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Effect of Transmit Beamforming on Clutter Levels in Transthoracic Echocardiography

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Abstract

Transmit beamforming has a strong impact on several factors that govern image quality, field-of-view, and frame-rate in ultrasound imaging. For cardiac applications, the visualization of fine structures and the ability to track their motion is equally important. Consequently, beamforming choices for echocardiography aim to optimize these trade-offs.

Acoustic clutter can dramatically impact image quality and degrade the diagnostic value of cardiac ultrasound imaging. Clutter levels, however, are closely tied to the choice of beamforming configuration. This study aims to quantify the impact of transmit beamforming on clutter levels under *in vivo* conditions. The performance of focused as well as plane wave transmit configurations in fundamental and harmonic modes is evaluated under matched conditions. Contrast between the cardiac chambers and the interventricular septum is used as a surrogate for the level of clutter in a given imaging scenario.

Under *in vivo* conditions, contrast was found to improve incrementally across the four beamforming configurations in the following order : fundamental-plane, fundamental-focused, harmonic-plane, and harmonic-focused. Using the fundamental-focused configuration as a reference, the harmonic-plane and harmonic-focused cases showed improvements in median contrast of 2.97 dB and 6.1 dB, respectively while the fundamental-plane case showed a contrast deterioration of 1.23 dB. Contrast was also found vary systematically as a function of imaging depth. Median contrast for the right ventricle (shallow chamber) was measured to be 2.96 dB lower than that in the left ventricle (deep chamber). The improved clutter suppression in both harmonic cases as well as the decay of clutter magnitude with imaging depth suggest that a substantial amount of clutter for transthoracic imaging is generated from reverberation between near-field tissue layers.

Keywords

echocardiography; transthoracic; clutter; beamforming; harmonic; plane wave

Introduction

Ultrasound imaging of the heart, or echocardiography, is an extremely valuable tool in clinical cardiology. It provides structural as well as functional information about the heart in a real-time, cost-effective, and convenient manner. It can be performed via transthoracic, transesophageal, or intracardiac windows. When implemented transthoracically, it is completely noninvasive and widely used for initial screening, management, as well as follow-up exams. Transthoracic echocardiography (TTE) involves the use of two- and three-dimensional imaging as well as power- and color-Doppler modes to survey the heart for abnormalities such as enlargement of chambers (1), ventricular dyssynchrony (2), and regurgitant or stenotic valves (3; 4). It is also used to derive quantitative metrics of cardiac function such as chamber volumes (5), ejection fraction (6), and flow velocities through valves (7). The diagnostic potential of echocardiography is rapidly expanding with increasing clinical acceptance and standardization of ultrasound-based modes such as tissue-doppler imaging (8) and myocardial strain imaging (9).

However, the benefits of TTE are contingent upon the ability to obtain good quality images. Imaging artifacts can present in echocardiograms as obscured or distorted structures and severely impact their diagnostic value (10, 11). The reliability and reproducibility of quantitative echocardiographic metrics of cardiac function are also drastically reduced for patients with poor image quality (12, 13). In echocardiography, image quality is generally determined based on the visibility of endocardial borders, the extent of shadowing artifacts, and the ability to discern recognizable anatomical structures at depth. Poor quality cardiac ultrasound images, often referred to as “technically difficult echocardiograms”, are associated with factors such as obesity, scar tissue (from prior injury/surgery), and lung disease. These factors result in high acoustic clutter and limited depth of penetration.

Acoustic clutter is the diffuse haze that overlays ultrasound images. It degrades border delineation and diminishes the contrast of hypoechoic or hyperechoic targets (14). It can arise as a result of three basic phenomenon: reverberation, off-axis scattering (due to side-lobes or grating lobes), and phase aberration (15). While all three phenomena likely contribute to clutter under *in vivo* conditions, their relative contributions can be difficult to differentiate. Properties of the propagation path (i.e. tissue through which an ultrasound image is acquired) is also a determinant of the magnitude and spatial extent of acoustic clutter. Heterogeneity in the speed-of-sound and presence of layered structures or specular reflectors has been shown to be associated with higher levels of acoustic clutter (16; 17; 18). In the case of echocardiography, the propagation path is complex in nature and includes several tissue layers (skin, fat, muscle, and potentially ribs); these conditions can lead to the generation of a considerable amount of reverberation clutter. This form of clutter is generated in the near-field, i.e. close to the transducer face and is likely to decay with depth. Clutter can also arise from bright off-axis structures such as the pericardium and lung boundaries. Additionally, the clutter generating structures can be relatively stationary (chest wall) or ones that move with the cardiac or respiratory cycles (pericardium and lungs) (19).

