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Ultrasonic Backscatter Imaging by Shear-Wave-Induced Echo Phase Encoding of Target Locations

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Abstract

We present a novel method for ultrasound backscatter image formation wherein lateral resolution of the target is obtained by using traveling shear waves to encode the lateral position of targets in the phase of the received echo. We demonstrate that the phase modulation as a function of shear wavenumber can be expressed in terms of a Fourier transform of the lateral component of the target echogenicity. The inverse transform, obtained by measurements of the phase modulation over a range of shear wave spatial frequencies, yields the lateral scatterer distribution. Range data are recovered from time of flight as in conventional ultrasound, yielding a B-mode-like image. In contrast to conventional ultrasound imaging, where mechanical or electronic focusing is used and lateral resolution is determined by aperture size and wavelength, we demonstrate that lateral resolution using the proposed method is independent of the properties of the aperture. Lateral resolution of the target is achieved using a stationary, unfocused, single-element transducer. We present simulated images of targets of uniform and non-uniform shear modulus. Compounding for speckle reduction is demonstrated. Finally, we demonstrate image formation with an unfocused transducer in gelatin phantoms of uniform shear modulus.

I. Introduction

Typical ultrasound backscatter medical imaging systems (e.g., B-scan) use an aperture which is large compared with the ultrasound wavelength to generate a focused beam pattern on transmit and receive [1]. Geometric focusing, either mechanical, electronic, or a combination of the two, is used to determine the phase of the aperture motion on transmit, and to adjust the echo phase on receive [2], [3]. For linear propagation, the frequency, size, and apodization of the aperture determine the beam pattern [1]. Deviations in assumed sound speed ($c_1 = 1540$ m/s typically) of the media between the transducer and target can distort the beam pattern and reduce image resolution [4].

It is often the case that the target to be imaged can support shear waves in addition to the longitudinal waves used for ultrasound imaging. This is true in medical imaging, in which tissues support propagation of low-frequency shear waves. The shear wave speed in tissue is much lower than the longitudinal wave speed and shear wave motion in the target can be tracked ultrasonically, as in many methods of elasticity imaging [5]-[8]. The goal in elasticity imaging is to reconstruct the shear modulus of the object, rather than images of echogenicity. The shearing of scatterers tends to decorrelate echo signals and acts as a noise source in elastography [9]. Echo decorrelation under tissue strain has been used as a means of speckle reduction [10].

We have previously noted a strong, correlated noise component in sonoelastographic measurements of shear wave speed in homogeneous phantoms [11]. In these measurements, a test sample supported shear waves of a known frequency, and the phase difference between the tracked tissue displacement at two locations which were a known distance apart

was used to estimate shear wavelength. We noted a significant variance in these measurements as a function of position that was not caused by inadequate echo SNR, and hypothesized that this variance was caused by tracking of bright off-axis speckles causing the tissue tracking point to deviate from the geometric beam center. The correlated noise varied with frequency but was highly repeatable at a given frequency. These observations led us to consider whether an image might be formed from measurements of the change in the echo signals from a target undergoing deformation caused by shear wave propagation.

In this paper, we demonstrate a method to reconstruct an image of the ultrasonic echogenicity of the target under the assumption of uniform shear modulus. This method does not rely on conventional geometric focusing, but rather on the variation in echo signal received from a target subject to plane shear waves applied over a range of frequencies. We show that in this method the lateral resolution is independent of the aperture size and instead determined by the shear wave velocity in the target and by the range of shear wave frequencies that can be applied to the target. An analytical model for the echo signal and the method of reconstruction is developed in Section II. The effects of a discrete set of shear wave frequencies and variation in shear wave amplitude are determined. A description of numerical simulations of imaging of point and diffuse targets, and of measurements of wire targets in a gelatin phantom, are presented in Section III. Imaging in both the simulations and experiment are performed with a stationary, single-element unfocused transducer. Simulations of compounding with this method, wherein incoherent averaging of images obtained at multiple transducer locations are also presented. A discussion follows the presentation of results in Section IV.

