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Shear Modulus Imaging with Spatially Modulated Ultrasound Radiation Force

Stephen McAleavey, Manoj Menon, and Etana Elegbe

Abstract

The application of Spatially Modulated Ultrasound Radiation Force (SMURF) to shear modulus imaging is demonstrated in tissue mimicking phantoms and porcine liver. Scanning and data acquisition was performed with a Siemens Antares ultrasound scanner and VF7-3 linear array operating at 4.21 MHz. Modulus estimates in uniform phantoms of ZerdineTM with shear moduli of 5.1 and 12.4 kPa exhibited standard deviations within 6% of the mean value. Zerdine spheres 1 cm in diameter (nominally 2.7, 4.7, and 15 kPa) in a 8 kPa (nominal) background are clearly resolved. Cross sectional images of a soft conical inclusion in a gelatin-based phantom indicate a spatial resolution of approximately 2.5 mm. Images of the shear modulus of an ex-vivo sample of porcine liver tissue show an average value of 3kPa. A stiff lesion induced with 0.5 mL of 10% glutaraldehyde is clearly visible as a region of shear modulus in excess of 10 kPa. A modulus gradient associated with the diffusion of the glutaraldehyde is visible. Two pulse sequences were examined, differing only in the timing of the beams used to generate the shear waves. Details of the beam sequences and subsequent signal processing are presented.

1 Introduction

Methods for quantitative imaging of tissue shear modulus using acoustic radiation force induced shear waves are currently under development. Tissue shear modulus is thought to have significant diagnostic utility, and provides an image contrast mechanism distinct from that of conventional imaging modalities, e.g. B-mode ultrasound, X-ray CT [1]. The ability to reliably quantify tissue modulus may reduce the need for tissue biopsy, and can allow longitudinal tracking of tissue response to therapy [1]. The use of acoustic radiation force to generate shear waves in tissue for the purpose of tissue characterization was first proposed by Sarvazyan [2].

Supersonic Shearwave Imaging (SSI), described by Bercoff, Tanter, and Fink, estimates tissue viscoelastic properties by tracking the progress of radiation force induced shear waves with an ultrafast scanner [3]. Multiple ultrasound tone bursts focused at a sequence of depths and transmitted in rapid succession are used to create a pair of large amplitude shear waves. The propagating shear waves are distorted by variations in shear modulus. Inversion algorithms applied to the tracked motion data are used to recover the shear modulus.

Palmeri et al. propose the Time-To-Peak (TTP) algorithm for shear modulus estimation [4]. In this method a shear wave is generated by the acoustic radiation force of a short (10's of μ s) focused ultrasound toneburst. Tissue motion lateral to the push beam is then tracked at multiple locations. The time to peak displacement at each location is taken as the "arrival time" of the shear wave. Knowledge of the arrival times and distance between the tracking locations allows the shear wave speed c_s to be estimated. The shear modulus G is then

determined through the relationship $G = \rho c_s^2$, where ρ is the tissue density. A linear regression analysis of the measured arrival times is used to make the method more robust to noise.

The method of Shear Dispersion Ultrasound Vibrometry (SDUV) uses a time-varying radiation force to generate a harmonic shear wave [5]. Ultrasonic tracking then measures tissue motion at known distances from the vibration source. The phase shift as a function of distance from the point of excitation is estimated from the motion data to determine the shear wavelength λ . Coupled with the known vibration frequency f , the shear modulus can be estimated as $G = \rho(\lambda f)^2$.

In the TTP and SDUV methods, the propagation path length for the shear waves is assumed to be equal to the distance between the axes of the tracking beams used to measure the passage of the shear wave. With this assumption, estimation of shear wave speed would seem to be a simple matter of determining the time required by the shear wave to propagate past two observation points. However, a significant variance in the shear wave speed estimate is observed with this method, even within a homogeneous phantom [4,6]. The variation is consistent at a given observation point and not due to inadequate SNR. Therefore it cannot be reduced by temporal averaging, and spatial averaging of measurements is required to reduce the estimate variance. This spatial averaging, of course, reduces spatial resolution.

