonic. The increased SNR of the fundamental compared to the harmonic, a byproduct of non-linear propagation, resulted in more accurate corrective delay estimation for the fundamental than for the harmonic. In an attenuating environment, this difference in SNR will be increased due to increased attenuation of the higher frequency harmonic relative to the fundamental. However, the second harmonic will be less susceptible to attenuation than a pulse transmitted at the same frequency as the generated second harmonic. The increased SNR of the fundamental compared to the harmonic is only significant when the overall SNR of the ultrasound system is low at the fundamental and harmonic frequencies. For the experimental setup used, low SNR resulted from use of a transducer designed to operate at 3.57 MHz transmitting and receiving at 2.08 and 4.16 MHz. The results of correction with equalized SNR, simulating the case where the SNR gap between fundamental and harmonic is small relative to overall SNR, demonstrated that the harmonic provides superior correction to the fundamental. We estimate that if overall system SNR can be improved by 14 dB, such as by utilizing a transducer designed specifically for harmonic imaging, the increased SNR of the fundamental will no longer result in a significant advantage in

TABLE IV. Comparison of SNR, estimated rms error, and actual rms error for correction using the 2.08 MHz fundamental and 4.16 MHz harmonic. A speckle region was used as a correction target.

	2.08 MHz Fundamental	4.16 MHz Harmonic
Corrected peak amplitude (dB) ($\pm$ SD)	-14.6 ($\pm$ 2.0)	-11.5 ($\pm$ 1.1)

correction effectiveness compared to the harmonic. It must be noted that the SNR required for harmonic correction to surpass fundamental correction is highly dependent on ultrasound system properties as well as the delay estimator used. Additionally, simulations by Kaupang³³ suggest that biological aberration can be frequency dependent, providing additional incentive to use the second harmonic for aberration correction as it is the same frequency used for imaging. With reasonably high SNR for both the 2.08 MHz fundamental and 4.16 MHz harmonic, results show correction is improved by using signal from a more tightly focused 4.16 MHz harmonic beam for delay estimation instead of signal from the 2.08 MHz fundamental.

For the B-mode images in Fig. 8 aberration from an electronic aberrator acting as a phase screen was compensated using a single set of corrective delays regardless of scan angle or range. However, in general imaging situations a single correction is overly simplistic as focusing delay errors can vary with scan line angle and range.²⁰ For proper correction of a B-mode or 3-D image, harmonic correction must be performed for each target isoplanatic patch in the scan area. In the presence of a distributed aberrator, target isoplanatic patches may be valid for only a single focal point, necessitating aberration measurement and correction (as described in Fig. 2) to be performed at each of the many focal points in the scan. Correction at multiple target isoplanatic patches can be accomplished with this harmonic technique since the fundamental beam can be steered in azimuth and elevation and focused at a desired range, as was seen in the off-axis correction experiments.

It is important to note that the harmonic correction technique is separate from the correction estimation algorithm and can be used with correction estimation methods other than the correlation-based method used in these experiments. In addition to different delay estimation techniques, refinements of the harmonic correction technique are also possible. There is the potential to start with a very low transmit frequency and use it to correct for a much higher transmit frequency. Such a scheme could be accomplished by pumping significant energy into the low-frequency transmit beam so that higher-order harmonics ($n = 3, 4, \dots$) are generated and used as correction sources. In addition to correcting for higher frequencies, these higher harmonics will have more tightly focused beams than the second harmonic [Eq. (1)]. Another method would use the second harmonic of the initial low-frequency pulse to correct for a second pulse. The second pulse would then generate a harmonic beam that would be used to correct for a third pulse ($4 \times$ the frequency of the initial pulse), and so on. Finally, there is no reason harmonic correction can only be used to correct timing delays; with a harmonic source it is also possible to correct for amplitude distortions, which awaits further exploration.

The harmonic correction scheme described has effectively corrected for electronic and physical aberrators when point, specular, and speckle targets were chosen to reflect the second harmonic beam. Correction was effective utilizing both on-axis and off-axis correction targets. These experiments demonstrate that harmonic source wavefront aberration correction using the second harmonic is feasible.

Comparison of correction performance between fundamental and harmonic correction indicated that the SNR of the current experimental setup is too low to benefit from the more point-like source generated by the harmonic. However, it was demonstrated that for reasonably high SNR, harmonic correction should outperform fundamental correction. The initial results from use of the second harmonic beam as a known generated wavefront correction source are promising; however, further development is needed in improving delay estimation speed, loading time for correction delays, and transducer sensitivity at the fundamental and harmonic frequencies to permit correction in real-time as will be required in medical applications. We suggest that real-time image correction may have a significant impact on image quality in hard-to-image patients and thereby further enhance the clinical utility of ultrasound.

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