II. Theory

The echo $r(t)$ from a collection of scatterers in two dimensions can be modeled as

$$r(t) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} h(x, z, t) \gamma(x, z) dx dz, \quad (1)$$

where $h(x, z, t)$ is the pulse-echo response to a point scatterer at (x, z) , and $\gamma(x, z)$ is the object function, which describes the scattering properties as a function of position [12]. Here the scatterers are taken to be in the xz plane; ultrasound propagation is in the $+z$ direction. Although $h(x, z, t)$ is spatially varying in general, in the far field of an unfocused transducer, local spatial invariance can be assumed, and the impulse response modeled as

$$h(x, z, t) = h(x, t - 2z/c) = h_e(x, t - 2z/c) e^{j\omega_1(t - 2z/c_1)}, \quad (2)$$

where h_e is the pulse envelope, ω_1 the ultrasound radian frequency, and c_1 the acoustic wave speed. This model implies that displacement of a point target in the $\pm z$ direction induces a proportional change in echo arrival time but does not otherwise change the echo. The envelope signal $h_e(x, t)$ is taken to be centered around $t = 0$, i.e., pulse emission occurs at $t = 0$. The envelope h_e is assumed to vary slowly relative to the ultrasound frequency.

With this assumption, the effect of a small displacement in the z direction, Δ_z , can be modeled as a phase shift of the echo signal. A point target at location $(x, z + \Delta_z)$ yields echo

$$\begin{aligned} h(x, z + \Delta_z, t) &= h(x, t - 2(z + \Delta_z)/c) \\ &= h_e(x, t - 2(z + \Delta_z)/c) e^{j\omega_1(t - 2(z + \Delta_z)/c_1)}. \end{aligned} \quad (3)$$

If the displacement Δ_z is small, such that the extra delay $2\Delta_z/c$ induces a negligible change in the envelope signal, then the effect of the displacement is just a phase shift of the echo of the undisturbed target,

$$h(x, z + \Delta_z, t) = h(x, t - 2z/c) e^{-2jk_l \Delta_z}. \quad (4)$$

An elastic medium supports shear wave propagation, in addition to the longitudinal ultrasound waves. Such a shear wave in the target object displaces scattering centers. A z -polarized plane shear wave of frequency ω_s propagating in the $+x$ direction induces a displacement of the scatterers in the $\pm z$ direction of

$$\Delta_z = a \cos(\omega_s t - k_s x - \phi), \quad (5)$$

where a is the displacement amplitude, and k_s the wave-number of the shear wave. The phase of the shear wave source with respect to the ultrasound pulse emission is denoted by ϕ . The passage of the shear wave moves a point at z to position $z + \Delta_z$. As described in [13], when $\omega_s a/c_l \ll 1$, and the pulse duration is short compared with the shear wave frequency, the value of t at the time of pulse-scatterer interaction is well approximated by z/c_l , where z is the undisturbed scatterer range. In the present case, $\omega_s a/c_l$ is on the order of 10^{-4} . Using the approximation of (4), (1) can be written to include the effect of the shear wave as

$$r(t, k_s, \phi) = \int_{-\infty}^{\infty} \hat{\gamma}(x, t) e^{-j\beta \cos(\omega_s z/c_l - k_s x - \phi)} dx, \quad (6)$$

where $\beta = 2ak_l$ and

$$\hat{\gamma}(x, t) = \int_{-\infty}^{\infty} h(x, t - 2z/c) \gamma(x, z) dz, \quad (7)$$

$\hat{\gamma}(x, t)$ represents the object function convolved in the z direction with the impulse response, and shaded in the x direction by h . That is, $\hat{\gamma}$ is the object function blurred in the range direction and illuminated but unblurred in the lateral direction. $r(t, k_s, \phi)$ represents the complete echo as a function of time, modulated by the shear wave displacement of the scatterers. In the absence of any shear-wave-induced motion, (1) and (7) are equivalent.

Eq. (7) is a phase-modulated version of (1). For small values of β , i.e., $|\beta| \ll \pi$, we can make the approximation $e^{-j\beta \cos(\omega_s z/c_l - k_s x - \phi)} \approx 1 - j\beta \cos(\omega_s z/c_l - k_s x - \phi)$. Using this approximation and the identity $\cos(x - \pi/2) = \sin(x)$ we can write

$$r(t, k_s, \pi) - r(t, k_s, 0) = 2j\beta \int_{-\infty}^{\infty} \hat{\gamma}(x, t) \cos\left(k_s x - \frac{\omega_s z}{c_l}\right) dx, \quad (8)$$

and

$$r(t, k_s, \pi/2) - r(t, k_s, -\pi/2) = 2j\beta \int_{-\infty}^{\infty} \hat{\gamma}(x, t) \sin\left(k_s x - \frac{\omega_s z}{c_l}\right) dx. \quad (9)$$