One technique that has made a tremendous impact on image quality in echocardiography is harmonic imaging. Clinically relevant image quality improvements in echocardiography

using harmonic imaging have been reported by several studies (20; 21; 22; 23; 24; 25). Harmonic imaging in ultrasound relies on non-linear propagation of sound waves through the imaging media which leads to the generation of echoes at higher harmonics of the transmit frequency (26). Echoes at these harmonic frequencies (generally second harmonic) can be isolated using band-pass filtering, pulse-inversion, or coded harmonics (27) and reconstructed into images with improved resolution and reduced clutter. In simulations, harmonic imaging has been shown to suppress clutter due to reverberation by as much as 26 dB (15).

The clutter suppression effect of harmonic imaging can be attributed to the fact that harmonics are preferentially generated in regions where the transmit beam pressure at the fundamental frequency is highest i.e., around the transmit focus in depth and the beam axis in the azimuthal and elevation dimensions. Consequently, reverberant echoes from tissue layers in the near-field as well as echoes from off-axis targets and side-lobes are suppressed (28). However, harmonic imaging comes with its own set of unique challenges. Harmonic signals are 15–20 dB lower in amplitude compared to their fundamental counterparts (29). This limits their signal-to-noise ratio (SNR) and penetration depth. Consequently, focused beams have traditionally been used for this mode to aid harmonic generation and improve SNR. It also necessitates the use of broad-band probes to capture echoes at the higher harmonics.

Aside from image quality, frame-rate or temporal resolution is also critical when imaging a rapidly moving organ such as the heart. For a fixed field-of-view (FOV) and line density, the temporal resolution of ultrasound images is dependent on the number of receive lines being reconstructed per transmit and the number of transmits required to capture the entire FOV. These parameters can be modulated through beamforming. Beamforming involves manipulating the phase and amplitude of individual elements across an array so as to control the spatial profile of transmit or receive events. In conventional two-dimensional echocardiography, focused transmit beams are swept across the FOV and used to reconstruct one receive line per transmit. This yields frame-rates of 30–80 Hz. While these frame-rates may suffice to capture the bulk motion of the heart, they are too slow to capture phenomenon such as waves propagating in the myocardium due to valve closure and dynamics of valve motion (30).

Frame-rates can be improved using parallel receive beamforming, wherein multiple receive lines (typically 4–64) are acquired simultaneously following a single transmit (31). In this configuration, the transmit beam is broadened by shrinking the active transmit aperture or using defocused beams. Multiple receive beams are then captured across the insonified region. An extrapolation of this concept is used in “ultra-fast” imaging using plane waves (32). Plane wave imaging involves using a broad transmit pulse combined with extensive parallel receive beamforming. It has the benefit of being able to reconstruct entire FOVs with a single transmit event; thereby achieving extremely high frame-rates greater than 1000 Hz. Imaging at these high frame-rates is critical for the translation of transient elastography techniques into cardiac applications (33; 34). Another approach to improving frame-rates is through the use of multi-line transmits wherein focused beams are transmitted simultaneously at several lateral locations (35).

However, this improvement in temporal performance comes at the cost of image quality. The lack of transmit focusing results in decreased resolution, lower SNR, and higher levels of acoustic clutter (36). These effects can also be attributed to lower pressure amplitudes and higher off-axis energy. While coherent summation of multiple steered plane waves has been shown to alleviate some of these challenges (37), a fundamental trade-off between image quality and temporal resolution exists.

For harmonic imaging, frame-rates have traditionally been limited due to the use of focused transmits. However, the combination of plane wave imaging and harmonic imaging has been a topic of exploration in the past few years. Recent studies have reported on the benefits of using plane waves in harmonic mode in the context of shear wave elastography (38; 39). The improvements are largely attributed to the clutter suppression effect of harmonic imaging and the improved temporal resolution due to plane wave imaging. While image quality improvements using the combined mode have been shown in phantoms and in limited *in vivo* examples, a thorough analysis of clutter levels for the various beamforming configurations has not previously been conducted.

The work presented herein aims to characterize and quantify clutter levels under *in vivo* imaging conditions for a variety of factors such as beamforming configuration, imaging depth, and TTE view. The performance of focused and plane wave imaging in fundamental as well as harmonic modes is evaluated for matched imaging conditions to assess their relative benefits. These results serve to inform decisions on beamforming trade-offs so as to optimize image quality and frame-rate for ultrasound-based techniques in cardiac imaging.

Methods

Experimental Design

Twelve healthy volunteers were recruited and imaged under an IRB approved protocol (Duke IRB Pro00032068) at the out-patient Cardiology Clinic at Duke University Hospital. Study participants spanned an age range of 22–35 years and a body mass index (BMI) range of 20–24 kg/m². Imaging was performed by an experienced sonographer in two standard transthoracic echocardiographic (TTE) views: parasternal long axis (PLAX) and parasternal short axis (PSAX). As shown in Figure 1, both views capture the interventricular septum (IVS) at a depth of 50–60 mm. The chambers of the right ventricle (RV) and the left ventricle (LV) are located shallow and deep to the IVS, respectively.

