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C-plane Reconstructions from Sheaf Acquisition for Ultrasound Electrode Vibration Elastography

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Abstract

This paper presents a novel algorithm for reconstructing and visualizing ablated volumes using radiofrequency ultrasound echo data acquired with the electrode vibration elastography approach. The ablation needle is vibrated using an actuator to generate shear wave pulses that are tracked in the ultrasound image plane at different locations away from the needle. This data is used for reconstructing shear wave velocity maps for each imaging plane. A C-plane reconstruction algorithm is proposed which estimates shear wave velocity values on a collection of transverse planes that are perpendicular to the imaging planes. The algorithm utilizes shear wave velocity maps from different imaging planes that share a common axis of intersection. These C-planes can be used to generate a 3D visualization of the ablated region. Experimental validation of this approach was carried out using data from a tissue mimicking phantom. The shear wave velocity estimates were within 20% of those obtained from a clinical scanner, and a contrast of over 4 dB was obtained between the stiff and soft regions of the phantom.

I. Introduction

Needle-based percutaneous ablation procedures are commonly carried out using either a microwave antenna or radiofrequency (RF) electrode for treating patients suffering from hepatocellular carcinoma [1]. Localized heating of cancerous tissue results in cell death, and recurrence can be prevented if a sufficiently large volume of the affected area of the liver is ablated. It is therefore crucial to provide the clinician with live feedback of the size and extent of the ablation during the course of the ablation procedure.

Ultrasound electrode vibration elastography has the potential of providing pseudo-real-time information, and additionally has the advantage of being less harmful than radiation based imaging modalities. Three dimensional imaging using ultrasound has thus far proved to be a challenge both in terms of computation, and the requirement of high frame rates. However, the advent of phase locked acquisition schemes, plane wave imaging and increased computational capabilities of modern processors has resulted in growing interest in three dimensional (3D) ultrasound elastography.

Existing 3D elasticity imaging methods using 1D array transducers operate by sweeping or translating the imaging plane to obtain many elevational slices with the help of a wobbler transducer [2]. In contrast, the present paper proposes a faster approach which allows the clinician to reconstruct a 3D volume with very few imaging planes. Another method involves the use of 2D matrix transducers that image the entire volume [7] and elasticity properties can be estimated from these 4D datasets. However, full 3D acquisitions provide lower frame rate and may take many minutes to register and process.

This paper presents an algorithm that generates C-plane visualizations of ablated volumes reconstructed using traditional 2D ultrasound shear wave elastographic images. Data is acquired over a set of collinear planes by rotating a linear array transducer around the ablation needle which acts as the common line of intersection. Mathematically, this geometry is called a *sheaf* [4].

II. Materials and Methods

A. Experimental Setup

An oil-in-gelatin based tissue mimicking phantom [5] was used for this study. The phantom consisted of a stiff ellipsoidal inclusion surrounded by a softer “background” material, to mimic the presence of an ablated, stiffened volume embedded in healthy hepatic tissue. A steel rod firmly attached to the center of the inclusion acted as an ablation needle. A cross-sectional view of the structure of the phantom is shown in the top panel of Fig. 1. The stiff inclusion was clearly visible in the B-mode image of the phantom due to different acoustic echogenicities of the two materials used. The phantom was placed in an open top acrylic container and a layer of safflower oil was poured to prevent desiccation. In practice, it is not easy to visualize the ablated region using B-mode alone, which makes elastography a promising tool for stiffness based imaging [6].

The experimental setup used for generating shear wave pulses consisted of an actuator that was attached to the steel rod. This actuator was driven by a control module (Physik Instrumente, Germany) synchronized to the transmit sequence of the ultrasound scanner. The control module was programmed to produce a narrow half-sinewave pulse, 20 ms wide and 200 microns in amplitude, using the actuator.

