Elastographic Imaging Using a Handheld Compressor

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Elastography is an emerging imaging modality that allows noninvasive imaging of tissue stiffness changes and stiffness values associated with pathology or as a result of therapy. However, many currently-used systems for elastography rely on a fixed geometry transducer and compressor system for imaging. This configuration is disadvantageous for imaging difficult-to-reach regions that are currently accessible with conventional ultrasound. In this paper, we describe a handheld, portable stepper motor controlled system for elastography. This system may reduce motion and jitter errors that are prevalent in completely ‘freehand elastography’ that employs hand-induced compressions using the transducer. The latter also requires collection of large amounts of data and use of strain estimation algorithms that may not be sensitive to phase changes or use additional preprocessing to minimize decorrelation effects. The stepper motor controlled handheld system provides controlled compressions and synchronized data acquisition. Our technique yields elastograms of a low-contrast phantom that have contrast levels and contrast-to-noise ratios that are comparable to those obtained with a fixed geometry system.

KEY WORDS: Elastography; elastogram; elasticity; freehand; handheld; imaging; rf ablation; strain; ultrasound.

INTRODUCTION

Imaging of tissue elastic parameters for diagnosis and treatment is rapidly gaining attention because of its ability to provide noninvasive and new diagnostic information. Efficient application of elastography requires carefully controlled compression of the tissue along with synchronized data acquisition. Most of the current systems used in elastography are limited in the sense that the tissue and transducer used for imaging are constrained to a fixed geometry. In this paper, we present a handheld compression system that, while not constrained to a fixed geometry, provides both controlled compression of the tissue and synchronized data acquisition. Other freehand or handheld approaches to elastography involve manual hand-induced compressions using the transducer.

Completely freehand elastography was first proposed by Bamber et al, where the speckle decorrelation rate was used as an estimate of tissue displacement and strain. However, most of the freehand techniques are limited on several fronts, the foremost being their inability to control motion and jitter errors. This has resulted in the use of robust (noise resistant) and motion-invariant strain estimators or additional preprocessing algorithms that minimize decorrelation effects. These methods are less sensitive than the algorithms used with fixed geometry systems.

Freehand methods also require continuous acquisition of ultrasound data at frame rates that depend on the amount and rate of compression, leading to acquisition and storage of excessive amounts of data. Doyley et al developed a fast data acquisition system to capture raw analog intermediate frequency (IF) data at the full frame rate of the ultrasound scanner. The IF data were acquired over a duration that varied with the strain rate. The acquired data were preprocessed using a model-based global motion distortion compensation technique by
resampling the postcompression rf signal using \textit{a priori} knowledge of the applied strain. Elastograms were then computed using a 2-D normalized cross-correlation method.\textsuperscript{21} Another group that uses freehand imaging\textsuperscript{22} reports the collection of rf frames at a frame rate of 11 Hz, with the acquisition of about 15-20 IQ (baseband) frames for each freehand acquisition. Another limitation with completely freehand imaging is that the applied displacement and hence the strain varies through successive frames.\textsuperscript{19,22} Hiltawsky et al\textsuperscript{22} report variations in the applied displacement range from 50-133 μm, yielding applied strains in the range of 0.25 to 0.67% per frame.

To overcome the difficulties described in data acquisition with completely freehand imaging, Zhu and Hall\textsuperscript{23} proposed the incorporation of a real-time processing and display feature in the freehand data acquisition. Real-time processing significantly improves the outcome with freehand elastography, allowing for acquisition and storage of only the data where quality elastograms are generated.\textsuperscript{17, 23} Another factor that has to be accounted for in freehand elastography, however, is that the elastograms obtained also include effects induced by the viscous properties of tissue due to the finite time lag between compressions and acquisitions of the rf data.

