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In Vivo Application of Short-Lag Spatial Coherence Imaging in Human Liver

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Abstract

We present the results of a patient study conducted to assess the performance of two novel imaging methods, namely Short-Lag Spatial Coherence (SLSC) and Harmonic Spatial Coherence Imaging (HSCI), in *in vivo* liver environment. Similar in appearance to the B-mode images, SLSC and HSCI images are based solely on the spatial coherence of fundamental and harmonic echo data, respectively, and do not depend on the echo magnitude. SLSC and HSCI suppress incoherent echo signals and thus tend to reduce clutter. The SLSC and HSCI images of 17 patients demonstrate sharper delineation of blood vessel walls, suppressed clutter inside the vessel lumen, and show reduced speckle in surrounding tissue compared to matched B-modes. Target contrast and contrast-to-noise ratio (CNR) show statistically significant improvements between fundamental B-mode and SLSC imaging and between harmonic B-mode and HSCI imaging (in all cases $p < 0.01$). The magnitude of improvement in contrast and CNR increases as the overall quality of B-mode images decreases. Poor quality fundamental B-mode images (where image quality classification is based on both contrast and CNR) exhibit the highest improvements in both contrast and CNR (288 % improvement in contrast and 533 % improvement in CNR).

Keywords

Ultrasound; Ultrasonic Imaging; SLSC; HSCI; spatial coherence; short-lag; *in vivo* liver imaging; clutter

Introduction

Ultrasound is commonly used for surveillance and diagnosis of liver diseases because of its high specificity (75-98%), relatively low cost, and lack of ionizing radiation (Bolondi et al., 2001; Larcos et al., 1998; Sherman, 1999; Mishra and Younossi, 2007; Mottin et al., 2004; Chen et al., 2002). In many hospitals, asymptomatic patients with an incidental elevation of liver enzymes receive periodic abdominal ultrasound scans in an attempt to diagnose early onset of non-alcoholic fatty liver disease (NAFLD), the most common form of liver disease (Mishra and Younossi, 2007; Kirovski et al., 2010). Hepatorenal echo contrast, liver tissue brightness, and vascular blurring are indicators typically used to detect and stage NAFLD (Mishra and Younossi, 2007). In a similar manner, the American Association for the Study

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of Liver Diseases has recommended periodic ultrasound exams as a part of standard surveillance procedure for hepatocellular carcinoma (HCC) (Bruix and Sherman, 2010).

For both NAFLD and HCC, the role of ultrasound in diagnosing and staging the disease is sometimes limited by inadequate image quality. Most of the patients that are at risk of or are already suffering from NAFLD are categorized as over-weight or obese (based on their body mass index) and may yield low quality images. A study done on morbidly obese patients reports the sensitivity of ultrasound in diagnosing hepatic steatosis (fatty liver disease) to be only 49.1% (Mottin et al., 2004). The growing population of overweight and obese patients is also under increased risk of developing HCC. The American Association for the Study of Liver Diseases acknowledges that these patients are often not good candidates for HCC surveillance (Bruix and Sherman, 2010), where the reported values for sensitivity are as low as 65% (Chen et al., 2002). The body habitus of overweight and obese patients introduces a variety of challenges in ultrasonic imaging, one of which is an increased level of acoustical noise known as clutter.

Clutter is a significant reason for diminished quality of ultrasound images of the liver (Lediju et al., 2008; Pinton et al., 2011). It appears as an overlaying haze or fill-in that obscures *in vivo* structures such as lesions and blood vessels and leads to inadequate visualization of liver anatomy and function. The three main sources of clutter that have been identified are phase aberration, reverberation among tissue layers, and off-axis scattering (Pinton et al., 2011). Many of these sources of clutter are present in overweight and obese individuals due to thick layers of subcutaneous fat and fat/connective tissue interfaces. As a result, there is a growing clinical need for ultrasound imaging techniques that would reduce clutter or improve image quality and allow for more patients to be effectively surveyed for liver diseases such as NAFLD or HCC.