Define I and Q as

$$I(t, k_s) = r(t, k_s, \pi) - r(t, k_s, 0) \quad (10)$$

and

$$Q(t, k_s) = r(t, k_s, \pi/2) - r(t, k_s, -\pi/2). \quad (11)$$

Note that neither I nor Q is necessarily real. Then

$$I(t, k_s) + jQ(t, k_s) = 2j\beta e^{jc_r z k_s} \int_{-\infty}^{\infty} \hat{\gamma}(x, t) e^{-jk_s x} dx, \quad (12)$$

that is, the sum $I(k_s) + jQ(k_s)$ is the Fourier transform of $\hat{\gamma}(x, t)$ scaled by $2j\beta e^{jc_r z k_s}$, where c_r is the ratio of shear wave speed to longitudinal wave speed and $c_r z k_s = \omega_s z / c_l$. The term $e^{jc_r z k_s}$ represents the phase shift of the shear wave caused by the propagation delay of the ultrasound pulse from the transducer to depth z . The inverse Fourier transform may be applied, then, to recover $\hat{\gamma}(x)$ from the signals $I(t, k_s)$ and $Q(t, k_s)$,

$$\hat{\gamma}(x, t) = \frac{1}{4\pi j\beta} \int_{-\infty}^{+\infty} \{I(t, k_s) + jQ(t, k_s)\} e^{-jc_r z k_s} e^{jk_s x} dk_s. \quad (13)$$

Recovery of $\hat{\gamma}$, then, yields an image of the object γ blurred by the axial (time) component of the impulse response, and shaded laterally by the extent of h_e . For typical imaging depths and wave speed ratios, the term $e^{-jc_r z |k_s|}$ is negligible.

This development assumed that the vibration amplitude a , and therefore β , is independent of shear wave frequency and position. It is straightforward to show that variation in β with shear wave frequency introduces a lateral blurring, whereas variation with x introduces a lateral shading. Treating β as a separable function of space and shear wavenumber, $\beta = \beta_x(x)\beta_{k_s}(k_s)$, we can write (13) to include both effects as

$$\beta_x(x) \hat{\gamma}(x, t) *_x B(x) = \frac{-j}{4\pi} \int_{-\infty}^{+\infty} \{I(t, k_s) + jQ(t, k_s)\} e^{-jc_r z k_s} e^{jk_s x} dk_s, \quad (14)$$

where $*_x$ indicates convolution with respect to x and $B(x)$ is the inverse Fourier transform of β_{k_s} ,

$$B(x) = \frac{1}{2\pi} \int_{-\infty}^{+\infty} \beta_{k_s}(k_s) e^{jk_s x} dk_s. \quad (15)$$

Here we have used the Fourier transform property that multiplication in one domain is equivalent to convolution in the other. Thus, the convolution on the left-hand side of (14) clears the β_{k_s} term on the right-hand side. The β_x term, being independent of k_s is simply cleared by multiplication. Taking the right-hand side of (14) as the reconstructed image, it is clear that β_x shades the image as a function of lateral position, whereas the convolution with $B(x)$ introduces a lateral blurring.

Several effects may be anticipated from (14). As the shear wave propagates from its source, any attenuation with distance will lead to decreased image brightness. A laterally varying gain, similar to time-gain compensation in B-mode imaging, could be applied to eliminate

this shading. If β_k is a rectangular window function, such that $\beta_{k_s}(k_s) = \beta_0$ for $|k_s|$ less than some maximum value and zero otherwise, the effect is to convolve the ideal image with a scaled sinc, similar to the effect of a unapodized phased array. The strong lateral side lobes which result are undesirable for image contrast. A smoother β_{k_s} function may be used to reduce ringing. Because the effect of β_{k_s} on I and Q is multiplicative, actual scaling of vibration amplitude with wavenumber or scaling of I and Q after echo collection are equivalent.

Because $B(x)$ represents the lateral blurring function, it is straightforward to predict lateral resolution by calculating the Fourier transform in (15) and applying a suitable resolution metric. For instance, we may consider the full-width at half-maximum (FWHM) size of $B(x)$ as representative of point target resolution. In the case of a uniform weighting of shear wave frequency data up to a maximum spatial frequency $k_{\max}$, $B(x) = \sin(k_{\max}x)/\pi x$. $B(x)$ has a maximum at $x = 0$ of k/π . Solving $B(x)/B(0) = 1/2$ for x yields the FWHM value. Computing this expression numerically yields

$$\text{FWHM} = 3.79/k_{\max}. \quad (16)$$

Again, this resolution result is independent of the ultrasound wavelength and aperture size. For comparison, the FWHM beamwidth of a rectangular aperture of width ω focused at depth z is $1.21\lambda z/\omega$.

