ON THE USE OF ENVELOPE AND RF SIGNAL DECORRELATION AS TISSUE STRAIN ESTIMATORS

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Abstract—Bamber and Bush (1995) used the correlation coefficient for freehand elasticity imaging. Varghese and Ophir (1996) found it to be a biased estimator of strain with a large variability. In this study, we systematically investigate the effect of changes in various system and processing parameters on the performance of the correlation coefficient strain estimator, and demonstrate, using simulated data, that noise and frequency-dependent attenuation can introduce variable bias in this estimator. © 1997 World Federation for Ultrasound in Medicine & Biology.

Key Words: Bias, Decorrelation, Elastography, Frequency-dependent attenuation, Imaging, Precision, PSF, SNR, Strain, Ultrasound, Window size.

INTRODUCTION

Ultrasonic imaging based on tissue elasticity has been investigated for diagnosis of disease over the past several years (Alam et al. 1994; Krouskop et al. 1987; Lerner and Parker 1987; Lerner et al. 1990; O’Donnell et al. 1994; Ophir et al. 1991; Yamakoshi et al. 1990). Magnetic resonance imaging has also been used (Fowlkes et al. 1995; Muthupillai et al. 1995; Plewes et al. 1995). Ultrasonic techniques for the estimation of tissue elasticity are based on the estimation of strain due to external compression of the tissues (O’Donnell et al. 1994; Ophir et al. 1991). The local tissue displacements are estimated from the time delays of gated pre- and postcompression echo signals, which are then used to estimate the axial strain. In elastography (Ophir et al. 1991), time delays are estimated from the location of the peak of the cross-correlation function between windowed pre- and post-compression echo signals.

The tissue needs to be compressed to generate strain. The compression also introduces decorrelation in the RF echoes (Céspedes and Ophir 1993), which degrades the performance of delay and strain estimators. However, decorrelation itself has been used to estimate delay and/or strain. Dickinson and Hill (1982) studied motion estimation from the correlation coefficient between successive demodulated A-lines, but no rigorous theoretical basis was given. The same group later used the correlation coefficient to study liver tissue movement in response to cardiac cycles (Tristam et al. 1986, 1988). Dickinson (1980) conducted a thorough investigation of the performance, potentials and limitations of using the correlation coefficient for estimation of tissue motion using theory, simulations and experiments. Akiyama (1988) also used the magnitude of the cross-correlation coefficient to estimate motion. More recently, Bamber and Bush (1995) used the 2-D correlation coefficient of the envelope signals to compute and display elastograms for elasticity imaging. Although the displayed images were called “displacement images,” they have the typical characteristics of elastograms and not of displacement images due to tissue compression, and clearly show the hard inclusion in the tissue. However, the authors have acknowledged that the displayed images demonstrate strong dependence on strain (Bamber, personal communication 1997). Varghese and Ophir (1996) derived a simple theoretical expression showing that the strain can be estimated from the RF signal correlation coefficient at low strains. They presented some simulation results to show the utility of the correlation coefficient for the estimation of strain, and found it to be a biased estimator with large variability.

Despite the work on this strain estimator by various researchers, no published work has systematically characterized the performance of this estimator in various...
operating conditions. We have attempted to fill that void in this study. We investigated the effect of variation in the center frequency, bandwidth, signal-to-noise ratio (SNR) and the correlation window size on the correlation coefficient using controlled simulations. The simulation results support the conclusion by Varghese and Ophir (1996) that the correlation coefficient is a biased estimator of strain with large variability. We showed that certain phenomena, like frequency-dependent attenuation, can add additional undesirable variability to this estimator by changing the point-spread function (PSF) and the SNR.

THE CORRELATION COEFFICIENT AS A STRAIN ESTIMATOR

In elastography, tissue strains due to externally applied compression are estimated from the axial gradient of the time-delay estimates between the pre- and postcompression RF echo signals. Because the calculation of the correlation coefficient is simpler, the computation time could be significantly reduced if it could produce good quality estimates of strain.

