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NORMAL AND SHEAR STRAIN ESTIMATION USING BEAM STEERING ON LINEAR-ARRAY TRANSDUCERS

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Abstract

In current ultrasound elastography, only the axial component of the displacement vector is estimated and used to produce strain images. A method was recently proposed by our group to estimate both the axial and lateral components of a displacement vector following a uniaxial compression. Previous work evaluated the technique using both simulations and a mechanically translated phased array transducer. In this paper, we present initial results using beam steering on a linear array transducer attached to a commercial scanner to acquire echo signals for estimating 2-D displacement vectors. Single-inclusion and anthropomorphic breast phantoms with different boundary properties between the inclusion and background material are imaged by acquiring echo data along beam lines ranging from -15° to 15° relative to the compression direction. 1-D cross-correlation is used to calculate “angular displacements” in each acquisition direction, yielding axial and lateral components of the displacement vector. Strain tensor components are estimated from these displacements. Features on shear strain images generated for the inclusion phantom agree with those predicted using FEA analysis. Experimental results demonstrate the utility of this technique on clinical scanners. Shear strain tensors obtained using this method may provide useful information for the differentiation of benign from malignant tumors. For the linear array transducer used in this study, the optimum angular increment is around 3° . However, more work is required for the selection of an appropriate value for the maximum beam angle for optimal performance of this technique.

Keywords

Strain; Shear strain; Strain tensors; Elastography; Elastogram; Elasticity; Elasticity imaging; Stiffness; Ultrasound

INTRODUCTION

In recent years, ultrasound-based strain imaging has received considerable interest for use in diagnosing disease (Krouskop et al. 1987; Bertrand et al. 1989; Parker et al. 1990; Ophir et al. 1991, 1999; Cespedes et al. 1993; Odonnell et al. 1994; Muthupillai et al. 1995; Plewes et al. 1995, 2000; Insana et al. 2000; Doyley et al. 2001; Varghese et al. 2001; Emelianov et al. 2002; Nightingale et al. 2002). Elastography is an imaging modality that is based on mapping local internal strains that tissues experience after a quasistatic or dynamic compression. In this technique, strains are typically estimated along the axial direction corresponding to the beam propagation axis by taking the gradient of the tissue displacement following a uniaxial compression (generally in the direction of beam propagation) (Ophir et al. 1991, 1997; Garra

et al. 1997; Varghese et al. 2001). Local tissue displacements along the beam direction are measured using time delay estimation techniques (Quazi 1981).

Until now, lateral (perpendicular to the beam propagation axis and in the scan plane) and elevational (perpendicular to the beam propagation axis and to the scan plane) displacements are usually not measured in elasticity imaging. However, all the components of the strain tensor and displacement vector are required to characterize the deformation following compression because tissue motion inevitably occurs in three dimensions (Kallel and Ophir 1997; Bilgen 2000). Because the components of the strain tensor are coupled, accurate estimations of all components are necessary for a complete visualization of the strain incurred in tissue. In addition, without those components, other important parameters, such as shear strains and the Poisson's ratio cannot be estimated. In general, knowledge of the strain tensor and Poisson's ratio is necessary for Young's modulus reconstruction algorithms (O'Donnell et al. 1994; Kallel and Bertrand 1996).

Previous research in the estimation of multidimensional components of the tissue displacement vector for strain imaging has been reported (Lubinski et al. 1996; Konofagou and Ophir 1998). Lubinski et al. (1996) obtained lateral displacements utilizing assumptions of tissue incompressibility. In that work, lateral displacements were computed using the higher precision axial displacement estimates and assuming a Poisson's ratio of 0.495. However, the incompressibility assumption may not hold in some tissues such as lung tissue with a Poisson's ratio of 0.3 (Fung 1981) and cartilage with a Poisson's ratio of 0.17 (Jurvelin et al. 2003). In addition, the Poisson's ratio may not be constant, such as for poroelastic tissue when edema is present (Konofagou et al. 2001; Righetti et al. 2004). Konofagou and Ophir (1998) proposed the simultaneous estimation of both axial and lateral displacements and strains using a precision tracking algorithm based on weighted interpolation between neighboring RF A-lines in the lateral direction, along with an iterative correction of lateral and axial displacements. They applied a number of cross-correlation and correction stages for axial and lateral displacements. However, the extensive lateral interpolation between RF A-lines and the iterative nature of the algorithm increase its computational complexity.