A commercially available ultrasound system, Siemens Acuson SC 2000TM (Siemens Medical Solutions Inc., Malvern, PA, USA), and a adult cardiac phased array, Siemens 4V1c, were used for imaging. The probe has a center frequency of 3.2 MHz and a bandwidth of 2.5 MHz. Custom pulse sequences were developed to acquire B-mode as well as M-mode images in a sequential fashion. B-mode images were captured in a manufacturer-optimized imaging condition native to the system for a 90° azimuthal FOV and a maximum depth of 100 mm.

M-mode images were acquired for a single transmit beam (down the middle of the B-mode) with parallel receive beamforming to acquire five receive lines across a narrow lateral extent

(1°). Clutter level was measured using all receive lines, however, the M-mode images presented herein correspond to the central angle. M-mode images were acquired for four beamforming configurations comprising combinations of fundamental and harmonic mode using focused and plane wave transmit settings. Fundamental imaging was performed at 3.2 MHz while harmonic imaging (using pulse-inversion) employed a 2 MHz transmission corresponding to a 4 MHz receive frequency. In the focused setting, the full probe aperture (18.87 mm) was used to generate a focus at a depth coinciding with the IVS, yielding F-numbers in the range of 2.65 to 3.18. For the plane wave case, the full aperture was used again, however the focus was set to 1000 mm so as to achieve a near-planar transmit wavefront. Therefore, the four beamforming configurations were, fundamental-plane (FP), fundamental-focused (FF), harmonic-plane (HP), and harmonic-focused (HF). The mechanical index of the transmit pulses used in this study are in the range of 1.01–1.65.

For each configuration, raw M-mode echo data was acquired for a duration of 5 s at a pulse repetition frequency of 4 kHz. Beamformed, in-phase and quadrature (IQ) data (i.e. focused, summed, and demodulated signals) were acquired prior to envelope detection and log compression. Each “data set” was comprised of one B-mode and four M-mode images. M-mode images (one for each beamforming configuration) were acquired in quick succession following the acquisition of the B-mode to maintain a stable field-of-view (FOV) within each data set. Each data set was acquired over a duration of 35–40 s. Volunteers were not required to hold their breath given the prolonged duration of the acquisition. 3–8 repeated acquisitions were performed on each volunteer and in each view. Five data sets with independent speckle realizations were also acquired on a ATS imaging phantom, Model 549 (ATS Laboratories Inc., Bridgeport, CT, USA). Anechoic targets at a depth of 40 mm and 60 mm were imaged through a water path as well as in the presence of a reverberant tissue layer (pork belly).

Data Processing

B-mode images were used for anatomical reference to identify specific cardiac walls on the corresponding M-mode and also to confirm that standard TTE views were obtained. Spatiotemporal regions-of-interest (ROIs) representing the IVS were manually traced on each M-mode image. ROIs were also traced shallow and deep to the IVS to represent the RV and LV, respectively. These are shown as dotted green lines in Figure 2. ROIs were carefully chosen so as to exclude cardiac anatomical features such as valve leaflets, papillary muscles, and chordae tendineae.

Contrast was measured as the ratio of the mean envelope-detected IQ data within ROIs representing the cardiac chambers and that of the IVS, as shown in Equation 1. While the ratio was computed using linear data (representing pressure signals), the contrast is presented on a logarithmic scale. On an ideal image, the IVS is hyperechoic compared to the chambers. However, contrast was defined to be a positive quantity to make the analysis intuitive.

$$C_c = -20 \log_{10} \left(\frac{\bar{v}_c(z,t)}{\bar{v}_{IVS}(z,t)} \right), \quad (1)$$

where C_c refers to measured contrast, $\bar{v}_c$ stands for the spatiotemporal mean of the signal within an ROI corresponding to a particular chamber (RV or LV), while $\bar{v}_{IVS}$ represents the spatiotemporal mean over the interventricular septum. z and t represent the spatial (depth) and temporal dimensions of the M-mode image, respectively, and c represents the specific chamber (RV or LV).

M-mode images that displayed artifacts related to breathing, such as the invasion of lung tissue into the imaging plane or the clear presence of cardiac structures in the ROIs of the RV or LV, were excluded from analysis. The final analysis was conducted on 169 data sets across the 12 volunteers. M-mode data in the fundamental-plane configuration was not acquired for 90 data sets due to experimental limitations.

