Available online www.jocpr.com

Journal of Chemical and Pharmaceutical Research, 2013, 5(12):332-337

Research Article

ISSN : 0975-7384
CODEN(USA) : JCPRC5

Ultrasound lateral displacement and lateral strain estimation using a two-step strategy

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ABSTRACT

Ultrasound elastography has been shown to be useful technique in imaging information about the tissue's elastic moduli, which plays an important role in early diagnosis of several kinds of tumors. Usually, elastograms display only axial displacement and axial strain. Lateral displacement and strain estimation is difficult because there is no phase information and the precision of data is lacking too. Most of the lateral estimation methods previously reported are based on cross-correlation function (CCF). And several kinds of interpolations of high computational cost are often used to obtain more accurate results. In this paper, a two-step method combining two-dimensional sum of squared differences (SSD) and cubic spline representation for lateral displacement and strain estimation is introduced. This method can result directly in accurate and continuous estimates with no need for interpolation. Computer simulations and phantom experiments are performed to investigate the performance of proposed method in terms of bias and jitter errors and image quality.

Keywords: Elastography, Lateral, Displacement estimation, Speckle tracking, Spline

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INTRODUCTION

Elastography, or ultrasound strain imaging, is of great interest and has been well developed in past decades [1-4]. In this technique, tumors or cancers are distinguishable from normal tissue by their different magnitudes of strains, which reveal the change of the pathological tissues' elastic properties. In quasi-static strain imaging, an external mechanical pressure is applied to biological tissue. Then displacements and strains are estimated from pre- and post-compression ultrasonic signal to represent the elastic properties of the tissue [1]. By displaying these displacements and corresponding strains, some pathological information of the tissue could be observed in early clinical diagnosis.

Typically, only axial displacement and strain processing are performed, which are in the same direction as the external compression and the ultrasound transmission. However, most of the biological tissues are nearly incompressible [5], which means axial compression will cause tissue expanding in lateral direction which is perpendicular to the beamlines but parallel to the 2-D linear probe. So the imaging of displacement and strain in lateral direction can provide important additional information.

The accuracy of ultrasonic data in lateral direction is much lower than in the axial direction due to the lower sampling rate. Hence, most of the lateral displacement estimation methods use interpolation to get highly accurate results. An iterative method based on weighted linear interpolation was proposed in [6]. In this method, the original beamlines were interpolated before lateral displacement estimation so as to obtain more accurate lateral estimates. However, their interpolation operation results in enormous computational load and data space requirement.

In this paper, a two-step method was proposed to track continuous lateral displacement rapidly. Two-dimensional SSD of windows in pre- and post-compression data frames are utilized to calculate integral level displacement. Then the accurate lateral displacements are estimated by finding the minimum of the sum squared error (SSE) of the lateral data segments presented by cubic spline. Simulations and phantom experiments are performed to validate the estimates.

EXPERIMENTAL SECTION

2-D coarse displacement estimation

Our method begins with a block matching algorithm at coarse scale. Consider a pair of ultrasound data frames, representing sampled ultrasound echoes before and after the compression applied. A reference kernel $s_1[i,j]$ of size J × K within the pre-compression frame is processed with a comparison window $s_2[i,j]$ of size M × N within the post-compression frame to find where the best match is. SSD was generally used as a preferred similarity measure function than normalized cross-correlation (NCC) because of its balance between performance and computational cost [7-8]. A smaller reference kernel is used in our method to track 2-D motion, i.e. J < M and K < N. The SSD matrix is calculated while the reference kernel is shifted through the comparison window. This 2-D matrix is defined as:

$$\begin{gathered} R_{SSD}(k,l)=\sum_{i=1}^{M}\sum_{j=1}^{N}\left(s_1[i,j]-s_2[i+k,j+l]\right)^2 . \\ k=0 \ldots M-J+1, l=0 \ldots N-K+1 \end{gathered} \tag{1}$$

In this matrix, the position of minimal 2-D SSD value indicates where the best match is, and also where our 2-D coarse displacement estimate is.

