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IMPROVEMENTS IN ELASTOGRAPHIC CONTRAST-TO-NOISE RATIO USING SPATIAL-ANGULAR COMPOUNDING

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Abstract

Spatial-angular compounding is a new technique developed for improving the signal-to-noise ratio (SNR) in elastography. Under this method, elastograms of a region-of-interest (ROI) are obtained from a spatially weighted average of local strain estimated along different insonification angles. In this article, we investigate the improvements in the strain contrast and contrast-to-noise ratio (CNR) of the spatially compounded elastograms. Spatial angular compounding is also applied and evaluated in conjunction with global temporal stretching. Quantitative experimental results obtained using a single-inclusion tissue-mimicking phantom demonstrate that the strain contrast reduces slightly but the CNR improves by around 8 to 13 dB. We also present experimental spatial angular compounding results obtained from an *in vitro* thermal lesion in canine liver tissue embedded in a gelatin phantom that demonstrate the improved visual characteristics (due to the improved CNR) of the compound elastogram. The experimental results provide guidelines for the practical range of maximum insonification angles and estimates of the optimum angular increment. (E-mail: tvarghese@wisc.edu)

Keywords

Angular compounding; Compounding; Elastography; Elastogram; Elasticity; Elasticity imaging; Axial strain; Lateral strain; Strain imaging

INTRODUCTION

Elastography or elasticity imaging has developed into an imaging modality for the diagnosis of abnormalities in the breast, prostate, thyroid, liver and kidneys, as well as for intravascular atherosclerotic plaque characterization applications (Parker et al. 1990; Ophir et al. 1991; Cespedes 1993; O'Donnell et al. 1994; Chen et al. 1995; Bamber and Bush 1996; de Korte et al. 2000; Pesavento et al. 2000; Insana et al. 2001; Nightingale et al. 2001; Varghese et al. 2001). Elastography also enhances the capability for imaging and monitoring thermal ablative therapies for cancer treatment in organs such as the liver, kidney and prostate that are not clearly visualized under conventional ultrasound (US) imaging (Stafford et al. 1998; Righetti et al. 1999; Varghese et al. 2002, 2003; Souchon et al. 2003). Different types of perturbation or mechanical stimuli, such as quasistatic compression (Ophir et al. 1991; O'Donnell et al. 1994; Hall et al. 2003), dynamic vibration (Parker et al. 1990; Dutt et al. 2000) or acoustic radiation force (Nightingale et al. 2001; Fatemi et al. 2002; Lizzi et al. 2003), have been used to calculate local tissue displacements or strains, revealing a profile of local stiffness changes in biologic tissue. These stiffness variations provide new information on the pathologic state

of tissue, enabling its application for diagnosis of disease. One of these methods, termed elastography, introduced by Ophir et al. (1991), computes the displacements using windowed cross-correlation analysis under a quasistatic compression. This technique is utilized in our angular compounding approach.

Many algorithms have been developed to minimize correlation artifacts and to improve the elastographic signal-to-noise ratio (SNR_e), such as temporal stretching (Cespedes and Ophir 1993; Alam and Ophir 1997), adaptive temporal stretching (Alam et al. 1998), multicompression averaging (Varghese et al. 1996) and wavelet denoising (Techavipoo and Varghese 2004). Temporal stretching refers to the stretching of the time domain postcompression signal to match the precompression state or signal. Generally, temporal stretching assumes that the average applied compression or strain is known *a priori*, and uses this average compression value or “stretch factor” to stretch the postcompression signal before estimation of the displacements and axial strains. This algorithm is effective; however, it requires *a priori* knowledge of the applied compression as previously described and works most efficiently when imaging homogeneous and uniformly elastic media. Adaptive stretching, on the other hand, iteratively varies the stretch factors that are applied to the postcompression signals to maximize the correlation coefficients between the pre- and stretched postcompression windows. These stretch factors were then used as the strain estimates. Multicompression averaging reduces noise by averaging elastograms generated using successive small compressions and wavelet denoising smoothes the displacement estimates in the wavelet domain without losing edge information. Spatial angular compounding can be used in conjunction with any of the methods described above. In this article, we will also present results obtained with global temporal stretching.