The purpose of this paper is to describe a handheld technique for elastography that incorporates inline stepper motors into the transducer assembly. The mechanized technique described here may minimize motion and jitter errors associated with purely handheld compression. In addition, it provides more efficient synchronized data acquisition using a stepper-motor controlled compression. Since data acquisition is synchronized to the compression, we acquire only a pair of rf frames (i.e., the precompression and postcompression frames), similar to the fixed-geometry elastography system.\textsuperscript{7}

**MATERIALS AND METHODS**

In this section, we discuss the application of the handheld compression device to obtain elastograms in \textit{ex-vivo} situations for imaging a single-inclusion elastography phantom and for delineating thermal lesions generated by rf ablation.\textsuperscript{26} The handheld approach is compared to the commonly-used fixed geometry system for elastography under similar conditions.

For handheld elastographic imaging, the system incorporates a stepper motor controlled compression device as illustrated in figure 1. The handheld compression system is a Vexta PX 245-01 A. A stepper motor and a low profile unislide assembly with a 1.5-inch slider. The motion of the system is controlled using a programmable stepper motor controller. The transducer is attached to the unislide using a plexiglass frame that incorporates a compression plate (larger than the sample) to provide a uniform stress distribution. The handheld compression system is portable, lightweight and provides a uniform, controlled compression of tissue. For the fixed geometry elastography system, the stepper motor controlled compressor is mounted on a rigid frame. A personal computer controls the operation of both the handheld and fixed geometry elastography system.

**Phantom tests of the handheld system**

Elastographic imaging performance using the fixed geometry system and the handheld system was quantitatively evaluated using an Aloka SSD-2000 real-time scanner (Aloka Co. Ltd., Tokyo, Japan). The scanner operates with dynamic receive focusing using a 5 MHz linear array transducer (40 mm) with a 70% bandwidth at -6dB. The ultrasound rf signals were digitized using an 12-bit data acquisition board (Gage Applied Sciences, Toronto) at a sampling rate of 50 MHz. A single transmit focal zone was employed.
Tests were done by imaging a low contrast elastographic phantom immersed in safflower oil. The phantom is constructed from water-based gels, with tissue-like scatter and attenuation. The gelatin phantom (90 x 90 x 90 mm) contains a 10 mm diameter cylindrical inclusion whose axis is 2.5 cm from, and parallel to, the top surface of the phantom. The elastic modulus of the material comprising the inclusion is 1.6 times that of the surrounding background material, with both materials assumed to possess a Poisson's ratio of 0.495. The phantom's properties are representative of soft tissue, with a speed of sound of 1,510 m/s, and an attenuation coefficient of 0.7 dB/cm-MHz.

Rf echo signals were acquired before and after 1% compressions of the phantom, first using the handheld system and then the laboratory based fixed-geometry elastography system. A large compressor (larger than the phantom surface) on both systems simulates uniform stress conditions in the phantom. Placement of the phantom in a safflower oil bath helps satisfy slip boundary conditions, i.e., the phantom is free to move in both the lateral and elevational directions during compression.

Offline analysis of the rf echo-signal frames acquired before and after compression was performed using a coherent cross-correlation algorithm that compares the echo-signals vector-by-vector. A 3 mm window length with a 50% overlap of the data segments was used in this analysis. Local tissue displacements were estimated from the echo-shifts between pre- and postcompression echo signals. A 3-point least squares algorithm was used to obtain the gradient of the displacement estimates to estimate the local strain. Soft tissues deform more than stiffer tissues, and these differences are quantified in images of tissue strain, termed

*Measurement of the elastic properties of phantom materials was performed by Dr. Thomas Krouskop at the Baylor college of Medicine using an Instron mechanical testing machine.
elastograms. Noise spikes in the elastograms were reduced using a 5x5 median filter. No other filtering or smoothing techniques were used while processing the rf data frames. A discussion on the resolution of both the fixed geometry and hand-held elastographic system is presented in the Appendix.

Ex-vivo handheld elastography on liver tissue

The performance of the handheld system was also evaluated on thermal lesions generated in ex-vivo canine liver tissue during rf ablation.73 Lobes of liver tissue with approximate dimensions of 40 mm by 40 mm and 20 mm thickness were encased in a gelatin cube with dimensions of 80 mm. Liver tissue was encased in the gelatin phantom to provide a regular surface for compression. The gelatin concentration was regulated so that its stiffness was approximately 1.6 times that of the normal liver tissue. This estimate was obtained using elastograms of normal liver samples encased in gelatin for a 1% applied compression using the fixed geometry system.

