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Vibro-acoustography and Multifrequency Image Compounding

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Abstract

Vibro-acoustography is an ultrasound based imaging modality that can visualize normal and abnormal soft tissue through mapping the acoustic response of the object to a harmonic radiation force at frequency Δf induced by focused ultrasound. In this method, the ultrasound energy is converted from high ultrasound frequencies to a low acoustic frequency (acoustic emission) that is often two orders of magnitude smaller than the ultrasound frequency. The acoustic emission is normally detected by a hydrophone. Depending on the setup, this low frequency sound may reverberate by object boundaries or other structures present in the acoustic paths before it reaches the hydrophone. This effect produces an artifact in the image in the form of gradual variations in image intensity that may compromise image quality. The use of tonebursts with finite length yields acoustic emission at Δf and at sidebands centered about Δf . Multiple images are formed by selectively applying bandpass filters on the acoustic emission at Δf and the associated sidebands. The data at these multiple frequencies are compounded through both coherent and incoherent processes to reduce the acoustic emission reverberation artifacts. Experimental results from a urethane breast phantom are described. The coherent and incoherent compounding of multifrequency data show, both qualitatively and quantitatively, the efficacy of this reverberation reduction method. This paper presents theory describing the physical origin of this artifact and use of image data created using multifrequency vibro-acoustography for reducing reverberation artifacts.

Keywords

Ultrasound; Radiation force; Vibro-acoustography; Imaging; Reverberation; Frequency compounding

1. Introduction

In recent years, several new imaging techniques based on radiation force of ultrasound have been developed. Vibro-acoustography (VA) is a method that uses acoustic response (acoustic emission) of object to the harmonic radiation force of ultrasound for imaging and material characterization [1, 2].

In this method two co-focused ultrasound beams of slightly different frequencies f_1 and f_2 ($\Delta f = f_1 - f_2 \ll f_1, f_2$) are transmitted and intersect at their joint focal point. The radiation

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force from these two beams has a component at Δf (called dynamic ultrasound radiation force), which vibrates the object. In this method the ultrasound energy is converted from high ultrasound frequencies to a low acoustic frequency (acoustic emission) that is often two orders of magnitude smaller than that ultrasound frequency. The acoustic emission is normally detected by a hydrophone and is used to construct the image of the object. The ultrasound raster scans the object, and the resulting acoustic signal is recorded. The image of the object is formed by modulating the brightness of each image pixel proportional to the amplitude of the acoustic signal from the point of the object.

Vibro-acoustography images have some unique characteristics that set them apart from traditional ultrasound imaging. This is in part due to the non-linear phenomenon of frequency conversion in this method. For example, VA images are speckle free, which is a significant advantage over conventional pulse-echo imaging. More recent contributions have examined the use of multiple simultaneous ultrasound signals and the resulting stress fields for image formation [3, 4]. The strength of this method is that more information can be gained at multiple difference frequencies, thereby reducing scan times for a given amount of information acquired.

VA may be used for a variety of imaging and characterization applications, including medical and industrial applications. Vibro-acoustography has been studied intensively in breast imaging for detection of abnormal lesions [5–8]. Recent human study results indicate that various types of breast lesions as well as microcalcifications can be detected by VA. Vibro-acoustography has also been found to be valuable in therapeutic applications in the prostate gland, particularly in detecting brachytherapy seeds and monitoring cryotherapy [9–13]. Vibro-acoustography has also been used for imaging mass lesions in liver, calcifications in arteries in the breast and peripheral circulation, and bone [14–17].

Vibro-acoustography has also found use as an imaging modality in the nondestructive testing field. Different modes such as continuous wave, toneburst, and chirp excitation methods have been reported for surface flaw detection and surface roughness measurements [18–22].

As any other imaging techniques, VA may also produce some image artifacts which depend on the application and imaging setup. An artifact that is specific to VA is reverberation of the acoustic emission in the medium. In many cases, the object that is being investigated is smaller or on the same order as the wavelength of the acoustic emission (typically 20–50 mm). The acoustic emission will traverse multiple paths from the focal region of the transducer to the hydrophone used for measurement. This reverberation of the acoustic emission can create artificial spatial variations of image intensity that may affect overall image quality.

This paper presents mathematical models for the frequency composition of the VA signal as well as the process that leads to reverberation artifact. Finally, we present a method based on frequency compounding for reducing reverberation artifacts in VA images.