Techavipoo et al. (2004) recently proposed a method to estimate components of a displacement vector using RF echo signal data acquired along multiple angular insonification directions of the ultrasound beam. Tissue displacements at each spatial location in the compressed medium are measured along each beam direction using time-delay estimation techniques (Quazi 1981). A linear model of the relationship between these directional displacements and components of the actual displacement vector is constructed. The different components of the displacement vector are then estimated by using a least-squares solution (Techavipoo et al. 2004). The strain tensor components are computed from the gradient of the displacement vector components. The Poisson's ratio can be approximated from the ratio of the lateral strain to the axial strain under the plane strain assumption.

The above technique has yielded accurate normal and shear strain tensors for simulated pre and post compression RF echo signals (Techavipoo et al. 2004). In that study, angular data were simulated by rotating the transducer around the simulated phantom; however, the compression direction was maintained the same for all the angular data. Initial experimental tests were done by acquiring angular data using a phased array transducer that was mechanically translated in the lateral direction.

Implementation of this technique on a clinical ultrasound system will require the use of controlled beam steering on stationary linear or phased array transducers to obtain RF echo signal frames from a given location at different insonification angles. Elimination of the transducer translation and of the need for rearranging data as done in early tests will

significantly speed up data acquisition, improve the positional accuracy to assure that echo data used to measure tissue displacements from a given location originate from that location and reduce the computational requirements.

In this paper, we experimentally investigate this technique for estimating the lateral and axial components of displacement vectors using beam steering on a linear array transducer. A unidirectional quasistatic compression force is applied to a sample and pre- and postcompression RF echo signals are acquired at different beam steering angles. Angular displacements computed at different angles are used to estimate axial and lateral displacement vectors. Normal and shear strain images are then estimated from the gradient of displacements vectors. Experimental results from tissue-mimicking (TM) phantoms are presented and compared with the theoretical results.

MATERIALS AND METHODS

Data acquisition

A single-inclusion TM phantom of size $9 \times 9 \times 9\text{-cm}^3$ with a cylindrical inclusion (Madsen et al. 2006) and an anthropomorphic breast phantom (Madsen et al. 2006) with a diameter of 12 cm and a height of 14 cm were used to evaluate our method. The inclusion phantom contains a 1-cm diameter cylindrical object encased in a uniform background. The inclusion is three times stiffer than the background. The TM anthropomorphic breast phantom contains seven different inclusions of varying types, shapes and sizes, mimicking breast fat, cancers and fibroadenomas. A 1.4 cm diameter spherical TM fibroadenoma and a TM cancerous inclusion with an irregular shape were imaged in our experiments using this phantom. The unbound TM fibroadenoma simulated the case of a benign tumor, 2.3 times stiffer than the background, round with smooth edges and loosely bound to the surrounding material (Krouskop et al. 1998; Konofagou et al. 2000). The TM cancerous inclusion has simulated infiltrating spicules, 1.7 times stiffer than the background and fully bound to the surrounding medium.

Both phantoms were scanned using an Ultrasonix 500RP (Ultrasonix Medical Corporation, Bothell, WA, USA, and Vancouver, BC, Canada) real-time scanner equipped with a 5-MHz linear-array transducer with an approximately 60% band width. The Ultrasonix 500RP is also equipped with an ultrasound research interface (URI) that readily allows an expert programmer to enter the research environment and alter operating conditions of the system and also to introduce new echo signal processing techniques. To acquire RF data at different beam angles, we developed a URI client program to communicate with the Ultrasonix URI and software server and to control the beam steering algorithm. The client program enables the operator to input a maximum angle and an angular increment; the machine then automatically scans the phantom along the specified angular sweep. The beam steering program used in our experiment enables beam steering within a range -15° to 15° , starting from the axial direction of the nonsteered case. The phantoms were scanned over this angular range with an angular increment of 0.75° . The maximum beam-steered angle of 15° was chosen because of possibilities of grating lobes with the linear array transducer used in this study, and the angular increment of 0.75° was chosen based on previous work (Techavipoo et al. 2004). A schematic illustration of the angular data acquisition is shown in Fig. 1.