2-D recorrelation

The complicated 2-D in-plane motion of tissue under axial compression can result in dramatic descend of correlation between windows in pre- and post-compression frames, which is so-called decorrelation. To improve accuracy of estimation, several method has been proposed, including sligning ad stretching [3,9]. Realignment of reference kernel and its corresponding dataset with respect to each other is an important way to improve correlation. This can be done by translating the center of reference kernel inside the comparison window using integer displacement results from 2-D SSD.

Continuous sub-sample lateral displacement estimation

Our aim is to obtain directly continuous and accurate lateral displacement from lateral discrete ultrasonic data segments. Cubic spline is used to continuouslyrepresent the discrete segments. Cubic spline is chosen because of its balance between convenience and accuracy [10]. Several spline-based methods were introduced to perform time delay estimation (TDE) [11-12]. However, the performance of spline-based method used in ultrasound lateral displacement estimation has not been studied thoroughly. Considering the lack of phase information and data accuracy, spline-based method is expected to be particularly well suited for ultrasound lateral displacement estimation.

Assuming g_1 and g_2 are two discretely sampled lateral ultrasonic signal segments (i.e. pre- and post-compression segments). They are denoted as:

$$g_1[n]=g_1(n\cdot\chi), \tag{2}$$

$$g_2[n]=g_2(n\cdot\chi), \tag{3}$$

where n is an integer, $g_1[n]$and $g_2[n]$ are with the same length K, and χ is the spatial sampling interval in lateral direction (i.e. the distance between axial beamlines). The pre-compression segment is represented as continuous function using cubic spline, which is:

$$\hat{g}_1(x)=\left\{\begin{array}{cc} \chi \le x \le 2\cdot\chi & f_1(x)=a_1x^3+b_1x^2+c_1x+d_1 \\ \vdots & \vdots \\ n\cdot\chi \le x \le (n+1)\cdot\chi & f_i(x)=a_nx^3+b_nx^2+c_nx+d_n \\ \vdots & \vdots \\ (K-1)\cdot\chi \le x \le K\cdot\chi & f_{N-1}(x)=a_{K-1}x^3+b_{K-1}x^2+c_{K-1}x+d_{K-1} \end{array}\right\}, \tag{4}$$

where $f_i(x)$ is the polynomial representation which is imposed to be continuous at joining points, including the function, its first derivative and its second derivative. The coefficients can be determined through several methods, including matrix pseudo inverses and infinite or finite impulse response filters. In this paper, spline function in MATLAB is used to calculate them.

In our method, SSE of the continuous reference segment and the discrete comparison segment is used to find where the best match is. The SSE of these segments can be represented as:

$$\varepsilon(x) = \sum_{n=1}^{K-1} \left(\hat{g}_1(n\chi + x) - g_2[n]\right)^2 . \tag{5}$$

A continuous SSE function can be obtained by substituting the cubic splines representation of g_1 into (5).The minimum of $\varepsilon(x)$ can be found at a value x' which is also our estimate of lateral motion between reference segment and comparison segment. In this paper, the $x'$is calculated simply by finding the zero point of the first derivative of $\varepsilon(x)$.

In traditional displacement estimation methods, the reference kernel is shifted through the comparison window to find the minimal value of a pattern matching function such as normalized cross-correlation, SAD, or SSD [11-12]. Usually, this is a process of high computational cost [13].However, the first step (i.e. 2-D SSD) of our method has low computational density. After aligning of the lateral segments, the second step (i.e. SSE) is needed to be performed only once to obtain the sub-sample and continuous result within the range of $-\chi$-$+\chi$. After the lateral displacement is estimated, the lateral strain is computed using a 1-D low-pass digital differentiator (SGDD) described by Luo in [14]:

$$y(n) = \sum_{k=1}^{M} \frac{3k}{(2M+1)(M+1)(M)} \left(x(n+k) - x(n-k)\right) . \tag{6}$$