Harmonic imaging and various phase aberration correction methods are examples of techniques developed to reduce clutter. Harmonic imaging is based on nonlinear wave propagation, which generates frequency components at integer multiples of the transmitted frequency. This phenomenon is negligible close to the surface of the transducer, which results in a reduction of near-field reverberation clutter (Christopher, 1997; Averkiou et al., 1997). However, harmonic imaging does not suppress clutter originating from phase aberration; in fact, it is more susceptible to phase aberration than fundamental B-mode due to the higher receive frequency (Pinton et al., 2011). Phase aberration correction methods can reduce side-lobe levels and the effects of off-axis scattering (Flax and O'Donnell, 1988; Nock et al., 1989; Fink, 1992; Liu and Waag, 1994). However, their potential to improve quality of fundamental B-mode images is limited because of overlaying reverberation clutter (Pinton et al., 2011).

Adaptive imaging methods based on spatial coherence of the pressure field present an alternative approach to improving image quality. Walker and Trahey developed a translating aperture technique to improve speckle correlation and thus facilitate near-field phase aberration correction (Walker and Trahey, 1997). Mallart and Fink introduced the Coherence Factor, defined as a ratio of the coherent sum to the incoherent sum of the received echoes, in an attempt to quantify focusing characteristics of ultrasonic imaging systems (Mallart and Fink, 1994). The Generalized Coherence Factor (Li and Li, 2003) is a coherence measurement similar to the Coherence Factor and was developed primarily to suppress phase-aberrated signals. The Phase Coherence Factor and the Sign Coherence Factor are based on the variance of phase of the received echoes across the aperture and are intended to reduce the off-axis scattering (Camacho et al., 2009). The Generalized Coherence Factor, Phase Coherence Factor, and Sign Coherence Factor are all used to weight the B-mode on a pixel-by-pixel basis in order to improve B-mode image quality.

We have recently developed a beamforming method called Short-Lag Spatial Coherence (SLSC) imaging (Lediju et al., 2011) that suppresses the incoherent portion of the backscattered ultrasound field and has the potential to reduce clutter. We have also adapted SLSC imaging to the backscattered harmonic field and called it Harmonic Spatial Coherence Imaging (HSCI) (Dahl et al., 2011b). Both SLSC and HSCI are based purely on the spatial coherence of backscattered echoes and do not depend on the echo magnitude. We have demonstrated through simulations that SLSC imaging performs as well as conventional B-mode imaging in detecting lesions under noise-free conditions while it dramatically improves lesion detectability in a noisy environment (Dahl et al., 2011a). Sample SLSC and HSCI images of *in vivo* liver, heart, and kidney also show improved visibility of targets and indicate that SLSC and HSCI could have significant clinical impact (Dahl et al., 2011b). In this paper, we report the results of a pilot patient study conducted to assess the clinical performance of SLSC and HSCI in liver imaging. Image metrics (contrast and CNR) are computed for fundamental B-mode, harmonic B-mode, SLSC, and HSCI images to quantify and compare visibility of *in vivo* targets across different imaging techniques. Based on our previous results, we expect SLSC and HSCI to demonstrate greater improvement in low quality fundamental and harmonic B-mode images, respectively, because acoustical noise is not present in significant amounts in high-quality B-mode images (Dahl et al., 2011a).

Methods

Short-Lag Spatial Coherence Imaging

In pulse-echo ultrasound imaging, the pulse scattered from a point in tissue spreads out as it propagates back to the transducer. The spatial coherence of the echo is a measure of how similar the echo is at any two points in space. The spatial coherence is often expressed as the spatial correlation which can be estimated by

$$\widehat{R}(m) = \frac{1}{N-m} \sum_{i=1}^{N-m} \frac{\sum_{n=n_1}^{n_2} s_i(n) s_{i+m}(n)}{\sqrt{\sum_{n=n_1}^{n_2} s_i^2(n) \sum_{n=n_1}^{n_2} s_{i+m}^2(n)}}. \quad (1)$$

Here, N is the total number of elements in the array, $s_i(n)$ is the sampled signal on the i^{th} element of the array, n is a sample number in the time dimension, and the difference $n_2 - n_1$ is typically on the order of a wavelength, in samples. m is the spatial lag given in terms of number of elements. Due to the variances of the signals $s_i(n)$ and $s_{i+m}(n)$ that appear in the normalizing term of the equation 1, the spatial correlation R does not depend on the echo magnitude.