Our previous results suggest that the measurement noise is due to a speckle-induced variation in the effective distance between the track beams [6]. Interference of the echo signal from discrete scattering sources within the tissue gives rise to areas of strong and weak reflection. The tracked echo along a given line will reflect the motion of the scatterers that most constructively interfere. These scatterers are not necessarily located on the beam axis, but are potentially displaced from it. The distance between the tracked points is not, then, simply the distance between the axes of ultrasound tracking beams, but rather a function of the two-way beam pattern and scatterer distribution at a given depth. A similar effect was noted by Ophir et al. in pulse-echo estimates of ultrasonic sound speed estimation [7]. Tracking beam spacings are on the order of 1–4 mm, while beamwidths can be on the order of 0.2–0.5 mm. Thus, this speckle-induced uncertainty in the path length can be a substantial fraction of the path length.

In contrast, the SMURF method performs tracking along a single A-line while pushing at two or more locations. The shear wave speed is estimated from the difference in arrival times of the pulses, and the path length difference is assumed to be equal to the offset in the push beam locations. The critical difference between this method and those described above is that the tracking takes place at a single location. The location of a speckle at a particular depth may bias the arrival time estimate of a shear wave from one pushing location. However, the same bias will be applied to *all* shear waves propagating in the same direction. Estimation of the arrival time difference will subtract away any constant bias in the measurement. A speckle-induced uncertainty in the location of the push beams does not exist, because the radiation force is overwhelmingly determined by the attenuation of the medium rather than its scattering¹ [8]. The scattering contribution to the radiation force is small, and the location of the push beam is well represented by the push beam axes.

A challenge for the SMURF method is the viscoelastic nature of tissues. Our previous development of the SMURF method assumed a linear elastic model that neglected shear wave dispersion. We have found that dispersion effects complicate the application of

¹A possible exception is at strongly reflecting boundaries, as between organs. In the bulk of tissues, scattering is assumed to be weak and to make a negligible contribution to the radiation force.

SMURF in strongly viscoelastic media, e.g. liver tissue. We have developed and present here a modified sequence, denoted as SMURF-mod. Briefly, the SMURF-mod method introduces a time delay between each push beam spatial peak. This delay allows individual shear wave peaks generated by the push beam sequence to be more easily detected, while retaining the advantages of a single tracking location.

The goal of this paper is to demonstrate shear modulus imaging with SMURF. The standard SMURF method is compared with the modified sequence in elastic phantoms. We demonstrate that the two methods provide equivalent results in elastic media. Images of modulus are found to be similar with both methods in elastic spherical and conical phantoms. Modulus estimates in homogeneous phantoms are also shown to agree within the estimate standard deviations. SMURF-mod is found to be superior in liver tissue, while SMURF with similar push beam spacing does not yield a useful modulus estimate. As described in the Methods section, SMURF-mod is slightly more time consuming, as it requires more tracking echoes to generate an estimate.

2 Methods

All imaging was performed with a Siemens Antares scanner (Siemens Medical Solutions, USA, Ultrasound Group). Echo data were collected using the Axius Direct™ research interface [9], which allows acquisition of beamformed RF echo data sampled at 40 MHz and digitized with 16-bit resolution. A linear array transducer (Siemens VF7-3) was used for scanning in all cases. This transducer array consists of 192 elements of approximately 0.2 mm pitch and 7.5 mm height. The elevation focus of the array, set by a fixed cylindrical lens, is 3.8 cm.

2.1 Imaging Sequences

Images of shear modulus are developed from SMURF pulse ensembles. Each ensemble provides an estimate of modulus as a function of range along a ray emanating from the transducer. The grouping of a set of transmitted pulses and corresponding echo signals into an ensemble is conceptually similar to the grouping used to describe a Color Doppler pulse sequence [10]. Multiple ensembles are transmitted with a uniform lateral translation between each ensemble to sweep out a two dimensional area and allow for image formation. In the experiments described here with a linear array transducer, a simple translation in x is applied between one ensemble and the next. In the case of a phased array probe, beam steering would be used to sweep ensembles over the region of interest.

