PHASE ABERRATION CORRECTION FOR ULTRASOUND IMAGING IN TEMPORAL-SPATIAL FREQUENCY DOMAIN

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ABSTRACT

Phase aberration due to tissues with inhomogeneous acoustic speeds is a major source for image degradation in medical ultrasound. In most phased array pulse-echo ultrasound systems, the delay used to steer and focus the beams are calculated assuming constant speed. In practice, however, the acoustic speed varies for different types of tissue. In this paper, we present a method to estimate the phase errors between the elements of a linear array, based on signal representation in the spatial-temporal Fourier domain. Compared to the standard cross-correlation methods used for time delay estimation, the proposed technique shows better performance.

1. INTRODUCTION

The correction of phase errors due to propagation in inhomogeneous media has been the focus of a lot of research in medical ultrasound. These phase errors affect the quality of the images. In particular, the point spread function is broadened which lowers the resolution and blurs the image. In coherent imaging systems using linear arrays, beamforming is used to generate a single line (an A-line) in the image from each two-dimensional snapshot (the two dimensions being time and distance along the array axis). The first step in the beamforming procedure consists of aligning the radio frequency (rf) signals received by the elements of the array, by compensating for the geometric delays. The envelope of the linear summation of the geometrically aligned signals is logarithmically compressed to give an A-line. A constant speed of 1540 m/s is usually assumed when calculating the geometric delays used to steer the ultrasound beam during transmission and to focus during the reception. It has been shown that the acoustic speed ranges from 1430 m/s in fat up to 1665 m/s for collagen [1]. As a result, the signals at the different array elements are not aligned after the geometric delay compensation.

Several methods have been proposed to solve this problem. In [2], O'Donnell et al proposed an iterative method to measure the time delay, by calculating the cross-correlation between the signals received by adjacent array elements. A computationally efficient version of this method was proposed in [3], in which the delays are estimated by minimizing the sum of absolute differences for pairs of adjacent elements. Trahey et al. [4], [5] also suggested a method that would adaptively maximize the speckle brightness by adding or subtracting small amounts of delays in each iteration. A frequency domain method for narrowband signals was presented in [6], which requires the complete two-dimensional scan to be acquired before estimating the time delay profile and also involves matrix inversions. These methods are generally too slow to be implemented in real time. Due to the limited success of these methods, ultrasound imaging systems used in practice have yet to implement any phase aberration correction algorithm.

In this paper, we present a different approach to the phase aberration problem, based on signal representation in spatial-temporal frequency domain. The phase errors are estimated from the two-dimensional spectrum and used to align the received signals. The main computational load for the proposed method is the calculation of Fast Fourier Transforms (FFT) of the received signals, which can be implemented in hardware.

2. PROBLEM FORMULATION

The geometric delay compensation can be seen as mapping the received circular wavefronts into plane waves orthogonal to the array axis. Figure 1-(a) and 1-(b) show a wideband pulse with and without phase aberration. The corresponding signals after geometric delay compensation are shown in Figure 1-(c) and 1-(d). It is to be noted that, signals arriving at different times are echos from different depths in the tissue. Therefore the phase delay profile changes with time (depth) and the signals are jointly non-stationary processes. Therefore, the signals from two adjacent elements in the array are windowed such that within the window, the signals can be considered jointly wide-sense stationary and the relative time delay is constant.

Let $x_i(t)$ and $x_{i-1}(t)$ be the windowed signals received by two adjacent elements in the array after correcting for the geometric delays. Assuming the phase aberration causes one of the signals to be a delayed version of the other, then we can write,

$$x_i(t) = x_{i-1}(t - \tau_i) + n_i(t), \tag{1}$$

where τ_i is the delay between the two signals, and $n_i(t)$ is the additive white Gaussian noise. The problem is to estimate the delay τ_i for $i = 2, \ldots, M$, where M is the number of elements in the array.