To facilitate a better understanding of spatial coherence, the spatial correlation can be thought of as a continuous function of lag l , where $l = m/N$ to make R a function of a fraction of the transmit aperture width, with maximum lag equal to 1. An image metric, called the short-lag spatial coherence value, V_{slsc} , is then defined as the integral of $\widehat{R}(l)$:

$$V_{slsc} = \int_0^Q \widehat{R}(l) dl \approx \sum_{m=1}^M \widehat{R}(m) \Delta m \quad (2)$$

where Q is typically in the range of 5 – 30 % of the transmit aperture width, corresponding to the region of short lags. The approximation on the right hand side is used to implement calculation on real data. The short-lag coherence metric V_{slsc} is used as the pixel brightness in the SLSC image. To construct a complete SLSC image, V_{slsc} is computed according to equations 1 and 2 at every depth and image line across a field of view.

Application of the SLSC imaging method to the second-harmonic signals is called Harmonic Spatial Coherence Imaging (HSCI). Second-harmonic signals are obtained by performing either band-pass filtering (Christopher, 1997) or pulse inversion (Chapman and Lazenby, 1997) on the channel signals. It is important to note that the order of operations in HSCI is different than in harmonic B-mode imaging, where the harmonic signals are computed after the channels signals have been delayed and summed.

Clinical study

Unprocessed, radio-frequency (RF) ultrasound data was collected *in vivo* on liver vasculature, common bile duct, and other hypoechoic/anechoic hepatic structures of interest in 17 patients who were referred to the Department of Radiology, Duke University Hospital for ultrasound exam of the abdomen in the time period between September 2010 and September 2011. Both hospitalized patients and out-patients were included in the study. Written consent was obtained from all participants and the study was approved by the local Institutional Review Board.

Data was collected using a Siemens 4C1 transducer (Siemens Medical Solutions USA, Inc., Issaquah, WA) attached to a modified Siemens Acuson S2000 Ultrasound scanner (Siemens Medical Solutions USA, Inc., Issaquah, WA). The transducer used a 2.5 MHz center frequency with a 60 % fractional bandwidth. The individual channel signals needed to create SLSC and HSCI images were recorded using a custom synthetic-receive-aperture sequence in combination with the Axius Direct™ Ultrasound Research Interface (Brunke et al., 2007). The transducer had a total of 192 elements but, due to the limited system programmability, individual-channel RF data was saved for only 64 elements from 54 A-lines, where the 64 elements were centered about the transmit aperture for each beam. For comparison, conventional beamformed RF signals were acquired concurrently with the single-channel-acquisition sequence to construct full B-mode images. Full aperture (192 elements) was used on the transmit end with the transmit frequency set to 1.82 MHz and the focal depth adjusted to optimally capture the liver vasculature and other structures of interest.

Harmonic channel data was obtained using the pulse-inversion method (Chapman and Lazenby, 1997). Individual channel data was collected for positive and negative transmit pulses and the two data sets were added together to produce a harmonic channel data set. The signals obtained from the positive transmit pulse were used to construct fundamental B-mode and SLSC images. Because the single-channel acquisition sequence for harmonic data was developed after the start of the study, harmonic data was collected for the final 12 of the 17 subjects.

Data Processing and Statistical Analysis

SLSC and HSCI images were created from the fundamental and harmonic individual channel data, respectively, using equations 1 and 2. The size of the correlation kernel used to compute the spatial correlation function ($n_2 - n_1$ in Eq. 1) was chosen to be one wavelength in order to maintain axial resolution similar to B-mode imaging and, at the same time, yield stable coherence functions. Contrast and contrast-to-noise ratio (CNR) were calculated in the B-mode and SLSC/HSCI images for the structures of interest according to the equations below:

$$Contrast = -20 \log_{10} \left(\frac{S_i}{S_o} \right), \quad (3)$$

$$CNR = \frac{S_i - S_o}{\sqrt{\sigma_i^2 + \sigma_o^2}} \quad (4)$$

In equations 3 and 4, S_i and S_o are the mean signal magnitudes of the same-size regions inside and outside of the lesion, respectively, and σ_i^2 and σ_o^2 are corresponding signal variances. User-selected regions inside and outside the anatomy of interest were the same for the matched B-mode and SLSC/HSCI images.

