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Plane wave imaging improves single track location shear wave elasticity imaging

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Abstract

Single track location shear wave elasticity imaging (STL-SWEI) is immune to speckle bias, but the quality of the images is depth-dependent. We hypothesize that plane wave imaging can reduce the depth dependency of STL-SWEI. To test this hypothesis, we developed a novel technique known as plane-wave single track location shear wave elasticity imaging (pSTL-SWEI). To evaluate the pSTL-SWEI's potential, we performed studies on phantoms and excised murine pancreatic tumors. The mean shear wave speeds (SWS) measured with STL-SWEI and pSTL-SWEI were similar. However, the elastographic signal-to-noise ratio (SNR_e) of pSTL-SWEI elastograms was noticeably higher than that produced with STL-SWEI. Specifically, we observed an improvement in SNR_e ranging from 39.9%–55.1%, depending on tissue stiffness. The spatial resolution of pSTL-SWEI elastograms was 2.7%–12.1% lower than that produced with STL-SWEI. pSTL-SWEI elastograms displayed higher contrast-to-noise ratio (CNR_e) than those produced with STL-SWEI, especially when imaging was performed with low push pulse intensities and low pulse durations.

I. Introduction

Shear wave elasticity imaging (SWEI) measures tissue mechanical properties in real-time [1], but speckle bias affects most estimates of shear wave speed [2], [3], [4]. Shear wave elasticity imaging uses a four-step approach to measure shear modulus. First, an acoustic radiation force induces shear waves within the tissue under investigation. Second, diagnostic ultrasound (plane-wave or focused ultrasound imaging) tracks the propagating waves. Third, differences in arrival times of shear waves at known coordinate locations within the tissue provide an estimate of local shear wave speed (SWS). Fourth, combining the local tissue density (ρ) and shear wave speed provides local shear modulus (μ) estimates ($\mu = \rho \times SWS^2$). Shear wave elasticity imaging is now standard on most commercial ultrasound

scanners [5], which investigators use to conduct either pre-clinical [6] or clinical evaluations [7], [8]. Tracking shear waves at multiple spatial locations (MTL-SWEI, **Fig. 1a**), the standard tracking approach [2], [4], [9] employed in commercial ultrasound systems, randomly biases the arrival time estimate of the shear wave due to random variations in speckle brightness ([10]). For large homogeneous organs, such as liver [11], SWS are measured over large propagation windows compared to the size of speckle, which reduces the impact of speckle bias. However, this is not the case for small heterogeneous organs [6] or during ablation imaging [12] - estimating local SWS over small propagation windows does not reduce speckle bias, thus degrading the quality of the resulting images [13].

Single track location shear wave elasticity imaging (STL-SWEI) reduces the random bias incurred in the track locations, arrival times, and subsequent shear wave speed estimates [11]. STL-SWEI uses a pair of push beams and a single tracking beam to estimate shear wave speed, as illustrated in **Fig. 1b**. The first push beam induces shear waves within the tissue, whose arrival time is estimated at a fixed distance. The second push beam is transmitted at a different location (displaced laterally) within the tissue or phantom, and the arrival time required for the induced shear wave to traverse the increased path length [10] is measured at the same coordinate location as the first push beam. Traditionally, STL-SWEI uses a focused ultrasound beam for all tracking, but other approaches such as supersonic shear imaging (SSI) and comb-push elastography (CUSE) employ plane wave imaging because it simultaneously imaging all of the required tracking locations, which would otherwise need to be imaged sequentially. The lateral resolution of plane wave images is lower than that acquired using focused ultrasound because no focusing is performed during the transmission. Compounding plane wave images acquired from multiple transmission angles minimizes this problem [14]. The data redundancy associated with the parallel beamforming method employed in compounded plane wave imaging should allow us to synthesize multiple STL-SWEI datasets from a single acquisition. Synthesizing multiple STL-SWEI data is advantageous because it will allow us to make numerous estimates of SWS from the fabricated dataset at each coordinate location, which when averaged will reduce the variance incurred in SWS and enhance elastographic performance.

Hollender *et al.* [15] were the first to develop STL-SWEI based on plane wave imaging, which they implemented on a partially parallel receive ultrasound system. Our approach differs from this early work in two ways. First, we used compounded plane wave images to track the propagation of shear waves, whereas Hollender *et al.* performed no compounding in their approach. Although we have not compared the impact of using compounded rather than non-compounded plane wave images for tracking, researchers have demonstrated that compounding improves the accuracy and precision of SWS estimates, noticeably [16]. Second, we implemented pSTL-SWEI on a complete parallel receive ultrasound system equipped with a pixel-based software beamformer that allows us to synthesize track lines for a given push beam at any lateral position, which provides more tracking lines for averaging. Hardware based beamforming systems employed by Hollender *et al.* often utilize a two-stage beamforming process to reduce the computational burden of the beamforming process, but this may produce grating lobes, which reduces the accuracy of SWS [13], [17].

This article reports the results of studies conducted with calibrated phantoms and excised murine pancreatic tumors to evaluate the performance of plane wave single-track-location shear wave elasticity imaging (pSTL-SWEI) relative to STL-SWEI and MTL-SWEI. The results demonstrate that pSTL-SWEI elastograms offer better elastographic performance than their STL-SWEI and MTL-SWEI counterparts. Additionally, pSTL-SWEI produces useful shear modulus images with less intense push beams than standard STL-SWEI.