Varghese and Ophir (1996) have shown that for pre- and postcompression signals expressed as follows:

\[ r_1(t) = s_1(t) + n_1(t) = s(t) \ast p(t) + n_1(t) \]  

and

\[ r_2(t) = s_2(t) + n_2(t) = s(t) \ast p(t) + n_2(t) \]  

the strain \( \varepsilon \) can be estimated for small strains using the following expression

\[ \varepsilon \approx 1 - \rho_{12}. \]  

In eqn (2), \( \rho_{12} \) is the correlation coefficient between the pre- and postcompression echo signals. We will refer to the parameter \( (1 - \rho_{12}) \) as the decorrelation coefficient for the rest of this paper. In eqn (1), \( s(t) \) is the 1-D scattering distribution of the elastic target, \( p(t) \) is the impulse response of the ultrasonic system, \( n_1(t) \) and \( n_2(t) \) are uncorrelated renditions of random noise and \( \ast \) denotes convolution. The constant

\[ \alpha = 1 - \varepsilon \]  

is close to unity, because the applied strain \( \varepsilon \) is generally small in elastography (\( \varepsilon \leq 0.01 \)) (Ophir et al. 1991). Simulation results have shown that the RF signal decorrelation coefficient is a biased estimate with large variability, even at low strains (Varghese and Ophir et al. 1996). Bamber and Bush (1995) used the correlation coefficient of the signal envelope to generate elastograms. The correlation coefficient, however, may vary due to changes in the center frequency, bandwidth, SNR and window size, even if the strain is held constant. We attempt to address these issues in this paper.

METHOD

We performed a 1-D simulation using MATLAB® (MathWorks, Inc., Natick, MA) to investigate the variability of the correlation coefficient with respect to the parameters under investigation. We did not perform a more involved 2-D or 3-D simulation, because 1-D simulation provides the best-case scenario for the estimator performance because the effects due to lateral and elevational movements of the scatterers are ignored. We simulated a line of 512 uniformly spaced random amplitude (Gaussian-distributed) irresolvable scatterers within a transducer beam. An average of 10 scatterers per wavelength were simulated. We simulated a Gaussian round-trip transfer function. We investigated the effect of only 1 parameter in every experiment. The other parameters were held constant at their default values (center frequency = 5 MHz, bandwidth = 60%, correlation window size = 1 mm, no additive noise). The RF A-line was computed by convolving the scatterer profile with the impulse response of the system. The A-lines were sampled at 50 MHz. The scatterer spacings were then linearly reduced to simulate tissue strain, and the A-line was recomputed. When investigating the effect of noise on the strain estimates, white noise was added to obtain a predetermined SNR value. After the pre- and postcompression A-lines were computed, we evaluated the correlation coefficient between the windowed pre- and postcompression signals. We repeated this procedure for 128 independent realizations and calculated the mean and standard deviation of the correlation coefficient for that set of parameters. We then changed the applied strain and repeated the simulation. Finally, we changed the parameters of interest and repeated the simulation experiment.

RESULTS

In Fig. 1a, the mean decorrelation coefficient for the RF signals is plotted vs. strain. Figure 1b shows the same graph magnified for strains \( \leq 1\% \). Default parameters were used. Obviously, the mean decorrelation coefficient is a monotonically increasing function of strain, behaving quite linearly at low strains. At strains below 0.5%, the strain and the mean decorrelation coefficient are also nearly equal. Varghese and Ophir (1996) have shown that the decorrelation coefficient and the strain are approximately equal at low strains. Thus, we have also
also exhibits a large variability that increases with increasing strain. However, compared to the RF, the envelope decorrelation coefficient appears to be a better strain estimator for large tissue strain. An adaptive method can be used to combine the RF and envelope processing similar to the one proposed by Cáspedes (1993) and later implemented by Varghese and Ophir (1997b) for conventional elastography. The smaller decorrelations at larger strains in the envelope compared to the RF supports the conclusions in Varghese and Ophir (1997b). In elastography, the applied strain is typically less than 1%. However, the tissue strain in an inhomogeneous tissue may span a wider range (typically $\leq 5\%$).

