

Spectral estimators in elastography

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Abstract

Like velocity, strain induces a time delay and a time scaling to the received signal. Elastography typically uses time delay techniques to indirectly (i.e. via the displacement estimate) measure tissue strain induced by an applied compression, and considers time scaling as a source of distortion. More recently, we have shown that the time scaling factor can also be spectrally estimated and used as a direct measure of strain. Strain causes a Doppler-like frequency shift and a change in bandwidth of the bandpass power spectrum of the echo signal. Two frequency shift strain estimators are described that have been proven to be more robust but less precise when compared to time delay estimators, both in simulations and experiments. The increased robustness is due to the insensitivity of the spectral techniques to phase decorrelation noise. In this paper we discuss and compare the theoretical and experimental findings obtained with traditional time delay estimators and with the newly proposed spectral methods. © 2000 Published by Elsevier Science B.V. All rights reserved.

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1. Introduction

Throughout the past 15 years, time-domain techniques have dominated the area of motion estimation in ultrasound, more specifically in blood flow and strain estimation. Both strain and velocity cause a time shift in the signal whose temporal and spatial derivatives yield strain and velocity, respectively. As a result, in blood flow estimation for example, methods for velocity estimation are divided into two distinct groups: (a) the Doppler frequency shift (direct approach); (b) the time/phase shift (indirect approach). The latter is the technique most widely used in the field of velocity estimation. A similar division also exists in elastography, i.e. strain can be estimated through the time shift [1] and frequency shift techniques [2,3]. Current estimators of tissue motion in elastography, such as time-domain cross-correlation-based tracking algorithms [1] and Fourier-based phase-tracking techniques [4] are coherent estimation techniques, i.e. they rely on the accurate estimation of tissue displacement afforded by considering changes in the phase of the signals due to compression. The coherent estimation techniques gen-

erally have the advantage of being highly precise but less robust, i.e. they are more sensitive to decorrelation noise. On the other hand, incoherent methods, such as time-delay estimation using the envelope of echo signals [5,6], are less precise but more robust.

Recently, we developed incoherent tissue strain estimators that estimate strain directly, i.e. without involving any time-domain estimation of displacement or noise-amplifying gradient calculations [3], using the power spectra of the signals. The main idea is based on the Fourier scaling property, which implies that a compression or expansion of the time-domain signal should lead to an expansion or compression of its power spectrum, respectively. Spectral estimators can be divided into two main groups: (a) the spectral shift methods; and (b) the spectral bandwidth methods [3]. For the purpose of this paper we focus primarily on the spectral shift methods; we estimate the relative shift in the spectral centroid caused by the compressive tissue strain using and comparing two distinct estimators: the centroid shift method [3] and the spectral cross-correlation method [7] in simulations and experiments.

2. Theory

In the signal model typically used in elastography the pre- and post-compression echo signals are given as

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follows:

$$r_1(z) = h(z) * e(z) + n_1(z) \quad (1)$$

$$r_2(z) = h(z) * e(az - z_0) + n_2(z) \quad (2)$$

where $*$ denotes convolution, z is a spatial variable, $r_1(z)$ and $r_2(z)$ are the received RF signals before and after compression, respectively, $h(z)$ is the impulse response of the ultrasound system or point-spread function (PSF) (assumed shift-invariant), $e(z)$ is the scattering function, a is the compression coefficient (or scaling factor) linked to strain through Eq. (3), z_0 denotes the time shift and $n_1(z)$ and $n_2(z)$ are independent zero mean white noise sources [3]. In elastography the compression coefficient is related to the strain s via the following equation [8]:

$$a = \frac{1}{1-s}. \quad (3)$$

Assuming Gaussian modulated sine and cosine pulses for both the ultrasound PSF and the scattering function, we have recently derived a direct spectral strain estimator [3] that estimates the strain in Eq. (3) directly via

$$\hat{s} \cong A \left(\frac{\hat{f}_{c2} - \hat{f}_{c1}}{\hat{f}_{c1}} \right) \quad (4)$$

where \hat{s} is the strain estimate, \hat{f}_{c1} and \hat{f}_{c2} are the centroid estimates of the pre- and post-compression power spectra, respectively, and A is a constant that depends solely on the spatial frequencies and bandwidths of the PSF and scattering spectra [3]. So, the strain in the tissue can be determined via the relative centroid shift of the pre-compression centroid. Eq. (4) yields two methods of estimation: (a) the centroid estimator that estimates the pre- and post-compression centroids \hat{f}_{c1} and \hat{f}_{c2} separately and then calculates the difference to estimate the shift $\hat{f}_{c1} - \hat{f}_{c2}$ [3]; and (b) the spectral cross-correlation estimator that estimates the shift via cross-correlation of the spectra [7]. In Sections 3 and 4 the two estimators and their performance are described

and compared to the previously described RF [1] and envelope [5,6] cross-correlation estimators.