Statistical Analysis

As discussed above, contrast data was acquired across three main parameters: beamforming configuration, cardiac chamber, and TTE view. A multivariate analysis was performed using a mixed model with fixed effects for the three parameters mentioned above and random effects across volunteers and repeated acquisitions. Magnitude as well as direction of the effects of the three main parameters on contrast and their two-way interactions were isolated through this analysis. Repeated measures were incorporated into the model to assess inter- vs. intra- volunteer variability.

Contrast data was modeled as:

$$C_{ijklm} = \mu + c_i + v_j + b_k + p_l + (c.v)_{ij} + (v.b)_{jk} + (c.b)_{ik} + \epsilon_{ijklm}, \quad (2)$$

where C_{ijklm} is the measured contrast for the i^{th} chamber (RV or LV), j^{th} view (PLAX or PSAX), k^{th} beamforming configuration (FP, FF, HP, or HF) from the l^{th} volunteer (1–12) and the m^{th} repeated measurement (1–8). Note that an unequal number of repeated measurements were performed for each condition. The model in Equation 2 approximates contrast in terms of a baseline mean (μ), which corresponds to the contrast of the RV, in PLAX using the fundamental-focused (FF) configuration. The model goes on to include the main effects of the chamber (c_i), the view (v_j), and the beamforming configuration (b_k). It also includes a random effect of the volunteer (p_l) and pairwise interactions between the three main factors. The random effect, p_l , is assumed to have a zero-mean Gaussian distribution with standard deviation of σ_p . Lastly, ϵ_{ijklm} represents measurement error, assumed to have an independent zero-mean Gaussian distribution with a standard deviation σ_e . The model was fit to the data

by the method of restricted maximum likelihood (REML) using the *nlme* package in the R computing platform (40).

Results

Table 1 presents mean and standard deviations of five contrast measurements made on an imaging phantom (ATS Model 549) using the four beamforming configurations. The anechoic targets were 8 mm in diameter and centered at depths of 40 mm and 60 mm. The 12 mm background speckle region between the two targets was used for reference. When imaging through a water path, the plane wave configurations showed similar performance in fundamental as well as harmonic mode with mean contrast (across the targets) of 20.67 dB and 20.88 dB, respectively. Mean contrast was significantly higher in the fundamental-focused configuration, 30.99 dB, and highest in the harmonic-focused case, 35.01 dB. Contrast for the deep target was lower than the shallow one; on average the difference between the two was 3.92 dB.

When imaging through a reverberant tissue layer (pork belly), contrast was found to be lower for all targets and in all beamforming configurations. However, mean contrast (across the targets) in this case was found to improve across beamforming configurations. Fundamental-plane showed the lowest contrast, 6.77 dB, followed by fundamental-focused, 9.97 dB. Mean contrast in harmonic-plane and harmonic-focused configurations was 10.31 dB and 19.22 dB, respectively.

Figure 2 portrays a comparison of high and low clutter imaging environments *in vivo*. B-mode and M-mode images shown here were acquired in the PSAX view on two separate volunteers. Clutter, which appears as a diffuse haze, is noticeable within the cardiac chambers. While clutter is observable to some degree across all four images, it is especially pronounced within the RV in Figure 2(a) and (b). Clutter arising from stationary sources appears in the M-mode (Figure 2(b) and (d)) as horizontal streaks through time within the RV chamber. Contrast measurements for the RV (shallow) and LV (deep) are superimposed on the M-mode images. In the high clutter case, Figure 2(b), the LV shows substantially higher contrast compared to the RV.

Figures 3 and 4 demonstrate the effect of beamforming on clutter levels in PSAX and PLAX views, respectively. Each figure shows a full data set comprising the B-mode, (a), as well as M-mode images for all four beamforming configurations, (b)–(e). Contrast measurements for both cardiac chambers as a function of beamforming configuration are tabulated below the B-mode images.

A qualitative comparison of the M-mode images, in both examples, shows similar performance for both fundamental configurations (focused and plane). A reduction in clutter is observed when using the harmonic-plane mode. The harmonic-focused configuration further enhances clutter suppression and improves image quality. Decreased clutter is accompanied by improved border delineation of the IVS and a cleaner representation of myocardial walls against the background of the hypoechoic cardiac chambers. Contrast measurements, presented in the tables, capture this reduction in clutter levels across

beamforming configuration in a quantitative sense. Comparing fundamental-plane with harmonic-focused, the mean contrast improvement (across cardiac chambers) was 7.70 dB in PSAX and 5.56 dB in PLAX.

Contrast measurements as a function of beamforming configuration, for all repeated acquisitions, and across the twelve volunteers in the study are shown in Figure 5. Within each beamforming configuration, the data are segmented by cardiac chamber; RV is presented in the light gray markers while LV is in dark gray. Box plots for the respective distributions are superimposed. The primary trend of note is that median contrast increases monotonically in response to beamforming configuration from left to right or from fundamental-plane to harmonic-focused. This effect is observed for both the RV as well as the LV. Comparing fundamental-plane to harmonic-focused, median contrast was found to increase for the RV from 6.71 dB to 13.91 dB and for the LV from 9.62 dB to 16.71 dB.