II. Methods

A. Plane wave single track location shear wave elasticity imaging (pSTL-SWEI):

A pSTL-SWEI beam sequence consists of multiple push beams each followed by plane wave imaging to track the resulting shear waves (**Fig. 2**). These push beams are located at the central region of the aperture with equal lateral intervals. Like CUSE, pSTL-SWEI uses multiple focused transmit beams to induce shear waves, and plane wave imaging to track the propagating waves. However, in pSTL-SWEI, the push beams are transmitted sequentially rather than simultaneously as in CUSE. pSTL-SWEI uses the STL-SWEI approach to estimate SWS in the central region (between two push beams) of the elastogram, and the MTL-SWEI technique in the flanking regions. In the central region, we synthesize an STL-SWEI dataset consisting of pair of push beams with a common tracking line (located at distance T), as illustrated in **Fig. 2**. At each coordinate location we synthesize numerous STL-SWEI datasets by varying the distance T between the push pair and their common track lines. From each STL-SWEI dataset, we use the normalized cross-correlation technique to estimate SWS. To aggregate the estimated SWS values into a single SWS estimate, we weighted each measure of SWS by the square of the normalized cross-correlation coefficient (peak value), then computed the average of the weighted values. STL-SWEI only measures SWS between push beams; therefore, we use the MTL-SWEI approach to extend the width of the SWS image. Specifically, we synthesize multiple MTL-SWEI datasets from the same acquisition. We average (weighted) SWS values estimated from the synthesized MTL-SWEI datasets. We used the two-dimensional auto-correlation method (3×2 pixels in axial and slow-time direction) described in [18] to estimate the shear wavefields. To remove high frequency jitter, we applied a second order bandpass Butterworth filter (cut-off frequencies at 50 Hz and 1 kHz) to particle velocities measured along the slow-time direction. To suppress reflection artifacts, we applied 2D directional filters [19] to wave data acquired from each push beam. We estimated SWS by applying the GPU-based cross-correlation tracking method described in [20] to slow-time wave profiles corresponding to either a push pair (STL-SWEI and pSTL-SWEI) or a track pair (MTL-SWEI). We performed all SWS reconstruction on a Nvidia Graphics Processing Unit (Tesla k20X, Nvidia Corporation, Santa Clara, CA, USA).

B. Elastographic data acquisition

We implemented three shear wave elasticity imaging approaches (pSTL-SWEI, STL-SWEI, and MTL-SWEI) on a research scanner (Vantage 64, Verasonics Inc., Kirkland, WA, USA) equipped with an L7-4 (Phillips Healthcare, Andover, MA, USA) linear transducer array. **Table I** provides the key push and tracking beam parameters used in all studies, unless otherwise stated. During MTL-SWEI and pSTL-SWEI, we performed compounded plane

wave imaging (CPW) with a pulse repetition frequency of 7 kHz. We acquired plane wave images from three different transmission angles (-4° , 0° , and 4°). During STL-SWEI, we set the focus of the tracking beam to 30 mm. To ensure that the pulse repetition frequency (PRF) of pSTL-SWEI and MTL-SWEI were similar to those achieved during STL-SWEI, we performed CPW with the sliding window compounding approach described in [21]. We transferred all in-phase and quadrature (IQ) echo data from the scanner and stored it on an external computer for off-line post-processing. We implemented STL-SWEI as described in [10] where shear waves generated from a pair of push beams are tracked at a fixed location. The fixed tracking location and the push pair were translated over the entire field-of-view. We kept the distance (P) between push beams constant (3 mm unless otherwise stated) during imaging.

For most studies, we performed STL-SWEI with 30 pairs of push beams (total 60 push beams) to cover a field-of-view (FOV) of 18 mm. We can potentially reduce the number of push beams in STL-SWEI by keeping tracking location fixed while translating the push beams as mentioned in [15], but the increasing distance between the push and tracking beams would increase shear wave attenuation noticeably. For example, the second and the third push beams in **Fig. 1b** would have been the same if their track beams did not change their positions, which would enable the use of fewer push beams. On the other hand, the tracking lines in pSTL-SWEI are spatially invariant and the pushbeam pair synthesis is performed in post-processing. Thus, pSTL-SWEI requires fewer push beams than STL-SWEI of [10]. For most studies, we used 35 push beams (30 synthesized push pair) in pSTL-SWEI. Push beams were delivered laterally 0.6 mm apart. Unless otherwise stated, the distance T between push pairs and track beams was varied from 4.2 mm to 8.7 mm in order to synthesize 16 STL-SWEI data sets (maximum number of tracking lines allowed on the left). The distance T was measured with respect to the left push beam in a push pair. Thus, the maximum distance between push and track was 11.7 mm (maximum T + spacing between pushes), which may need to be minimized in the case of highly attenuating tissue. Tracking lines were always synthesized on the left side of the push pair, however they can also be synthesized on the right. Within the flanking regions, we extracted pairs of track lines that are 3 mm apart. For each tracking pair, we processed and averaged shear wave profiles from 5 push beams at distances of 4.2 mm to 6.6 mm from the tracking lines. The MTL-SWEI technique was implemented with two push beams at 20.4 mm apart with plane wave tracking for each push beam. The MTL-SWEI data set was also processed using the cross-correlation technique. The distance between track lines (3 mm) used in the time-of-flight calculation was equal to the spacing between push beams in a push pair of STL-SWEI and pSTL-SWEI.

C. Acoustic field measurement

We used a calibrated broadband hydrophone (Model HGL-0085, Onda Corporation, Sunnyvale, CA, USA) to measure the intensity of the push beam pulses. All hydrophone signals were digitized with a digital oscilloscope (Model DS1104Z, RIGOL Technologies, Beijing, China) for transmit voltages of 11.9 V, 15.3 V, 22.1V, 29.0 V, 35.8 V, 42.7V, 52.9 V and 63.2 V. When calculating spatial peak intensity of the push beam ($I_{sppa,0.3}$), we took tissue attenuation (0.3 dB/cm/MHz) into account as described in [3].

D. Data analysis

We assessed the elastograms qualitatively by visually inspecting the images, and quantitatively using the elastographic signal-to-noise ratio (SNR_e), elastographic contrast-to-noise ratio (CNR_e), and resolution (R_{20-80}) performance metrics.

We defined SNR_e as follows [15]:

$$SNR_e = \frac{\mu}{\sigma}, \quad (1)$$

where μ represents the mean SWS in an elastically homogeneous region of the tissue, and σ represents the variance of SWS values within the region.

We defined CNR_e as follows [22]:

$$CNR_e = \frac{|\mu_L - \mu_B|}{\sqrt{\sigma_L^2 + \sigma_B^2}} \quad (2)$$

where μ_L and μ_B represent the mean SWS in the inclusion and surrounding tissue respectively; and σ_L and σ_B represent the variance of SWS in the inclusion and surrounding tissue respectively.