A RITA model 1500 electrosurgical device (RITA Medical Systems Inc., Mountain View, CA) was used for the ablation. The rf probe was inserted into the liver tissue through the gelatin phantom at a depth of 3 cm. Ultrasound imaging was performed in a direction perpendicular to the rf ablation probe axis. Rf ablation of the target tissue was performed for 10 minutes, raising the target temperature to 90 °C at a 50 W power level.18, 26 Elastograms were generated about 10 minutes after the ablation was completed to allow the lesions to cool off. Frames of rf echo-signals were acquired before and after compressive increments of 1%, first using a standard fixed geometry elastography system18 and then using the handheld device described in this paper.

RESULTS

Elastograms of the single-inclusion low-contrast elastography phantom using both the fixed-geometry and the handheld system are shown in figure 2. Elastograms using the two acquisition modes exhibit similar contrast and contrast-to-noise ratio (CNR)31 characteristics, as illustrated in table 1 (see Appendix for definitions of these parameters). The contrast between the inclusion and the background and the $CNR_e$ in each elastogram were computed for a small 7x7 mm$^2$ region in the inclusion, and small 7x7 mm$^2$ regions at the four corners of the images of the phantom. Using data from the corners reduces the impact of stress concentration artifacts on the $CNR_e$. The mean and median values from the four estimates of the $CNR_e$ were then calculated. The values shown in table 1 are the mean and standard deviation over twenty-five independent elastograms. Observe that we obtain higher values of the

<table>
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<th>Contrast Mean</th>
<th>CNR$_{e}$ (dB) Mean</th>
<th>CNR$_{e}$ (dB) Median</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>s.d.</td>
<td>s.d.</td>
<td>s.d.</td>
</tr>
<tr>
<td>Fixed geometry</td>
<td>1.2953</td>
<td>0.0676</td>
<td>9.9158</td>
</tr>
<tr>
<td>Handheld system</td>
<td>1.3039</td>
<td>0.0574</td>
<td>8.8704</td>
</tr>
</tbody>
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TABLE 1 Contrast and contrast-to-noise ($CNR_e$) ratios for the phantom. The mean and standard deviations of the strain estimates over a 7x7 mm$^2$ region, were computed over 25 different independent elastograms of the same phantom.
Using the fixed geometry system than with the handheld system. This is expected because the latter is subject to jitter errors. However, the handheld system still provides comparable results to the fixed-geometry system.

Mean strains and elastographic signal-to-noise ratios (SNRe) obtained from the same 7x7 mm² region in the inclusion and the small 7x7 mm² regions at the four corners of the elastogram of the phantom are presented in table 2. Note that small reductions in the SNRe are apparent when comparing the values obtained for handheld and fixed geometry elastography. A point that is illustrated by this table is that there is a reduction in mean strains obtained in both the background and inclusion for the handheld approach. The recoil of the handheld system may cause this reduction in applied strain during compression, resulting in effectively smaller applied strains on the phantom than the strain obtained with the fixed-geometry system. This reduction in the applied strains may also be a contributing factor to the reduction in the CNRe and SNRe observed in tables 1 and 2.

The elastographic phantom used in this study provides a challenging test object due to the low-contrast nature of the inclusion. The results in figure 2 and tables 1 and 2 illustrate that

**FIG. 2** B-mode gray scale image and elastogram of the test phantom obtained using (a) a fixed geometry elastography system and (b) the handheld stepper-motor controlled elastography system.
the strain sensitivity (see Appendix) of the handheld system would be comparable to that of the fixed geometry system, even though mean strains induced by the handheld compressor were somewhat smaller.

Elastograms and B-mode images of the liver lesions, obtained with the conventional fixed geometry elastography system, are illustrated in figure 3. Note that on B-mode images we observe increased echogenity near the ablated region and some shadowing below the region of high echogenicity. The increased echogenity may be due to bubbles formed during the ablation procedure. Nevertheless, the lesion boundary is only faintly seen in the B-mode image because of the hyperechoic core and a hypoechoic region surrounding it. On the other hand, the thermal lesion is clearly observed in the axial-strain elastograms. The thermal lesion is depicted as a stiffer (darker or lower strain) region in the elastogram.