A compression plate with a rectangular slot for the transducer face was mounted on a linear stage driven by a stepper motor. The compression plate is larger than the phantom surface, to provide a uniform compression of the phantom, as shown in Fig. 1. The compression was applied along the axial direction, *i.e.*, at a 0-degree angle with respect to the nonsteered case. Echo signals were acquired over a total depth of 4 cm centered about the inclusion imaged. Data were acquired before and after a compression of 1% of the phantom height. The stepper-motor-induced compression process is controlled by the URI program on the Ultrasonix 500RP

system, enabling the pre- and postcompression RF data sets to be synchronized with the actual compression cycle.

Displacement estimation

Each of the angular RF pre- and postcompression frame pairs was analyzed separately, to determine tissue displacements at the specified beam angle. These are referred to as angular displacement frames. Angular displacements were computed using a 1-D cross-correlation algorithm with a window size of 3 mm and 75% overlap of consecutive windows. The window size and overlap chosen were a compromise between improved spatial resolution and smoothness of the displacement images. In this study, we used typical values for the window size (3 mm) and overlap (75 %) for the cross-correlation algorithm described in the literature.

Linear interpolation was then applied for image registration of the angular displacement data to a Cartesian spatial grid. Since the displacement images at different angles have different pixel grid locations and some may occur between the Cartesian spatial grid, bilinear interpolation was utilized to register the angular displacement estimates. By assuming that the angular displacement images are locally smooth, linear interpolation can be applied to interpolate the angular observations at the pixels onto the zero-angle grid (Techavipoo et al. 2004).

The displacement vectors at each pixel on the zero-angle grid can be estimated by using a least-squares approach. The relationship between the actual displacement vectors and the measured angular displacement is (Techavipoo et al. 2004):

$$q = A\bar{d} + n, \quad (1)$$

where

$$q = \begin{bmatrix} q_{\theta_1} \\ q_{\theta_2} \\ \vdots \\ q_{\theta_m} \end{bmatrix}, \quad A = \begin{bmatrix} \cos \theta_1 & \sin \theta_1 \\ \cos \theta_2 & \sin \theta_2 \\ \vdots & \vdots \\ \cos \theta_m & \sin \theta_m \end{bmatrix},$$

$$\bar{d} = \begin{bmatrix} d_z \\ d_x \end{bmatrix}, \quad \text{and } n = \begin{bmatrix} n_{\theta_1} \\ n_{\theta_2} \\ \vdots \\ n_{\theta_m} \end{bmatrix}.$$

The term q_{θ} represents an observation of the displacement vector $\bar{d}$ at beam angle θ_i for $i = 1, \dots, m$, where m is the total number of beam steering angles; n_{θ_i} is the noise in the observation at angle θ_i . For simplicity, the noise in the angular strain estimates is assumed to be similar for all measurements using this model. d_z and d_x denote, respectively, the axial and lateral displacement for the zero-angle condition. We can minimize the squared error between the measurement q and the linear model $A\bar{d}$ with respect to $\bar{d}$ to estimate the value of $\bar{d}$. The solution is the least-squares solution (Scharf 1991), which is given by:

$$\bar{d} = (A^T A)^{-1} A^T q \quad (2)$$

where $\bar{d} = [\bar{d}_z, \bar{d}_x]^T$ is a vector whose elements represent estimates of the axial and lateral displacement.

The number of angular displacement observations used to calculate displacement vectors is different at different pixel positions. Figure 2 shows an example of the map of the number of angular displacement observations used at each pixel to compute displacement vectors. The total number of displacement observations is 41, with the angle range from -15° to 15° and angular increment 0.75° . Note that only in the central part of the ROI (brightest display region) are the angular displacement from all angular acquisitions used. Fewer observations are available at the edges of the ROI.