We used the R_{20-80} performance metric [23] to measure the resolution in a transition region, which was defined as follows:

$$R_{20-80} = 2 \ln(4)\lambda, \quad (3)$$

where λ represents the width of a transition region. We estimated λ by fitting a sigmoid profile to a transition layer. The sigmoid curve was defined as follows:

$$c(x) = c_2 + \frac{(c_1 - c_2)}{1 + e^{\frac{x - x_1}{\lambda}}} \quad (4)$$

where c_1 and c_2 are SWS at two sides of the region and x_1 is the transition boundary. For a given SWS profile at a transition region, c_1 , c_2 , x_1 and λ are obtained from non-linear least square fitting.

III. Phantom studies

In this section, we describe studies we conducted on groups of phantoms (homogeneous, heterogeneous, two-layer, and viscoelastic). The goals of these studies were to assess how (a) stiffness influences the mean SWS and SNR_e of elastograms produced with STL-SWEI,

MTL-SWEI, and pSTL-SWEI; (b) averaging influences mean SWS and SNR_e of pSTL-SWEI elastograms; (c) push-beam spacing influences performance (resolution, CNR_e , mean SWS) of STL-SWEI and pSTL-SWEI elastograms; (d) shear attenuation influences elastographic image quality. For all studies unless otherwise stated, we conducted shear wave imaging with a push beam spacing (ΔP) and tracking distance (ΔT) of 3 mm and 6 mm, respectively. The default push beam spacing (3 mm) represented the median value of the range of push beam spacing explored in this work. Similarly, the decision to use a 6 mm tracking distance was guided by the results of an empirical study (not discussed here). We observed that 6 mm represented the optimum spacing for tracking shear wave propagation over the stiffness range (3.4 to 50.4 KPa) of the phantoms used in this study. The reader should also assume that, unless otherwise stated, a total of 60 (30 pairs), 35 (30 synthesized pairs), and 2 push beams were employed during STL-SWEI, pSTL-SWEI and MTL-SWEI, respectively.

(a) Evaluating STL-SWEI, MTL-SWEI, and pSTL-SWEI elastograms:

We performed studies with four commercially available homogeneous phantoms (Model 039, CIRS Inc., Norfolk, VA, USA) with stiffness values ranging from 3.4 kPa to 50.5 kPa. The stiffness values reported by the manufacturer of the calibrated phantoms differed from elastographic measurements and are well reported in the literature [15], [24]; therefore, we used the Virtual Touch Quantification (VTQ) method implemented on the Siemens Acuson 300 scanner (Siemens-Acuson, Mountain View, CA, USA) to provide independent estimates of the SWS of the phantoms. We performed all measurements on the Siemens scanner with a 8 MHz 9L4 probe. For each phantom, we conducted 20 VTQ measurements with 30 mm transmit focus. For pSTL-SWEI, homogeneous phantoms were scanned with fewer push beams (25 push beams, 20 synthesized pairs) to illustrate the FOV gain through hybrid processing.

(b) Evaluating the robustness of STL-SWEI and pSTL-SWEI:

To evaluate the robustness of pSTL-SWEI and STL-SWEI to random noise (low shear SNR, electrical noise, and tracking errors), we performed these studies on the 50.4 KPa phantom. Specifically, we varied the transmission voltage from 12 to 63 volts and the temporal width of the push-beam from 15 μ s to 200 μ s. We also assessed how averaging influenced pSTL-SWEI image quality. Specifically, we gradually increased the number of track lines that were averaged from 1 to 16. While increasing the number of averaging, track lines at gradually increasing distances from push were included.

(c) Evaluating lesion detectability:

To assess how well STL-SWEI, pSTL-SWEI, and MTL-SWEI can detect inclusions, we performed studies on a heterogeneous phantom (Model 049A, CIRS Inc., Norfolk, VA, USA) containing cylindrical inclusions of different diameters and contrasts. We performed studies on two cylindrical inclusions with diameters of 7.55 mm and 6.0 mm. One inclusion was stiffer than the background and the other was softer. Like the homogeneous phantoms, we used the VTQ method to quantify the stiffness of the inclusions and surrounding background. All SWEI parameters, except the number of push beams, were identical to those used in the studies with the homogeneous phantoms. Specifically, to ensure the

inclusions were inside the central portion of the elastograms, we transmitted 35 push beams (30 synthesized push pairs) during pSTL-SWEI.

(d) Evaluating the resolution of STL-SWEI and pSTL-SWEI elastograms:

To quantify the resolution of STL-SWEI and pSTL-SWEI elastograms, we fabricated a two-layer phantom (60 mm × 60 mm × 60 mm) from a suspension consisting of 2% (by weight) cornstarch, porcine skin gelatin (300 bloom, Type A, Sigma-Aldrich Corporation, St. Louis, MO, USA), and 18 MΩ high purity water. We fabricated the layers of the phantom from two different concentrations of gelatin: 13% and 10% by weight. We acquired STL-SWEI and pSTL-SWEI datasets from the two-layer phantom. In both approaches, we used push beam spacings of 1.8, 2.4, 3.0, 3.6 and 4.2 mm. Effect of averaging was evaluated in the same manner as the homogeneous phantoms.

(e) Evaluating the effects of shear wave attenuation:

We used a commercially available viscoelastic phantom (Serial no. 2095.1–1, CIRS Inc., Norfolk, VA, USA) to assess how shear wave attenuation impacts elastographic image quality. The shear wave attenuation coefficient was approximately $0.606 \pm 0.063 \text{ Np mm}^{-1} \text{ kHz}^{-1}$, which we estimated with a “first order” dispersion model [25], [26]. During pSTL-SWEI imaging, SWS were averaged over a large number (up to 16) of tracking lines; however, in attenuating tissues, only a limited number of tracking lines may contain useful shear waves. To understand this effect, we performed studies on the viscoelastic phantom evaluating the impact of averaging.