Each ensemble consists of two types of pulses — “tracking” and “pushing” pulses. Tracking pulses are standard B-mode-style pulses used to follow the motion of tissue in response to applied radiation forces. These pulses were generated with a single cycle, 4.21 MHz sinewave excitation. Transmit focal depths of 1.5, 2, 2.5 and 3 cm were used, and in all cases apodized to an F-number of 1.8. Dynamic focusing and apodization was applied to the received echoes. Pushing pulses for generating displacements were 200 cycle (47.5 μ s) tonebursts of 4.21 MHz center frequency. The focal depth of the push beam was matched to that of the tracking beam, and a uniform F/3.5 apodization was used. Field II simulations of the push beam pattern are illustrated in figure 1[11].

Two ensemble types were evaluated. The pulse sequence for each ensemble type is illustrated schematically in figure 2. In the first type (SMURF), a tracking pulse is transmitted at one scan line location. The echo, dynamically focused and apodized, is collected to generate a reference A-line. Two pushing pulses are then transmitted in rapid succession. The focus of each push beam is at the same depth z as the tracking beam but translated laterally by Δp . The distance between the beam axis of the tracking line and the

axis of the first push beam, Δ_T , is independent of the spacing between push beams. A series of tracking pulses is transmitted and echoes collected along the same scan line as the reference echo. Cross correlation processing of the resulting echo signals, as described below, allows the tissue motion in response to the two pushing pulses to be measured. Unless otherwise noted, in the experiments described here the pulse period (time between start of each pulse) was $134\mu\text{s}$, Δ_P was 2.48 mm, and Δ_T was 1.8 mm. The value of Δ_P was chosen such that the first lateral null of the one push beam aligns with that of the other push beam, maximizing the intensity variation in the region of interest. For other values of Δ_P the push beam F-number should be adjusted to maintain the alignment of the push beam nulls. Since the depth of field is proportional to the F-number squared, there is a trade-off between push beam spacing (i.e. lateral resolution) and depth of field. The value of Δ_T was selected so that the propagation delay of the shear wave from the first push location to the tracking location was greater than the time required to transmit the second push beam and two tracking beams. This ensured that the entire shear wave would be observed.

The second ensemble type (SMURF-mod) differs from the first in that tracking pulses are inserted between the first and second push pulses. Following the first push pulse, 30 tracking pulses are transmitted and echoes collected to measure the tissue motion. The second push pulses is then transmitted, translated laterally by Δ_P relative to the first push pulse beam axis and $\Delta_P + \Delta_T$ relative to the tracking beam axis. Forty tracking pulses are then transmitted and echoes collected at the tracking beam location. The extra 10 tracking pulses are added to allow for the extra transit time needed for the shear wave from the second push pulse to reach the tracking beam axis, compared to the shear wave from the first tracking pulse. In both ensemble types all tracking is performed along a single A-line.

SMURF imaging sequences used in this study were made up of 60 ensembles, with a lateral translation of 0.53 mm between ensembles, yielding a lateral field of view of 3.2 cm. SMURF-mod imaging sequences contained 40 ensembles with the same 0.53 mm translation between ensembles, resulting in a 2.1 cm field of view. RF echo data was collected over a 0–4 cm axial window.