2.1. Analogy to the Doppler effect

An interesting observation through Eq. (4) is that strain has a similar effect on the bandpass echo power spectrum as tissue velocity, i.e. it produces a Doppler frequency shift and a change in the bandwidth. In the case of Doppler the time compression (or scaling) factor is

$$a = 1 - \frac{2|v|\cos\theta}{c} \quad (5)$$

where v is the velocity of the scatterers, θ is the angle of insonification relative to the flow direction, and c is the speed of sound [10]. The velocity is estimated via the well-known Doppler equation, i.e.

$$\frac{f_d}{f_0} \cong -\frac{2\cos\theta}{c}v \text{ or } v \cong A' \frac{f_d}{f_0} \quad (6)$$

where f_d is the Doppler frequency and A' is equal to $-c/(2\cos\theta)$. The analogy between the second part of Eq. (6) and Eq. (4) is evident. However, the two comparable effects are due to different physical effects. Strain causes scatterers to move closer together (or further apart), thereby changing the center frequency and bandwidth of the stationary post-compressed signal. A downshift is expected if tension (negative strain) occurs and an upshift if compression (positive strain) occurs. An example of spectral upshift in the power spectrum following a 1% applied strain is illustrated in Fig. 1. The Doppler effect, on the other hand, links frequency and bandwidth changes to the velocity of moving scatterers. Depending on the direction of the velocity vector, a Doppler up- or downshift occurs.

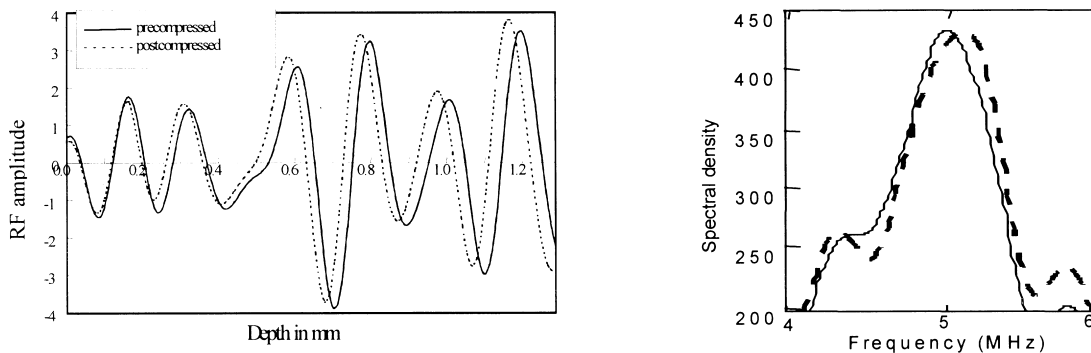


Fig. 1. The centroid shift due to strain: example of 1% tissue strain. (a) Pre- (solid) and post-compressed (dotted) gated simulated signals; (b) detail of the pre- and post-compressed power spectra. The post-compressed spectrum is upshifted by an amount proportional to the tissue strain incurred.

3. Simulation methods and results

Simulation results using a 1D scattering model [3] are used in this section to compare the performances of the centroid shift estimator [3] and the spectral cross-correlation estimator [7] with the standard time-domain methods of the precise RF cross-correlation [1] and the envelope cross-correlation [6]. Spectral strain estimation [following Eq. (4)] was performed using pre- and post-compressed power spectra of windowed RF signals. Fig. 2 presents the elastograms obtained using both these methods along with the ideal elastogram (i.e. true strain image) for three different amounts of applied compression. The estimators used were the RF cross-correlation [Fig. 2(b)] [1], the envelope cross-correlation [Fig. 2(c)] [5,6], the centroid method [Fig. 2(d)] [3] and the spectral cross-correlation method [Fig. 2(e)] [7]. Note that at the low strain value of 1% [Fig. 2(i)] all estimators perform with high elastographic signal-to-noise ratio (SNR_e) with the RF cross-correlation algorithm providing the closest correspondence to the ideal elastogram. On the other hand, for larger applied strains [5% and 10%, Fig. 2(ii) and (iii)] the time-domain cross-correlation algorithms fail to accurately estimate tissue strain due to the increased signal decorrel-

ation errors [Fig. 2(iib) and (iiib), respectively]. On the other hand, the good quality elastograms generated using the spectral method at 5% and 10% compression [Fig. 2(ii/iiid) and (ii/iiie), respectively] depict the robustness of the spectral methods while the highest signal-to-noise ratio is obtained using the spectral cross-correlation estimator.