IV. Ex vivo murine tumor study

To further assess the performance of pSTL-SWEI elastograms, we conducted studies with a murine pancreatic tumor model [27]. To develop the model, we cultured murine pancreatic tumor cells (Pan02) in Dulbecco’s Modified Eagles Medium (DMEM) supplemented with 10% phosphate buffered saline (PBS), which we implanted in the pancreas of two C57BL/6 mice (The Jackson Laboratory, Bar Harbor, ME, USA) with protocols approved by the Institutional Animal Care and Use Committee (IACUC) of the University of Rochester. We grew the tumors by injecting 1×10^5 tumor cells in Matrigel™ (BD Biosciences, San Jose, California, USA) and 50% media into the tail of the pancreas of five C57BL/6 mice with a tuberculin syringe. We allowed the tumors to grow until their volume was between 125–175 mm³, the smallest range of tumor volumes that our shear wave elastographic imaging system can detect.

All tumors were surgically removed and encased in a 60 mm × 60 mm × 60 mm gelatin block as described in [28]. We manufactured each gelatin block from a suspension consisting of 10% by weight porcine skin gelatin (300 bloom, Type A, Sigma-Aldrich Corporation, St. Louis, MO, USA), 2% by weight corn starch (Spectrum Pharmaceuticals, Henderson, NV, USA), and 18 MΩ high purity water. To assess our ability to detect the embedded tumors in the elastograms, we computed CNR_e as defined in equation (2). To perform this analysis, we manually outlined the tumors that were visible in the sonograms, and assumed the remaining image was the background.

V. Results

Comparison of pSTL-SWEI, STL-SWEI and MTL-SWEI

Qualitatively, STL-SWEI and pSTL-SWEI elastograms were generally less noisy and contained fewer artifacts than those produced with MTL-SWEI (**Fig. 3**). Reflection artifacts were visible in STL-SWEI elastograms acquired from high contrast heterogeneous phantoms. Specifically, we noticed higher shear modulus on the left side of the inclusion than on the right side, which is consistent with previously reported research [19]. The directional filter employed in pSTL-SWEI and MTL-SWEI suppressed reflection artifacts; however, this filter could not be applied to the STL-SWEI approach used in [10] because it doesn't capture the complete wave field due to the translation of the common track beam. Statistical analysis (Student's t-test) revealed no significant difference in the mean SWS estimated with MTL-SWEI, STL-SWEI, and pSTL-SWEI ($p < 0.05$).

Despite the equivalence of SWS estimated with the three approaches, pSTL-SWEI elastograms provided higher SNR_e than those created with either STL-SWEI or MTL-SWEI (**Fig. 4**). SNR_e was 39.9% to 55.1% higher in pSTL-SWEI than STL-SWEI. Closer examination of pSTL-SWEI elastograms revealed that the SNR_e of the central portion of the image was noticeably higher (177.8% to 263.5%) than that of the flanking regions (**Fig. 5**). Although this had negligible effect on the mean SWS, the noise (variance) generated by the two elastographic imaging approaches were different (**Fig. 5**).

For inclusion phantoms, all three methods have similar SWS values in the background (within 5% of the VTQ-measured SWS). However, SWS estimated by STL-SWEI inside the stiff inclusion was higher than other two methods (see **Fig. 6**). Specifically, the mean SWS estimated with STL-SWEI was 4.4%, 10.2% and 4.9% higher than those achieved using pSTL-SWEI, MTL-SWEI and VTQ, respectively. pSTL-SWEI produced a CNR_e improvement of 39.9% and 85.6% compared to STL-SWEI in stiff and soft inclusion phantoms, respectively. **Table II** reports the nominal and VTQ values of the homogeneous and heterogeneous phantoms. Shear moduli of the CIRS phantoms (homogeneous and heterogeneous) estimated with the VTQ and pSTL-SWEI differ from those reported by the manufacturer. Specifically, pSTL-SWEI estimates of shear modulus agree more with those measured with VTQ than the nominal values. Differences between the VTQ measurements and those made with STL-SWEI, MTL-SWEI, and pSTL-SWEI could be due to differences in the frequency content of the shear wave [29], push F-number, transducer, and system dependent variations [30].

Impact of averaging

The averaging procedure employed in pSTL-SWEI influenced SWS (**Fig. 7a**) negligible, but impacted SNR_e more significantly. For all phantoms, SNR_e improved noticeably with increasing averaging (**Fig 7b**). For example, the SNR_e increased from 45 to 119 when the number of tracking lines averaged was increased from 1 to 16 in the softest phantom.

Impact of averaging on resolution

To assess how averaging influences the spatial resolution of pSTL-SWEI elastograms, we performed studies with a bilayer phantom. We expect that the wider lateral beam profile of plane wave imaging may have an impact on the lateral resolution of pSTL-SWEI; therefore, we restrict our analysis to evaluating the lateral resolution only. **Figure 8 (a-b)** shows lateral SWS profiles taken from the focal region for $P = 1.8$ mm and $P = 4.2$ mm, respectively. Increasing push beam spacing increases the smoothness of the STL-SWEI curve. Visually, the resolution of elastograms produced with both approaches was similar. However, **Fig. 8c** demonstrates that STL-SWEI has better lateral resolution than pSTL-SWEI. Specifically, we estimate that the lateral resolution of the STL-SWEI elastograms was approximately 2.7% to 12.1% higher than the resolution of pSTL-SWEI. We attribute the difference in resolution to the wider lateral beam profile of plane wave images [16]. We could use coherent compounding with a large number of steering angles to solve this problem, however it would lower the frame rate. Increasing the push beam spacing improved the performance of both methods, but the CNRe of pSTL-SWEI was consistently higher than that achieved using STL-SWEI (**Fig. 8d**). In **Fig. 8e** CNRe and lateral resolution are plotted against each other as a parametric function of push beam spacing. pSTL-SWEI achieved better CNRe at the expense of lateral resolution. Higher CNRe allows us to perform pSTL-SWEI with smaller push beam spacing, which improves spatial resolution. For example, averaging 16 tracking lines when performing pSTL-SWEI with 1.8 mm push spacing produced images with similar CNRe to those achieved with STL-SWEI using 4.2 mm push spacing (**Fig. 8(f)**). The relationship between averaging over tracking lines and resolution was found to be complex, but in general there was a trend of improving resolution with an increasing degree of averaging (**Fig 8f**).