Pseudo-high frame rate RF ultrasound data was acquired using a research mode ultrasound scanner (Analogic-Ultrasonix Corp. Richmond, BC, Canada) capable of generating beamformed echo data for post-processing. A 9L4 linear array transducer operating at 5 MHz center frequency was used for imaging the phantom to a depth of 4.5 cm. Each RF data frame consisted of 128 lateral A-lines sampled at 40 MHz axially. A phase locked acquisition scheme was used to attain high frame rates necessary for tracking the shear wave generated after needle vibration [8].

The sheaf pattern of acquisition was achieved by rotating the transducer around the needle, while ensuring that the needle was always in the image plane. The angle of the image plane was aligned with the help of guide markers placed along the walls of the plastic container that held the phantom.

B. Reconstruction Algorithm

A crosscorrelation based frame-to-frame displacement estimation routine was used to estimate the local displacement of each pixel in the image plane. Axial displacements were tracked using a 1D correlation window of size 2 mm and a 75% axial overlap between successive windows. This information was then used to analyze the axial displacement trajectory as a function of time for each pixel in the image plane. The peaks of these trajectories were used as an estimate of the time of arrival (TOA) of the shear wave pulse at each pixel [9].

Assuming that the shear wave pulse travels in a direction perpendicular to the needle (laterally along the face of the transducer), the reciprocal of the slope of the TOA curves as a function of distance from the needle can be used to estimate the shear wave velocity (SWV). This is shown in the bottom panel in Fig. 1. To mitigate the amplification of noise that is a concomitant of finite differencing of noisy data, a 5-sample sliding average filter was applied to the data prior to finite differencing. The final SWV image was median filtered using a rectangular kernel with 5 axial samples and 2 lateral samples. This was repeated for every imaging plane in the sheaf to obtain a collection of shear wave images.

The next step of the algorithm involved estimation of the shear wave velocities on different C-planes. Since each C-plane is perpendicular to the image planes, SWV estimates form a “star pattern” as shown in Fig. 2. These data points are used to infer the shear wave velocities on a fine rectangular grid over the entire C-plane. Let $\mathbf{v}$ denote the vector of SWV estimates (data points), and $\mathbf{x}$ denote the vector of unknown values on the grid. The final reconstruction must obey two intuitive requirements: (a) it should agree with the data points and (b) it should be smooth to some extent. The first requirement is enforced by assuming that each data point is approximately equal to the value obtained from bilinear interpolation of the (unknown) values at the four nearest grid points. This can be compactly represented using a matrix $\mathbf{B}$ which contains the bilinear interpolation coefficients for each grid point in $\mathbf{x}$. The linear system of equations that must be solved can be written as $\mathbf{B}\mathbf{x} = \mathbf{v}$. Note that this system may not have a unique solution in case $\mathbf{B}$ has more columns than rows. This situation will arise when one attempts to reconstruct the C-plane on a very fine grid of points, exceeding the number of data points. Imposing additional smoothness constraints is one way of bypassing this issue.

For this paper, a smoothness constraint is applied by calculating the sum of second derivatives (Laplacian) along both axes of the C-plane. A small value of the Laplacian is desirable to ensure a smooth reconstruction. The Laplacian is approximated using 2nd order central differences; this finite differencing operation for each point on the grid can be compactly represented using another matrix $\mathbf{D}$, where $\mathbf{D}\mathbf{x}$ represents the vector of Laplacian values on the grid. The complete C-plane reconstruction can now be posed as the following optimization problem:

$$\underset{\mathbf{x}}{\text{minimize}} \|\mathbf{B}\mathbf{x} - \mathbf{v}\|^2 + \lambda \|\mathbf{D}\mathbf{x}\|^2 \quad (1)$$

where the “tuning parameter” λ controls the effect of the smoothing term. For this study, this value was manually chosen as $\lambda = 0.006$.

C-planes reconstructed at different depths were then stacked to reconstruct and visualize the ablation in 3D.