4. Experimental methods and results

In simulations we tested the robustness of the estimators to decorrelation noise due to 1D motion. In the experimental setting, we test the same property in the presence of decorrelation noise due to 3D motion. In order to generate the experimental results, a gelatin phantom ($90 \times 90 \times 90 \text{ mm}^3$) containing a cylindrical inclusion with a 20 mm diameter, positioned at the center of the phantom and three times stiffer than the background was used. The ultrasound system and parameters used for acquiring the data were the same as those used in Ref. [3]. Comparison of the estimation performance using RF and envelope cross-correlation techniques, the centroid strain estimator and spectral cross-correlation method is illustrated qualitatively using

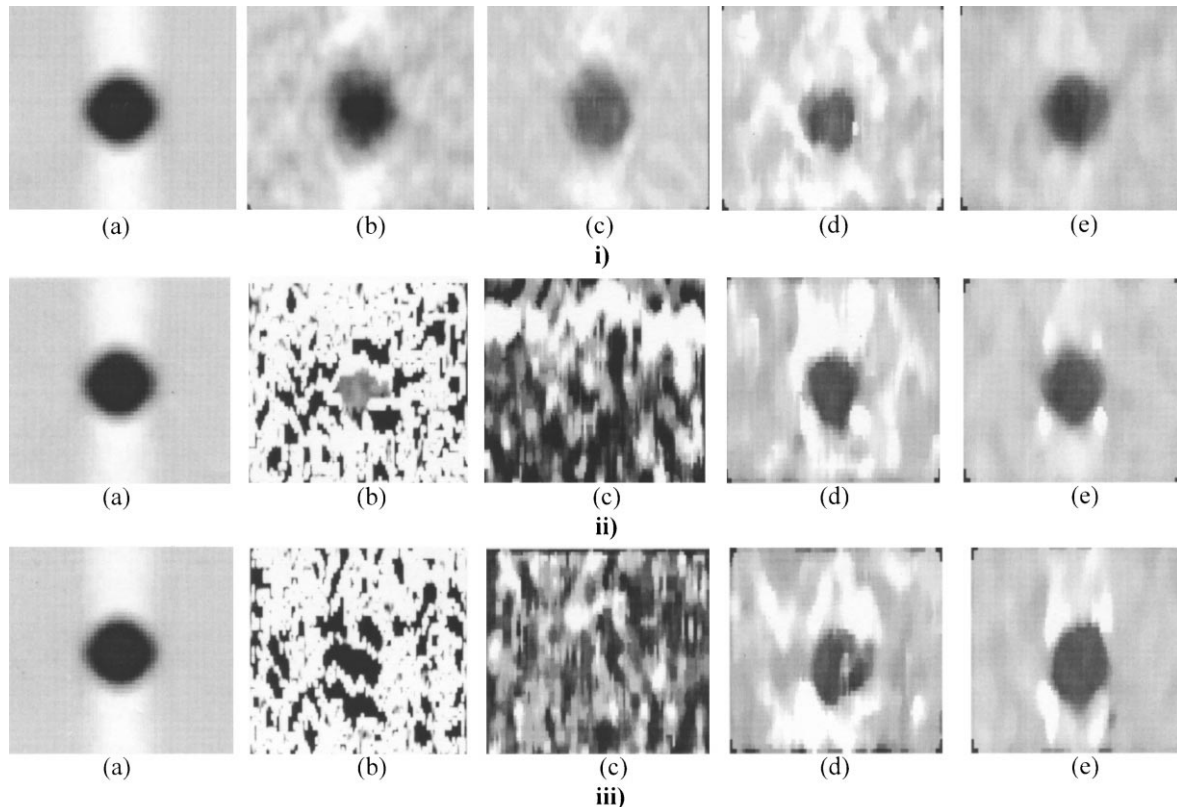


Fig. 2. Elastograms obtained using the 1D model for scatterer motion using the FEA simulation model for (i) 1%, (ii) 5% and (iii) 10% applied compression. The simulated phantom contains an inclusion that is three times stiffer than the background. (a) Ideal elastogram, (b) elastogram obtained using the RF cross-correlation method, (c) elastogram obtained using the envelope cross-correlation method, (d) elastogram obtained using the centroid method, (e) elastogram obtained using the spectral cross-correlation method.