Based on contrast and CNR values, B-mode images were classified as high, medium, or poor quality. High quality images were classified as having contrast higher than 10 dB and CNR higher than 1.1. Low quality images were classified as having either contrast lower than 6 dB or CNR lower than 0.8. Any image that did not fall into one of the two categories was classified as medium quality. The above criteria were determined empirically. In our experience, high quality images had large values for both contrast and CNR, and either contrast less than 6 dB or CNR less than 0.8 (or both) were enough to result in inadequate or poor visualization of the target and yield a low quality image.

Hypothetical improvements in contrast and CNR between different imaging methods were investigated using a matched-pair t-test. To compare fundamental B-mode to SLSC, $n = 17$ matched sample pairs (from all subjects) were used to calculate the test statistic. To compare harmonic B-mode to HSCI, SLSC to HSCI, and fundamental B-mode to harmonic B-mode $n = 12$ matched sample pairs (from the subjects that had harmonic data) were used to find the test statistic. Adjusted p values were calculated by using Holm-Bonferroni method to control the type I error rate arising from multiple testing (Holm, 1979). Adjusted p values of less than 0.05 were considered to indicate a significant difference. In addition, the magnitude of improvement was computed for each tested sample. Samples were also divided into high, medium, and poor quality images and improvements were computed for individual categories as well.

Results

An example of a matched B-mode and SLSC image is presented in Figures 1 through 4. Because a limited amount of individual channel data can only be saved in order to create an SLSC image (due to programmability limitations of the ultrasound scanner), the SLSC image covers a narrower field of view than the B-mode image. For display purposes, the SLSC/HSCI images are overlaid on the corresponding B-mode images between the arrows. B-mode images are displayed on a decibel scale while a linear gray-scale colormap is used to display SLSC/HSCI images. The dynamic range of the B-mode images were chosen to optimally visualize structures of interest while the background brightness of the SLSC and HSCI images were scaled to match that of the B-mode images. The regions used to calculate the contrast and CNR are indicated by white contour lines in the B-mode images (the same regions are used in the matched SLSC/HSCI images).

Figure 1 demonstrates a high quality fundamental B-mode image (left) and its corresponding SLSC image (right). A large blood vessel appears at the center of field of view in both images. In the B-mode image, the lumen of the vessel contains low-level acoustic noise. In the SLSC image, the clutter in the lumen is reduced and the surrounding tissue is smoother, which results in slightly higher contrast and CNR; 13.5 dB contrast and 1.86 CNR obtained for the SLSC image compared to 11.3 dB contrast and 1.4 CNR obtained for the B-mode image.

An example of a medium quality fundamental B-mode and its corresponding SLSC image is shown in Figure 2. A blood vessel spanning the depth of the image at the center of field of view can be seen in both images, with the SLSC image demonstrating significant improvements in target visualization. Contrast and CNR values were calculated to be 8.9 dB and 2.05 for SLSC imaging, respectively, and 7.8 dB and 0.95 for B-mode imaging. Matched harmonic B-mode and HSCI images of the same structure in Figure 2 are shown in Figure 3. Following the criteria (for contrast and CNR) outlined in section II, the harmonic B-mode image is classified as medium quality. The overall level of noise in the image is reduced when compared to the fundamental B-mode (Figure 2). In the HSCI image, clutter is reduced further than in the harmonic B-mode. The CNR was found to be 2.13 for the HSCI image and 1.16 for B-mode. Contrast is comparable between the two images; 9 dB for HSCI and 9.6 dB for B-mode.