Corresponding handheld elastograms and B-mode images of the thermal lesion are presented in figure 4. Figures 4(a) - (c) show three different sets of gray-scale B-mode images along with the corresponding elastograms of the ablated tissue. The thermal lesions are slightly off-center due to the handheld nature of the compressions. Again, the lesions are clearly visualized as stiffer regions in the elastograms. Comparisons of the elastograms in figure 4 illustrate the repeatability of the handheld experiment. Since the stiffness ratio in the liver-in-gelatin phantoms may change across liver samples from different animals, no quantitative results are presented for the thermal lesions.

<table>
<thead>
<tr>
<th>Inclusion</th>
<th>Background</th>
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<tr>
<td>Mean strain</td>
<td>Mean (SNR_e)</td>
</tr>
<tr>
<td>Mean</td>
<td>s.d.</td>
</tr>
<tr>
<td>Fixed geometry</td>
<td>0.0085</td>
</tr>
<tr>
<td>Handheld system</td>
<td>0.0057</td>
</tr>
</tbody>
</table>

FIG. 3 B-mode gray scale image and elastogram of a liver sample containing a thermal lesion obtained using the fixed geometry elastography system.
FIG. 4 Handheld B-mode gray scale image and elastograms of liver tissue as in figure 3. (a) to (c) represent three different instances of elastograms generated using the handheld stepper-motor controlled elastography system.
The thermal lesion depicted by the handheld technique has a size and shape similar to that obtained using the fixed-geometry elastography system. In addition, there are no striking differences in the apparent noise properties in the elastogram in figures 3 and 4. A comparison between the elastograms in figures 3 and 4 demonstrates no significant disadvantages associated with the use of the handheld system. Under the method of handheld elastography proposed in this paper, the clinician only has to hold the compression system steady at a fixed position. Errors due to unsteady motion of the transducer and variations in the tilt/angle of the transducer with respect to the imaged surface are expected to be lower for this method than those presented using a completely freehand acquisition. In addition, due to the synchronized acquisition of the data, we acquire significantly less data than that required with completely freehand elastography. The reduction in the data collected provides advantages in storage and in the processing time required to select viable data sets that generate quality elastograms.

**DISCUSSION AND CONCLUSIONS**

Completely freehand imaging in elastography would significantly enhance the appeal of the technique in clinical situations. However, there are several problems associated with the use of hand-induced transducer motion and compression. First, it is difficult to avoid rotational and translational motion of the transducer during compression, and this can lead to out-of-plane motion. Acquisition of adequate ultrasound rf data pairs under these conditions will require real-time elastography to accommodate the amount and range of compression. Secondly, any out-of-plane motion will induce additional decorrelation that would negatively impact the sensitivity and $SNR$ of strain estimates; it may also introduce additional artifacts in the elastogram. Thirdly, completely handheld transducer motion may lead to larger compressions or displacements, which in turn would require the use of more robust but less sensitive signal processing approaches for strain estimation.

Another factor that is ignored in freehand elastography is the viscoelastic nature of the imaged tissue. Because an instantaneous compression along with synchronized acquisition of the rf data is not employed, some amount of viscous relaxation of tissue would contaminate the elastogram. The handheld technique presented in this paper allows for carefully-controlled compressions that minimize motion errors caused by rotation or translation of the ultrasound probe. The clinician only has to hold the compression system steady during the examination. The system approaches the flexibility of completely freehand elastography while minimizing extraneous noise sources. The compression induced is instantaneous, and the data acquired are synchronized to the compression, thereby minimizing tissue viscous effects in the generated elastogram. The handheld compression system illustrated in figure 1 can be made even more compact using inline compressors that would provide specific compression increments. In addition, the compression plate can be removed to access difficult-to-reach regions. Synchronization of the data acquisition with the stepper-motor controlled tissue compressor provides the most efficient means of acquiring rf data. Finally, the use of this method of compression allows the use of previously-developed, sensitive strain estimation algorithms for elastography, thereby allowing the generation of high quality elastograms.