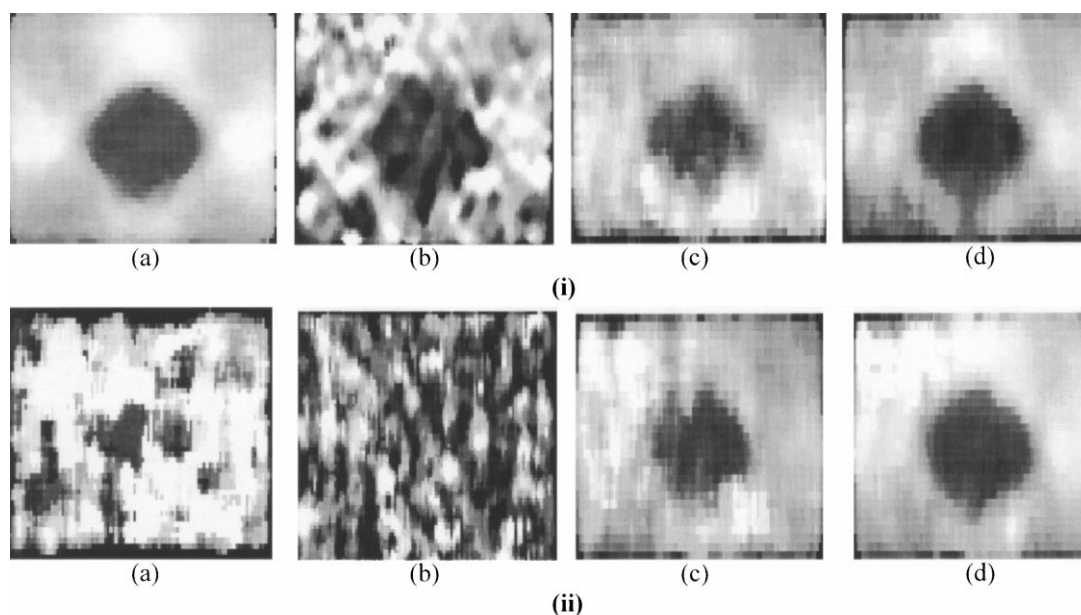


Fig. 3. Elastograms obtained using coherent and incoherent strain estimation algorithms, for a phantom with an inclusion that is three times stiffer than the background. The RF signals are obtained using a 7.5 MHz linear array with a 50% bandwidth. The elastograms are obtained using (i) 0.5% and (ii) 3% applied compression. (a) Elastogram obtained using RF cross-correlation method, (b) elastogram obtained using the envelope cross-correlation method, (c) elastogram obtained using centroid method, (d) elastogram obtained using the spectral cross-correlation method.

elastograms obtained at both low (0.5%) and high (3%) applied strains in Fig. 3. Note that coherent strain estimation provides the elastogram with the highest SNR_e for the low compression of 0.5% [Fig. 3(i)], when compared to the envelope and spectral methods. However, for the large applied compression of 3% when decorrelation noise due to 3D motion is higher [Fig. 3(ii)], the RF as well as envelope strain estimators fail completely when compared to the spectral methods, which produce quality elastograms. So, although the envelope is typically more robust than the RF cross-correlation method [11], the results of Figs. 2 and 3 clearly show that the spectral approach is significantly more robust than the envelope cross-correlation method. Furthermore, the spectral method yields similar results in both cases of low and high applied strain, demonstrating its robustness in a 3D scenario as well. Note that in the experimental setting spectral cross-correlation also performs better than the centroid estimator. These results are comparable to what has been obtained with the recorrelation method [9] and may indicate, despite the lower precision, a more robust method of estimating axial strain.

5. Conclusions

The new concept described here is based on the direct estimation of tissue strain from the relative frequency shift in the power spectrum. The centroid shift and spectral cross-correlation estimators estimate the relative

centroid shift in the power spectrum resulting from the applied compression. These estimators have three major characteristics: they are (a) direct and (b) spectral strain estimators, i.e. operate in the frequency domain. The direct strain estimation assures that no noise is added through the use of gradient operators, as is the case in time-delay based elastographic techniques. The spectral characteristic makes the spectral methods more robust, since they are phase independent and therefore suffer less from motion-induced decorrelation noise, as shown both in a 1D and an experimental 3D setting. The spectral cross-correlation achieves a higher signal-to-noise ratio than the centroid method and this issue has also been investigated theoretically [7]. Spectral strain estimation has been shown particularly useful for obtaining good elastograms in noisy jitter environments produced by unpredictable tissue and/or system motion [3]. In addition, the incoherent spectral methods were shown to be more robust than the previously described incoherent time-domain method of envelope cross-correlation, both in simulations and experiments. Due to their robustness, potential useful applications of spectral strain estimation are in hand-held, intravascular [12] and abdominal elastography. Furthermore, the time-domain and frequency-domain estimators are complementary and can be combined into a dynamic range expansion method for simultaneous low and high strain estimation [5]. Finally, the estimation of strain in the spectral domain may be proven to be essential for compensation in backscattering and attenuation measurements, where strain is typically a noise source [13].

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