Spatial-angular compounding for elastography was recently introduced by Techavipoo et al. (2004b). Local strain estimates around the same region-of-interest (ROI) acquired from different insonification angles are averaged to obtain a compound elastogram. This technique utilizes the same concept to reduce speckle noise as in conventional US imaging with angular compounding of the B-mode signals referred to as sono-CT (Burckhardt 1978; Trahey et al. 1986; O'Donnell and Silverstein 1988; Wagner et al. 1988; Ping 1997; Entekin et al. 2001; Tanter et al. 2002). However, because strain is a tensor, strain estimates obtained from different angular views have to be appropriately weighted before compounding (Techavipoo et al. 2004b). The results presented by Techavipoo et al. (2004b) also illustrate the improvement in the elastographic SNR_e obtained with angular compounding. Another advantage with spatial angular compounding is that the spatial resolution in the compounded images is not significantly degraded by the compounding process because the compounded signals arise from the same spatial location, but at different insonification angles. The variation in the insonification angle provides sufficient decorrelation to enable the improvement in the SNR_e of the averaged strain estimate (Trahey et al. 1986; O'Donnell and Silverstein 1988; Wagner et al. 1988).

Axial resolution in elastography depends primarily on the window length and overlap used to obtain the strain estimates (Ophir et al. 1999; Alam et al. 2000; Varghese et al. 2001). Spatial-angular compounding provides the ability to improve spatial resolution by using smaller window lengths and increased overlaps between data segments by trading off the increased SNR_e obtained with angular compounding for improved spatial resolution. Smaller window lengths are more susceptible to mismatches in the signal segments, providing noisier strain estimates, and an increased overlap introduces distortions in strain estimates, due to correlation artifacts (Ophir et al. 1999; Varghese et al. 2001).

In this article, we investigate the strain contrast and elastographic contrast-to-noise ratio (CNR_e) (Varghese and Ophir 1998) variations using spatial-angular compounding in

elastographic imaging. Ultrasound RF echo signals at different angular insonification directions are acquired by translating a phased-array US transducer in the lateral direction. We also examine the use of global temporal stretching in conjunction with spatial-angular compounding. Variations in the strain contrast and CNR_e with angular compounding were not previously evaluated. Although the SNR_e describes the appearance of a uniformly elastic region on an elastogram, lesion or tumor detectability is clearly described using the CNR_e (Varghese and Ophir 1998) and the strain contrast parameters. We also present experimental results obtained using a single inclusion phantom and phantoms containing an *in vitro* thermal lesion created in liver tissue.

MATERIALS AND METHODS

Elastographic imaging performance using angular spatial compounding was assessed using a single-inclusion elastographic tissue-mimicking phantom of size $10 \times 10 \times 10$ cm, manufactured in our laboratory (Madsen et al. 2003). The single-inclusion phantom contains a 1-cm diameter cylindrical inclusion with a modulus contrast of 3.8 (expected strain contrast of 2.4 using the contrast transfer efficiency expression; Kallel et al. 1996) to the background. The Young's modulus of the background material was 20 kPa, and the inclusion possessed a Young's modulus of 76 kPa, measured using the ELF 3200 mechanical testing system (EnduraTEC, Minnetonka, MN, USA) in our laboratory.

Imaging performance was also assessed using a thermal lesion created in normal canine liver tissue and encased in a gelatin phantom. This phantom was used to evaluate the potential of applying elastographic techniques for evaluating stiffer regions (thermal lesions) in biologic tissue (liver). The thermal lesion was created using RF ablation. A RITA model 1500 electrosurgical device (RITA Medical Systems Inc., Mountain View, CA, USA) was used for the RF ablation procedure. The RF electrode consists of a 15-gauge shaft through which multiple sharp prongs, each 0.021 inches (0.53 mm) in diameter (25 gauge) can be deployed. Fully extended, the prongs are in an "umbrella" configuration at 45° intervals. The last 1 cm of the electrode tip and each prong constitutes the electrically active surface. The electrode is inserted into a liver lobe and the prongs are deployed, keeping them within the liver parenchyma. The thermal lesion was created before encasing it within the gelatin phantom. RF ablation of liver tissue was conducted for 10 min, raising the target temperature to 100°C at a 150-W power level.

