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Effects of Dual Apodization With Cross-Correlation on Tissue Harmonic and Pulse Inversion Harmonic Imaging in the Presence of Phase Aberration

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Abstract

Dual apodization with cross-correlation (DAX) is a relatively new beamforming technique which can suppress side lobes and clutter to enhance ultrasound image contrast. However, previous studies have shown that with increasing aberrator strength, contrast enhancements with DAX diminish and DAX becomes more prone to image artifacts. In this paper, we propose integrating DAX with tissue harmonic imaging (THI) or pulse inversion harmonic imaging (PIHI) to overcome their shortcomings and achieve higher image contrast. Compared with conventional imaging, our experimental results showed that DAX with THI allows for synergistic enhancements of image contrast with improvements of more than 231% for a 5-mm pork aberrator and 703% for a 12-mm pork aberrator. With PIHI, improvements of 238% and 890% were observed for the two pork tissue samples. Our results suggest that the complementary contrast enhancement mechanism employed by the proposed method may be useful in improving imaging of technically difficult patients in clinics.

I. Introduction

A. Tissue and Pulse Inversion Harmonic Imaging

Among many recent developments in medical ultrasound, tissue harmonic imaging (THI), first introduced by Averkiou *et al.* in 1997 [1], is considered one of the more notable techniques and is widely used in clinics today. THI has been proven useful in detecting subtle lesions in organs such as the thyroid and breast, delineating endocardial borders, and visualization of cardiac chambers, as well as in abdominal imaging [2]–[5]. The simplest and most common method used in THI to extract harmonic content from a received echo is harmonic band filtering [1], [6], [7], in which the received signal is band-pass filtered with a center frequency equal to the second harmonic. However, this method may not completely separate the harmonic band from fundamental band because overlap between the two bands is inevitable, especially when a broadband pulse is used for better axial resolution. It is possible to reduce the amount of this overlap by generating a narrower transmit bandwidth, but at the expense of reduced axial resolution.

Another attempt to overcome the limitations of harmonic band filtering is the use of pulse inversion harmonic imaging (PIHI), which was first introduced in 1997 [8] and has been widely studied for its clinical benefits. This method requires two transmit pulses along the same propagation path, one of which is shifted 180° out of phase with respect to the other. The echoes from the two transmit firings are then summed to obtain a beam with a cancellation of signals at the fundamental frequency. Hence, with the pulse inversion technique, it is possible to remove the spectral overlap while maintaining a wide bandwidth of the transmit pulse.

Both THI and PIHI utilize nonlinear distortion of the transmitted frequencies within the body to generate an ultrasound image that is sharper and higher in contrast than the conventional B-mode image. This is primarily due to the fact that the ultrasound beam formed using the second-harmonic signal has reduced beamwidth, which leads to improved resolution [9]. Moreover, side lobes and clutter, which may be prominent at the fundamental frequency, are suppressed in tissue harmonic and pulse inversion harmonic images [10]. Reverberation artifacts are also reduced in THI and PIHI because the harmonic content of reverberation echoes is lower than signals at the fundamental frequency [11]. Finally, because the harmonic signals only pass through the aberrating layers once on receive, as opposed to twice for fundamental signals, the effects of aberration on the harmonic beam are reduced as well [10].

Despite numerous benefits that THI and PIHI offer, they may not provide sufficient contrast enhancement in many clinical situations, especially those that involve large patients who pose technical challenges. Therefore, a novel technique for improving the quality of tissue and pulse inversion harmonic images would be a valuable tool to fully benefit from the useful information provided by harmonic images.

B. Dual Apodization with Cross-Correlation

Dual apodization with cross-correlation (DAX) is a relatively new technique developed by Seo and Yen that can effectively enhance ultrasound image contrast by suppressing side lobes and clutter [12]. The novelty of this technique is in using two receive (RX) apodizations, rather than one apodization as used in conventional delay-and-sum (DAS) beamforming, to create two distinct point spread functions having similar main lobe signals but very different clutter patterns. By differentiating the clutter signals from main-lobe-dominated signals and suppressing them, DAX has shown contrast enhancements of 139% in simulation and 123% experimentally in a tissue-mimicking phantom without reduced frame rate, increased computational burden, or a sacrifice in either lateral or axial resolution [12].