It is possible that the handheld elastograms presented in this paper (Figs. 2, 4) may include some effects of motion-induced artifacts and jitter errors. Nevertheless, the use of the controlled compression system appears to reduce these errors as shown by the contrast and $CNR$ values in table 1 and the mean strain and $SNR$ values in table 2. The strain algorithm used to generate the elastograms in this work is the coherent cross-correlation method, which is very
sensitive to signal decorrelation errors. Also, no additional preprocessing is performed prior to strain estimation.

We observed a small decrease in the $\text{CNR}_{e}$ and $\text{SNR}_{e}$ with the handheld system compared to the fixed geometry system. This is shown in tables 1 and 2 for the elastograms generated using the phantom. Nevertheless, the contrast values are similar for both methods, and the standard deviations illustrate the precision of the results. In addition, $\text{ex-vivo}$ elastograms obtained on ablated canine liver tissue demonstrate qualitative agreement between elastograms taken with the handheld and fixed geometry elastography systems.

The availability of handheld systems for elastography may enhance the appeal of elastography for clinical applications. The ease of imaging that is currently available with an ultrasound system is maintained while providing additional diagnostic information about tissue stiffness, such as that associated with pathology or resulting from interventional procedures. The approach described in this paper can be viewed as a bridge between totally-freehand elastography and bench-top systems.

**ACKNOWLEDGEMENTS**

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**APPENDIX**

**Signal-to-noise ratio ($\text{SNR}_{e}$)**

The elastographic $\text{SNR}_{e}$ characterizes the noise of the elastographic system at which a value of strain is estimated. It is defined by

$$\text{SNR}_{e} = \frac{m_s}{\sigma_s}$$

where $m_s$ denote the mean strain estimate and $\sigma_s$ denotes the standard deviation of the strain noise estimated from the elastogram.

**Contrast-to-noise ratio ($\text{CNR}_{e}$)**

The $\text{CNR}_{e}$ encompasses the contrast and noise characteristics of the output elastogram and is given by:

$$\text{CNR}_{e} = \frac{2(s_1 - s_2)^2}{\sigma_{s1}^2 + \sigma_{s2}^2}$$

where $s_1$ and $s_2$ represent the mean values of the strain in two regions, and $\sigma_{s1}^2$ and $\sigma_{s2}^2$ denote the strain variances in the two regions respectively.
Sensitivity

Strain sensitivity\textsuperscript{35,37} is defined as the smallest value of the strain ($s_{\text{min}}$) in the elastogram depicted at a specified value of the SNR, i.e.,

\[
\text{Sensitivity} = s_{\text{min}} \left| \begin{array}{c}
\text{SNR} \\
\text{e, ref}
\end{array} \right.
\]

Resolution

The length of the cross-correlation window was initially used as a measure of the axial resolution in elastography. Recently, it was found that axial resolution in elastography is ultimately limited by the duration of the ultrasound pulse, i.e., it is inversely proportional to the absolute bandwidth of the ultrasound system.\textsuperscript{31} Thus, the attainable resolution is ultimately limited by the impulse response of the ultrasonic system, in a similar manner to the limits of resolution of the underlying sonographic process. Elastographic resolution has also been heuristically determined for a controlled simulation experiment\textsuperscript{30} of a low-contrast 1-D wedge phantom. These results illustrate that the window overlap factor (in comparison to the window length) dominates resolution.

Previous simulation results for the elastographic resolution\textsuperscript{30,31} assumed the gradient method\textsuperscript{1-6} for strain estimation, along with global temporal stretching.\textsuperscript{6} In our paper, we have used a 3-point least squares method\textsuperscript{29} to obtain strain estimates without temporal stretching, and the corresponding resolution values have not yet been derived for this method. Nevertheless, the point to be noted is that both the fixed-geometry and handheld elastographic systems presented in this paper have the same resolution since everything other than the data acquisition strategy are identical in these systems.

REFERENCES