Both phantoms were scanned using an Aloka SSD 2000 real-time scanner (Aloka Inc., Tokyo, Japan) using a 3.5-MHz phased-array transducer with an approximate 60% bandwidth. To simulate the effects of compound data acquisition of RF echo signals, the phased-array transducer was linearly translated in the lateral direction over the phantoms, using a precision linear stage, so that each location in the sample was scanned from multiple angles, as shown in Fig. 1. Pre- and postcompression data to obtain elastograms from the inclusion phantoms were obtained over eight independent realizations, to obtain statistical results. The same scanning plane at the axis of symmetry of the phantom was utilized for all the realizations; however, the precompression of the phantom was varied to obtain independent data realizations.

A compression plate with a rectangular slot to match the transducer face was mounted on a linear stage driven by a stepper motor. The compression plate was larger than the phantom surface and provided a uniform compression of the phantom. Echo signals were acquired, originating from the top of the phantom to a depth of 6 cm, before and after an axial compression of 1% of the phantom height. The compression increment of 1% was selected, based on strain filter results (Varghese and Ophir 1997) that provide a high SNR_e for the strain estimates around an applied compression value of 1%. The selection of the compression increment does

not significantly impact the strain-estimation performance, as long as it lies within the usable range of strain in the strain filter framework. The precompression for each data set was varied at 0.5% increments for a total pre-compression of 4% for the data set used in this article. Because a phased-array transducer was used to acquire the pre- and postcompression data at different insonification angles in this article, the data-acquisition procedure required the translation of the combined transducer-compression plate fixture in the lateral direction along the 2-D scanning plane (Techavipoo et al. 2004a, 2004b). The pre- and postcompression signal data sets were acquired at 1/6 mm lateral increments over a 40-mm distance, resulting in the acquisition of multiple (*i.e.*, 240) pre- and postcompression data sets in a sector form at different lateral positions, as shown in Fig. 1. The phantom was compressed after each translation, to obtain the pre- and postcompression data. A more elegant method would be the use of the beam steering feature with a linear-array transducer for data acquisition, enabling the use of a single compression of the phantom.

Each sector data set used in this article consisted of 120 A-lines arranged over a 90° sector. The RF signals were digitized using a 12-bit data-acquisition board (Gage Applied Technologies Inc., Montreal, QUE, Canada) at a sampling rate of 50 MHz. During off-line analysis, the sector data sets were rearranged so that the A-lines along the same angle from all the sector images were grouped together to form RF frames at the corresponding angle. This is illustrated in Fig. 1, where the distance between the A-lines in the regrouped data was 1/6 mm. Acquisition of the data in this manner provided RF frames over an angular range of $\pm 45^\circ$, with a minimum angular increment of 0.75° , which is limited by the phased-array transducer used.

Each of these regrouped RF pre- and postcompression echo signal frame pairs were analyzed separately to generate an elastogram at each specified angle, referred to as angular elastograms. Angular elastograms were computed using the following procedure. Tissue displacements along the angled A-lines pairs (the angular displacement data includes both axial and lateral motion of the scatterers) were estimated (without resampling the RF signals) using a cross-correlation algorithm using a window size of 3 mm and 75% overlap of consecutive windows. A five-point, 1-D median filter was then applied to eliminate the outliers in the angular displacement estimates. A three-point least-squares strain estimator (Kallel and Ophir 1997) was used to generate local angular strain estimates. Angular elastograms were then filtered, using a 5×5 median filter to suppress strain outliers. Note that these processing parameters were chosen based on parameters reported in the literature.