Strain tensor estimation

Components of the strain tensor, including normal strains along the axial and lateral directions and shear strains, are obtained from the displacement vectors. The normal strains are defined by:

$$e_{zz} = \frac{\partial d_z}{\partial z} \text{ and } e_{xx} = \frac{\partial d_x}{\partial x} \quad (3)$$

where e_{zz} and e_{xx} are the strains in z and x directions, respectively, and the shear strain is defined by

$$e_{zx} = \frac{1}{2} \left(\frac{\partial d_z}{\partial x} + \frac{\partial d_x}{\partial z} \right) \quad (4)$$

In the example illustrated here, partial derivatives are approximated by using a least-squares strain estimator (LSQSE) (Kallel and Ophir 1997). The size of the kernel used in LSQSE is around 3 mm in this study. Finally, the computed normal and shear strain elastograms were filtered using a 5×5 median filter, corresponding to 3×3 mm in size, to suppress strain outliers.

RESULTS

Single-inclusion phantom

Figure 3 shows examples of the angular displacement and angular strain images of the inclusion phantom at the -12° , 0° and 12° beam angles with respect to the direction of compression. Displacement and strain images were displayed on the scale that provided good contrast. Some of the displacement or strain values encountered in the image were not within the color-bar ranges, leading to saturation of some pixels. The angular displacement images correspond to q_θ in eqn (1). Here, linear interpolation has been applied for image registration of the angular displacement data to a Cartesian spatial grid and the inclusion center is forced to be at the center of the zero-angle Cartesian grid coordinates in all images. Consistent with previous results using a mechanically translated phased array (Techavipoo et al. 2004), we observe increased noise artifacts in the displacement and strain images at the larger echo acquisition angles. The elastograms obtained at larger angles with respect to the direction of compression suffer from significant decorrelation of the pre- and postcompression RF echo signals as a result of greater motion of scatterers in the cross-beam direction for these acquisition angles.

Figure 4 presents plots of the measured displacement at three different locations in the single inclusion phantom (indicated in Fig. 2) vs. the beam angle. The theoretical prediction of displacement was obtained by modeling single inclusion phantoms using finite element analysis (FEA) software (ANSYS Inc., Canonsburg, PA, USA). The inclusion and the background were taken to be bound together and the inclusion was assumed to be three times stiffer than the background. The measured displacement (dots), least-squares fit of the measured data (dashed lines) and the theoretical prediction (solid lines) are all plotted vs. the beam angle. Results are shown for angular increments of 0.75° (top row) and 3° (bottom row). As mentioned earlier, the number of observations available in the least-squares solution may

be less than the total number of angled RF frames, depending on the pixel location. In Fig. 2, point I and point II are scanned by ultrasound beams from every angular insonification within the -15° to 15° angular data acquisition range. For point III, however, ultrasound beams from angles around 5° to 15° do not propagate through this region. Thus, as shown in Fig. 4c, the axial and lateral displacement data for point III are generated using fewer angular displacement observations. Figure 4 demonstrates that experimental results for the angular displacements follow closely the theoretical values obtained using the FEA model. The slight discrepancy between theory and experiment is likely because the compression plate was not precisely parallel to the phantom surface resulting in the induced stress on the top surface of the phantom not being uniform.

Analytical and estimated displacements at points I, II and III in Fig. 2 are shown in Table 1. Analytical values of the displacements were obtained using FEA simulation and these are regarded as the theoretical prediction. The three locations or points in the image are chosen to represent regions in the phantom that have zero, positive and negative lateral displacement, respectively. Because point I is along the A-line that is at the center of the compressed surface, the lateral displacement at this position should be zero. Points II and III, on the other hand have opposite signs for the lateral displacement. In addition, point III is at the edge of the ROI, where fewer observations are available, as shown in Fig. 2. As shown in Table 1, the estimated displacement vector components are very close to the predicted values. Results are similar for data generated using angular increments of 0.75° and 3° .