2. Methods

2.1 Reverberation Artifact and Frequency Compounding

The acoustic emission is a propagating compressional wave and can undergo one or more reflections from the object boundaries, or other nearby structures, before being received by the hydrophone. The reflected signal can interfere constructively or destructively depending on the relative phases of the multiple signals. This phenomenon, called reverberation, can manifest itself in images as dark and bright areas of low spatial frequency, which can affect

overall image contrast and detectability of abnormal tissue. This section describes the mathematical principles of reverberation and presents methods for reducing reverberation artifacts.

To perform vibro-acoustography, finite length duration ultrasound tonebursts are used. In addition to the component at Δf , the short duration of tonebursts gives rise to sidebands centered about Δf . Images may be formed at Δf and the associated sidebands. The data at these multiple frequencies can be compounded, or combined, using coherent or incoherent processes to reduce the artifacts associated with reverberation of the acoustic emission. We define coherent compounding as the sum of data of different frequencies without performing any envelope calculations. This is accomplished physically by the summing of waves of different frequencies at the hydrophone position. This composite signal contains multiple weighted frequency components. The only process that is then performed is application of a bandpass filter of a specific bandwidth, retaining the sums of the signals within the specified bandwidth. The root-mean-square (rms) of the filtered signals is then computed for display of the data in an image. Incoherent compounding involves narrowband filtering of the same data and computing the rms to form images from distinct frequency components and then summing those formed images. The distinction between coherent and incoherent is when the sum of the different frequency components is performed, before or after the rms calculation.

The use of the term compounding needs to be clarified and put in context with respect to diagnostic B-mode ultrasound and its efficacy in the problem of speckle reduction. Two types of compounding have been addressed in the B-mode ultrasound literature: spatial and frequency compounding. Spatial compounding involves acquisition of multiple views of the same tissue using steered ultrasound [23, 24]. The envelope data for the different views are summed together to create a composite image. Spatial compounding with N views increases the signal-to-noise ratio of the composite image by a factor of $\sqrt{N}$ [25]. Recently, spatial compounding has been performed by coherently summing images created using plane wave insonifications at different angles [26].

Frequency compounding has also been performed in B-mode imaging. This typically involves either transmitting different frequencies and forming images from different ultrasound frequencies, and then performing compounding [27–29]. Another method is to filter the received data into different bands to generate images for compounding [30]. Frequency compounding has been shown to reduce speckle in ultrasound images.

Previous work in multifrequency vibro-acoustography has indicated that incoherent combinations of the image data yielded images with improved contrast and detail [3, 31]. This study will incorporate both coherent and incoherent combinations of data as well as an analysis of a previously unstudied source of multiple excitation frequencies. We will adopt an approach similar to that used in frequency compounding used in B-mode ultrasound imaging where the time-domain signal is filtered into different bands which are then used for image formation and compounding.

2.2 Frequency Composition of Acoustic Emission

In vibro-acoustography, two ultrasound beams with finite temporal durations are transmitted and focused at a common focus point, r_0 . The total pressure from these two beams at a point r near the focus is

$$p(r, t) = w(t)p_1(r)\cos(2\pi f_1 t + \varphi_1) + w(t)p_2(r)\cos(2\pi f_2 t + \varphi_2), \quad (1)$$

where $w(t)$ is a temporal window function of the transmitted ultrasound, $p_i(r)$ is the spatial pressure distribution function for the beam with frequency f_i ($i = 1, 2$) and $\Delta f = f_1 - f_2$. For simplicity here we assume that the same window function applies to $p_1(r)$ and $p_2(r)$. The pressure could also be written as $p(r, t) = w(t)p_w(r, t)$ where $p_w(r, t) = p_1(r) \cos(2\pi f_1 t + \varphi_1) + p_2(r) \cos(2\pi f_2 t + \varphi_2)$. The radiation force that produces the acoustic emission is proportional to the short-term time average, denoted by $\langle \cdot \rangle$, of the energy density [2],

$$\begin{aligned}\langle e(r, t) \rangle &= \langle p^2(r, t) / \rho c^2 \rangle \\ &= \langle w^2(t) p_w^2(r, t) / \rho c^2 \rangle,\end{aligned}\quad (2)$$

where ρ and c is the density and speed of sound in the medium, respectively. This relationship can be expanded as

$$\langle e(r, t) \rangle = \frac{w^2(t)}{\rho c^2} [1 + p_1(r)p_2(r)\cos(2\pi\Delta f t + \varphi_1 - \varphi_2)].\quad (3)$$