Sampling of $I(k_s) + jQ(k_s)$ at discrete values of k_s allows image reconstruction over a limited field of view. In this case, k_s must be sampled densely enough to avoid aliasing (wrap-around) artifacts in the reconstructed image. Given a wavenumber step size of Δ_k , the inverse discrete-time Fourier transform of $R(k_n)$,

$$\sum_{n=-\infty}^{\infty} R(k_n) e^{j\Delta_k n x},$$

is periodic with a period of $2\pi/\Delta_k$. To avoid wrap-around aliasing of the image, Δ_k must be small enough to ensure that the entire illuminated extent of the object is less than π / Δ_k . If the illuminated object is of lateral extent L , then the Nyquist sampling requirement is satisfied if $\Delta_k \leq 2\pi / L$. This inequality may be expressed in terms of the vibration frequency ω through the identity $k = \omega / c$ as

$$\Delta_{\omega} \leq \frac{2\pi c_s}{L}.$$

In those cases where the term $e^{-jcrz|k_s|}$ is negligible, the need to collect echoes for both positive and negative values of k_s is eliminated. This is so because, when $e^{-jcrz|k_s|}$ is one, $I(k_s) = I(-k_s)$ and $Q(k_s) = -Q(-k_s)$. In this case, the echo data generated from backward-going shear waves is redundant, and reconstruction from only forward-going waves is possible.

III. Methods

A. Simulation

1) Uniform Modulus Targets—Point and diffuse scattering targets subject to displacement by a traveling, plane shear wave in a uniform, lossless elastic material were

modeled using Field II [14]. Key simulation parameters are listed in Table I. These parameters were chosen in part to match the phantom experiments described in Section III-B. A single-element transducer, comparable to a single element of a typical 6-MHz linear array (0.2 mm wide), was modeled as the source and receiver. Shear wave propagation was modeled as a plane wave over the frequencies listed in Table I. Scatterers were translated according to (5) and echoes calculated at shear wave phases of $\phi = -90^\circ, 0^\circ, 90^\circ$, and 180° . Intermediate signals $I(t, k_s, \phi)$ and $Q(t, k_s, \phi)$ were calculated as described previously, and a 1-D inverse FFT applied along the k_s direction to generate the images. Zero-padding of k_s domain data was used to achieve interpolation in the image domain.

A wire target phantom was modeled as four point scatterers in the $y = 0$ plane along the diagonal of a 2.5-mm grid, similar to the experimental phantom described in Section III-B. A shear wave frequency range of 50 to 1500 Hz was used to simulate imaging of the wire targets. Image reconstruction was performed with both uniform and Hamming-window weighting of the I and Q signals with respect to shear wave frequency.

A diffuse scatterer target was formed by randomly distributing 5000 point targets over a 1×1 cm region centered 3 cm from the transducer. The scatterers were confined to the y plane. A hyperechoic lesion was simulated by increasing the reflectivity of scatterers within a 4-mm-diameter circle by a factor of 4, for a 12 dB increase in echogenicity. Imaging simulations were performed over shear wave frequency ranges of 50 to 1500 Hz and 50 to 3000 Hz. As with the point target phantom described previously, reconstructions with uniform and Hamming-window weighting were performed.

2) Stiff Inclusion Targets—A second set of point and diffuse targets was simulated to observe the effects of shear modulus variation on image quality. These targets differed from those described above in that the shear modulus was not uniform throughout. Rather, these models consisted of a 2×2 cm block with a centered 1-cm-diameter inclusion. In all cases, background had a shear modulus of 5 kPa, whereas the inclusion was modeled as either 5 kPa (i.e., matched) or 7.5 kPa, 50% greater than the background. Comsol Multiphysics (Comsol, Inc. Burlington, MA) finite element software was used to calculate the propagation of shear waves in the $+x$ direction (left to right) in the model, assuming free boundaries for the top and bottom. A matched lossy layer was added on the right hand side to eliminate reverberations of the shear wave. The deformations calculated by the finite element simulation at each of the four phases of the shear wave were applied to the scatterers in Field II simulation. Echoes were calculated and the image reconstructed as for the other simulations. The wire target phantom in this model consisted of a 5×5 grid of point scatterers with a pitch of 5 mm. The diffuse phantom consisted of approximately 20 000 scatterers randomly distributed throughout the 2×2 cm area of the model. Those scatterers within the 1-cm-diameter inclusion had a reflectivity 12 dB greater than that of the background scatterers.