While contrast for both chambers exhibit the same trend across beamforming configuration, their distributions are distinct from one another. On average, over the four beamforming configurations, median contrast for the RV is 2.96 dB lower than that for the LV; this difference is largest in the harmonic-plane configuration.

Figure 6 illustrates the relative trend of contrast across beamforming configuration. Relative contrast was derived in a matched sense (within each data set) by referencing the raw contrast measurements using each beamforming configuration to their counterparts in the fundamental-focused configuration. Cardiac chamber specific results are presented for the RV and the LV in light and dark shades, respectively. Relative contrast distributions highlight the dominant effect of beamforming. Compared to fundamental-focused, the fundamental-plane configuration shows a subtly lower contrast; median difference of -0.79 dB. The harmonic configurations, however showed more pronounced effects with the plane and focused cases showing improvements of 2.97 dB and 5.28 dB respectively. The difference between cardiac chambers is almost negligible in the case of relative contrast across beamforming configurations (compared to absolute contrast, i.e. Figure. 5).

Figure 7 shows a scatter plot of contrast measurements for each beamforming configuration plotted against their counterpart in the fundamental-focused case, i.e. the reference mode. In this format, the abscissa indicates the contrast in the reference beamforming mode (FF, in gray), going from high clutter on the left to low clutter on the right. The ordinate indicates contrast (for a matched view from the same data set) for the other three beamforming configurations (FP, in green; HP, in red; HF, in blue). A strong positive correlation is observed between contrast in the baseline mode compared with the same in each of the other three modes. This relationship is highlighted by the linear fits derived from the multivariate model and reflected in the Pearson correlation coefficients; 0.97 for FP, 0.80 for HP, and 0.85 for HF. Vertical offsets from the diagonal indicate changes in clutter levels from the baseline. In the case of fundamental-plane, this change is subtle and in the negative direction; suggesting slightly higher clutter. Alternatively, the two harmonic modes show large positive offsets with harmonic-focused being considerably larger than harmonic-plane. This is suggestive of clutter reduction due to the use of harmonic imaging.

Contrast measurements as a function of beamforming configuration for individual study participants is presented in Figure 8. Within each participant, beamforming configurations are placed in the following order: (from left to right) fundamental-plane, fundamental-focused, harmonic-plane, and harmonic-focused. The thin gray lines represent individual data sets while the thicker black line represents the mean. Contrast values show considerable variability between multiple acquisitions within a participant as well as across different participants. However, the general trend of increasing contrast (decreasing clutter) across beamforming, going from left to right, was observed in all cases.

Multivariate statistical modeling was used to isolate the impact of individual factors, assess their interactions, and compare these to random effects. Results from this analysis are presented in Table 2. The multivariate model described by Equation 2 accounts for fixed effects of chamber, view, and beamforming configuration as well as random effects across study participants and repeated measurements. The baseline condition (row 1) for this analysis was RV, PLAX, and fundamental-focused. Rows 2–6 represent changes in contrast (from baseline) as a result of independent contributions of each individual factor. Rows 7–13 represent pairwise interaction effects; these can be interpreted as added effects due to the combined contribution of two factors relative to the sum of their individual contributions. Effects were considered statistically significant for p -values under 0.01.

The effect of view, chamber, and beamforming configuration were all found to be statistically significant with p -values less than 0.0001. However, the magnitudes and directions of these effects varied. The standard deviation of contrast measurements across study participants (σ_p) was found to be 2.51 dB; the same across repeated measurements, i.e. within study participants (σ_e) was 2.57 dB. From a baseline contrast value of 6.44 dB, contrast for the LV was found to be 5.05 dB higher. Switching to the harmonic-focused configuration was shown to result in a 5.16 dB improvement in contrast. The effect of view and the two plane wave cases (fundamental and harmonic) also yielded statistically significant, albeit smaller effect sizes of 2.2 dB, -1.94 dB, and 1.94 dB, respectively. All interaction terms modeled herein, with the exception of the interaction of view and chamber, were found to be statistically insignificant. The combined effect of switching to LV and PSAX was 3.29 dB lower than the sum of their individual contributions.

Discussion

The diagnostic value of ultrasound images can be drastically degraded in the presence of acoustic clutter. This form of image degradation can be detrimental for clinically relevant tasks where border delineation is critical such as measuring thickness of tissue layers, quantifying sizes of anechoic regions, and visualizing subtle structures such as vegetations. Artifacts due to clutter from stationary as well as moving sources can not only obscure the presentation of anatomical structures but also impede the accuracy of ultrasound-based motion tracking algorithms (41).