Ideal and experimental images of the displacements in both the z (axial) and x (lateral) directions for the single inclusion phantom are shown in Fig. 5. We also estimated the displacement images by applying a 2-D cross-correlation algorithm to the zero-angle RF data and this is shown in Fig. 5c. The quality of the lateral displacement shown in Fig. 5c is very poor compared with the lateral displacement obtained by the beam steering technique (Fig. 5b, lower). This is because the lateral displacement is very small and less than the pitch between the A-lines. Figure 5a and b show that, as the phantom is compressed in the z direction, it also expands in the x direction. For the experimental results with the beam steered acquisition, noise artifacts in the axial displacement image (Fig. 5b, top) are lower than in the axial displacement image obtained only at zero degrees (Fig. 3b) because of the use of spatial compounding inherent in this method. The lateral displacement estimates (Fig. 5b, lower) are also significantly noisier than the axial displacement estimates because they are much smaller. The axis of symmetry can be observed to be at a depth of 2 cm for the axial displacement images and at a lateral position of 2 cm for the lateral displacement images.

Figure 6 presents the ideal and experimental strain images along the z and x directions. For the axial strain image obtained experimentally and shown in Fig. 6 (top), the inclusion shape and local strain values match those observed in the ideal image quite well. The lateral strain elastogram in Fig. 6 (bottom) has lower strain values than the axial strain image, which is expected because of the axial direction of the applied compression. The lateral strain image obtained using a 2-D cross-correlation algorithm without any lateral interpolation of the RF beam lines (Fig. 6c, bottom) is also shown. However, this image is very noisy and the inclusion is not clearly visualized. In contrast, the outline of the inclusion can be clearly seen in the lateral strain image obtained using the beam steering technique (Fig. 6b, bottom). Values of the CNR_e were calculated using the rectangular ROI within the inclusion and in the background region, shown in Fig. 6b, for the experimental strain images. The CNR_e of the axial and lateral strain images in Fig. 6b were 49 dB and 18 dB, respectively. For the strain images obtained using 2-D cross-correlation, the CNR_e obtained was 37 dB (axial) and -11 dB (lateral).

Shear strains obtained using eqn (4) are shown in Fig. 7, where comparisons between the ideal shear strain images (top) and their experimental counterparts generated from the angular

displacement data (bottom) are presented. Images for the first pair shown in Fig. 7a are almost identical, except for the increased noise in regions around the edges of the image and around the inclusion. Images in the second pair shown in Fig. 7b illustrate similar patterns of stress concentrations around the inclusion, but the experimental image is noisier because of greater variance inherent in the lateral displacement data. These noise artifacts lead to the poorer quality of the shear strain image in Fig. 7c. Nevertheless, the boundary of the inclusion still can be observed in this image.

Anthropomorphic breast phantom

Figure 8 shows the ideal and experimental normal strain images of the unbound fibroadenoma mimicking inclusion in the breast phantom. Similar to the results presented in Figs. 4 to 7, ideal strain images were obtained by modeling single inclusion phantoms using FEA software. In this case, the inclusion was not bound to the background material. It was assumed to be three times stiffer than the background, following the design of the experimental phantom. The lateral elastogram appears to be darker than the axial elastogram, which implies that a smaller strain was incurred in the lateral direction. Also, increased noise artifacts are observed in the lateral strain image. Shear strain images of the TM fibroadenoma inclusion are shown in Fig. 9. Note that regions of the black-and-white patterns that indicate shearing strains around the inclusion shown in Fig. 9a are smaller than those for the cylindrical inclusion shown in Fig. 7a. This is in agreement with the theoretical results for the shear strain patterns of unbounded inclusions. The strain magnitudes of the shearing patterns around the TM fibroadenoma are larger than in the case of the cylindrical inclusion, which implies a larger mobility of the unbound inclusion. This is easily explained by the spherical shape of the unbound TM fibroadenoma inclusion, which makes it more likely than the cylindrical inclusion to slip in the lateral and elevational directions.

Axial and lateral strain images of a TM cancerous lesion in the phantom are shown in Fig. 10. In this case, the simulated lesion is firmly attached to the background tissue-mimicking material. The irregular shape of the lesion can be observed clearly in both elastograms. The strain contrast appears lower than that of the TM fibroadenoma because of the smaller Young's modulus value of the materials used for the TM cancerous inclusion than those used in the simulated fibroadenoma.