3) Scan Conversion—The simulated imaging system (as well as the experimental system described subsequently) uses a single unfocused transducer element. The scan geometry is illustrated in Fig. 1. In contrast to the model developed in Section II, here echo arrival time corresponds to range r , rather than depth z . Thus, the reconstructed echo data obtained by applying the 1-D Fourier transform along the k_s direction yields an image in an (r, x) coordinate system. No depth-dependent magnification is present [as in an (r, θ) sector scan data set], but an x -dependent range distortion is present. Scan conversion to a geometrically correct (x, z) space is obtained using the coordinate transformations $x_i = x_o$ and $r_i^2 = z_o^2 + x_o^2$, where the subscripts i and o denote input and output (pre- and post-) scan conversion

coordinates. Aside from the somewhat unusual geometric transformation, the scan conversion process is typical.

4) Compounding—Echo data may be acquired from various locations of the ultrasound transducer to obtain several looks at the target. As the orientation of the ultrasound wavefront to the scatterers is altered, it is expected that the speckle pattern of two scan-converted images taken at different look angles will be not be identical. Incoherent averaging of scan-converted images might then be used to reduce speckle.

In conventional ultrasound imaging with a linear array, spatial compounding may be achieved by subdividing the array into independent apertures and incoherently averaging the images obtained with each subaperture. Likewise, multiple look-angles can be achieved with lateral translation of a phased array [15]. A disadvantage of this method is the loss of lateral resolution caused by the reduced f-number [16]. In the present case, the lateral resolution is determined through the range of shear wavenumbers applied to the target; lateral resolution is independent of the properties of the aperture. Collecting echo data at multiple angles therefore does not entail any loss of lateral resolution in the present scheme, while potentially providing speckle reduction.

Compounding was modeled by simulating echoes for transducer locations from $x_0 = -2$ to 2 cm in 1-mm steps. After envelope detection and scan conversion, images were averaged. The correlation coefficient of speckle patterns as a function of lateral displacement of the transducer was calculated and presented in the Section IV.

B. Experiment

Gelatin block phantoms containing nylon monofilament wire targets were imaged using the setup described below. The blocks were $10 \times 4 \times 6.5$ cm (x, y, z) in size and composed of 7.5% gelatin by weight. Lengths of nylon monofilament (0.13 mm diameter) were embedded in the gelatin block parallel to the y -axis, arranged along the diagonal of a 2.5-mm square grid. The shear modulus of the phantom was determined by unconfined compression testing to be approximately 5 kPa. A second gelatin block phantom contained a 6-mm-diameter hyperechoic cylindrical inclusion in place of the monofilament targets. This inclusion was made of the same gelatin concentration as the rest of the phantom, but included powdered cellulose particles as an ultrasound scattering agent.

The experimental setup for imaging of the gelatin blocks is illustrated schematically in Figs. 1 and 2. A digital function generator (AFG3021B, Tektronix Inc., Beaverton, OR) drove an electromagnetic shaker (4810, Bruel & Kjaer, Nærum, Denmark) through a small amplifier (2706, Bruel & Kjaer). This function generator was synchronized to a second (33220A, Agilent Technologies, Santa Clara, CA) operating at four times the vibration frequency, allowing for echo acquisition at 0° , 90° , 180° , and 270° relative to the shear wave. This $4\omega_s$ signal triggered both the pulser/receiver and digital oscilloscope. Pulse-echo RF data were acquired using a single element of a 7-MHz linear array (Aloka, Tokyo, Japan) driven by a pulser-receiver (DPR300, JSR Ultrasonics, Pittsford, NY). A digital oscilloscope (44MXi, LeCroy Corp., Chestnut Ridge, NY) was used to record the echo signals. Echo averaging was used to improve SNR and the effective resolution of the oscilloscope. For each vibration frequency, 1024 echoes were obtained, 256 for each phase. The large numbers of echoes acquired reflect the limitations of our experimental setup rather than an inherent limitation of the technique. Shortcomings of our experimental system include the lack of synchronization between the oscilloscope sampling clock and the low (8-bit) resolution of the signal. Echo averaging was helpful in mitigating these limitations.