Ignoring the constant term, the dynamic part of the energy density, $e_d(r, t)$, at frequency $\Delta\omega$ is

$$e_d(r, t) = \frac{p_1(r)p_2(r)}{\rho c^2} w^2(t) \cos(2\pi\Delta f t + \varphi_1 - \varphi_2).\quad (4)$$

The frequency domain representations of the window functions are $w(t) \leftrightarrow W_1(f)$ and $w^2(t) \leftrightarrow [W_1(f) \otimes W_1(f)] = W_2(f)$, where $\otimes$ denotes convolution in the frequency domain. The two co-focused ultrasound beams exert a radiation force on the object. The dynamic component of the radiation stress is proportional to $E_d(r, f)$ [2]. In the frequency domain, the dynamic radiation stress $F_d(r, f)$ can be expressed as

$$F_d(r, f) = \frac{Ap_1(r)p_2(r)}{2\rho c^2} [W_2(f - \Delta f) + W_2(f + \Delta f)],\quad (5)$$

where A is a constant. This radiation force vibrates the object, which in turn produces an acoustic field (acoustic emission) in the medium. This acoustic field propagates through the medium and can be detected by a hydrophone.

Let us consider the ultrasound beams focused at r_0 in the object. Then, the acoustic emission P_d at point r' (hydrophone location) can be written in the general form as

$$P_d(r', r_0, f) = K_0 M(r', r_0, f) [G(r, f) \otimes_r F_d(r, f)],\quad (6)$$

where the K_0 is a constant, $G(r, f)$ represents the dynamic response of the object to a force at point r , the symbol $\otimes_r$ represents two-dimensional convolution in the spatial (r) domain, analogous to scanning the force over an object. The result of this convolution is an acoustic pressure (acoustic emission) which is directly related to the object. This result is then

multiplied by $M(r', r_0, f)$, which represents the transfer function for propagation of the acoustic field from the excitation region at the ultrasound focus, r_0 , to the hydrophone at r' . This function describes the acoustic environment that the acoustic emission propagates through to the measurement point at the hydrophone. The function $M(r', r_0, f)$ will change based on the acoustic environment between and around the object and the hydrophone. The relationship in (6) is the wideband version of a similar equation that was presented in [32] for a single frequency (Δf) excitation.

2.3 Reverberation of Acoustic Emission

Normally, the VA image is constructed from the amplitude of the acoustic emission signal at Δf . Commonly, the image is intended to represent spatial variations of the object function $G(r, \Delta f)$, and the influence of the propagation medium function $M(r', r_0, f)$ on the image is neglected. Here we consider cases where the effect of the propagation medium $M(r', r_0, f)$ is significant.

A typical case is when the acoustic field emitted from the focal region is reflected by the boundaries or other structures present in the propagation path. An example of such a situation in Figure 1 where the sound from the focus point r_0 is reflected by the object boundaries before reaching the hydrophone at r' .

Depending on the path length, l_n , each reflection arrives with a different delay (or phase). This phenomenon is called reverberation. For this purpose, the function $M(r', r_0, f)$ may be expressed as

$$M(r', r_0, f) = M_0(r', r_0, f) \sum_{n=1}^N a_n e^{-j2\pi f l_n / c}, \quad (7)$$

where l_n , $n = 1, \dots, N$, is the n th path length between r_0 and r' , a_n is the amplitude of the signal associated with the n th path, and $n = 1$ represents the direct path. The sum in (7) shows that the reverberation contribution is a function of position (r_0 and r') and frequency f . Depending on the values of a_n and l_n , spatial variations of $M_0(r', r_0, f)$ can produce significant artifactual variations in $P_d(r', r_0, f)$, and hence in image intensity.

Vibro-acoustography measures the acoustic response of an object stimulated by radiation force. One of the advantages of the method is that it does not exhibit the speckle with high spatial frequency variations associated with B-mode images. However, in some cases, the acoustic environment that VA is being performed, an organ or a phantom in a water tank, can be acoustically reverberant. The acoustic signal can propagate over multiple paths before being recorded by the hydrophone. This situation is somewhat similar to B-mode pulses being scattered and received by an array transducer. In both cases, the signals interfere and are summed coherently. When an envelope operation is performed over the spatial extent, intensity variations can arise, speckle in the case of B-mode images, and a slowly varying background in VA images.