Figure 11 presents shear strain images of the TM cancerous inclusion. As shown in Fig. 11a, the shear strain patterns around the lesion are not as obvious as those observed for the TM fibroadenoma shown in Fig. 9a. The absolute value of the strain around the TM cancerous lesion is also lower than that of the unbound TM fibroadenoma, which is consistent with a lower mobility of the TM cancerous lesion.

DISCUSSION

Both axial and lateral displacement images can be obtained by using the proposed technique. Axial and lateral strain images can then be derived from these displacement vector components. As shown in Fig. 6b (top), the axial elastogram obtained using this technique has a higher image quality than the "traditional" strain images shown in Fig. 3b because the displacement noise has been reduced by the linear fitting procedure. The technique, therefore, can also be used as a noise reduction method for elastography (Rao and Varghese 2005). The lateral strain image obtained by this technique, however, has a lower CNR_e than the axial strain image. This result can be explained by three factors. First, the applied compression is along the axial direction and the lateral displacement is significantly smaller than the axial displacement. Errors or noise artifacts, therefore, are larger in the lateral displacement measurements than in axial displacement determinations. The second reason is the limited range of insonification angles along which angular RF data could be acquired with the linear array probe. For the

linear-array transducer used in this study, the maximum beam steering angle is limited to 15° . From eqn (1), we find that the lateral displacements d_x contribute a very small component to the observed angular displacement because of the small magnitude of $\sin \theta$ ($-15^\circ \leq \theta \leq 15^\circ$), say 0 to ± 0.26 , whereas the axial displacement dominates the observed displacement. It is expected that better lateral strain images can be obtained by using larger insonification angles. However, decorrelation noise also increases with beam angle because of the cross-beam motion of the scatterers following the applied compression. More work is required to select an appropriate value of the maximum beam angle for optimal performance in strain imaging. The third reason for the low quality of the lateral strain image is the low bandwidths (around 60%) of the Ultrasonix system used in our experiments. Better results can be obtained using scanners exhibiting better bandwidths.

Other important strain tensor components that can be estimated from the axial and lateral displacements are shear strains. Shear strain elastograms demonstrate, for example, any slippage of a mass (Konofagou et al. 2000) that may occur during compression, as the shearing strains patterns in Figures 9 and 11 vividly show. Experimental results presented in this study suggest that shear strain may have useful implications in the differentiation of fibroadenomas from cancerous tumors, if the designs of the phantom truly represent such tumors. Different patterns can be observed in the shear strain images of the TM fibroadenoma and the TM cancerous inclusion, illustrated in Fig. 9 and Fig. 11. Smaller magnitudes of the shearing strain around the TM cancerous lesion imply that it is less mobile and that it does not slip or move during compression, as do the TM fibroadenomas.

The algorithm used in this paper does not utilize the incompressibility assumption proposed by Lubinski et al. (1996). The primary trade-off involves the collection of multiple angular RF data sets using our approach, whereas the Konofagou and Ophir (1998) method uses a single pair of pre- and postcompression echo signal frames. However, the use of an optimal angular increment would help to reduce the computational burden and yet obtain similar image quality. For very small angular increments (0.75°), the computational efficiency is low because the angular RF data are highly correlated. From Fig. 4 and Table 1, we also find that the angular increment of 3° provides similar results as that obtained using an angular increment of 0.75° , but the efficiency has been improved by a factor of four because fewer angular elastograms are utilized. This indicates that the optimum interval is probably close to 3° for the experimental system used in this study.

Because the compression used in elastography causes 3-D displacements, 3-D strain tensor imaging is necessary. The normal and shear components of the strain tensor estimation algorithm proposed in this paper can be easily extended to the 3-D case to estimate elevational displacements and strains. For this case, 3-D datasets, where the transducer beams are steered along both lateral and elevational directions, which could be obtained using 1.75 D or 2-D array transducers, can be incorporated into the algorithm.