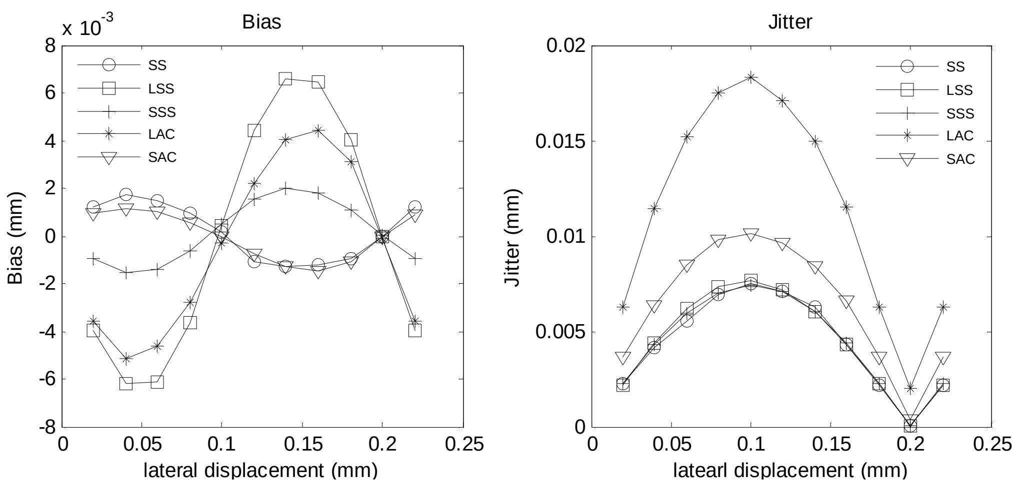

Fig. 1. Bias (left) and jitter (right) as a function of varying lateral displacement. The pitch is 0.2 mm. Ultrasound envelope data was used in five different methods. Lateral and axial 1-D kernel length in estimation is 2.4 and 2.5 mm, respectively

Lateral rigid motion simulation

To evaluate the performance of our method, a lateral rigid motion simulation is performed. The simulated phantom is displaced only laterally with respect to the probe. In each position, ultrasound data frame is generated by convolving random scatterers inside the phantom with complex point spread function (PSF) [15].The RF data is then converted to its complex analytic representation and demodulated into a baseband I/Q signal. The scan area is 40 mm deep and 40 mm wide. The total amount of scatterers is set to be 2×10^5. Based on the characteristics of the imaging ultrasound system Saset C21 (Saset medical Lt., Chengdu, Sichuan, China), which is used in experiments, a 5 MHz center frequency is assumed and a 40 MHz sampling frequency is used to digitize the RF signal. Total 200 beam lines are simulated. This resulted in a line spacing of 200 μm, which is also called as the pitch. The speed of sound in the medium is set at 1540 m/s. The phantom is displaced laterally at a sub-pitch step of 0.02 mm. This procedure is repeated 11 times until a maximum motion of 0.22 mm is reached.