Impact of noise

To assess how image quality varies with depth when transmit voltage and push width were varied, we divided the elastogram into five regions as illustrated in **Fig. 9(a)**. We used **Fig. 10** to convert the transmit voltages to acoustic intensity and mechanical index by means of hydrophone measurement. For both STL-SWEI and pSTL-SWEI, SWS increased with increasing spatial peak intensity up to 240 watt/cm² then plateaued (**Fig. 9b**). We observed a similar trend when we varied pulse duration and kept the transmit voltage fixed. Specifically, beyond a threshold duration (20 μ s), SWS was independent of pulse duration (**Fig. 9c**). The push beam's spatial peak intensity and duration also influenced the SNRe of the resulting elastograms (see **Fig. 11**). In all regions, pSTL-SWEI with 16 track line averaging performed better than STL-SWEI. At the shallowest region (region A), the performance of pSTL-SWEI elastograms was better than elastograms created with STL-SWEI even without averaging. pSTL-SWEI had better depth performance because plane wave images had higher sonographic SNR over a larger depth. **Figure 11** also illustrates that in all regions, except region E (beyond focus), we can achieve the same performance as STL-SWEI operating at 200 μ s by using pSTL-SWEI operating at a 75 μ s push pulse. This represents a 63% reduction in pulse width. These results demonstrate that pSTL-SWEI achieves depth gain from plane wave imaging and is robust to noise associated with weaker shear waves compared to STL-SWEI.

Ex-vivo tumor study

In general, pSTL-SWEI elastograms obtained from pancreatic murine tumors were less noisy compared to STL-SWEI elastograms (**Fig. 12**), and provided better SWS estimates deeper within the tissue. **Table III** demonstrates that both methods produce similar mean SWS inside the tumor. CNRe of pSTL-SWEI was found to be marginally better than STL-SWEI. The second tumor was also scanned with a low push beam intensity (236.2 watt/cm² as opposed to 361.9 watt/cm²) and a low push pulse duration (50 μ s as opposed to 200 μ s). From visual inspection (**Fig 13**), it would appear that reducing the intensity increased the variance of STL-SWEI elastogram noticeably. Similarly, reducing the pulse duration had a more profound effect on STL-SWEI elastograms than their pSTL-SWEI counterpart. Pulse durations of 50 μ s and 200 μ s produced visually similar pSTL-SWEI elastograms (**Fig. 13**), but noise corrupted the STL-SWEI elastograms. CNRe of pSTL-SWEI was marginally affected by low push duration and was reduced by 24.5% due to low push intensity (**Table III**). CNRe of STL-SWEI was 23.8% and 25.7% lower than pSTL-SWEI for low intensity and low push duration cases, respectively.

Impact of shear attenuation

Impact of shear attenuation on pSTL-SWEI image quality was evaluated on a viscous phantom. **Fig. 14(a)-(b)** shows temporal profiles of shear waves acquired from elastic and viscoelastic phantoms using the STL-SWEI experiment. Shear attenuation and pulse broadening, synonymous with dispersion, are apparent in data collected from the viscoelastic phantom (**Fig 14(a)**). However, neither significant shear wave attenuation nor pulse broadening was apparent in the data acquired from the purely elastic phantom (**Fig. 14(b)**). In this phantom, the shear waves attenuated as they traveled further from the push beam. Consequently, STL-SWEI datasets synthesized with an increasing distance between push and track (T) produced SWS maps with increasing variance. **Fig. 14(c-h)** demonstrates that the averaged SWS image has better quality than each individual image synthesized with different tracking distance and standard STL-SWEI. **Fig. 14(i)** shows that the quality of these individual synthesized elastograms degrades with increasing distance between push and track. The degradation happens more rapidly in viscous media than purely elastic media. For example, SNRe degraded from 31 to 12 when T was changed from 6 mm to 10 mm. Image quality degraded slightly when averaging was performed with more than 6 tracking lines (see **Fig 14(j)**). However, the performance with greater than 6 averaging was still better than no averaging (tracking distance = 6 mm) and standard STL-SWEI with tracking distance of 6 mm. Reducing the tracking beam should minimize this problem, but further studies are needed to confirm this expectation.

VI. Discussion

In this study, we investigated the feasibility of using plane wave imaging to improve the performance of STL-SWEI. We can summarize the primary findings of this study as follows.

- 1) The mean SWS estimated with all three shear wave imaging approaches were comparable (**Fig. 4a**), but pSTL-SWEI and MTL-SWEI produced elastograms with the highest and lowest SNRe, respectively (**Fig. 4b**).

- 2) At any given spacing, the spatial resolution of pSTL-SWEI elastograms was slightly inferior to that produced with STL-SWEI (**Fig. 8**).
- 3) pSTL-SWEI can be performed at a lower push beam spacing to recover the resolution while maintaining a CNR_e comparable to STL-SWEI (**Fig. 8e**).
- 4) The spatial peak intensity and duration of the push beams influenced the elastographic image quality (**Fig. 11**).
- 5) The quality of pSTL was higher than STL-SWEI at all depths and the most improvement was found at the shallow region (**Fig. 11**).
- 6) The unfocused nature of tracking in pSTL-SWEI provides depth gain, and utilization of parallel beamforming (hybrid processing) provided a gain in FOV (**Fig. 11 & Fig. 3**).
- 7) pSTL-SWEI can produce elastograms of equivalent accuracy and SNR_e as STL-SWEI with a less intense push beam (**Fig. 11**).
- 8) pSTL-SWEI produced better elastograms (higher CNR_e) of murine pancreatic tumors with lower spatial peak intensity and longer pulse duration (**Figs. 12, 13, Table III**). The overall low CNR_e of pancreatic tumors was caused by inherent tumor heterogeneity [31].