2.2 Signal Processing

All signal processing subsequent to echo collection was implemented in MATLAB (The Mathworks, Natick, MA). Tissue displacement relative to the position prior to radiation force excitation was measured using normalized cross correlation of windowed segments of the RF echo signals, as described in [6]. The window segments were $1.5\mu\text{s}$ long, corresponding to a range window of 1.2 mm. Low-pass interpolation was used to increase the RF sample frequency to 320 MHz. Following calculation of the normalized cross correlation of each echo window with the reference echo, parabolic interpolation of the correlation function was used to determine the location of the peak with sub-sample precision. The resulting discrete-time measurements of displacement $d[z, m]$, a function both of target range z and echo index m were converted to a velocity estimate $v[z, m]$ by the finite difference approximation $v[z, m] = (d[z, m + 1] - d[z, m]) - PRF$, where PRF is the tracking pulse repetition frequency. In the case of SMURF echo sequences, the tracking process results in a single two dimensional matrix v of velocity estimates vs. depth and time. The modified-SMURF pulse sequences are tracked identically but split into two matrices, v_1 and v_2 , corresponding to velocity estimates associated with the first and second pushing pulses.

The method used to generate modulus estimates varied slightly depending on the pulse sequence used. For SMURF sequences the autocorrelation method described in [6] was used. The velocity sample was upsampled by a factor of ten by low-pass interpolation to improve the precision of the shear wave delay estimates. For each depth z the autocorrelation R of the upsampled discrete time velocity estimate $v_z[n]$ was calculated as

$$R_v[m] = \sum_n v[z, n]v[z, n+m]. \quad (1)$$

The lag corresponding to the period of the vibration signal at depth z was taken to be the lag of the peak of $R_{v_z}[m]$ in the range $m_{\min} < m < \infty$, where $m_{\min}$ is the lag of the minimum $R_{v,v}$ for a given z . Identification of this peak was improved by suppression of low-frequency components of the velocity signal (due either to transducer motion or the low-frequency, non-spatially modulated component of the velocity signal [12]) by subtracting a second order polynomial fit from the autocorrelation of the velocity signal at each depth. The frequency f of the SMURF-induced shear wave is taken to be the reciprocal of the lag of the identified local maximum of R . The shear modulus G was then calculated as [6]

$$G = \rho(\Delta_p f)^2 \frac{1}{(1 - fT)^2}, \quad (2)$$

where T is the time between push pulses, and ρ the material density (assumed to be 1.0 g/cm³). The factor of $1/(1 - fT)^2$ is a correction term arising from the propagation of the shear wave due to the first pushing pulse before the transmission of the second pulse. This early propagation increases the effective spacing between beams.

Processing of the SMURF-mod data is similar in principle. Cross-correlation of the velocity signals v_1 and v_2 at each depth z was used to determine the difference in arrival time τ due to the extra distance Δ_p the shear wave due to the second push pulse must propagate to arrive at the tracking location. Cross correlation was calculated as

$$C_{v_1, v_2}[m] = \sum_n v_1[z, n]v_2[z, n+m]. \quad (3)$$

The lag M (in samples) of global maximum of $C_{v_1, v_2}[m]$ was taken as the arrival time difference between the two shear waves. Dividing this sample lag by the pulse repetition frequency yields the time difference τ . The shear wave speed c_s is taken to be the ratio $\Delta_p = \tau$, and the modulus obtained through the relation

$$G = \rho c_s^2. \quad (4)$$

Note that, in contrast to the SMURF calculation, no correction factor is applied. The velocity signals $v_1[z, n]$ and $v_2[z, n]$ have their temporal origin set with respect to their corresponding pushing pulses; no time difference would be observed in the case where $\Delta_p = 0$.

For both pulse sequences the shear wave propagation paths are identical except for the difference Δ_p . The estimated modulus thus corresponds to the modulus of the region between the pushing pulses. Because the tracking of the tissue takes place along the same A-line in all cases, any decorrelation or speckle-induced noise is identical for both measurements. The differential nature of the time delay measurement ensures that biases common to both pulses are suppressed.

Similar to conventional B-mode imaging, multiple transmit foci were used to improve the depth of field of the image. Focal depths of 1.5, 2, 2.5 and 3 cm were used. The final image was a composite of the focal zones of each of these sub-images. In the results presented here, no blending of images from each focal zone was performed. The image sections were merely stacked together to form the composite image.