We then investigated the effect of change in the center frequency on the RF and envelope decorrelation

![Figure 1](image1.png)

**Fig. 1.** (a) RF decorrelation coefficient vs. strain. Parameters: $f_0 = 5$ MHz; bandwidth (BW) = 60%; window size = 1 mm; SNR = $\infty$. (b) Magnification of (a) at low strains.

plotted an "ideal" line on which the decorrelation coefficient equals strain. But, the standard deviation of the decorrelation coefficient is comparable to its mean at every strain level, and increases with increasing strain. Thus, a given decorrelation can be produced by a wide range of strains due to its large variability, resulting in poor precision.

In Fig. 2a, the decorrelation coefficient for the envelope of the echo signals is plotted vs. strain (default parameters used). Figure 2b shows the same graph magnified for strains $\leq 5\%$. Compared to the RF decorrelation coefficient, the mean envelope decorrelation coefficient remains linear for larger strains. At strains below 3%, the strain and the mean decorrelation coefficient are also nearly equal. The envelope decorrelation coefficient

![Figure 2](image2.png)

**Fig. 2.** (a) Envelope decorrelation coefficient vs. strain. Parameters: $f_0 = 5$ MHz; BW = 60%; window size = 1 mm; SNR = $\infty$. (b) Magnification of (a) at lower strains.
with increasing center frequency but, as expected, the trend was more significant in the RF. Thus, any variation in the center frequency during the propagation (due to frequency-dependent attenuation or other reasons) may introduce a variable bias and further ambiguities in determining the strain from the decorrelation coefficient. For example, for the simulation in Fig. 3a, with other parameters kept constant to their default values, an expected value of the RF decorrelation coefficient of 0.0159 may be due to 0.5% strain at a center frequency of 7.5 MHz, or 1% strain at a center frequency of 3.4 MHz, or another combination.

In Fig. 4a, we plotted the RF decorrelation coefficient vs. bandwidth for 0.5% and 1% strains. In Fig. 4b, the envelope decorrelation coefficient is plotted vs. bandwidth for 1% and 2% strains. Except for extremely
coefficient. The increase in the center frequency manifests itself as an increase in the rate of change of phase in the RF signal whereas, in the envelope, the speckles become finer containing higher frequencies. Because strain makes faster varying signals decorrelate quicker, variation in the center frequency is expected to introduce a larger variation in the RF decorrelation coefficient than in the envelope decorrelation coefficient. Figure 3a shows how the mean RF decorrelation coefficient varies with the change in center frequency at 0.5% and 1% strains. Figure 3b shows the effect of changing the center frequency on the mean envelope decorrelation coefficient at 1% and 2% strains. A constant bandwidth of 3 MHz was used. Both the mean RF and envelope decorrelation coefficient exhibit a monotonic increasing trend

Fig. 3. Effect of changing the center frequency. Parameters: BW = 3 MHz; window size = 1 mm; SNR = ∞. (a) RF signals; (b) envelope signals.

Fig. 4. Effect of changing the bandwidth. Parameters: $f_0 = 5$ MHz; window size = 1 mm; SNR = ∞. (a) RF signals; (b) envelope signals.
narrow bandwidth (20%), the RF and envelope decorrelation coefficients show little variation with respect to bandwidth.

Figures 5a and b show the variation in the RF and envelope decorrelation coefficient vs. SNR, respectively. The SNR is found to have a rather significant effect on both the RF and envelope decorrelation coefficients, especially at low SNRs. The variation in the correlation coefficient with respect to SNR is expected. Friemel (1994) and Céspedes et al. (1997) have shown that $\rho$ and SNR can be related using the relationship $\rho = \text{SNR}/(1 + \text{SNR})$. We have added a plot of $(1 - \rho)$ vs. SNR using this relationship in Fig. 5a ("theory", no strain present). It almost follows the graph for 0.1% strain, but has slightly lower decorrelation because there was no additional decorrelation due to strain. Higher strains introduce additional decorrelation as expected (Fig. 5a). It is nearly impossible to have any a priori knowledge about the SNR in the RF echoes, especially because it is depth-dependent. In any tissue, both the center frequency and the SNR decrease with depth due to various factors like frequency-dependent attenuation and beam spreading. So, their effects on the decorrelation coefficient partially offset each other. But, because it is very difficult to have any a priori knowledge about the center frequency and SNR at any depth, the uncertainty in the strain estimates obtained from the decorrelation coefficient remains.