Shear wave elastography possesses distinct advantages over quasi-static elastography [32] in its ability to provide quantitative measures of tissue elasticity without the need for solving the inverse problem [33], and it is less operator-dependent [34], [35]. However, the multiple track location (MTL) approach employed in clinically available systems suffers from speckle bias. Specifically, the random nature of ultrasound speckle patterns bias shear wave speed estimates ([36], [2], [3], [37]) due to uncertainties in defining the measurement location precisely. **Figure 4** demonstrates that speckle bias does not affect the mean SWS over a 2D region (i.e., the mean SWS measured with MTL-SWEI and STL-SWEI were similar), but it does increase the variance of SWS estimates, which degrades SNR_e . STL-SWEI is less susceptible to speckle bias because it uses the same set of scatterers to estimate the arrival times generated with the push beams ([10]). Consequently, the bias incurred when estimating arrival times cancels when the difference in arrival times (i.e., shear wave speed) is computed.

Although STL-SWEI reduces speckle bias, attenuation in shear wave and low sonographic signal-to-noise ratio degrade image quality (i.e., increase measurement variability). **Figure 11** demonstrates that increasing the intensity of the push beams or their duration reduces measurement variability and increases SNR_e . Maximum intensity and pulse width are, however, limited by the tissue and transducer face heating. An alternate approach to reducing measurement noise is to perform multi-acquisition averaging. Although this has been an effective noise reduction approach for quasi-static elastography [38], it reduces frame rate. Another alternate approach is to perform STL-SWEI with plane wave rather than conventional imaging. Doing this allows us to synthesize multiple STL-SWEI measurements from a single plane wave dataset without increasing push intensity or reducing frame rate.

Figure 7 demonstrates that averaging SWS over multiple realizations of synthetic STL-SWE measurements reduces the variance, which increases SNR_e .

This result suggests that averaging more lines would improve performance further. One approach to doing this would be to increase line density during beamforming. However, there is a limit on how we can improve performance by increasing line density. We predict that beyond a certain point, performance will plateau. Specifically, decreasing the line density beyond the ultrasound resolution cell will cause SWS estimates to become more correlated, which will negate the effects of averaging.

Montaldo *et al.* [14] demonstrated that coherent compounding of a large set of steered plane waves achieves uniform virtual focus comparable to a focused transmission. We expect compounding with more beam steering angles will further enhance the performance of pSTL-SWEI, but excessive compounding will degrade frame rate and could render the technique impractical. Using the Nyquist sampling theorem, we estimate that SWEI should be performed with a minimum frame rate of 2 kHz. We based this estimate on published results that demonstrate that shear waves contain significant spectral components up to 1 kHz [22]. Averaging that improves pSTL-SWEI cannot be applied to MTL-SWEI to the same effect. Averaging will reduce motion estimation jitter noise, but it cannot reduce “first order” effects such as speckle bias.

High shear attenuation in real tissue may reduce the number of viable tracking lines used in the averaging process of pSTL-SWEI. Using an attenuating phantom, we demonstrated that a high number of averaging may reduce performance gain. In our experiment, the reduction was only moderate and still produced better images than standard STL-SWEI. This is partly due to the correlation coefficient-based weighting process that suppresses the contribution of SWS maps synthesized with high tracking distance. Also, in tissues with significantly high attenuation, the performance reduction will be higher when averaging size is high. However, we still expect pSTL-SWEI to outperform standard STL-SWEI. **Fig 14h** demonstrated that by averaging as few as 5 tracking lines, SNR_e can be improved by 3.2 times compared to standard STL-SWEI. In our case, this required the use of tracking lines spaced over a distance of 1.2 mm, which is a significantly smaller propagation distance than normal practice (often on the order of cm) for *in vivo* imaging [11]. Although our weighted averaging process is capable of suppressing the contribution of low quality tracking lines, it may be useful to implement a track line selection algorithm that selectively removes lines with low shear SNR from the averaging set. Such an algorithm can be implemented by applying thresholding on the correlation coefficient or the energy of the tracked shear wave.

The distance between push and tracking lines influences pSTL-SWEI performance. It should be large enough to capture the entire shear waveform [39], but small enough to minimize shear wave attenuation. A fixed track line spacing (transducer pitch) was used in all work reported in this study; therefore, we expect performance will vary when SWEI is conducted with other parameters. We plan to assess the impact of track line spacing on SWS estimation in a future study. Specifically, we will determine the smallest track line spacing that SWEI can be utilized without negating the effects of averaging. Determining the optimum tracking parameters (track beam spacing, track beam locations, and track beam count) will be

challenging because they are likely to vary with changes in physiological conditions. We plan to conduct an empirical study to assess how physiological and tissue properties impact the optimum SWEI parameters.

A limitation of this study is that we investigated only one approach to STL-SWEI. Researchers have proposed two different approaches to STL-SWEI. One approach [10] produces two-dimensional elastograms by translating a pair of push beams laterally while maintaining a fixed distance between the tracking location and the push pair (as illustrated in **Fig. 1b**). The another approach is to create elastograms by only translating the push beams with a fixed tracking beam [15]. Shear waves attenuates less in the first approach because the distance between push and track beams is smaller. Additionally, since this first approach does not acquire the full wave field (shear wave propagation in space and time), directional filters that reduce reflection artifacts are difficult to implement. An important question that was not addressed in this paper is whether the performance gained with the second approach is similar to that obtained with the first. This a question that will be explored in future studies. For STL-SWEI and pSTL-SWEI, we have assumed that the SWS estimates spatially register at the mid-point of a push pair [10]. However, researchers [15] have demonstrated that the hourglass shape of the push beam can create a depth-dependent lateral shift in STL-SWEI images. For pSTL-SWEI, the square of the correlation coefficient was used as the averaging weight. However, we do not know if this was the optimum weight. In a preliminary study on the 3.4 KPa homogeneous phantom, we also tested the normalized energy of the shear wave and normalized correlation coefficient times normalized shear wave energy as weights for averaging. None of these resulted in a significant difference in the SNR_e for the homogeneous phantoms. However, in highly attenuating materials, some types of weight might be better at suppressing the contribution of low quality estimates than others. Thus, the use of weights needs to be carefully studied. Additionally, we did not assess how well we could visualize lesions in the flanking regions of pSTL-SWEI elastograms. We expect inclusions in the flanking areas will be less discernible than inclusions with equivalent contrast located in the central portion of the image because speckle noise is lower in the central region. We plan to conduct additional studies to corroborate this expectation. Our future goal is to utilize the noise robustness of pSTL-SWEI in clinical settings. Since pSTL-SWEI works reliably even with weak shear waves, it might assist in imaging patients with higher body mass index. Clinical and pre-clinical imaging of pancreatic tumors are also under consideration.