Despite the promising improvements DAX has demonstrated, its performance was shown to decrease, often with undesirable image artifacts, as the aberration strength was increased [13]–[15]. This suggests that DAX could potentially lead to misdiagnosis if used to visualize large patients with thick layers of subcutaneous fat that defocus the ultrasound beam and mask the signals from deep-seated structures.

One approach to remedy this problem is to apply DAX on tissue or pulse inversion harmonic signals. Because the effects of aberration are suppressed at the second harmonic, we expect better performance of DAX without creating any artifacts. This anticipated improvement in the performance of DAX is not necessarily due to possible higher correlation between the two harmonic RF signals with different receive apodizations. Rather, the novelty of this approach is in using the two independent contrast enhancement mechanisms employed by harmonic imaging and DAX that work together synergistically in minimizing the effects of aberration, side lobes, and clutter. In addition, because harmonic imaging is arguably the standard mode for many clinical imaging applications [16], [17], performance evaluation of DAX on THI and PIHI is doubly important. Similar investigations have been made previously by Fedewa *et al.* [18], who demonstrated transmit apodizations that generate a greater spatial coherence of the backscatter at the second harmonic than at the fundamental. More recently, Dahl *et al.* [19] introduced and evaluated the performance of a harmonic version of the short-lag spatial coherence (SLSC) imaging technique, which relies on lateral spatial coherence of the backscatter to enhance image contrast [20]. In this paper, we report our findings on the performance of DAX when applied to harmonic images and demonstrate potential benefits from their complementary contrast enhancement mechanisms.

II. Methods

A. Experiments in Tissue-Mimicking Phantoms

For our experimental setup, RF data sets were acquired from an ATS ultrasound phantom (Model 549, ATS Laboratories, Bridgeport, CT) containing 6-mm-diameter anechoic cysts located at 50 mm in depth and sampled at 45 MHz for offline processing. This serves as the control data set for comparison. To mimic near-field aberrating layers, pork belly samples of thickness 5 and 12 mm were placed between the transducer and the ATS cylindrical lesion phantom. These RF signals were collected using a Verasonics data acquisition system (Verasonics, Redmond, WA) with a 128-element C4-2 curvilinear array with a pitch of 409 μm . A 2-cycle transmit pulse with a center frequency of 1.96 MHz and a subaperture size of 64 elements were used. The parameters for the C4-2 curvilinear array used in this study are summarized in Table I. To compare PIHI with conventional imaging and THI, data were collected again with a transmit pulse of the same amplitude but inverted, and were summed together to generate a new data set containing pulse inverted harmonic signals. For each acquisition, data from each channel were collected 12 times and averaged. All individual channel RF signals were band-pass filtered using a 128-tap finite impulse response (FIR) band-pass filter with fractional bandwidth of 50% for fundamental and 30% for tissue harmonic and pulse inversion harmonic data sets. Offline beamforming was then performed using Matlab (The MathWorks Inc. Natick, MA) with a constant f-number of 2. For the control data set, the transmit focus was set to a depth of 50 mm where the 6-mm-diameter anechoic cyst is located, but when a pork tissue was used to create aberration effects, the transmit focus was shifted down by the thickness of the tissue. Dynamic receive focusing was used with focal updates every 1 mm in range. The experimental imaging parameters are summarized in Table II.