In echocardiography, the dominant sources of clutter are presumed to be reverberation and off-axis scattering. Echoes generated through these two mechanisms get superimposed over echoes from structures along the beam axis because they arrive at the receive aperture

simultaneously. The additive nature of clutter can, thus, make it difficult to quantify its effects under *in vivo* conditions where the FOVs consist mostly of speckle generating tissue e.g., liver or thyroid imaging. However, clutter levels can be estimated in cases where the FOV contains substantial anechoic or hypoechoic regions.

TTE provides one such opportunity; while the myocardial walls are composed of muscular structures, the cardiac chambers are largely filled with blood. At typical frequencies used for transthoracic imaging (2–5 MHz), echoes from blood are 40–100 dB lower in magnitude compared to those from myocardial walls such as the interventricular septum (IVS) (42). Thus, in the ideal “clutter-free” image the cardiac chambers would appear nearly anechoic compared to the myocardial walls. Varying levels of contrast loss and lack of image conspicuity due to clutter is evident in all M-mode images presented here. Measuring contrast between cardiac chambers and the IVS can, therefore, provide an estimate of clutter levels. A low contrast measurement is indicative of a high clutter environment and vice-versa. Moreover, depth dependent changes in clutter levels can be readily studied in TTE by comparing clutter levels in different cardiac chambers.

The impact of clutter can be mitigated through the use of beamforming. For applications such as echocardiography where motion tracking can be just as important as the ability to visualize fine structure, beamforming strategies seek to find an optimum over a complex and interrelated set of parameters that impact image quality (spatial resolution, SNR, and clutter suppression) as well as temporal resolution, and field-of-view.

Phantom Trends

In the phantom, when imaging through a water path the mean contrast for the anechoic targets using the fundamental-focused configuration was measured to be 30.99 dB. Compared to this, the fundamental-plane and the harmonic-plane cases showed substantial deteriorations of 10.32 dB and 10.11 dB, respectively, while the harmonic-focused case yielded a 4.02 dB improvement. The marked decrease in contrast for the plane wave cases compared to their focused counterparts can be attributed to an increase in the contribution of off-axis scattering when using plane waves. Harmonic imaging is expected to yield improvements primarily through the suppression of reverberant echoes (15) and to a lesser extent by limiting the contribution from the side-lobes. Given the lack of sources of reverberation in the phantom, the improvement in contrast when using harmonic-focused (compared to fundamental-focused) is most likely a consequence of the suppression of echoes from the side-lobes. However, the degradation of contrast in the harmonic-plane case suggests that, in this scenario, the increase in off-axis clutter due to the use of plane waves has a stronger impact compared to the suppression of reverberant clutter afforded by harmonic imaging. This result provides a validation of the expected relationships between the various beamforming configurations in a low reverberation environment.

In the presence of a reverberant tissue layer, mean contrast for the anechoic lesions in the fundamental-focused configuration was dramatically reduced from 30.99 dB to 9.97 dB. However, the trends across beamforming configurations in this case matched the ones observed under *in vivo* conditions. The harmonic-plane and harmonic-focused cases were shown to improve contrast by 0.34 dB and 9.25 dB, respectively; while the fundamental-

plane case showed a 3.2 dB reduction. While a monotonic trend was observed in the contrast of the shallow lesion, the deep lesion showed a slight reduction in contrast when using the harmonic-plane configuration. This could be attributed to a decrease in SNR as a function of depth due to lower harmonic generation when using plane waves. These results illustrate the effect of reverberant clutter as a function of beamforming configuration in a controlled environment.

In vivo Trends

Under *in vivo* imaging conditions, contrast values varied over a wide range for each beamforming configuration. Using the fundamental-focused configuration, contrast values ranged from -0.87 dB to 19.63 dB. Measurements varied not only across study participants but also between repeated acquisitions within a particular participant, as illustrated in Figure 8. This variability can be attributed to changes in the propagation path used for imaging. Additionally, all volunteers in this study had BMIs within a fairly narrow range. Considerably higher clutter levels (and lower contrasts) would be expected for patients with higher BMIs (14).

Using the fundamental-focused configuration, median contrast (across chambers) was found to be 9.99 dB. The fundamental-plane showed a relatively minor deterioration of 1.23 dB, while the harmonic-plane and harmonic-focused modes showed improvements of 2.97 dB and 6.1 dB, respectively, over the fundamental-focused. These results align well with the phantom results in the presence of a reverberant tissue layer.