2.3 Phantoms

Three phantom types were used in this study. Homogeneous phantoms, with uniform shear modulus, acoustic wavespeed, attenuation, and backscatter characteristics, were used to verify the calibration of the imaging sequences. A phantom containing spherical lesions of varying shear modulus was used to evaluate the effect of contrast on shear modulus images. Last, a phantom with a conical inclusion was scanned to assess the spatial resolution of the imaging sequences.

Homogeneous Phantoms—Two cylinders of Zerdine™ tissue-mimicking material (Computerized Imaging Reference Systems, Inc, Norfolk, VA) were used as homogeneous phantoms. The cylinders had a diameter of 5.1 cm and a height of 2.4 cm. The nominal sound speed and attenuation were 1540 m/s and 0.5 dB/cm/MHz, respectively. The shear modulus of the compliant ($G=5.5$ kPa) and stiff ($G=12$ kPa) cylinders determined through unconfined compression was taken as the gold-standard value [6].

Spherical Inclusion Phantom—A CIRS Model 049 Elastography Phantom was used as a test target for imaging (Computerized Imaging Reference Systems, Inc, Norfolk, VA). This phantom is ultrasonically tissue mimicking with a sound speed of 1540 m/s and attenuation of 0.5 dB/cm/MHz. The phantom contains spherical inclusions that appear iso-echoic with respect to the background in B-mode but differ significantly in shear modulus. The nominal moduli, as specified by the manufacturer, of the background and scanned inclusions are listed in Table 1. Spheres 1 cm in diameter, centered 2.5 cm below the scanning surface, were imaged. Each sphere was scanned individually, as the field of view was insufficient to include multiple lesions. Scans of the phantom were performed with the scan plane aligned with the diameter of the spherical inclusions.

Conical Inclusion Phantom—To ascertain the effect of lesion diameter on modulus estimate, a conical phantom was fabricated. This phantom, sketched in figure 3 contained conical and cylindrical inclusions of low shear modulus in a relatively stiffer background. The conical section has a height of 33 mm and base diameter 12 mm. The phantom was gelatin based and used cornstarch as a scattering agent [6].

Circular cross sectional scans of the conical inclusion were obtained at nine locations in 4 mm increments along the axis of the cone. Translation of the transducer was performed by a linear stage. The phantom was acoustically coupled to the transducer through a mineral oil at a separation of 1–2 mm. The base of the cone was taken as the zero point for scan plane translation along the axis of the cone. Alignment of the scan plane with the base was achieved by inspection of B-mode images; the scan plane which just included a clear specular reflection from the cylindrical base of the conical section was taken as the starting scan plane. Longitudinal scans of the conical section were obtained with the cone axis aligned within the scan plane.

2.4 Porcine Liver

Porcine liver tissue was obtained fresh at slaughter from a local butcher. The liver was kept immersed in chilled 0.9% saline until scanning. A single lobe of the liver was excised from the whole and kept immersed in saline for scanning. The chilled saline was replaced with

room temperature saline, and the liver allowed to warm for 30 minutes. The liver was scanned to find a plane with uniform B-mode appearance, free from obvious large vessels, and SMURF data collected. A stiff lesion was then induced by injection of 0.5 mL of 10% glutaraldehyde solution. Shear modulus images were obtained at four and eight minutes after the injection.

3 Results

3.1 Homogeneous phantom

Shear modulus images of the uniform Zerdine phantoms are presented in figure 4. These images have not been subjected to any median or spatial average filtering. The mean and standard deviation of the modulus estimate over the field of view for each phantom is listed in Table 2. The mean modulus values agree within the associated standard deviation across each of the beam sequences. No bias in the modulus estimate appears to be introduced by the use of the second-order fit autocorrelation filtering of the SMURF data. The filtering does eliminate some false-peak detection errors that lead to gross underestimates of the modulus, as seen in unfiltered modulus estimates in figure 4. Sample auto- and cross-correlation traces for each method are shown in figure 5.