B. Data Processing

Fig. 1 describes the process of generating the different images that are presented in this study to investigate the effects of DAX on THI and PIHI. The conventional B-mode ultrasound image [Fig. 1(e)] was formed by first band-pass filtering the time-delayed RF data [Figs. 1(a) and (b)] at the fundamental frequency, f_0 , summing them across the elements [Fig. 1(c)], and then performing envelope detection using Hilbert transform and log compression [Fig. 1(d)]. The tissue harmonic image [Fig. 1(e)] was formed from the same RF data in the same manner, except the RF data was band-pass filtered at the second harmonic, $2f_0$. The PI harmonic image [Fig. 1(e)] was also generated from second-harmonic signals, but the RF data was added to another set of RF data acquired with an inverted transmit pulse before band-pass filtering at the second harmonic [Figs. 1(a) and (b)]. To investigate the effect of DAX on these imaging modes [Fig. 1(g)], DAX was applied to channel RF data after band-pass filtering [Fig. 1(f)]. For all results, receive apodizations with a 10–10 alternating pattern similar in terms of wavelengths to those reported in previous studies [13]–[15] were utilized. The 10–10 alternating pattern consists of a set of two complementary apodization functions with 10 alternating elements enabled on one but disabled on the other, as shown in Fig. 2. Use of such apodization functions creates grating lobes that are 180° out of phase with respect to one another such that the main-lobe-dominated signals can be distinguished from clutter signals by means of normalized cross-correlation.

A segment size of 2.5 wavelengths was empirically chosen for zero-lag, normalized cross-correlation (NCC) between the two beamformed RF signals obtained with the two complementary receive apodization functions. By cross-correlating the two beamformed RF signals at zero lag at every axial sample location followed by a thresholding operator, a weighting matrix composed of values ranging from the threshold value $\epsilon = 0.001$ to 1 was

created. If the coefficient is greater than or equal to the threshold value $\epsilon = 0.001$, the coefficient is kept, but if the coefficient is smaller than the threshold value $\epsilon = 0.001$, it is set to 0.001 to obtain $20\log_{10}(0.001) = -60$ dB reduction in the amplitude. Because main-lobe-dominated signals, such as those coming from speckles, are highly correlated, coefficients associated with these signals are close to 1. Uncorrelated clutter signals, such as those associated with cystic lesions, have coefficients close to the threshold value of 0.001 [12]. To minimize any artifacts, the weighting matrix was median filtered. In general, smaller segment and median filter sizes tend to yield artifacts because of the random nature of speckle, whereas larger sizes have a tendency to obscure the boundaries of the anechoic cyst and make the cyst look smaller than its actual size. After investigating the performance of several segment and filter sizes, a segment size of 2.5λ and a median filter size of $2\lambda \times 4\lambda$ have been found to yield optimal results in terms of percent CNR improvement and lesion detectability. By multiplying the resulting weighting matrix by the combined RF data, DAX improves image contrast by attenuating signals dominated by clutter. Image contrast was assessed for fundamental, THI, and PIHI, each with and without DAX applied, using the equation for contrast-to-noise ratio (CNR) [21]:

$$\text{CNR} = \frac{\bar{S}_t - \bar{S}_b}{\sigma_b},$$

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 of the envelope-detected, log-compressed image in decibels.

III. Results

Fig. 3 shows experimental images of a 6-mm-diameter anechoic cyst from the ATS tissue-mimicking phantom with no aberrator using conventional imaging, THI, and PIHI with and without DAX. Similar sets of images of the same anechoic cyst acquired with 5-mm and 12-mm pork tissues are shown in Figs. 4 and 5, respectively. All images are shown on a 50 dB dynamic range. The CNR values were calculated and are shown in Table III. The CNR values for conventional imaging decreased from 5.32 with no aberrator to 3.42 for 5-mm pork tissue, and to 0.94 for 12-mm pork tissue. DAX increased the CNR value to 9.15 for no aberrator, 4.75 for 5-mm pork tissue, and 1.84 for 12-mm pork tissue. The CNR values for tissue and pulse inversion harmonic images with no aberrator were 6.35 and 7.05, respectively, and were reduced to 6.30 and 6.31 with a 5-mm pork aberrator. Further degradation in image quality was observed with a 12-mm aberrator; these CNR values were lowered to 3.84 and 4.31. DAX with THI achieved synergistic enhancements of image contrast, yielding a CNR of 13.10 with no aberrator, 11.32 with a 5-mm pork aberrator, and 7.55 with a 12-mm pork aberrator. DAX with PIHI resulted in a CNR of 16.18 with no aberrator, 11.57 with a 5-mm pork aberrator, and 9.31 with a 12-mm pork aberrator.