When imaging the phantom through a water path, the harmonic-plane configuration showed a large deterioration in contrast with reference to the fundamental-focused case. However for phantom imaging in the presence of a reverberant tissue layer as well as under *in vivo* conditions it yielded an improvement in contrast. This evidence suggests that harmonic imaging using plane waves is only likely to show contrast improvements in the presence of strong reverberation clutter. For environments where reverberation is not the primary source of image degradation, the harmonic-plane configuration is likely to yield lower contrast as a result of lower SNR of the harmonic signals as well as increased contribution of off-axis echoes due to the use of plane waves.

Clutter levels for *in vivo* images were also found to decay substantially with depth away from the chest wall, i.e. clutter in the shallow chamber (RV) was substantially higher than that in the deep chamber (LV). The statistical significance and appreciable magnitude of this effect (2.96 dB) are also suggestive of the presence of strong clutter from near-field reverberations. Even though absolute clutter levels between the two chambers differed from each other; the trend of relative changes in clutter as a function of beamforming was found to be consistent across cardiac chambers (as shown in Figure 6). This suggests that beamforming driven effects dominate over other confounding factors that differentiate clutter levels between the two cardiac chambers.

The highest contrast, however, was observed in the harmonic-focused configuration; the median across chambers in this configuration was measured to be 16.09 dB. This result is consistent with the findings on the phantom and indicates that the benefits of harmonic

imaging are maximized when using focused beams. This effect can be attributed to the fact that higher pressures are generated when using a focused transmit configuration which in turn yields greater harmonic generation and thus improved SNR of the received echoes.

While the results presented portray a consistent relationship between clutter levels and beamforming configuration, there are a few confounding factors that were not addressed in this study. Given that clutter can arise from stationary as well as moving structures, its magnitude is likely to fluctuate over the cardiac cycle. Contrast data presented here were averaged over 4–5 cardiac cycles to overcome potential bias from transient changes, however, the variation in clutter levels through time was not explicitly addressed. Changes in SNR as a function of beamforming were also not studied here. SNR can be estimated on phantoms and stationary tissue by tracking speckle correlation through time. However, similar analysis on cardiac images is more likely to represent motion-induced decorrelation as opposed to the effects of SNR.

Beamforming configuration was treated as an ordinal variable in this analysis. In doing so, the interactions between fundamental vs. harmonic and focused vs. plane wave were not directly interrogated. The apparent non-linearity in the contrast improvement between adjacent pairs of beamforming configurations warrants a closer analysis into their interplay. Lastly, the comparison of focused beams and plane waves represents only a small subset over a wide spectrum of transmit beamforming configurations. Several other conditions such as diverging waves, coherent summation of steered plane waves, multi-line transmits, and other transmit configurations used in synthetic aperture techniques also aim to address the trade-offs between image quality, frame-rate, and field-of-view. Future work will look to expand the analysis of clutter levels in fundamental as well as harmonic modes to the other transmit geometries, over a wider range of BMIs, and also across more *in vivo* imaging environments such as fetal and abdominal imaging.

Conclusion

This study investigated the effects of transmit beamforming on clutter levels in the context of transthoracic echocardiography. Contrast between the cardiac chambers and the interventricular septum was used to estimate clutter levels under *in vivo* imaging conditions. Focused and plane wave transmit configurations were evaluated in fundamental as well as harmonic modes to assess their susceptibility to acoustic clutter. Trends in the magnitude of clutter were analyzed, in an absolute as well as relative sense, across parameters relevant to TTE imaging such as view (PLAX, PSAX), chamber (RV, LV) and beamforming configuration.

Clutter levels for the four transmit beamforming configurations were found to be significantly different from one another. In both, fundamental as well as harmonic modes, the focused case yielded higher contrast compared to its plane wave counterpart demonstrating the impact of off-axis scattering. Additionally, the harmonic-plane configuration outperformed the fundamental-focused case. Contrast was also found to improve with increasing depth away from the chest wall. These results suggest a strong contribution of reverberant clutter under *in vivo* transthoracic settings.

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Figure 1.

Representative B-mode and M-mode images for the two transthoracic echocardiographic (TTE) views used in the study, presented on a 35 dB dynamic range. B-mode shows a spatial map of echogenicity, while M-mode is a spatiotemporal representation along the center line of the B-mode image (dotted green line) as a function of time and spans multiple cardiac cycles. (a) and (b) show the heart in the parasternal long axis (PLAX) view while (c) and (d) depict it in parasternal short axis (PSAX). The interventricular septum (IVS) appears as the bright tissue layer at a depth of 50–60 mm in all four images. The hypoechoic regions shallow and deep to the IVS represent the chamber of the right ventricle (RV) and the left ventricle (LV), respectively.

Figure 2.