Following the estimation of the angular elastograms, spatial compounding for elastography was performed. Compound elastograms were obtained by a weighted averaging of the angular elastograms obtained at a specified angular increment ($\Delta\theta$). The value of the angular increment $\Delta\theta$ utilized in the compounding process and the largest or maximum insonification angle determine the number of angular elastograms (N) that were averaged. For example, in our data set, the smallest angular increment $\Delta\theta$ was 0.75° (smallest angle between the A-lines of the phased array). Therefore, for spatial angular compounding, we could use this angular increment value or a multiple of 0.75° . However, the number of angular elastograms (N) available for compounding changed with the choice of the angular increment value, with smaller values of the angular increment leading to larger N and larger values of the angular increment leading to smaller N . These two parameters (*i.e.*, the angular increment $\Delta\theta$ and the number of strain estimates N) had a significant impact on the enhancement of the SNR_e in the compounded elastogram. The compounded elastogram is, therefore, given by:

$$C(\Delta\theta, N) = \frac{1}{N} \sum_{i=-n}^n w_x(v, i\Delta\theta) A(i\Delta\theta), \quad (1)$$

where $C(\Delta\theta, N)$ represents a compound elastogram for specified values of $\Delta\theta$ and N , where N is equal to $2n + 1$, where n is the number of angular elastograms in either the positive or negative angular direction. $A(\theta)$ and $w_x(v, \theta)$ are, respectively, the angular elastogram and the weighting factor for axial strains at angle θ . The weighting factor is given by Techavipoo et al. (2004b) as follows:

$$w_x(v, \theta) = 1 / (\cos^2 \theta - v \sin^2 \theta), \quad (2)$$

where v is the Poisson's ratio, which is assumed to be 0.495 (Fung 1981) for incompressible soft tissue. The spatial averaging of the angular strain estimates in the compound elastogram can be described using the following steps. In the first step, each angular elastogram was spatially registered so that the pixels of each angular elastogram lie on the same rectangular grid. This registration was performed using bilinear interpolation (Berkhoff et al. 1994; Richard and Arthur 1994). The registered angular elastograms were then appropriately weighted as shown in eqn (1) and then spatially averaged. Spatial averaging was performed over the weighted strain estimates obtained from the angular strain at each pixel in the rectangular grid. Due to the different insonification angles, some of the angular elastograms may not have contributed to all of the pixels in the rectangular grid. Each pixel in the rectangular grid or compound elastogram, therefore, may have contained different numbers of strain estimates obtained from the angular elastograms.

Only cases where the compounding of both symmetrical positive and negative angles were considered, as shown in the summation index of eqn (1). Therefore, the total number of angular elastograms used in compounding was $2n + 1$, as described before. In our notation, a compounded elastogram is identified by its maximum angle, which is equal to $n\Delta\theta$, with $\Delta\theta$ (angular increment) values of 0.75° , 1.5° , 3.75° and 7.5° being applied. The minimum $\Delta\theta$ value was limited to 0.75° for our phased-array transducer as previously described.

For statistical analysis, we acquired eight sets of pre- and postcompression RF data along the same imaging plane of the inclusion phantom using different pre-compression, as previously described. Note that the different precompression amount had a small impact on the calculated strain contrast, due to the hardening of the phantom (Varghese et al. 2000). The values of the strain contrast C_0 and the CNR_e were analyzed for both non-compounded and compound elastograms. The strain contrast C_0 and the CNR_e are defined as follows (Varghese and Ophir 1998):

$$C_0 = s_2 / s_1 \quad (3)$$

and

$$CNR_e \text{ (dB)} = 20 \log_{10} \left(\frac{2(s_1 - s_2)^2}{(\sigma_1^2 + \sigma_2^2)} \right) \quad (4)$$

where s and σ denote the mean and SD, respectively, of the strains in a selected ROI. The subscripts 1 and 2 represent the regions inside and outside (background) the inclusion, respectively.