IV. Discussion

For different levels of aberration, our results consistently showed that DAX increases CNR for conventional imaging, THI, and PIHI. For all percent CNR improvements presented in this study, conventional imaging with no DAX serves as the reference. As shown in Table III, the benefit of using THI or PIHI over conventional imaging in terms of contrast enhancement is roughly 30% or less. However, when DAX is used, THI and PIHI both achieved CNR values above 13.

The utility of THI and PIHI becomes more apparent when a weak aberration is introduced. Because of the suppression of aberration effects at the second harmonic, THI and PIHI

restored some of the image contrast, with CNR improvements of 84% and 85%, respectively. When combined with PIHI, DAX suppressed most of the remaining aberration effects and achieved a CNR of 11.57. DAX combined with THI resulted in a CNR of 11.32, which is comparable to, but slightly lower than that with PIHI because of the spectral overlap between the fundamental and the second-harmonic bands that could not be completely separated by harmonic band filtering, as shown in Fig. 6. For THI, a filter bandwidth of 30%, as opposed to 50% for conventional imaging, was used to minimize the contributions of the spectral overlap to the harmonic images. If a larger bandwidth was chosen for the band-pass filter such that the spectral overlap is not removed, it could lead to ripple-like artifacts in the tissue harmonic images. A filter bandwidth of 30% was also used for PIHI to be consistent with THI. The CNR values for DAX combined with THI and PIHI were both higher than that for DAX combined with conventional imaging, which resulted in a CNR of 4.75.

Finally, the value of using DAX on THI or PIHI is evaluated in the presence of strong aberration effects from a 12-mm pork tissue. Using conventional imaging, the cyst is hardly visible with a CNR of 0.94. In this case, DAX achieves a 96% improvement in CNR, resulting in a CNR of 1.84. However, the circular shape of the cyst cannot be clearly identified and small cyst-like artifacts are observed in the surroundings. Despite the CNR improvements of 309% and 359% achieved with THI and PIHI, the cyst was still masked by aberration effects, and the CNR was kept below that of an unaberrated cyst. Thus, THI, PIHI, and DAX with conventional imaging all failed to fully suppress effects of the 12-mm pork and improve image contrast without creating unwanted artifacts and cyst shape distortions. However, DAX combined with THI showed a greater CNR improvement of 703%, with most of the aberration effects suppressed within the cyst. Also, the cyst was circular in shape and free of any artifacts that were seen with conventional imaging. PIHI combined with DAX showed a CNR improvement of 890% and resulted in a circular cyst with high lesion detectability that was free of any artifacts within and around the cyst.

Because harmonic imaging generally suffers from low SNR because it is based on a fraction of the transmitted pulse energy that has been converted into second-harmonic signals, adverse effects on the performance of DAX may be present because of low SNR of harmonic signals. Cross-correlation between two beamformed RF signals having decreased SNR would result in a wider range of decreased cross-correlation coefficients. Hence, DAX, in its current form, may be over-attenuating main-lobe-dominated harmonic signals at increased depth. However, a previous study by Seo and Yen [13] has shown that a CNR improvement of more than 100% compared with conventional DAS beamforming was achieved so long as a system SNR of 15 dB or higher was maintained. In addition, application of a weighting function as a function of depth similar to the concept of time-gain compensation (TGC) to the cross-correlation coefficient matrix may be offered as a possible solution to this problem.

V. Conclusion and Future Work

In this work, we compared the performance of DAX with conventional imaging, THI, and PIHI in terms of CNR improvement in the presence of different levels of aberration. We have shown that decreased performance of DAX with increasing aberration strength could be overcome with THI or PIHI as reduced aberration effects on harmonic beams allow DAX to better distinguish main-lobe-dominated signals from side-lobe-dominated signals. The improvements in cystic contrast, as well as the visualization of the lesion boundaries, obtained using the proposed method were presented in this work.

In conclusion, we demonstrated in this study that DAX with THI or PIHI enhances ultrasound image contrast. Our results show that even in the presence of a strong aberration from a 12-mm pork tissue sample, both THI and PIHI greatly benefit from DAX with CNR improvements of over 703% and 890%, whereas conventional imaging yields 96% with irregularly-shaped cysts and unwanted artifacts. Although the best performance is expected with PIHI, THI with harmonic band filtering would still allow for a decent amount of enhancement in image contrast even in the presence of a strong aberrator. Future work will involve evaluating the performance of the proposed method *in vivo* and making modifications to DAX parameters for maximal improvement.