C. Data Analysis

Cuboid regions of interest (ROI) were manually selected in the three different regions of the volume to calculate the mean (μ) and standard deviation (σ) of the SWV. These measurements were compared against values obtained from a commercial Supersonic Imagine (Aix-en-Provence, France) scanner using the shear wave imaging mode (SSI). Additionally, signal to noise ratio and contrast values were also calculated using the following equations:

$$SNR(dB) = 20 \log_{10} \frac{\mu}{\sigma}$$

and

$$C(dB) = 20 \log_{10} \frac{\mu_1}{\mu_2}$$

where subscripts denote two different stiffnesses present in the tissue mimicking phantom.

Mean and standard deviations were calculated from 5 independent datasets obtained by repeating the phantom experiment.

III. Results

As seen from Table I, the SWV estimates obtained from the 3D C-plane reconstructions were within 20% of those obtained using the SSI method. The SNR was over 10 dB for all three regions of interest.

Contrast between different pairs of regions is shown in Table II. The best contrast of 4.8 dB was obtained between the stiff ablated region and the soft background; the partially ablated region and the stiff region, however, were the most difficult to distinguish.

Two different imaging plane reconstructions with corresponding B-mode images of the phantom are shown in Fig. 3. A lower velocity partially ablated region can be seen in the left half of Fig. 3(b). The C-plane reconstructed at half the imaging depth is shown in in Fig. 4(a), and Fig. 4(b) shows a slice-plane view of the 3D reconstruction obtained after stacking all the C-planes together. In the C-plane view, a small portion of the partially ablated region is visible in the right half of the image.

IV. Conclusion

This paper presented a method for extending 2D shear wave imaging methods to obtain C-plane and 3D reconstructions with very few 2D imaging planes. The C-plane reconstruction algorithm can be extended to other imaging geometries as well. It is also not limited to SWV

reconstruction alone. It can be used for ablation monitoring by reconstructing and visualizing other physical properties of interest, such as strain or temperature.

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Fig. 1.

A cross-sectional view of the phantom in an imaging plane passing through the partially ablated region is shown in the top panel. The needle is firmly bound to the stiff ellipsoidal inclusion, which is embedded in a softer background material. The region mimicking a partially ablated region is attached on one side of the ellipsoid. The bottom panel shows an idealized plot of shear wave pulse arrival time plotted as a function of distance away from the needle along an imaginary line at constant depth in the imaging plane. Change points in this plot correspond to material boundaries whereas the shear wave velocity is equal to the reciprocal of the slope of this function.

Fig. 2.

The sheaf pattern of acquisition involves rotating the transducer through different planes that share a common axis aligned with the ablation needle. SWV images are reconstructed on each imaging plane separately. A C-plane at approximately half the imaging depth is shown. Cross-hairs indicate locations of data points (shear wave velocity values) available on this C-plane that are used for approximating SWV values over a fine grid of points on this plane.

Fig. 3.

B-mode images and SWV maps for two different image planes are shown in this figure. The partially ablated region is visible in (a) and (b), but not in (c) and (d).

Fig. 4.

(a) One of the C-planes reconstructed at half the imaging depth. (b) A 3 slice view of the 3D reconstruction obtained after stacking all the C-planes.

TABLE I

SNR in dB and shear wave velocity estimates in m/s

	SNR (dB)	SWV (m/s)	SSI SWV (m/s)
background	12.8 ± 1.7	0.6947 ± 0.04	0.9 ± 0.07
irregular region	14.4 ± 0.7	0.9843 ± 0.02	1.1 ± 0.05
ellipsoid	11.5 ± 1.9	1.2199 ± 0.12	1.2 ± 0.03

Values of SWV for the three regions of interest are shown. The number of equi-angular imaging planes used for 3D reconstruction is 16. These values are within 20% of the shear wave velocity values for the background and 5% for the ellipsoid, measured using a commercial imaging system (SSI).

TABLE II

Contrast (dB)

	Contrast (dB)
irregular/background	3.04 ± 0.33
ellipsoid/irregular	1.85 ± 0.64
ellipsoid/background	4.87 ± 0.35