3.2 Spherical Inclusion Phantom

Matched B-mode, SMURF, and SMURF-mod images of Type I, II, and III spherical inclusions are shown in Figure 6. Modulus values for the lesions are consistent for both estimators. The horizontal band of slightly lower modulus is thought to be a seam (visible in the matched B-mode images) in the phantom introduced in the manufacturing process. The low-modulus artifact in the lower-right quadrant of the SMURF image of Type III lesion is a failure of the autocorrelation estimator. In this region a “false peak” in the autocorrelation with greater lag than the correct value has been identified. This artifact illustrates an advantage of the SMURF-mod method, namely freedom from false-peak errors of this type.

Using the SMURF-mod sequence, it is possible to use a smaller value of Δ_P and still distinguish resulting two shear waves peaks, due to their greater separation in time. This fact leads naturally to the question of whether lateral resolution can be enhanced by closer spacing of push beams of a given size. To address this question, a series of images were formed with varying push beam spacing, shown in figure 7. Modulus estimates are consistent across all push beam spacings but show increased noise with decreasing beam spacing. An increase in measurement variance is expected with decreasing beam spacing, as a given error in the arrival time difference τ becomes a greater fraction of the time difference. While a slight sharpening of the lateral edges is evident, the sharpening is not as large as would be suggested by the reduction in Δ_P . A possible explanation for the modest enhancement in resolution with reduced push beam spacing is the width of the pushing beams, which remained unchanged as the beams were moved together. At the focus the distance between the lateral nulls of the intensity pattern is 2.48 mm. Thus, as the beam spacing is reduced from this value, the beam overlap becomes significant. Furthermore, the variance in the estimate increases, as the uncertainty in the arrival times becomes a larger fraction of the difference in arrival times.

3.3 Gelatin Phantom

Modulus images of cross sections through the gelatin phantom are presented in Figure 8. The diameter in the scan plane varies from 12 to 1.5 mm in the images shown. The modulus estimates at the center of the lesion remain consistent except for the smallest lesions, when blurring with the background results in elevated values. The asymmetry of the smaller diameter sections is due to the anisotropic resolution of this method. The lateral resolution is

strongly determined by the push beam width and spacing, while axial resolution is primarily dependent on the tracking pulse length and tracking window size. In the present case the push beam spacing is 2.48 mm while the tracking pulse and window together are 1.2 mm, implying a 2–1 resolution asymmetry.

An apparent shear modulus variation is visible between 20 and 30 mm depth radiating downward from the left and right edges of the inclusion. This is due to refraction of the push-beam by the ultrasound wave speed differences in the two regions. Pulse echo measurements subsequent to the SMURF scans of this phantom indicated longitudinal wave speeds of 1590 m/s in the background and 1640 m/s in the conical section. This is considered in the discussion section.

Matched B-mode and modulus images of the cone phantom with the scan plane aligned with the cone axis is shown in Figure 9. Both the raw estimate and estimate with a 1 mm × 1 mm spatial median filter and 2× spatial interpolation are presented for comparison.

3.4 Liver

Matched B-mode and SMURF-mod images of porcine liver tissue are shown in Figure 10. Prior to injection of glutaraldehyde the modulus images are relatively uniform with an average estimated shear modulus of 3 kPa. Four minutes after injection of 0.5 mL 10% glutaraldehyde a stiff lesion, approximately 4 mm in diameter, is clearly visible in the modulus image. A more modest increase in stiffness is visible over a larger area surrounding the injection site. This larger region shows greater stiffness eight minutes after injection. The B-mode images are essentially unchanged over the course of the experiment.

Standard SMURF images are not shown, as the method failed in liver tissue. While displacement images showed that both push beams displaced the liver tissue, the initially distinct peaks in displacement merged into a single, broader peak before any significant shear wave propagation occurred. Thus attenuated, the spatially modulated component was too small to be detected successfully at adjacent tracking locations. As the SMURF-mod results suggest, increasing the spacing between pushing beams would allow for distinguishable spatial modulation peaks to be detected, but at the cost of lateral resolution.