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References

1. Averkiou, MA.; Roundhill, DR.; Powers, JE. A new imaging technique based on the nonlinear properties of tissues. Proc. IEEE Ultrasonics Symp; 1997. p. 1561-1566.
2. Kornbluth M. Native tissue harmonic imaging improves endocardia border definition and visualization of cardiac structures. J Am Soc Echocardiogr. 1998; 11(7):693–701. [PubMed: 9692526]
3. Shapiro RS, Wagreich J, Parsons RB, Stancato-Pasik A, Yeh H, Lao R. Tissue harmonic imaging sonography: Evaluation of image quality compared with conventional sonography. AJR Am J Roentgenol. 1998; 171(5):1203–1206. [PubMed: 9798848]
4. Szopinski KT, Pajk AM, Wysocki M, Amy D, Szopinska M, Jakubowski W. Tissue harmonic imaging: Utility in breast sonography. J Ultrasound Med. 2003; 22(5):479–487. [PubMed: 12751859]
5. Szopinski KT, Wysocki M, Pajk AM, Slapa RZ, Jakubowski W, Szopinski M. Tissue harmonic imaging of thyroid nodules: Initial experience. J Ultrasound Med. 2003; 22(1):5–12. [PubMed: 12523604]
6. Christopher T. Finite amplitude distortion-based inhomogeneous pulse echo ultrasonic imaging. IEEE Trans Ultrason Ferroelectr Freq Control. 1997; 44(1):125–139. [PubMed: 18244110]
7. Ward B, Baker AC, Humphrey VF. Nonlinear propagation applied to the improvement of resolution in diagnostic medical ultrasound equipment. J Acoust Soc Am. 1997; 101(1):143–154. [PubMed: 9000731]
8. Chapman, CS.; Lazenby, JC. Ultrasound imaging system employing phase inversion subtraction to enhance the image. US Patent. 5 632 277. May 27. 1997
9. Starritt HC, Duck FA, Hawkins AJ, Humphrey VF. The development of harmonic distortion in pulsed finite-amplitude ultrasound passing through liver. Phys Med Biol. 1986; 31(12):1401–1409. [PubMed: 3809241]
10. Desser TS, Jeffrey RB. Tissue harmonic imaging techniques: Physical principles and clinical applications. Semin Ultrasound CT MR. 2001; 22(1):1–10. [PubMed: 11300583]
11. Averkiou MA. Tissue harmonic imaging. Proc IEEE Ultrasonics Symp. 2000; 2:1530–1541.
12. Seo C, Yen JT. Sidelobe suppression in ultrasound imaging using dual apodization with cross-correlation. IEEE Trans Ultrason Ferroelectr Freq Control. Oct; 2008 55(10):2198–2210. [PubMed: 18986868]
13. Seo C, Yen JT. Evaluating the robustness of dual apodization with cross-correlation. IEEE Trans Ultrason Ferroelectr Freq Control. Feb; 2009 56(2):291–303. [PubMed: 19251516]
14. Shin, J.; Yen, JT. Improved image quality using phase aberration correction and dual apodization with cross-correlation. Proc. IEEE Ultrasonics Symp.; 2011. p. 393-396.
15. Shin J, Yen JT. Synergistic enhancements of ultrasound image contrast with a combination of phase aberration correction and dual apodization with cross-correlation. IEEE Trans Ultrason Ferroelectr Freq Control. Sep; 2012 59(9):2089–2101. [PubMed: 23007784]