Figure 4 shows a poor quality fundamental B-mode and its matched SLSC image. Here, the B-mode image has diminished diagnostic value as part of the blood vessel in the center of field of view is not visible. In the SLSC image, acoustic noise is suppressed and the structure is clearly visualized. This is reflected in tremendous improvement in contrast and CNR from B-mode to SLSC. Contrast was found to be 4.44 dB for SLSC compared to 0.26 dB for B-mode imaging. CNR was estimated to be 0.79 for SLSC imaging compared to 0.03 for B-mode imaging.

Contrast and CNR values calculated from matched B-mode and SLSC/HSCI images for all patients are displayed as scatter plots in Figures 5 and 6 respectively. Data obtained from one patient is represented by a single point on the scatter plot. The ordinate of the point is the contrast/CNR value calculated from the B-mode image and the abscissa is the contrast/CNR value calculated from the matched SLSC/HSCI image. The main diagonal of the graph (denoted by the solid line) indicates where B-mode and SLSC/HSCI images yielded the same contrast/CNR. Points above the main diagonal indicate improvement in contrast/CNR from B-mode to SLSC/HSCI. Different markers are used to indicate subjects with low, medium, and high B-mode-image quality and dotted lines are used to indicate the boundaries of the contrast/CNR values used to classify B-mode images. Scatter plots are shown for both fundamental (Fig. 5 (a) and 6 (a)) and harmonic (Fig. 5 (b) and 6 (b)) data. Because there were subjects for which only the fundamental data was collected, a light gray color markers were used on the fundamental-data plots to distinguish the subjects for which harmonic data existed as well.

In Figure 5, it can be seen on both plots that, with exception of a single subject, all SLSC/HSCI images have higher contrast than their B-mode counterparts. For one subject, contrast becomes slightly worse from B-mode to SLSC/HSCI. In Figure 6, all subjects (including both fundamental and harmonic data) demonstrate an improvement in CNR from B-mode to SLSC/HSCI. Because both contrast and CNR are used to classify B-mode images, a few points in Figures 5 and 6 appear to fall outside the contrast or CNR demarcation of low/mid/high quality.

A summary of the results in Figures 5 and 6 is provided in Tables 1 and 2. Mean values for contrast and CNR and their ranges for the different imaging methods are presented in Table 1. For fundamental B-mode and SLSC imaging, mean values and ranges are presented for two sets of data; the first set includes all of the subjects ($n = 17$), and the second set includes only the subjects for which harmonic data existed as well ($n = 12$). For harmonic B-mode and HSCI, mean values and ranges are shown for all subjects that had harmonic data ($n = 12$). Overall, for both contrast and CNR, SLSC obtains the highest mean values followed by HSCI, harmonic B-mode, and fundamental B-mode. Also, all of the mean values obtained

for fundamental B-mode and SLSC on the subset of data ($n = 12$) are slightly lower than those obtained on the full set of data ($n = 17$) (Table 1).

A more detailed comparison is presented in Table 2 which shows average improvements (expressed as percentages) of contrast and CNR achieved between different imaging methods. For each pair of imaging methods, separate improvements for images of high, medium, and poor quality are shown in addition to the overall improvement. Due to the insufficient number of samples for some of the categories, statistical significance is shown only for the overall improvements (shown are adjusted p values and number of sample pairs, n , used to calculate the test statistic). Overall, SLSC and HSCI show significant improvements in target detection when compared to fundamental and harmonic B-mode, respectively; adjusted p values are smaller than 0.001 for both contrast and CNR. In addition, the results obtained from the individual image-quality categories suggest that the improvements increase as the image quality decreases with the highest percentage improvements (between fundamental B-mode and SLSC and between harmonic B-mode and HSCI) observed for poor quality B-mode images. Also, greater improvements are observed between fundamental B-mode and SLSC than between harmonic B-mode and HSCI. The improvements between fundamental B-mode and harmonic B-mode and between HSCI and SLSC are statistically insignificant (adjusted p values are larger than 0.05 for both contrast and CNR). As such, these improvements are not reported in Table 2.