VII. Conclusion

In this paper, we present a novel approach to STL-SWEI known as pSTL-SWEI that use compounded plane wave imaging to track shear wave propagation. During pSTL-SWEI, we synthesize different STL-SWEI datasets and SWS maps by varying the distance between the push beam pair and the tracking location. Aggregating the SWS maps allows us to produce a single SWS map with higher SNR_e and CNR_e than those produced with standard STL-SWEI. pSTL-SWEI can probe deeper depths because it is more resistant to shear wave attenuation than STL-SWEI. Additionally, we expect the superior CNR_e of pSTL-SWEI will enhance the clinical utility of SWEI because this could allow clinicians to perform SWEI with smaller push beams spacing and less intense push beams.

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Fig. 1:

Schematic diagram illustrating the general principles of (a) MTL-SWEI and (b) STL-SWEI. Grey and black arrows represent the push and tracking beams, respectively. The black dots represent lateral locations where SWS is estimated. In plane wave imaging based MTL-SWEI, propagation of shear wave is tracked over the entire aperture. Since SWS cannot be reconstructed within push beam regions, multiple push beams (two in this case) are used to cover the blind region. In STL-SWEI, shear waves from a pair of push beams at ΔP distance apart are tracked at a common location L_1 . The ratio of ΔP to arrival time difference provides local SWS estimates. The push pair along with the common tracking line at ΔT distance apart is translated laterally to form a 2D image.

Fig. 2:

The basic principles of pSTL-SWEI. Multiple push beams (red arrows) are transmitted within the tissue, and plane wave (PW) images (black arrows) are acquired after each push beam. The time-of-flight of shear waves generated by two push beams, P apart, is estimated with a common tracking line located at T to the left for each pair of push beams. Different STL-SWEI datasets are synthesized from the acquired data by varying T (T_1 , T_2 , T_3 etc.). Each dataset is processed separately as illustrated in Figure 1(b) to generate a

unique SWS map. A final SWS map is computed from the weighted average of SWS maps generated from the synthetic STL-SWEI dataset.

Fig. 3:

Sonogram and elastograms acquired from homogeneous (a-h) and heterogeneous (i-p) phantoms. Elastograms were produced with (b,f,j,n) STL-SWEI, (c,g,k,o) pSTL-SWEI, and (d,h,l,p) MTL-SWEI. The first and second rows correspond to 3.4 KPa and 50.4 KPa homogeneous phantoms, respectively. The heterogeneous elastograms were obtained from phantoms containing stiff (third row) and soft (fourth row) inclusions. The pSTL-SWEI has a central region (with reduced speckle-bias) indicated by solid vertical lines. The width of STL-SWEI elastograms is smaller than pSTL-SWEI and MTL-SWEI elastograms because the axes of the push beams with a finite aperture size cannot be placed beyond a certain central region. Inclusion phantoms were scanned with a higher number of push beams in pSTL-SWEI to widen the central region. However, the total size of the image remains the same due to the hybrid nature of pSTL-SWEI. The dotted rectangular areas and circular regions were used for calculating mean SWS, SNR_e and CNR_e in **Figs 4 and 5**. The dotted circle represents 50% of the diameter of the inclusions as determined from the sonograms.

Fig. 4: Bar plots of the mean (a) SWS, and (b) SNR_e computed from four homogeneous phantoms (CIRS model 039) with STL-SWEI, MTL-SWEI, pSTL-SWEI. Mean values of VTQ measurements are provided for reference. The error bars represents 1 standard deviation computed over three independent acquisitions.

Fig. 5:

(a) Mean SWS and (b) SNR_e computed from the central and flanking regions of pSTL-SWEI elastograms obtained for four homogeneous phantoms. In the central region, SWS was estimated using STL-SWEI approach, whereas in the flanking regions we estimate SWS with the MTL-SWEI approach. The ROI used to compute these parameters are illustrated in **Fig. 3**. Standard deviation (indicated as an errorbar) was computed over three separate acquisitions.

Fig. 6:

(a) Mean SWS and (b) CNR_e computed from the inclusion phantom elastograms. The inclusion background ROIs used to compute these parameters are illustrated in **Fig. 3**. Standard deviation (indicated as an errorbar) was computed over three separate acquisitions.

Fig. 7: Plots of (a) SWS, and (b) SNR_e computed from four homogeneous pSTL-SWEI elastograms as a function of averaging. These performance metrics were calculated from the central region of pSTL-SWEI elastograms. Standard deviation (indicated as an errorbar) was computed over three separate acquisitions.

Fig. 8:

Results of a two-layer phantom experiment. Lateral SWS profiles for STL-SWEI and pSTL-SWEI with 16 averaging lines using push beam spacing of (a) 1.8 mm and (b) 4.2 mm.

Resolution measured from five lateral profiles within 2 mm of focus were averaged. Dotted and solid vertical lines indicate the lateral resolution (R_{80-20}) of STL-SWEI and pSTL-SWEI, respectively. (c) Relationship between push beam spacing and lateral resolution for STL-SWEI, pSTL-SWEI with no averaging and pSTL-SWEI with 16 averaging lines. (d) Relationship between push beam spacing and CNR_e . Increasing the distance between the pair of push beams increases the transition region width, but also improves the CNR_e . (e) Lateral resolution and CNR_e computed at different push beam spacings (1.8, 2.4, 3, 3.6 and 4.2 mm) are plotted against each other. In general, pSTL-SWEI has better contrast but wider resolution compared to standard STL-SWEI. (f) Effect of averaging over tracking lines on the pSTL-SWEI at different beam spacings.