The echo ensemble average was calculated for each phase, and $I(k)$ and $Q(k)$ calculated as in (10) and (11). Note that because $I(t, k) = I(t, -k)$ and $Q(t, k) = -Q(t, -k)$, it is not necessary to generate $-x$ -going shear waves (i.e., waves with negative wavenumbers).

IV. Results

Fig. 3 presents reconstructed images of simulated point targets. The lateral profile of the point targets demonstrates both the correct position of the wires and the predicted form of the lateral component of the point spread function. As predicted by theory (14), the lateral width of the wire images is determined by the range or bandwidth of the shear wave excitation. In the simulation, frequencies up to 1500 Hz are applied. The measured FWHM size of the point targets in the uniform-weighting reconstructions (0.88 mm) agrees with the value predicted by (16). Because of the truncation of the collected data in the frequency (k_s) domain, strong ringing artifacts are associated with the point response when vibration amplitude is constant up to the maximum shear wave frequency. This is most clearly visible in the 40-dB response, seen in Fig. 3(b). The application of a Hamming window to the I and Q signals along the k_s dimension is effective in controlling the side lobe levels, with a consequent tradeoff in main lobe width. Thus, main lobe/side lobe balance may be controlled, just as apodization of the ultrasound aperture would be used in conventional delay-and-sum beamforming.

Simulated images of the diffuse and point target phantoms with and without a stiff inclusion are presented in Fig. 4. The apparent spatially varying tilt of the point targets is due to the scan conversion processes described earlier. The uniform 5 kPa target, both diffuse [Fig. 4(a)] and point [Fig. 4(c)], are rendered with the correct geometry. The targets with the stiff inclusion [Figs. 4(b) and 4(d)] show a distortion caused by the increased shear wave speed in the 1-cm lesion. Point targets for which the stiff inclusion is in the path between the shear wave source and target appear to be shifted to the left. This is consistent with the smaller shear wave phase shift with distance in the stiffer lesion. In principle, this distortion could be overcome if the true shear wave field was known; for instance, if it were to be estimated using conventional B-mode imaging methods, such as sonoelastography [5]. Such reconstructions are beyond the scope of this paper.

Images of the wires in the gelatin phantom are shown in Fig. 5, using the same dynamic ranges as for the simulated wires. The greater lateral blurring, especially evident in the 40-dB grayscale image, is due to deviations from ideal, constant amplitude plane wave shear propagation in the phantom. Although an adjustment was made for the expected decrease in shear wave amplitude with frequency caused by attenuation and mass-loading of the shaker, additional, non-monotonic variations in shear wave amplitude with frequency were observed. This additional modulation leads to the observed distortion.

The reconstructed image of the 6-mm hyperechoic inclusion is shown in Fig. 6. The strong, broad echo at the top of the image is due to a small air bubble trapped at the interface between the cylinder and background. The circular profile of the target is visible and well-localized.

Images of the simulated diffuse target with uniform shear modulus are presented in Fig. 7. Images were simulated for maximum shear wave frequencies of 1.5 and 3 kHz, with either uniform (rectangular) or Hamming-window weighting of the k_s -space data. The high side lobe levels associated with uniform weighting are again visible, particularly in the anechoic regions of the phantom. The curvature of the side lobe artifacts is a result of the scan conversion process. As in the wire target images of Fig. 3, application of the Hamming window to the I and Q signals reduces side lobe levels at the expense of main lobe width.

The improvement in resolution with shear wave bandwidth is clearly visible between the 1.5 and 3 kHz maximum frequency images.

Compounded images of these same targets are shown in Fig. 8. The images are calculated over the same range of shear wave frequencies and using the same windowing functions as those in Fig. 7 but represent the average of 41 images obtained by moving the ultrasound transducer from $x = -2$ to 2 cm in 1-mm steps. The ratio of the mean to the standard deviation (μ/σ) of the echo envelope in the uniform speckle region indicated in Fig. 8 is reported for each of the four compounded images in Table II. The expected value of μ/σ in a fully developed speckle image is 1.9, and N independent speckle patterns would yield a $\sqrt{N}$ increase in μ/σ . Because all computed values of μ/σ are less than $1.9\sqrt{41}$, the speckle patterns obtained at each location are not independent. Correlation coefficients of the speckle patterns in this same region of interest as a function of distance between each transducer position is plotted in Fig. 9 for each imaging configuration. As suggested by the relative values of μ/σ for each configuration, the speckle correlation drops most rapidly with translation of the transducer element for the low-frequency shear wave excitation data with a Hamming window applied, and most slowly for the high shear wave frequency, rectangularly weighted data.