4 Discussion

The homogeneous elastic phantom results indicate that both SMURF methods yield comparable modulus estimates. As expected, the resolution of both SMURF and SMURF-mod is similar, based on images of the spherical inclusion phantom. Identical push-beam spacing was used for both methods, and dispersion was negligible. Under these conditions both sequences produce similar results. The SMURF-mod sequence simplifies somewhat the signal processing needed to reconstruct a modulus image. The increased temporal separation makes it easier to time the arrival of the two shear waves and avoid the false autocorrelation peak possible with SMURF. As demonstrated here, however, false peaks can be effectively eliminated by high-pass filtering of the autocorrelation signal. Because of the extra time required to collect the SMURF-mod ensemble, sensitivity to physiological and transducer motion can be expected to increase, and the maximum frame rate adversely affected.

Where the two methods differ significantly in performance is in viscoelastic media. Here the SMURF-mod sequence is superior. Dispersion rapidly attenuated the high-frequency, spatially modulated component of the shear wave generated by the SMURF sequence, rendering it undetectable. While an increased push beam spacing would lower the induced shear wave frequency and allow detection, this would also adversely affect lateral resolution. Introduction of the time delay between push beams in the SMURF-mod sequence

allowed for successful timing of shear wave propagation. The advantages of tracking at a single location are retained in the SMURF-mod sequence.

In Figure 8 an image artifact is visible distal to the left and right edges of the circular inclusion. This artifact is visible as lines of low apparent shear modulus relative to the background. These lines are due to the difference in the longitudinal (ultrasound) wave speed between the inclusion and background. While sound speed error in the inclusion causes a refraction of all the beams, it is significant in the modulus reconstruction only to the degree that it causes a change in the effective push-beam spacing Δp . Refraction of the tracking beam (unless extreme) affects the shear wave signals from both push beams equally and does not bias the result. Refraction-induced variation in Δp , on the other hand, leads to an underestimate of modulus if the beams refracted such that Δp is increased leads to an underestimate of the true shear modulus. While undesirable, this artifact would also exist in methods that track the propagation of a single shear wave at multiple locations, and is not inherent to SMURF. A related concern is that variations in the tissue absorption and scattering, along with the separate propagation paths of the push beams, may induce shear modulus measurement variance. While the literature indicates that absorption is the dominant mechanism of attenuation in liver [8] and thus one might expect radiation force to be likewise determined primarily by tissue absorption properties, we have not demonstrated this here.

5 Conclusion

A demonstration of shear modulus imaging using the SMURF method has been presented. Modulus images of uniform phantoms show values consistent with those obtained by unconfined compression. The standard deviation of the estimates over the field of view was 3–6% of the mean estimated value. Cross sectional images of a conical inclusion with half the modulus of the background indicate that structures comparable in size to the push-beam spacing (2.5 mm) may be resolved. Both the SMURF and SMURF-mod sequences yielded comparable modulus estimates in elastic phantoms. In the more viscoelastic liver tissue it was found that the SMURF-mod sequence yielded superior results. Future work will investigate the effect of push-pulse timing and beam spacing on measurements in viscoelastic materials.

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Figure 1. Simulated push-beam patterns. The Siemens VF7-3 transducer was modeled using Field 2. The relative intensities of a 4.21 MHz, 100 cycle toneburst in a 0.5 dB/cm/MHz medium was modeled for the focal depths indicated. The images show the total intensity pattern over the two push beam transmissions. The grayscale in each panel is scaled to the peak intensity within the field of view.

Figure 2.