RESULTS

Elastograms obtained with and without angular compounding using the single inclusion phantom are shown in Fig. 2. Elastograms obtained without temporal stretching are shown in Fig. 2a to c and those with temporal stretching are shown in Fig. 2d to f. Comparison of the

elastogram obtained without compounding in Fig. 2a with the angular compounded elastograms in Fig. 2b and c using the maximum angles of 12° and 18.75° , respectively, demonstrates the significant reduction in noise artifacts. This reduction in noise artifacts is clearly visible in the regions both inside and outside the inclusion, as shown by the reduction in strain fluctuations and subsequent improvement in the visual quality of the inclusion after spatial angular compounding. For these compounded elastograms, the visual quality improved with an increase in the maximum angle used (using either smaller angular increments or larger number of angular elastograms). This improvement in the inclusion detectability is quantified using the CNR_e parameter plotted and described in the following figures. Note that the region demarcated by the two dashed lines represents locations that are common (*i.e.*, that contain all the averaged angular estimates in each pixel of the rectangular grid) to all angular elastograms compounded. This common region reduces with an increase in the maximum insonification angle. Moreover, angular compounding can be used in conjunction with temporal stretching, which further improves the elastogram quality, as shown in Figs. 2e and f. In addition, the lesion boundaries visualized on the compounded elastograms are smoother and better defined than those without compounding.

Quantitative variations in the strain contrast and the CNR_e were calculated using the rectangular ROI within and outside the inclusion, shown in Fig. 2b. The strain contrast plotted against the maximum angle is shown in Fig. 3a without, and Fig. 3b with global temporal stretching. The plots in Fig. 3 show the mean of the inclusion contrast with respect to the background, across eight sets of compounded elastograms. The error bars denote the SDs only at selected angles, to reduce the clutter in the graphs. With increasing amounts of precompression, the strains both inside the phantom inclusion and in the surrounding background decrease, with the inclusion strain decreasing at a slower rate, due to its increased stiffness, thereby introducing a small reduction in strain contrast at higher precompression levels. In addition, the strain contrast also slightly decreases with an increase in the maximum angle. The strain contrasts with and without stretching are almost identical, with slightly smaller error bars for the estimates obtained with stretching. Different values of the angular increment also do not significantly affect the strain contrast.

The CNR_e curves plotted against the maximum angle are shown in Fig. 4. These CNR_e values are the mean across the same eight sets of the compounded elastograms as in Fig. 3. The small impact of strain hardening due to the different precompression levels is also present in the mean CNR_e plots. In general, the CNR_e curves increase for small maximum angles, with a saturation or leveling-off and subsequent decrease after a maximum angle of around 15° . The CNR_e without stretching in Fig. 4a for an angular increment of 0.75° increases from 30 to 43 dB over the maximum angle from 0 to 12° , providing an improvement of 13 dB. On the other hand, the CNR_e with stretching in Fig. 4b increases from 40 to 48 dB over the maximum angle of 0 to 18.75° , providing an improvement of 8 dB. The different values of the angular increment appear to have an effect only on the plots of CNR_e without temporal stretching, where the CNR_e reduces with larger angular increments. The angular increment between 1.5 and 3.75° appears to contain the optimum angle, which would minimize the number of angular elastograms required for compounding while maintaining similar improvements in the CNR_e . The same CNR_e curves in Fig. 4 are alternatively plotted against the number of angular elastograms used in the compounding process, as shown in Fig. 5. These plots illustrate that an increase in the angular increment effectively reduces the number of angular elastograms used in the compounding process, consequently reducing computational time. As shown, using about 7 to 11 angular elastograms with the 3.75° angular increment improves the CNR_e from 30 to 42 dB without stretching (40 to 48 dB with stretching).

Elastograms obtained from the thermal lesion in canine liver encased in a gelatin phantom for 0.5% compression without temporal stretching are shown in Fig. 6. An elastogram without

compounding is shown in Fig. 6a, and compound elastograms are shown in Fig. 6b and c using the maximum angles of 12° and 18.75° , respectively. These compound elastograms utilize the 0.75° angular increment. Figure 6d to f is similar to those in Fig. 6a to c; however, global temporal stretching was applied before generating these elastograms. Even though the use of global temporal stretching suppresses additional noise artifacts and provides smoother elastograms, we also clearly observe the blurring of the edges of the thermal lesion. The lesion also appears to be smaller, with a thicker transition region between the lesion and surrounding normal tissue. These artifacts are probably introduced due to the use of the same stretch factor corresponding to the applied compression over all the angular RF data used to generate the angular elastograms. In addition, because the temporal stretching algorithm used was 1-D, artifacts due to lateral motion of the scatterers in the angular RF data also increase with the insonification angle. The stretch factor utilized also does not account for these changes in the tissue displacement. The angular strain e_θ depends on both the axial strain e_x and the lateral strain e_y , as follows (Techavipoo et al. 2004b):