Tissue compression results in decorrelation, which depends on strain (as well as center frequency and SNR). However, for any given strain, the decorrelation also depends on the correlation window size. This is due to the fact that displacements are depth-dependent in the presence of strain. We generally want a small window for better resolution. However, sometimes it is necessary to have larger windows for better noise performance. In Fig. 6a, we plotted the variation in the RF decorrelation coefficient with the change in correlation window size for 0.5% and 1% strains. Figure 6b shows the dependence of envelope decorrelation coefficient on the correlation window size for 1% and 2% strains. As expected, both the mean RF and envelope decorrelation coefficients appear to have significant dependence on the correlation window size. Although the correlation coefficient varies with window size, the latter is controlled by the operator. As long as the window size is incorporated in estimating the strain, it should not pose a problem.

**DISCUSSION AND CONCLUSIONS**

Bamber and Bush (1995) proposed to use the decorrelation coefficient of the envelope signals for elasticity imaging. Varghese and Ophir (1996) have demonstrated the decorrelation coefficient to have a poor precision as a strain estimator. In this study, we have attempted to characterize the performance of the decorrelation coefficient between the pre- and postcompression echo signals as a strain estimator, and have demonstrated, using simulated data, that changes in the center frequency and SNR introduce unknown variable bias.

As an ultrasonic pulse propagates through the tissue, both the PSF and the SNR vary due to frequency-dependent attenuation, changes in the beam, etc. To illustrate the problems due to the variability in the correlation coefficient, separately for changes in the center frequency and SNR, we take a hypothetical case where the center frequency of the transmitted pulse is 5 MHz. Let the near-field SNR be 40 dB and the attenuation coefficient for the tissue be 0.56 dB/(cm-MHz). Assum-
impossible to precisely determine the center frequency and SNR.

For any estimator, it is important to know its precision and bias, if any. The computer simulations of Varghese and Ophir (1996) showed that the correlation coefficient estimation has large variability—signifying low estimator precision. They have also shown this estimator to be biased but, when the bias is known, it can be corrected for. The additional uncertainties reported in this paper suggest that, due to variation in the parameters such as center frequency and SNR, when the decorrelation coefficient is used to estimate the strain, the strain estimates may have unknown bias and low precision.

To compare the precision of the decorrelation coefficient estimator, we performed some simple calculations. For the RF correlation coefficient, the elastographic signal-to-noise ratio (SNR) is 2.9 and 3.3, at 0.5% and 1% strains, respectively. For the envelope they are 1.3 and 1.2, respectively. For the conventional displacement gradient strain estimator, using the strain filter formulation (Varghese and Ophir 1997b) they are 2.5 and 2.8, respectively. These are comparable numbers; however, no variable bias is associated with the conventional gradient estimator, and various techniques have been proposed to improve the SNR (Alam and Ophir 1997; Céspedes and Ophir 1993).

Large variability in the decorrelation coefficient can also pose an additional problem. Because the post-compression RF line is the echo from the tissue that has been compressed, the prior estimates of the strain are required to place the correlation window on the pre- and post-compression RF lines at identical tissue locations. Because the strain estimates using the decorrelation coefficient have low precision, the correlation windows may not be precisely at the same location for the pre- and postcompression signals, resulting in inaccurate computation. In our investigation, we rid ourselves of this variability by using only the RF line segments closest to the transducer. An alternative method may be an average correction based on the applied compression (Bamber and Bush 1995).

When a tissue is compressed, tissue also moves in the lateral and elevational directions. Any movement of the tissue within the beam in either direction will introduce additional drop in the correlation. Such decorrelation may increase the errors in the strain computation, because all decorrelation is attributed to strain in this estimator. We did not address this issue in this study.

Because of its simplicity, this estimator may be a valuable tool for freehand elasticity imaging. But, the disadvantages need to be recognized and care should be taken in using this estimator.
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