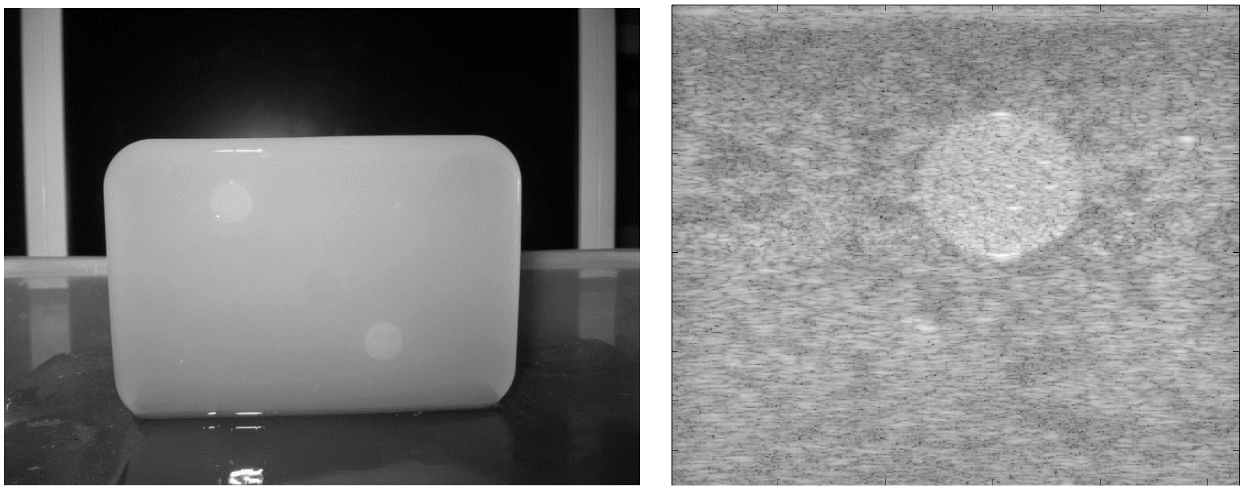

Fig. 2. 2-D experimental setup: (left) the photo of agar-based phantom. The surrounding area was colored by red ink to distinguish the harder inclusion more clearly, (right) a sample sonogram acquired by iMago C21 ultrasound system

Lateral displacement and strain estimation

In previously proposed 2-D displacement estimators, cross-correlation function was used as pattern-matching function between 1-D or 2-D kernels [3] and [16]. In 1-D axial kernel case, original signals were usually interpolated first. After the best match was found, sub-sample lateral displacement estimation was obtained by using 1-D cosine fitting or polynomial interpolation [2,3,6]. The cosine fitting was confirmed to outperform the polynomial interpolation. Hence, only cosine fitting is performed in this paper. These estimators which perform 16:1 linear and spline interpolation first on original envelope data are called as LAC and SAC hereafter, respectively. To evaluate the effects of different interpolation methods, proposed method, which is referred to as SS (SSD and Spline), were implemented using interpolated data also. The interpolated versions of proposed method are referred to as LSS and SSS, respectively.

In this paper, the performance of each estimator is considered in terms of its bias and jitter. Bias is defined as the mean of error of calculated lateral displacement and jitter is defined as the standard deviation of the estimation error [12].

RESULTS AND DISCUSSION

Simulation results

Figure 1 presents the results from different estimators in lateral rigid motion simulation. In case of lateral rigid motion, bias and jitter of lateral displacement estimates undergo periodic variations with a period equal to the pitch [13], which is 0.2 mm in our simulation. In Figure1, all the lateral displacement estimates are calculated using ultrasound envelope data. In left plot, it can be seen that SS, SSS, and SAC have nearly the same level of bias. And the spline interpolated version of proposed method has reverse magnitudes with respect to the other two. LAC shows larger biases, whereas LSS shows the worst. In right plot, all three versions of proposed method show almost the same and much smaller jitter than the axial 1-D methods. As we known in strain imaging, jitter has a much more significance than bias [13].

Experiment results

A number of experiments are also performed on an agar-based phantom. An inhomogeneous phantom [110 (axial) × 80 (lateral) × 30 (elevational) mm] made of gelatin (agar) was generated. The concentration of the agar in the surrounding regions is 1.0%. Two harder cylindrical inclusions both with a diameter of 10 mm and agar concentration of 3.8% are embedded at depth of 15 mm and 55 mm, respectively. The phantom is immersed in a tank full of water and placed on the bottom of the tank. A linear ultrasound probe is immersed in the water too, which is parallel to the phantom but had no immediate contact with the up boundary of the phantom. The phantom is compressed on both lateral boundaries with two plate compressors which are forced freehand.