Discussion

SLSC and HSCI tend to reduce clutter and improve the overall image quality compared to B-mode liver images. As seen in the sample images (Figures 1 - 4), the most significant improvements are achieved in visualization of small-size, negative intrinsic-contrast structures such as blood vessels. While the source of contrast in B-mode images is the difference in echo brightness, the contrast in SLSC/HSCI images arises from the difference in signal coherence (Lediju et al., 2011; Dahl et al., 2011a). This means that in B-mode images, incoherent signals inside of a vessel lumen appear similar to surrounding tissue while in SLSC/HSCI images, these signals are highly suppressed. As a result, when compared to their matched B-mode images SLSC/HSCI images display crisper vessel boundaries and reduced clutter inside of a vessel lumen.

The surrounding tissue appears smoother and more uniform in SLSC/HSCI images compared to their matched B-mode images (Figures 1 - 4). We hypothesize that these differences arise because the mechanisms responsible for speckle noise in SLSC/HSCI images are different than those in B-mode images. The grainy texture of B-mode images is due to variations of the speckle brightness, resulting from the constructive and destructive interference of reflections from sub-wavelength particles. The variance of the speckle brightness increases with average echo magnitude often reaching large values (Wagner et al., 1983). Speckle noise in the SLSC/HSCI images is due to variations of the spatial coherence function, which does not depend on echo magnitude and exhibits relatively high uniformity at low lags (Walker and Trahey, 1997).

Improvements in target visualization between fundamental B-mode images and SLSC images and between harmonic B-mode images and HSCI images are supported by statistically significant differences in target contrast and CNR. For both contrast and CNR, greater relative improvements are observed between fundamental B-mode and SLSC than between harmonic B-mode and HSCI. We hypothesize that this is due to the fact that harmonic B-mode already reduces near-field reverberation clutter, thus leaving HSCI with less incoherent signal to suppress.

Similarly, an increase in relative contrast/CNR improvements (between B-mode and SLSC/HSCI) is observed with the decrease in B-mode image-quality (Table 2) because lower quality B-mode images have a stronger incoherent component, meaning that the relative improvements made by SLSC/HSCI would be higher. As hypothesized, the largest improvements are seen for the poor quality B-mode images. Some B-mode image-quality categories lack the number of samples to show statistical significance of these improvements. Therefore, hypothesis testing was done only for the overall improvements.

As previously stated, the results did not show significant differences (in contrast or CNR) between SLSC and HSCI and between fundamental B-mode and harmonic B-mode images. A power analysis indicated that these tests lacked power due to insufficient number of samples. As the primary focus of our study was to determine if SLSC and HSCI produced higher quality images than the delay-and-sum beamforming methods currently used in clinic and for that purpose, our sample size was sufficiently large.

This study faces several limitations. First, for all study participants, we were unable to acquire any images of focal lesions in the liver. We are aware of the importance of lesion detection/visualization in ultrasound liver imaging and acknowledge that this specific population needs to be included in order to assess the performance of SLSC/HSCI in fulfilling this task. Second, we are unable to create B-mode images that have similar high-quality displays as commercial, state-of-art US scanners because access to the sophisticated post-processing filters developed by major scanner manufacturers is not readily available. In addition, the lack of hardware/computational power that would allow faster acquisition and on-site processing of large amounts of individual-channel data prevented us from obtaining real-time feedback of SLSC/HSCI images. These issues could be alleviated with new software-based-beamforming platforms that would allow quick acquisition and access to large amounts of single-channel data. Nevertheless, the results presented in this paper are encouraging because the improvements in blood vessel visualization demonstrate that SLSC/HSCI could be well suited for *in vivo* detection of other anechoic/hypoechoic targets in the liver (such as focal lesions).

Conclusions

A study was performed that showed SLSC and HSCI imaging produce higher quality *in vivo* liver images than conventional delay-and-sum beamforming methods. SLSC and HSCI images of the livers of 17 randomly selected patients display improved visibility of blood vessels (i. e. sharper delineation of vessel walls and suppressed clutter in the vessel lumen) as well as higher uniformity of the surrounding tissue when compared to their matched B-mode images. These observations are supported by statistically significant improvements in target contrast and CNR from fundamental B-mode to SLSC and from harmonic B-mode to HSCI. In addition, the magnitude of the improvements obtained by spatial-coherence imaging increases as the quality of B-mode images decreases; the highest relative improvements are observed for poor quality fundamental B-mode images. In the future, we plan to acquire real-time SLSC/HSCI images by using a computationally powerful, software-based beamforming system. We also anticipate further exploration of clinical applications of SLSC/HSCI imaging by recruiting additional patients and conducting other task-specific studies.