$$e_\theta = e_x \cos^2 \theta + e_y \sin^2 \theta. \quad (5)$$

Assuming that $e_y = -\nu e_x$ and that the Poisson's ratio ν of compressible tissue is 0.495, now, if e_x is known, e_θ can be estimated and used as the stretch factor for stretching the postcompression signals at different insonification angles. Moreover, the signals corresponding to the stiffer thermal lesion that undergoes smaller amounts of deformation may be overstretched with global temporal stretching, resulting in blurring at the edge of the lesion; see Fig. 6d. After spatial averaging of the stretched angular elastograms, the blurring artifact is more pronounced in Fig. 6e and f. This effect is also more noticeable in Fig. 6d to f than those in Fig. 2. However, it is clearly evident, from the results presented with spatial angular compounding, that the noise artifacts are significantly suppressed, providing the subsequent improvement in the visual quality of the elastograms without temporal stretching.

DISCUSSION AND CONCLUSIONS

The utility of spatial angular compounding for elastographic imaging was illustrated by Techavipoo et al. (2004b) for reduction of noise artifacts and the subsequent improvement in the SNR_e . The SNR_e improved by a factor of 2.4 (around 3.8 dB) for maximum angles in the range of 18.75° to 30° in experiments with a uniformly elastic phantom and compared with an elastogram without compounding. In this article, we illustrate that spatial angular compounding provides a CNR_e improvement of around 8 to 13 dB, with a maximum angle ranging from 12° to 18.75° , using a single-inclusion phantom model, and compared with an elastogram obtained without compounding. The maximum angle that can be used for the CNR_e improvement reduces to around half of that previously reported for the SNR_e improvement (Techavipoo et al. 2004b). The strain contrast after spatial angular compounding also reduces by around 11%.

The cost incurred with spatial angular compounding is the increase in the computation time and the corresponding reduction in the frame rate of the US system. Although, in our experiment, the acquisition of the RF signals was preformed by laterally translating a phased-array transducer, for clinical data acquisition, a scanner that provides beam steering with a linear-array transducer would provide the best and most efficient approach. In addition, the use of an optimum angular increment could further reduce the computation time. In our experiment, the optimum angular increment was around 1.5° to 3.75° . Utilization of this optimum angular increment would reduce the number of angular elastograms that have to be utilized in compounding to obtain similar improvements in the SNR_e and CNR_e of the elastogram. In our experiments, the CNR_e plots vs. the maximum angle illustrate that the largest maximum angle that could be used for compounding is less than 12° without stretching and 18.75° with temporal stretching.

Spatial angular compounding can also be applied in conjunction with the global temporal stretching algorithm. The results in this article demonstrate a further improvement in CNR_e . However, using a constant value of the stretching factor can blur the edges of regions with rapid changes in the strain profile. This effect was demonstrated in Alam et al. (1998) and also in this article. The inclusion appears to be smaller than that observed without stretching. The use of adaptive stretching instead of uniform stretching may reduce this artifact.

The resolution of the compound elastogram was previously discussed by Techavipoo et al. (2004b). Because the spatial compounding algorithm does not improve the noise quality by sacrificing spatial resolution, the resolution of a compound elastogram is almost the same as that of the noncompounded elastogram. Interpolation for the pixel registration of the angular strain estimates before compounding, however, may slightly lower the resolution of the compound elastogram.