V. Discussion

In contrast to conventional ultrasound imaging, the two-way beam transducer beam pattern does not influence lateral resolution. It does, however, determine the spatial extent of target illumination. In the simulations and experiments presented here, a small aperture, comparable to the ultrasound wavelength, was used. The targets all lay within the far field of the aperture and received relatively uniform illumination. A more tightly focused beam applied to an extended target would produce images of only the area illuminated by the beam, but would not otherwise alter the image. In this method, the aperture serves only to determine the area that is imaged, not to determine the image resolution.

Comparisons of the correlation coefficient of speckle versus aperture translation with results obtained using conventional phased array imaging, as in [15], are useful. In the results obtained here, the images with the lowest spatial resolution, i.e., the 1.5-kHz, Hamming-weighted data, showed the largest reduction in speckle, and the 3-kHz, rectangularly weighted data showed the least reduction. In conventional phased-array imaging, these would correspond to small and large apertures, respectively. The speckle reduction and rate of falloff in correlation with aperture translation are in line with the results obtained by Trahey [15]. The use of spatial compounding with this method is particularly useful. The maximum lateral resolution is set by the smallest shear wavelength used, not by the aperture size. A multi-element array instead serves to reduce speckle noise.

It is important to note that, in contrast to typical Doppler measurements of blood flow, the measured frequency of a vibrating target does not vary with the angle between the ultrasound beam and direction of motion [13]. The amplitude of the echo phase shift varies with the motion/wavefront angle, but the frequency does not. This angle does not result in any apparent shift of targets, but does add an image brightness component. The apparent brightness of a target will vary with the amplitude of the phase shift. This will vary as the sine of the motion/wave-front angle, with zero induced phase shift, and thus image brightness, for motion along the wavefront (i.e., no change in target range) and maximum brightness with motion perpendicular to the ultrasound wavefront. In the simulations and experiments here, the value of sine is very nearly 1, and the effect on the resulting images is negligible.

The development presented in Section II assumed a small-amplitude shear wave, such that higher-order terms of the expansion of the modulation exponential were negligible. Although the approximation is valid for the experiments presented here, it is not required. The use of higher-amplitude shear waves, for which higher-order terms of the expansion are significant, may allow greater resolution. Higher-order terms induce phase modulation components at integer multiples of the shear wave frequency, and corresponding integer multiples of the wavenumber. The tradeoff between effects of shear wave attenuation and feasibility of large amplitude vibration generation are not obvious and require further investigation.

One interesting observation is that, because lateral resolution does not depend on the phase of echoes at various points across the aperture (and in fact can be performed with a point source), the effects of an aberrating layer between the target and transducer are greatly reduced. Although an aberrating layer may distort the overall illumination of the target, the image shape will remain relatively undistorted. A constant shift in the arrival time of an echo from a particular scatterer caused by the aberrator will not distort the lateral position estimate in this method. A possible application is to aid B-mode imaging by using the method to enhance the lateral resolution of a conventionally focused beam.

The most significant obstacle to the realization of the proposed method is the ability to generate shear waves of a known wavelength in tissue. Challenges here include the large variation in shear modulus of the various tissues of the body, and the frequently strong viscoelastic component. The strong attenuation of shear waves with increasing frequency makes it infeasible to propagate them over large distances. These limitations might be addressed by using acoustic radiation force to generate shear waves close to the region of interest.

VI. Conclusion

A method for generating images of echogenicity in elastic targets distorted by traveling shear waves has been demonstrated. Rather than using conventional transmit and receive focusing, this method uses the phase shift induced in echoes from scattering points within the target by traveling shear waves of multiple frequencies. The ultrasound aperture determines the area imaged with this method, but not the lateral resolution. When the shear modulus of the material is known and plane waves can be generated over an adequate range of frequencies, images comparable to ordinary B-mode images may be created. Compounding of images collected from transducer elements at multiple locations has been shown to reduce coherent speckle artifacts and improve image SNR. This method decouples lateral image resolution from the properties of the ultrasound aperture, presenting interesting possibilities for further research.