Fig. 9:

Depth-dependent ROI in the pSTL-SWEI elastogram of the 50.4 KPa phantom used for quantitative analysis. The push transmit focus was set at 30 mm. Five different regions (Region A (10–15 mm), B (15–20 mm), C (20–25 mm), D (25–30 mm) and E (30–35 mm)) within the central region were used. The mean SWS were estimated for all regions. Showing (a) SWS estimated with STL-SWEI and pSTL-SWEI with increasing averaging as a function of $I_{sppa,0.3}$ with a fixed pulse duration of 200 μs; and (b) SWS as a function of push beam pulse duration for a fixed $I_{sppa,0.3}$ of 361.9 Wcm⁻². Profiles for the other four regions show a similar trend.

Fig. 10:

(a) Relations between push beam transmit voltages and spatial peak pulse average intensity and mechanical index measured from hydrophone experiments. (b) Acoustic waveform of the push beam (42.7 volts) recorded by hydrophone at the focal depth. Pulse average intensity was computed from the waveforms. (c) Intensity profile along the lateral direction through the point of maximum intensity. The peak of the profile corresponds to the I_{SPPA} . The intensity profile derated by an attenuation coefficient of $0.3 \text{ dB cm}^{-1} \text{ MHz}^{-1}$ is also shown.

Fig. 11:

Illustrating the effect of push beam intensity and push beam duration on the mean SNR_e of pSTL-SWEI and STL-SWEI elastograms corresponding to the 50.4 kiloPa homogeneous phantom. SNR_e was computed from the depths shown in **Fig 8**. Figures (a)-(e) show SNR_e computed from STL-SWEI and pSTL-SWEI elastograms with different numbers of averaging as a function of $I_{SPPA,0.3}$ at regions A-E, respectively. Figures (f)-(j) show SNR_e for STL-SWEI and pSTL-SWEI as a function of pulse width at regions A-E, respectively. Note that the y-axis scale differs between plots.

Fig. 12:

Elastograms of ex-vivo tumors. (a) Sonogram, (b) STL-SWEI elastograms, and (c) pSTL-SWEI elastogram obtained from tumor 1. (d) Sonogram, (e) STL-SWEI elastogram, and (f) pSTL-SWEI elastogram obtain from tumor 2. ROIs indicate tumor and background for quantitative analysis. Tumor ROI was segmented from the sonogram. The remaining background depth limited between 10 and 40 mm were used in the calculation of CNR_e . Elastograms correspond to a push $I_{sppa,0.3}$ of 361.9 W cm^{-2} and a push duration of $200 \mu\text{s}$. It would appear that the shear modulus of tumor 1 and the surrounding were similar, this was an unintentional consequence of not knowing the stiffness of the tumor prior to embedding it in gelatin.

Fig. 13:

Elastograms of tumor 2 scanned with either a low push beam or a low push duration (a) and (b) STL-SWEI and pSTL-SWEI elastograms corresponding to a push $I_{\text{sspa},0.3} = 236.2 \text{ W cm}^{-2}$ and push duration = $200 \mu\text{s}$. (c) and (d) show STL-SWEI and pSTL-SWEI elastograms corresponding to a push $I_{\text{sspa},0.3} = 361.9 \text{ W cm}^{-2}$ and push duration = $50 \mu\text{s}$.

Fig. 14:

(a) shows representative shear wave temporal profiles for two push beams of STL-SWEI at the focus on the viscous phantom. (b) Similar wave profiles from 50.4 KPa elastic phantom are shown for comparison. (c)-(f) correspond to SWS maps from individual STL-SWEI datasets synthesized from a pSTL-SWEI acquisition with tracking distances ΔT of 11.7, 9, 6 and 3 mm. (g) pSTL-SWEI elastogram obtained from averaging 16 SWS maps corresponding to tracking distances between 6 to 11.7 mm. (h) standard STL-SWEI (focused tracking) elastogram corresponding to a tracking distance of 6 mm. STL-SWEI datasets were synthesized from pSTL-SWEI by varying the tracking distances and (i) SNR_e was computed as a function of tracking distance from each individual SWS map. (j) SNR_e computed as a function of number of averaging lines. For all SNR_e computation, a fixed ROI was used over the entire pSTL-SWEI central region depth limited between 25 to 30 mm (focal region).

Table I:

Summary of push and tracking beam parameters used during shear wave imaging.

Parameters	Push beam	Tracking beam
Frequency (MHz)	5	5
Pulse duration (μ s)	200	0.2
F-number	2.0	2.0
Transmit voltage (V)	52.9	52.9
Focus (mm)	30	30

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Table II:

Stiffness of phantoms used in experimental studies.

Phantoms	Nominal Stiffness (KPa)	Nominal SWS (m/s)	VTQ SWS(m/s)	pSTL SWS (m/s)
Model 039 (1)	3.4	1.06	0.83 ± 0.01	0.88 ± 0.01
Model 039 (2)	13.9	2.11	1.67 ± 0.02	1.64 ± 0.01
Model 039 (3)	27.1	3.01	2.63 ± 0.06	2.60 ± 0.01
Model 039 (4)	50.4	4.10	3.37 ± 0.14	3.91 ± 0.02
Model 049A (Background)	25	2.89	2.41 ± 0.04	2.29 ± 0.01
Model 049A (stiff inclusion)	80	5.16	4.11 ± 0.16	4.13 ± 0.1
Model 049A (soft inclusion)	14	2.16	1.64 ± 0.02	1.68 ± 0.01

Table III:

Summary of the quantitative results obtained from *ex vivo* murine tumors scanned with different push intensity and push pulse width.

Timor	$I_{\text{sppa},0.3}$ (W/cm ²)	Pulse width (μ s)	pSTL SWS (m/s)	STL SWS (m/s)	pSTL CNR _e	STL CNR _e
1	361.9	200	2.84 ± 0.21	2.75 ± 0.23	2.33	1.64
2	361.9	200	5.00 ± 0.48	5.17 ± 0.62	2.00	1.91
2	236.2	200	5.11 ± 0.62	5.07 ± 0.65	1.51	1.15
2	361.9	50	4.98 ± 0.53	4.79 ± 0.67	1.91	1.42