Angular elastograms obtained in our experiment at larger angles of insonification are noisier than those obtained around the 0° angle (used for conventional B-mode imaging), as previously shown in Techavipoo et al. (2004b) using the SNR_e plots. This is due in part to our experimental protocol, that applies quasistatic deformations along the axial direction or the direction of the US beam propagation at the 0° angle. For incompressible material under slip boundary conditions, the lateral motion of tissue scatterers would be approximately half the axial displacement or compression. This complex movement (it includes axial, lateral and elevational scatterer motion) of the scatterers out of the US beam further decorrelates the RF echo signals, especially those acquired at larger insonification angles. Therefore, as the angular insonification direction of the US beam moves farther away from the direction of compression, scatterers move out of the US beam, due to the 3-D scatterer motion, thereby decorrelating the RF signals and subsequently corrupting the angular strain estimates. The relative contributions of the axial, lateral and elevational motions of the scatterers will change with the angle of insonification, with the axial and lateral components playing more prominent roles due to the 2-D nature of the data acquisition for spatial angular compounding. We also expect the axial motion and decorrelation of the scatterers to be more pronounced for insonification angles of less than 45° , due to the geometry.

Spatial angular compounding of other parametric images of ultrasonic tissue characterization parameters such as scatterer size (Gerig et al. 2004a, 2004b) and attenuation coefficient estimates (Tu et al. 2003) have also been shown to significantly reduce estimation noise. In addition, for elastographic applications, angular insonifications have also been utilized to image tissue displacement information in 2-D, thereby enabling estimation of both the normal and shear strain components, without assuming tissue incompressibility (Techavipoo et al. 2004a).

In summary, spatial angular compounding provides significant improvement in both the SNR_e and the CNR_e , which is an important metric for the evaluation of lesion detectability in elastograms. The penalty incurred with spatial angular compounding is the subsequent increase in the computational time and reduction in the frame rate of the US scanner. However, the use of optimal angular increments and subsequent independent angular elastograms would enable the use of fewer images to obtain similar improvements in the SNR_e and CNR_e . Spatial angular compounding also preserves spatial resolution, because we image the same region with independent data sets obtained by varying the insonification angle. Angular compounding can also be used in conjunction with other noise-reduction algorithms, such as temporal stretching shown in this article, multicompression averaging, wavelet denoising, etc. Clinical applications of angular compounding would, however, require the utilization of beam steering on linear-array transducers to obtain angular data sets, which is more efficient than translating the phased-array transducer as is described in this article.

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Fig. 1.
(top) RF data sets in sector format acquired at each lateral position. These sector scans are then rearranged (bottom) into angular RF data sets by grouping A-lines acquired for the same beam angle. Only three angular groups are displayed.

Fig. 2.

Elastograms of the inclusion phantom. The first row illustrates elastograms without global temporal stretching (Techavipoo et al. 2004b) and the second row shows those with global temporal stretching. The first column represents elastograms obtained without compounding, and the second and third columns show compounded elastograms using 0.75° angular increments with a maximum angles of 12° and 18.75° , respectively.

Fig. 3.

Plots of the mean strain contrast with SDs (error bars) over eight independent realizations of the compounded elastograms (a) Without and (b) With global temporal stretching. The strain contrast is plotted vs. the maximum angle over which the compounding was performed. Results obtained using 0.75°, 1.50°, 3.75° and 7.50 ° angular increments are shown. Note that the 0° angle refers to the elastogram without compounding. Error bars in the figure are shown at selected data points, to reduce clutter.

Fig. 4. Plots of the mean CNR_e with SDs (error bars) over eight independent realizations of the compound elastograms (a) Without and (b) With global temporal stretching vs. the maximum angle over which the compounding was performed. Note that the 0° angle refers to the elastogram without compounding. Error bars in the figure are shown at selected data points, to reduce clutter.

Fig. 5.

Plots of the mean CNR_e with SDs (error bars) over eight independent realizations of the compound elastograms (a) Without and (b) With global temporal stretching vs. the number of angular elastograms used for compounding. Error bars in the figure are shown at selected data points, to reduce clutter.

Fig. 6.

Elastogram obtained (a) Without compounding, compared with angular compounded elastograms using (b) 12° and (c) 18.75° maximum angles, respectively, and a 0.75° angular increment for a 0.5% applied compression. (d) to (f) The elastograms are generated using the same parameters as those shown in (a) to (c), except that global temporal stretching was also used. The region demarcated by the two dashed lines represents locations that include angular strain estimates along all the insonification angles used.