It should be noted that the reverberation in VA images should not be equated to speckle in B-mode images in either the origin or the nature of the signals. The speckle in B-mode ultrasound originates from the scattering of ultrasound within in the resolution cell of the ultrasound beam. The interference due to reverberation for VA imaging occurs outside of the resolution cell of the ultrasound beams and is related to the acoustic environment of the scan. The main similarity is that there is interference of propagating waves measured by a

phase sensitive transducer which results in variations in the background that can impair visualization of certain features in the image.

A spatial compounding approach to solving this problem would be difficult to implement for VA, however, due to windowing of the ultrasound toneburst, multiple frequency components can be produced in a single VA measurement. It should be noted that $G(r,f)$, the dynamic response of the object, is usually a smooth function with respect to frequency except perhaps in the case of the presence of lesions or distinct borders between different types of materials or tissues. So if frequency is varied by a few kilohertz, the changes are expected to be small. However, the function $M(r',r_0,f)$ can vary greatly with respect to frequency, that is, even small changes in frequency can cause large variations due to the phase dependence on frequency. These sideband frequencies typically have independent information, based on the strong dependence of $M(r',r_0,f)$ on frequency. This situation is analogous to having B-mode images that are generated from different frequency components. Compounding of these multiple VA images formed at different frequencies could be summed to attempt to cancel out reverberation effects, which manifest differently as the difference frequency is varied. Ideally, one would like to average truly independent images to achieve the most cancellation of uncorrelated parts of the images.

2.4 Frequency Compounding for Reducing Reverberation

Equation (7) shows that $M(r',r_0,f)$ varies as a function of position (relative distance between r_0 and r') and frequency. Here we try to exploit frequency to compensate for the spatial variations.

The sidelobes of the window function in (5) produce multiple sidebands around f_k , $k = 1, \dots, K$, frequencies, where K is the number of significant sidelobes in the acoustic emission in (6). Therefore, multiple images can be formed from these sidebands. Since these K images are constructed from different frequency components, according to Eqs. (6) and (7), the amplitude of the acoustic emission and hence the reverberation effect should be somewhat different in each image. Thus, by averaging such images, either coherently or incoherently, the effect of reverberation can be reduced or cancelled.

To form images at the sideband frequencies, narrowband filters with bandwidths much smaller than the width of each sidelobe are used to filter out the acoustic emission at f_k , $k = 1, \dots, K$. The filtered signal obtained at each focal position corresponding to pixel (x,y) is temporally gated and a rms metric is applied. The resulting K “spectral images” are denoted by $I_k(x,y)$, $k = 1 \dots, K$. Since the sidebands have different amplitudes, these images also have different relative amplitudes. Another factor that may change the relative amplitude of these images is the dependence of $G(r,f)$ and $M(r',r_0,f)$ in (6) on frequency.

To perform incoherent compounding, the K spectral images are first scaled such that the images have equal image energy. The energy in an image can be written as [33]

$$\varepsilon_k = \sum_{x=1}^X \sum_{y=1}^Y |I_k(x,y)|^2, \quad (8)$$

Thus, to match images in energy, each spectral image is multiplied by the scale factor $\gamma_k = \sqrt{\varepsilon_1/\varepsilon_k}$. The incoherent compounded image $I(x,y)$ is defined as the mean of the scaled spectral images,

$$I(x, y) = \frac{1}{K} \sum_{k=1}^K \gamma_k I_k(x, y). \quad (9)$$

Another way that the images can be compounded incoherently is using statistical metrics. For a set of images formed at different frequencies, a pixel-based approach can be used. On a pixel-by-pixel basis a statistical measure such as the median ($I_{MD}(x, y)$) is computed, as

$$I_{MD}(x, y) = \text{median } \gamma_k I_k(x, y)_{k=1, \dots, K}, \quad (10)$$

Another method for reducing reverberation artifact is coherent image averaging. The coherent image averaging is obtained by using a relatively wideband filter that allows the main component as well as the sidebands of the acoustic emission to pass through. The resultant signal is a wideband version of the acoustic emission. At each pixel located at (x, y) , the filtered signal is temporally gated and a rms metric is applied. The resultant image is denoted by $I_{WB}(x, y)$.