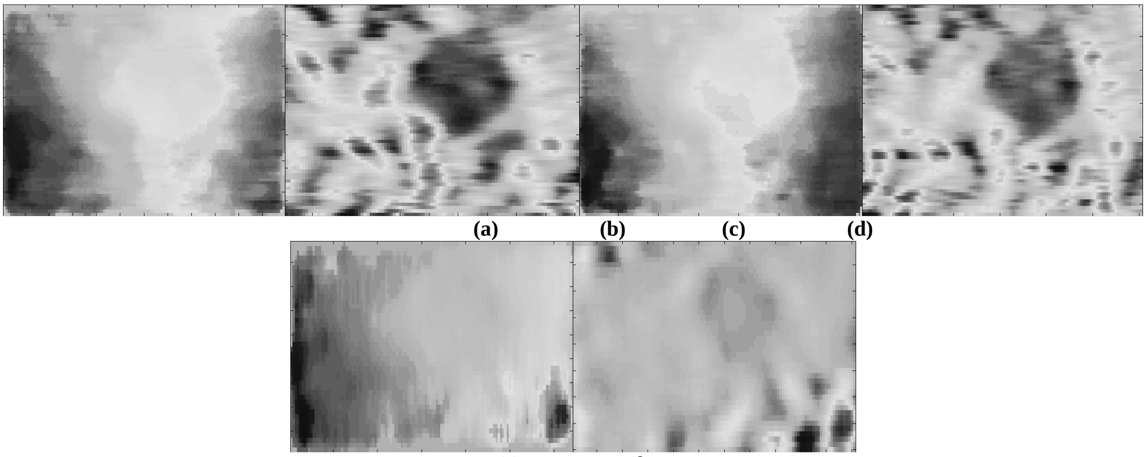

Fig. 3. Lateral displacement and strain image of part of the phantom. The size of imaging area is 3.0 mm * 3.8 mm.They are obtained using: (a) and (b) proposed method; (c) and (d) proposed method with 1:16 spline interpolation; (e) and (f) 1-D axial method with 1:16 linear interpolation. All results were calculated on envelope data.

Ultrasonic data are collected using a Saset iMago C21 ultrasound system with a 128-element SH7L38 linear array probe. 64 elements are used for transmit and receive, and a 7.5 MHz center frequency is used. A single transmit focus is set at 20 mm, and dynamic receive focusing is employed to acquire the signals. A frequency of 40 MHz is used to sample the ultrasound echoes for a depth of 30 mm. The data frames are composed of 264 lines for a width of 38 mm (i.e. a line spacing of 144 μm).The photo of the phantom and a sample of B-mode images acquired by the ultrasound system are shown in Figure2.

The estimated lateral displacements and strains are illustrated in Figure3. Because of the experimental error and electronic noise in the system, experimental results are noisier than those in simulation. The reference window size and comparison window size used in the first step of proposed method were 2.25 × 5.2 mm and 1.96 × 2.6 mm, respectively. Figure3 (a-d) show that the harder cylindrical inclusion is revealed clearly in the strain image obtained by proposed method. Furthermore, results show that the performance of the proposed method does not improve for the implementation of spline interpolation, which is in agreement with the simulation results reported in section 3.As illustrated in Figure3 (e-f), the results from classical 1-D axial method are highly corrupted by signal noise, resulting in erroneous values. Although results estimated by SAC are not presented here, they show a comparable pattern to LAC.

CONCLUSION

A two-step method for ultrasound lateral displacement and strain estimation is presented in this paper. Due to the lacking of phase information and data accuracy, methods using cross-correlation function result in poor performance. In simulations and experiments, proposed spline-based algorithm is shown to outperform the traditional ones in bias, jitter and image quality. It is shown that these improvements are accomplished with the absence of interpolation, which theoretically results in lower computing cost. The comparison of computational efficiencies of different methods is in our scope of future work. Moreover, simulations based on finite element method (FEM) and in *vivo* experiments will be performed too. Although this paper presents a preliminary work, we believe that this method has great potential because of its directly continuous and highly accurate results.

Acknowledgement
This work was supported in part by the Foundation of China and Natural Science Foundation of Sichuan Province (Grant No. 2013GZX0147-3).

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