B-mode and M-mode images illustrating a high clutter imaging scenario, (a) and (b), compared to a low clutter case, (c) and (d), in the parasternal short axis view. The dotted green lines on B-mode represent the lateral position of the M-mode, while the same on M-mode represent outlines of the ROIs for IVS, RV and LV. Measured contrast for each cardiac chamber (with respect to the IVS) over the spatio-temporal ROI is indicated on the M-mode images. Low contrast is indicative of a high clutter environment and vice-versa. Contrast, not only varies between the two imaging scenarios but also between the two cardiac chambers.

Figure 3. A representative data set in parasternal short axis. (a) shows the B-mode, (b) - (e) show M-mode images for the four beamforming configurations and the table (left bottom) shows contrast measurements in RV and LV as a function of beamforming configuration. Qualitative improvements in clutter levels are evident in the sequence of M-mode images (top to bottom) while quantitative trends are captured by the contrast measurements in the table.

Figure 4. A representative data set in parasternal long axis. (a) shows the B-mode, (b) - (e) show M-mode images for the four beamforming configurations and the table (left bottom) shows contrast measurements in RV and LV as a function of beamforming configuration. The trend of improvement in contrast (over beamforming configurations) appear to be similar in both TTE views and for both cardiac chambers.

Figure 5.

Contrast measurements as a function of beamforming configuration, split by cardiac chamber. For the box plots, the horizontal marker represents the median, the edges of the box represent the 25th and 75th percentile and the whiskers mark the furthest values not considered outliers. RV (light gray markers) shows lower contrast (higher clutter) as compared to LV (dark gray markers). A monotonic trend of incremental improvements in contrast (reduction in clutter) as a function of beamforming are observed in both RV and LV.

Figure 6.

Distributions and associated box plots for relative contrast (matched to within each data set) compared to the fundamental-focused configuration for each cardiac chamber. Contrast changes for the RV are represented by lighter shades and those for the LV are in darker shades. Box plots are as describe in Figure 4. The trend of improving relative contrast as a function of beamforming is highlighted here. No significant difference in relative contrast with beamforming configuration is observed between RV and LV.

Figure 7.

Scatter plot of measured absolute contrast for each beamforming configuration against their corresponding value in the fundamental-focused case. Linear fits are superimposed. Data points along the diagonal are contrast values in the fundamental-focused mode, the region above the diagonal represents improvements in contrast while the region below it represents degradation. The fundamental-plane case shows a small degradation in contrast while the two harmonic cases show substantial improvements.

Figure 8.

Trends of contrast as a function of beamforming split by volunteer. Contrast measurements for individual data sets are indicated in the light gray lines, and the median trend is shown in black. Despite the range of contrast measurements, the trend of incremental improvements across the four beamforming configurations was found to be consistent for all volunteers in the study.

Table 1.

Contrast between anechoic targets and speckle generating background on imaging phantom (ATS Model 549)

	Contrast (dB)			
	Water Path		Reverberant Tissue Layer	
	40 mm	60 mm	40 mm	60 mm
Fund. Plane (FP)	23.53 ± 0.65	17.81 ± 1.68	6.67 ± 1.87	6.87 ± 1.47
Fund Foc. (FF)	32.59 ± 0.67	29.39 ± 0.74	8.44 ± 1.46	11.51 ± 1.68
Harm. Plane (HP)	22.93 ± 1.44	18.83 ± 0.59	13.01 ± 2.39	7.62 ± 1.37
Harm. Foc. (HF)	36.85 ± 1.36	33.17 ± 0.92	19.85 ± 1.99	18.59 ± 2.90

ROI Diameter = 8 mm, n = 5

Table 2.

Multivariate analysis with mixed effects on contrast measurements.

Term	Effect	Standard Error	t-value	p-value
Baseline	6.44	0.83	7.75	
(RV, PLAX, Fund. Foc.)				
LV	5.05	0.47	10.76	< 0.0001
PSAX	2.2	0.49	4.53	< 0.0001
Fund. Plane	-1.94	0.65	-2.99	< 0.0001
Harm. Plane	1.94	0.51	3.8	< 0.0001
Harm. Foc.	5.16	0.51	10.09	< 0.0001
<i>LV : PSAX</i>	-3.29	0.45	-7.36	< 0.0001
<i>LV : Fund. Plane</i>	0.5	0.73	0.69	0.49
<i>LV : Harm. Plane</i>	0.66	0.58	1.14	0.26
<i>LV : Harm. Foc.</i>	0.15	0.58	0.27	0.79
<i>PSAX : Fund. Plane</i>	1.12	0.72	1.55	0.12
<i>PSAX : Harm. Plane</i>	1.08	0.57	1.9	0.06
<i>PSAX : Harm. Foc.</i>	0.09	0.57	0.16	0.87

RV = Right Ventricle, LV = Left Ventricle

PLAX = Parasternal Long Axis, PSAX = Parasternal Short Axis