In order to evaluate the effectiveness of the compounding methods described above in reducing reverberation artifact, one needs to define a metric that measures image intensity variations. For this purpose, we use a smoothness metric $R = \sigma^2 / \sigma^2 + L - 1$, where σ^2 is the variance of the image intensity levels and L is the number of gray levels in the image [34]. The minimum value of R is zero for a uniform image (constant gray level), and the maximum value of R is when σ^2 is at maximum of $L^2/4$. Assuming $L = 256$ for this study, R varies between 0 and 0.2. The R value for a given image also depends on the object structure and image details. However, in an image with reverberation artifact, σ^2 is often significantly influenced by the intensity variations caused reverberation. Therefore, R is expected to decrease after application of image compounding. Hence, we define a reverberation reduction factor, β , as the ratio of R values after image compounding (R') to that of original image (R):

$$\beta = R' / R, \quad (11)$$

where $\beta = 1$ represents no improvement and $\beta = 0$ represents a complete elimination of intensity variations (reverberation artifact) for a uniform object, and $\beta > 1$ represents that the compounded image is less smooth than the reference image.

3. Experiments

To explore the images formed at the difference frequency, Δf , and the associated sidebands, a phantom experiment was performed. The data from this experiment was used for coherent and incoherent compounding to reduce reverberation artifacts. A two element, confocal transducer with center frequency of 3.0 MHz with a spherical focus of 7.0 cm focus was used. Ultrasound frequencies of 3.10 and 3.05 MHz were used to produce acoustic emission at $\Delta f = 50$ kHz. The temporal window used in these studies is comprised of an exponential rise with rise time of 47 μs to a plateau for about 60 μs and then an exponential decay with decay time of 46 μs . The time window, $w(t)$, used in the studies as well as its square, $w^2(t)$ is shown in Figure 2(a). The reason we used a window with an exponential rise instead of a rectangular window is to avoid introducing a mechanical shock to the piezoelectric crystal of the transducer, which may create an artifactual acoustic signal. The frequency domain

representations of the signals $W_1(f)$ and $W_2(f)$ in Figure 2(a) are shown in Figure 2(b). The signal $W_2(f)$ gives frequency sidebands of ± 14 kHz and ± 23 kHz, which are the peaks of the first two sets of sidelobes. The heights of the sidelobes are 0.13 and 0.05 with respect to the main lobe amplitude of unity.

A urethane breast phantom (CIRS Model 013, Norfolk, CT) was placed in a large water tank and the transducer was mechanically raster scanned across the focal plane to create an image. The phantom by itself did not produce strong reverberation. To demonstrate the effectiveness of the compounding, a metal plate was placed about 6.5 cm behind the phantom to generate strong acoustic reverberation. The hydrophone was placed near the edge of the metal plate. An illustration of the experimental setup is shown in Figure 3. The hydrophone was moved in the x -direction to vary the acoustic environment and change the reverberation delay times for different experiments.

4. Results

Narrowband images from breast phantom scanned at two different hydrophone positions are shown in Figure 4. In this figure, $x = 0$ cm is the reference position near the edge of the metal plate. The top row was taken at $x = 0$ cm and the bottom row was taken at $x = 2.5$ cm. The narrowband images each obtained by a 1 kHz bandwidth filter centered at $\Delta f = 50$ kHz and two sidebands at $f = 36$ and 64 kHz were formed. Wideband images were also formed with varying bandwidths of 5, 10, and 20 kHz centered at $\Delta f = 50$ kHz. The wideband images are shown in Figure 5 for datasets acquired at $x = 0$ and 2.5 cm with their associated R and β values, where the β values were calculated using the $\Delta f = 50$ kHz with 1 kHz bandwidth as the reference image. For the images at $x = 0$ cm, the R and β values increased, while at $x = 2.5$ cm, as the bandwidth used increased, the R and β values decreased slightly. The decrease in R values at $x = 2.5$ cm qualitatively correlated with smoother images. Using (11), the values for β for the 5, 10 and 20 kHz images were 0.91, 0.96, and 1.01, respectively, for $x = 0$ cm and 0.72, 0.63, and 0.55, respectively, for $x = 2.5$ cm. The images from incoherent mean and median compounding were calculated and compared with the narrowband image (NB) formed at $f = 64$ kHz and $x = 2.5$ cm along with their corresponding R and β values in Figure 6. The compounding involved using the images at 36, 50, and 64 kHz. The image at 64 kHz was chosen as the reference because of its significant reverberation artifact. The compounded images have reduced R values and $\beta = 0.55$ and 0.78 for the incoherent mean and median compounding with respect to the NB image. These β values are indicative of a smoother image with consistent background such that the lesions could be differentiated from background. In the narrowband image, the lesions change their contrast with respect to the background, probably because of the reverberation. However, the processed images with coherent and incoherent compounding show lesions with consistent contrasts. It is noted that the incoherent compounding with the median metric has produced an artifactual U-shaped contour at the top of the image. This is due to the nonlinear nature of the median process. This type of artifact is expected to be reduced if more spectral images were used in the compounding process.