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Biography

Stephen McAleavey received the B.S. and M.S. degrees in electrical engineering and the Ph.D. degree in electrical and computer engineering from the University of Rochester in 1996, 1998, and 2002, respectively. From 2001 to 2004, he was a Research Associate in the Department of Biomedical Engineering at Duke University. In 2004, he returned to the University of Rochester, where he is currently an Associate Professor in the departments of Biomedical Engineering and Electrical and Computer Engineering. His research interests include Doppler methods, acoustic radiation force-based imaging, and applying tissue motion estimation to develop novel ultrasound imaging methods.

Fig. 1.

Sketch of the physical relationship between key components of the experimental setup. The electromagnetic shaker (top left) induces a $+x$ -going shear wave. The nylon wire targets, indicated by the dots, are illuminated by an unfocused transducer element, i.e., a single element of a linear array, which also receives the echo signals from all targets.

Fig. 2.
Block diagram representation of the signal flow for the data acquisition procedure described in the text.

Fig. 3. Simulated images of point targets in a uniform $G = 5$ kPa material. Images in the left column are shown on a 20-dB grayscale, whereas the right column is a 40-dB grayscale. No weighting is used in the upper image, whereas a Hamming window was applied in the bottom row.

Fig. 4.

Simulated images in regions of uniform and non-uniform shear modulus. The probe location (0, 0) corresponds to the top of the image. Images in (a) and (c) are simulations of a uniform 5-kPa shear modulus material with a 1-cm-diameter hyperechoic region (a) or point targets on a 5-mm grid. In (b) and (d), the shear modulus of a 1-cm-diameter region centered at (30, 0) (i.e., coincident with the hyperechoic lesion) is increased to 7.5 kPa, whereas the background maintains a 5 kPa modulus. Distortion of the hyperechoic lesion (b) and wire target locations (d) is visible.

Fig. 5.

Nylon monofilament wires in a gelatin block as imaged by the shear wave method. The images are displayed on a 20-dB (left) or 40-dB (right) dynamic range grayscale.

Fig. 6. Image of the hyperechoic cylinder in the gelatin block phantom reconstructed from experimental shear wave modulated data. The displayed dynamic range is 30 dB.

Fig. 7.

Simulated images of diffuse targets in a uniform 5-kPa shear modulus material. The maximum shear wave frequency used was 1.5 kHz in the left column (a) and (c), versus 3 kHz in the right column. No weighting of shear wave frequency components was used in the top row (a) and (b), corresponding to a rectangular weighting of the k-space data. A Hamming weighting was applied to the shear wave data for the reconstruction in the bottom row. The lower side lobe levels of the Hamming window are visible in the greatly reduced ringing artifacts. All images are depicted on a 40-dB grayscale.

Fig. 8.

Compounded images formed by averaging speckle images obtained at 41 ultrasound transducer locations from -2 to 2 cm. The maximum shear wave frequency used was 1.5 kHz in the left column (a) and (c), versus 3 kHz in the right column. No weighting of shear wave frequency components was used in the top row (a) and (b), corresponding to a rectangular weighting of the k-space data. A Hamming weighting was applied to the shear wave data for the reconstruction in the bottom row. All images are depicted on a 40-dB grayscale. The region used for computing μ/σ values is indicated by the dashed line in (a); the same region was used in all cases.

Fig. 9.

Speckle correlation coefficient versus aperture translation. Simulated echo data for the correlation coefficient calculations were collected from the right half of the speckle images depicted in Figs. 7 and 8. Coefficients were calculated as in [15]. Speckle images formed with higher shear wave bandwidth (3 kHz) exhibit higher spatial resolution and more gradual decorrelation with aperture translation than the lower bandwidth images.

TABLE I

Shear Wave Modulated Imaging Simulation Parameters.

Shear wave speed (c_s), background	2.2 m/s
Shear wave speed (c_s), lesion	2.2 or 2.74 m/s
Ultrasound frequency (f_0)	6.67 MHz
Shear wave frequency (f_s)	50 to 1500 Hz, 50-Hz steps
Shear wave amplitude (a)	7 μm

TABLE II

μ/σ Values for Several Window and Shear Wave Bandwidth Configurations.

Max frequency (Hz)	1500	3000
Rectangular	5.6	4.2
Hamming	6.0	4.4