16. Thomas JD, Rubin DN. Tissue harmonic imaging: Why does it work? *J Am Soc Echocardiogr.* 1998; 11(8):803–808. [PubMed: 9719092]
17. Varslot T, Masoy SE, Johansen TF, Angelsen B. Aberration in nonlinear acoustic wave propagation. *IEEE Trans Ultrason Ferroelectr Freq Control.* Mar; 2007 54(3):470–479. [PubMed: 17375817]
18. Fedewa RJ, Wallace KD, Holland MR, Jago JR, Ng GC, Rielly MR, Robinson BS, Miller JG. Spatial coherence of the nonlinearly generated second harmonic portion of backscatter for a clinical imaging system. *IEEE Trans Ultrason Ferroelectr Freq Control.* Aug; 2003 50(8):1010–1022. [PubMed: 12952092]
19. Dahl JJ, Jakovljevic M, Pinton GF, Trahey GE. Harmonic spatial coherence imaging: An ultrasonic imaging method based on backscatter coherence. *IEEE Trans Ultrason Ferroelectr Freq Control.* Apr; 2012 59(4):1203–1211.
20. Dahl JJ, Hyun D, Lediju M, Trahey GE. Lesion detectability in diagnostic ultrasound with short-lag spatial coherence imaging. *Ultrason Imaging.* Apr; 2011 33(2):119–133. [PubMed: 21710827]
21. Krishnan S, Rigby KW, O'Donnell M. Improved estimation of phase aberration profile. *IEEE Trans Ultrason Ferroelectr Freq Control.* 1997; 44(3):701–713.

Fig. 1.

The process of forming conventional, tissue harmonic (TH), and pulse inversion harmonic (PIH) images with and without dual apodization with cross-correlation (DAX).

Fig. 2.
The 10–10 alternating apodization functions used for dual apodization with cross-correlation (DAX).

Fig. 3. Experimental cyst images for conventional imaging at fundamental frequency, tissue harmonic imaging (THI), and pulse inversion harmonic imaging (PIHI): (bottom row) with and (top row) without dual apodization with cross-correlation (DAX) with no pork aberrator.

Fig. 4.

Experimental cysts for conventional imaging at fundamental frequency, tissue harmonic imaging (THI), and pulse inversion harmonic imaging (PIHI): (bottom row) with and (top row) without dual apodization with cross-correlation (DAX) in the presence of a 5-mm pork aberrator.

Fig. 5.

Experimental cysts for conventional imaging at fundamental frequency, tissue harmonic imaging (THI), and pulse inversion harmonic imaging (PIHI): (bottom row) with and (top row) without dual apodization with cross-correlation (DAX) in the presence of a 12-mm pork aberrator.

Fig. 6.

Frequency spectra of the received RF data for (left) conventional imaging at fundamental frequency of 1.96 MHz, (center) tissue harmonic imaging (THI), and (right) pulse inversion harmonic imaging (PIHI) at second harmonic. The frequency response of the band-pass filter (BPF) is also shown in all 3 cases.

TABLE I

Parameters for C4–2 Curvilinear Array.

Parameter	Value
Radius of curvature	40 mm
Azimuthal element pitch	0.409 mm
Elevation element height	13 mm
Number of elements	128
Aperture angle	75°

TABLE II

Parameters for Experimental Setup.

Parameter	Value
Transmit center frequency	1.96 MHz
Sampling frequency	45 MHz
Transmit focus	50 mm (no pork)
	55 mm (5-mm pork)
	62 mm (12-mm pork)
Lateral beam spacing	0.14° (0.225 mm at 50 mm depth)
	0.14° (0.238 mm at 55 mm depth)
	0.14° (0.255 mm at 62 mm depth)
Sound speed	1450 m/s (phantom)
Receive focal delay step	1 mm

TABLE III

Contrast-to-Noise Ratio (CNR) Values for Experimental Data.

Aberrator	No DAX		DAX	
	Conventional	THI	Conventional	PIHI
None	5.32	6.35	9.15	13.10
5-mm pork	3.42	6.30	4.75	11.32
12-mm pork	0.94	3.84	1.84	7.55
				9.31

CNR values for experimental anechoic cysts with pork aberrators are shown for conventional imaging at fundamental frequency, tissue harmonic imaging (THI), and pulse inversion harmonic imaging (PIHI) with and without dual apodization with cross-correlation (DAX). CNR values are shown for 5-mm (weak) and 12-mm (strong) pork aberrators. Results with no aberrator are also shown as a control for comparison. For each of the three cases of aberration, conventional imaging with no DAX serves as the reference value for CNR improvement.