5. Discussion

One of the first observable features of the multifrequency images are the different brightness and contrast of features in the images. Different features are better visualized at different frequencies, and this is quantitatively substantiated by the different CNR values for the different lesions in the breast phantom images formed using different frequency components. This phenomenon needs further explanation, but the reverberation effect as well as the frequency response of the object and the acoustic environment have roles in determining how the images appear.

In the breast phantom experiment, different degrees of reverberation artifacts are noticeable. These are manifested as bands or patterns of dark and bright regions in the image that have little to do with the actual features of the phantom. The lesions can be detected in all of the images but the contrast is affected by the reverberation artifacts. When the different types of compounding are performed, the β values were below 1 indicating that the images had reduced reverberation effects compared to the narrowband image formed at $\Delta f = 64$ kHz. The coherent compounding which involves using a bandpass filter with a wider bandwidth incorporates energy at many frequencies and provides the smoothest image in this case. The image formed through incoherent mean compounding yielded better results than the median compounding in terms of the quantitative measures reported.

The metal reflector was used to insure reflections. The images formed at different frequencies have different patterns originating from the reverberation. This independence of the different images poses a basis for summing the images so that the different patterns can provide some cancellation, producing a more uniform image in which the lesions are more distinct and the background is not a distracting factor for the image reader.

The bands in the breast phantom images seemed to change their spatial frequency as the frequency of the acoustic emission varied. Referring to Fig. 4, the spatial frequency of the bands is much higher at 64 kHz than at 36 kHz. The spatial wavelengths of at these frequencies are 23.1 and 41.1 mm. The images are 6×8 cm so only a few wavelengths could be visualized. One way to think of the interference pattern in these images is as a type of speckle, commonly associated with B-mode ultrasound images. To explore this type of comparison from a quantitative perspective an experiment was performed. We performed an experiment with a gelatin block of size $18 \times 8 \times 12$ cm in a large water tank. The setup was similar to that used in the breast phantom experiment. We scanned the phantom while varying Δf from 40–70 kHz in 10 kHz increments. The phantom was scanned with a series of reflectors placed close to the phantom to create reverberation. The images were analyzed with first-order statistics for speckle analysis [35]. Only first-order statistics were calculated for this analysis because only a few spatial wavelengths could be obtained in the VA images. The mean and standard deviation was calculated from the images for evaluation of the signal-to-noise ratio (SNR) which is the ratio of the image mean and standard deviation. For fully formed speckle that is modeled using a Rayleigh distribution, the $\text{SNR} = 1.91$ [35]. The images from this experiment are shown in Fig. 7. The SNR and mean SNR for the images are summarized in Table I. The SNR for the images formed in the reverberant environment had a mean close to the 1.91 associated with fully formed speckle.

These results must be placed in proper context. The image appearance of the homogeneous phantom and SNR results are similar to the characteristics of speckle observed in B-mode ultrasound imaging. The quantitative results are an interesting observation, in that the SNR for the VA images was close to that used for classifying fully formed speckle in B-mode ultrasound imaging. This is an interesting finding because the acoustic reverberation is exhibiting a characteristic similar to that found from scattering of ultrasound waves for B-mode ultrasound. Both processes cause interference of the waves which are measured by phase-sensitive detectors. However, the speckle in ultrasound images has its origin from scattering and interference from within the resolution cell of the ultrasound beams while the reverberation occurs from interference outside of the resolution cell of the ultrasound beams. The similarity between speckle in B-mode ultrasound imaging and the reverberation artifact in VA imaging needs further study, especially in light of the experimental observations described in this study.