Schematic illustration of pulse sequences used in generating shear modulus images. The relative locations of reference/tracking beams (R,T) in relation to the two push beams (P_1 , P_2) are shown in (a). The push beams are separated by lateral distance Δ_P while the track beam is displaced by Δ_T from the left push beam. The relative push and track locations are the same for both sequence types and all ensembles; the ensembles are walked across the aperture to build up an image. In (b) the timing of the two pulse sequence types are shown. At top (A) is the sequential SMURF sequence. Two B-mode style pulses (R_0 , R_1) establish the reference echo to which later tracking pulses (T_1 – T_N) are compared to determine displacement. Pushing pulses P_1 and P_2 are fired in rapid succession to generate the spatially varying radiation force. At bottom (B) is the SMURF-mod sequence. Following transmission of the reference pulses, a single push beam at P_1 is transmitted and tracked with B-mode pulses $T_{1,1}$ – $T_{1,N}$. The second pushing pulse is launched at location P_2 and tracked with pulses $T_{2,1}$ – $T_{2,M}$. All tracking occurs along the same a-line for a given ensemble.

Figure 3. Sketch of cone phantom. The shaded area represents the background, while the light areas indicate the inclusions.

Figure 4.

Unfiltered modulus images of uniform Zerdine phantoms. Images of the compliant phantom are in the left column; the stiff phantom is on the right. (a,b) SMURF without autocorrelation filtering. (c,d) SMURF with autocorrelation filtering. (e,f) SMURF-mod.

Figure 5.

Sample velocity vs time signals and correlation processing. Top Row: $v(z, t)$ signal obtained in the stiff (12.4 kPa nominal) uniform Zerdine phantom using (a) SMURF and (b) SMURF-mod sequences. Bottom Row: The autocorrelation of $v(z, t)$ for $z=2$ cm is plotted with (dot-dash), and without (solid) autocorrelation filtering. In (d) the cross-correlation signal is plotted for the SMURF-mod sequence. The lags of the peaks in (c) are greater than the lag of the peak in (d) due to the time between firing of the two push pulses in the SMURF sequence. This time is not included in the SMURF-mod sequence.

Figure 6.

Shear modulus images of 1 cm diameter inclusions in the CIRS Model 049 phantom. A push beam spacing of 2.48 mm was used for all images. Top row: Type I lesion, imaged with (a) B-mode, (b) SMURF, and (c) SMURF-mod sequence. Middle row: Type II lesion, imaged with (d) B-mode, (e) SMURF, and (f) SMURF-mod sequence. Bottom row: Type III lesion, imaged with (g) B-mode, (h) SMURF, and (i) SMURF-mod sequence. The modulus images have been subjected to a $1\text{ mm} \times 1\text{ mm}$ spatial median filter.

Figure 7. Images of Type I phantom obtained with varying push beam spacing using SMURF-mod sequence. Push beam spacings indicated above each figure. Noise in the modulus estimate increases with decreased push beam spacing.

Figure 8. Matched B-mode and SMURF cross-sectional images of the conical section within the gelatin phantom. A $1\text{ mm} \times 1\text{ mm}$ spatial median filter has been applied to the SMURF images.

Figure 9. Images of a longitudinal cross section of the conical phantom. (a) B-scan (b) SMURF image with $1\text{ mm} \times 1\text{ mm}$ spatial median filter.

Figure 10.

Matched B-mode and SMURF-mod images of porcine liver. Bmode (a) and SMURF-mod (b) images before injection of 0.5 mL, 10% gluteraldehyde solution. Bscans and SMURF-mod images at 4 minutes post-injection (c and d) and 8 minutes (e and f). Modulus images have been median filtered over a 1×1 mm window.

Table 1

Nominal shear moduli of scanned components of CIRS Model 049 phantom

Material Shear	Modulus (kPa)
Background	8.3 ± 1.3
Type I Lesion	2.7 ± 1
Type II Lesion	4.7 ± 1.3
Type III Lesion	15 ± 1.7

Table 2

Modulus data for uniform phantoms. Values given are the mean shear modulus values (in kilopascals) over the field of view in Figure 4. Values in parenthesis are the sample standard deviation of the estimates over the field of view.

	MTS	SMURF (none)	SMURF (filtered)	SMURF-mod
Compliant	5.1	5.51 (0.66)	5.59 (0.19)	5.73 (0.18)
Stiff	12.4	13.8 (1.8)	13.7 (0.86)	12.9 (0.32)