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Figure 1.

High quality fundamental B-mode image (left) and its matched SLSC image (right). Note that the SLSC image is the sector inserted into the B-mode image between the arrows. SLSC reduces clutter inside of blood vessels and increases smoothness of the surrounding tissue. The contrast of the large blood vessel located at the center of field of view at 10 cm depth is 11.3 dB and 13.5 dB, and the CNR is 1.4 and 1.86 for B-mode and SLSC images, respectively. The white contour lines in the B-mode image indicate the regions used to calculate contrast and CNR values.

Figure 2. Medium quality fundamental B-mode image (left) and the matched SLSC image (right). The blood vessel spanning the full depth of the image is detectable in both images. SLSC reduces clutter inside the vessel lumen and shows better delineation of its walls. The contrast is 7.8 dB and 8.9 dB and the CNR is 0.95 and 2.05 for B-mode and SLSC images, respectively.

Figure 3. Harmonic B-mode image (left) and matched HSCI image (right) of the same vessel as in Fig. 2. Even though the B-mode image is of medium quality, the HSCI image shows notable improvement in target visualization. The contrast is 9.6 dB and 9 dB and the CNR is 1.16 and 2.13 for B-mode and SLSC images, respectively.

Figure 4.

Poor quality fundamental B-mode image (left) and matched SLSC image (right). Noise has corrupted the B-mode image and does not indicate the presence of a vessel at the center of the field of view. In the SLSC image, the vessel is easily visualized. The contrast is 0.26 dB and 4.44 dB and the CNR is 0.03 and 0.79 for B-mode and SLSC images, respectively.

Figure 5. Scatter plots of contrast values calculated from (a) matched fundamental B-mode and SLSCI images and (b) matched harmonic B-mode and HSCI images for each patient. Points that are above the main diagonal indicate improvements in contrast from B-mode to SLSC/HSCI. Dotted lines represent the contrast limits used to classify B-mode images as low, medium, or high quality. Light-gray-colored points in (a) indicate the patients for which both fundamental and harmonic data was obtained.

Figure 6. Scatter plots of CNR values calculated from (a) matched fundamental B-mode and SLSCI images and (b) matched harmonic B-mode and HSCI images for each patient. Points that are above the main diagonal indicate improvements in CNR from B-mode to SLSC. Dotted lines represent the CNR limits used to classify B-mode images as low, medium, or high quality. Light-gray-colored points in (a) indicate the patients for which both fundamental and harmonic data was obtained.

Table 1

Mean and range for contrast and CNR values of hypoechoic/anechoic targets.

		Fundamental B-mode	SLSC	Harmonic B-mode	HSCI
Contrast (dB) <i>n</i> = 17	mean	5.58	12.36	—	—
	range	0.26 - 12.52	4.44 - 29.81	—	—
Contrast (dB) <i>n</i> = 12	mean	5.14	11.08	5.76	9.18
	range	0.26 - 10.37	4.44 - 24.15	0.57 - 9.72	1.50 - 19.33
CNR <i>n</i> = 17	mean	0.75	1.88	—	—
	range	0.03 - 1.39	0.66 - 4.56	—	—
CNR <i>n</i> = 12	mean	0.69	1.82	0.76	1.70
	range	0.03 - 1.28	0.66 - 4.56	0.08 - 1.28	0.28 - 2.50

Table 2

Average improvements achieved between fundamental B-mode and SLSC and between harmonic B-mode and HSCI.

		Fundamental B-mode to SLSC	Harmonic B-mode to HSCI
Contrast	High Medium Poor	22% 39% 288%	3.4% 22% 137%
	Overall	153% $p < 0.0001, n = 17$	68% $p < 0.01, n = 12$
CNR	High Medium Poor	63% 158% 533%	10% 102% 300%
	Overall	318% $p < 0.0001, n = 17$	177% $p < 0.001, n = 12$