The results from the experiment showed that different types of image compounding and different sets of frequencies may improve the image and aid in detection and visualization of

structures within the object. The post-processing described in this paper can be performed for any type of vibro-acoustography images with proper knowledge of the difference frequency and the frequency spectrum of the temporal window of the toneburst. Further research is needed to determine which type of processing would result in the optimal image quality for a particular group of objects. The temporal window could also be designed to enhance certain components and further study will be devoted towards this end.

When imaging phantoms or tissue *in vivo*, it is desirable in VA imaging to have images that have a uniform background and features should have contrast with respect to that background. We have found that reverberation creates low spatial frequency variations in the images that are not related to the object's response to the stimulating force. These spatial variations can hamper image examination because they can be distracting or can hide important features. The desire to have a uniform background led to the adoption of the smoothness metric. The features in the phantom images represent a relatively small number of pixels so the variance of the image is primarily determined by the background and is a good measure of the background smoothness.

The main contribution of this paper is the development of techniques for image compounding that can be applied for VA imaging in order to reduce the low spatial frequency variations introduced by acoustic reverberation to improve visualization of features in phantom images. Smoothing of the background patterns has been demonstrated with both incoherent and coherent frequency compounding. The optimal types of compounding, coherent versus incoherent, will need to be evaluated for frequency compounding for *in vivo* VA images.

6. Conclusion

Vibro-acoustography images are formed using low-frequency acoustic response of the object to a harmonic excitation induced by ultrasound. Depending on the setup, the acoustic response of the object may reverberate by object boundaries or other structures in the acoustic paths. This phenomenon produces an artifact in the image in the form of gradual variations in image intensity that may compromise image quality. A group of methods based on combining images obtained from different frequency bands were presented and shown to be effective in removing the reverberation artifact. While these methods improve image quality, they may also provide better identification of abnormalities in medical vibro-acoustography images.

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Fig. 1.
Illustration of multiple wave paths of acoustic emission from focal region to hydrophone.

Fig. 2. Time and frequency domain representations of gate function. (a) Time domain representations of $w(t)$ and $w^2(t)$, (b) Frequency domain representations of $W_1(f)$ and $W_2(f)$.

Fig. 3. Experimental setup for phantom experiment. The transducer scanned a urethane breast phantom in a large water tank. A metal plate and hydrophone were placed behind the phantom to create reverberation and measure the acoustic emission, respectively.

Fig. 4. Narrowband phantom images formed at different frequencies and at different hydrophone locations. The top row was taken at $x = 0$ and the bottom row was taken at $x = 2.5$ cm. All images are 6.0×8.0 cm.

BW = 5 kHz, $R = 0.0199$, $\beta = 0.91$ BW = 10 kHz, $R = 0.0209$, $\beta = 0.96$ BW = 20 kHz, $R = 0.0220$, $\beta = 1.01$

BW = 5 kHz, $R = 0.0069$, $\beta = 0.72$ BW = 10 kHz, $R = 0.0061$, $\beta = 0.63$ BW = 20 kHz, $R = 0.0054$, $\beta = 0.55$

Fig. 5.

Phantom images formed at with different bandwidths at different hydrophone locations. The top row was taken at $x = 0$ and the bottom row was taken at $x = 2.5$ cm. All images are 6.0×8.0 cm. The reference image for the β values are the images formed at 50 kHz with BW = 1 kHz shown in Figure 4 at $x = 0, 2.5$ cm, respectively.

Fig. 6. Phantom images with different formation conditions including narrowband filtering (*NB*) and compounding methods including incoherent (*I*), and median (*MD*) for $x = 2.5$ cm. The image contrast and brightness has been adjusted to display improvements. All images are 6.0×8.0 cm. The reference image for the β values is the *NB* image at $f = 64$ kHz.

Fig. 7.

Images of a large gelatin block in a reverberant environment at several difference frequencies. (a) $\Delta f = 40$ kHz, (a) $\Delta f = 50$ kHz, (a) $\Delta f = 60$ kHz, (a) $\Delta f = 70$ kHz. All the images are 8×8 cm.

Table I

SNR Values for Reverberation Images in Fig. 7

Δf , kHz	SNR
40	1.720
50	2.232
60	1.722
70	1.868
Mean $\pm$ SD	1.886 $\pm$ 0.241