CONCLUSIONS

Normal and shear strain images can be generated by using beam steering on a linear-array transducer. Experimental results on phantoms demonstrate the feasibility of this technique to function on clinical scanners for the computation of the normal and shear strain tensors. The angular increment and the maximum angle used are two important factors that affect the efficiency of the normal and shear strain estimation algorithm. The appropriate angular increments and the maximum beam steering angles for optimal performance in strain tensor imaging likely depend on the transducer, frequency, bandwidth and volume interrogated.

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Fig. 1.

A schematic diagram for elastographic imaging using a linear array transducer with beam steering. The z and x directions are defined as the axial and lateral directions of the ultrasound beam without steering.

Fig. 2.

Image illustrating the number of angular displacement observations available at each pixel to compute displacement vectors. The maximum number of displacement observations is 41 at points I and II, when the echo acquisition angle is from -15° to 15° and the increment between angular acquisitions is 0.75° .

Fig. 3. Displacement images (top) and strain images (bottom) of the single inclusion phantom under 1% uniaxial compression measured at (a) -12° , (b) 0° and (c) 12° , respectively. Note that the width and depth are in centimeters, while the displacement color bars are in millimeters.

Fig. 4.

Plot of displacement at three different locations in the inclusion phantom vs. beam angle, with an angular increment of 0.75° (top) and 3° (bottom). The case shown in (a) represents results point I in Fig. 2, where the applied compression results in an axial displacement of -0.45 mm, (b) (Point II) underwent an axial displacement of -0.52 mm and a lateral displacement of 0.032 mm, (c) Point III underwent displacements of -0.60 mm in the axial direction and -0.079 mm in the lateral direction.

Fig. 5.

(a) Ideal displacement, (b) experimental displacement images obtained from beam steering and (c) those obtained by applying a 2-D tracking algorithm to the zero-angle RF data in the axial (top) and lateral (bottom) directions for the inclusion phantom. The width and depth are in centimeters, while the displacement color bars are in millimeters.

Fig. 6.

(a) Ideal strain images, (b) experimental strain images obtained from beam steering and (c) those obtained by applying a 2-D tracking algorithm to the zero-angle RF data in the axial (top) and lateral (bottom) directions for the inclusion phantom. CNR_e were calculated using the rectangular ROI within the inclusion and in the background region for the experimental strain images. The CNR_e of the axial and lateral strain images in (b) are 49 dB and 18 dB, respectively. The CNR_e of the axial and lateral strain images shown in (c) are 37 dB and -11 dB, respectively.

Fig. 7.

Pairs of the ideal (top) and experimental (bottom) images of (a) $\partial d_z/\partial x$, (b) $\partial d_x/\partial z$ and (c) shear strain $= 0.5(\partial d_z/\partial x + \partial d_x/\partial z)$, where d_z and d_x are the displacements in the z (axial) and x (lateral) directions, respectively.

Fig. 8. Ideal (top) and experimental (bottom) strain images in the (a) axial and (b) lateral directions for the fibroadenoma inclusion.

Fig. 9.

Pairs of the ideal (top) and experimental (bottom) images of (a) $\partial d_z / \partial x$, (b) $\partial d_x / \partial z$ and (c) shear strain = $0.5(\partial d_z / \partial x + \partial d_x / \partial z)$, for the fibroadenoma mimicking inclusion.

Fig. 10.

(a) Axial and (b) lateral strain images for the cancer, mimicking inclusion.

Fig. 11. Images of (a) $\partial d_z / \partial x$, (b) $\partial d_x / \partial z$ and (c) shear strain = $0.5(\partial d_z / \partial x + \partial d_x / \partial z)$, for the cancer mimicking inclusion.

Table 1Displacements d (mm)

Direction	Point I		Point II		Point III	
	z	x	z	x	z	x
Position (cm)	2.0	2.0	2.8	2.5	3.5	0.5
Analytical d	-0.45	0	-0.520	-0.032	-0.603	0.079
Estimated d ($\Delta\theta = 0.75^\circ$)	-0.448	-0.008	-0.517	-0.040	-0.587	0.085
Estimated d ($\Delta\theta = 3^\circ$)	-0.448	-0.009	-0.517	-0.044	-0.587	0.086