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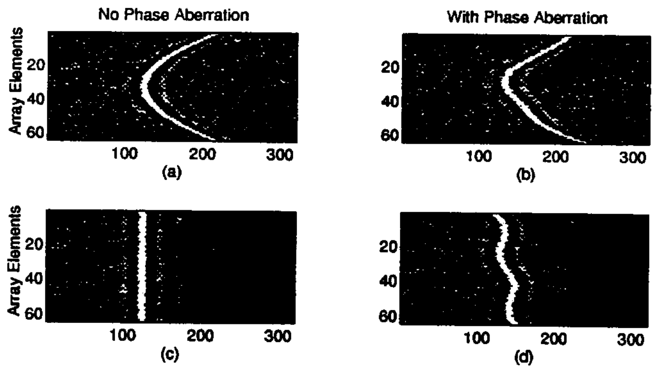

Figure 1: (a) A wideband pulse reflected from a point source, the horizontal axis is time (depth) and the vertical is the distance along the array, (b) same as (a) but in the presence of phase aberration. The corresponding signals after compensating for the geometric delays are shown in (c) and (d).

3. DELAY ESTIMATION

If we assume that the delay is changing linearly with the distance between the two elements, the signals arriving at the array can be seen as resulting from a plane wave with an angle of arrival θ_i from the array normal. The phase delay between the two elements is given by,

$$2\pi f \tau_i = 2\pi f \, \frac{D \sin\theta_i}{v} \tag{2}$$

or

$$\tau_i = \frac{D\sin\theta_i}{v}, \tag{3}$$

where v is the speed of propagation of the acoustic wave in the medium. The same delay can be written as a function of the spatial frequency along the array axis as νD. Therefore, the relationship between the spatial and temporal frequencies is [8],

$$\nu = f \, \frac{\sin\theta_i}{v}. \tag{4}$$

In order to estimate the delay τ_i, using Eqn. 3, the first step is to calculate $\sin\theta_i$ from Eqn. 4. Let $x(m,t)$ be a spatial-temporal two-dimensional variable of the signals received by the array, m being a discrete variable denoting the distance along the array (the position of elements). The two-dimensional frequency domain representation of $x(m,t)$ for $m = i-1, i$ is given by,

$$X_{i-1,i}(f,k) = \int_t \sum_{m=i-1}^{i} x(m,t) e^{-j2\pi km/2K} e^{-j2\pi ft} dt \qquad -K < k \le K, \tag{5}$$

where f is the temporal frequency, k is the discrete spatial frequency variable, and $2K$ is the number of points in the spatial Fourier transform. Omitting the subscripts $i, i-1$ for convenience, we can write,

$$\begin{aligned} X(f,k) &= \int_t \left[x_{i-1}(t)e^{-j\pi(i-1)k/K} + x_i(t)e^{-j\pi ik/K}\right] e^{-j2\pi ft} dt \\ &= e^{-j\pi(i-1)k/K} \int_t \left[x_i(t-\tau_i) + n_i(t)\right] e^{-j2\pi ft} dt \\ &\quad + e^{-j\pi ik/K} \int_t x_i(t) e^{-j2\pi ft} dt \\ &= e^{-j\pi(i-1)k/K} e^{-j2\pi f\tau_i} X_i(f) \\ &\quad + e^{-j\pi(i-1)k/K} N_i(f) + e^{-j\pi ik/K} X_i(f) \\ &= 2X_i(f) e^{-j\pi[ik/K + (f\tau_i - k/2K)]} \cos[\pi(f\tau_i - \frac{k}{2K})] \\ &\quad + e^{-j\pi(i-1)k/K} N_i(f), \end{aligned} \tag{6}$$

where $X_i(f)$ and $N_i(f)$ are the Fourier transforms of $x_i(t)$ and $n_i(t)$ respectively. The temporal spectrum is concentrated around the center frequency f_c and usually has a peak around f_c. Therefore, the peak value in the two-dimensional Fourier space lies on the line, representing a signal arriving at an angle θ, at or around f_c.

We will now show that the location of the maximum value in the 2D spectrum is an estimate of the time delay τ_i between the two signals. To find this maximum, we first select the one dimensional spatial spectrum $X(f_l, k)$ corresponding to a temporal frequency f_l such that $f_c - BW/2 \le f_l \le f_c + BW/2$, where BW is the bandwidth of the signals. In the following derivation, the noise term will be dropped, since it is a white Gaussian process with a flat spectrum which does not affect the peak location:

$$|X(f_l,k)|^2 = 4\,|X_i(f_l)|^2 \cos^2[\pi(f_l\tau_i - \frac{k}{2K})]. \tag{7}$$

Then we set the derivative of $|X(f_l,k)|^2$ with respect to k to zero,

$$\frac{d|X(f_l,k)|^2}{dk} = \frac{4\pi}{K}\,|X_i(f_l)|^2 \sin[2\pi(f_l\tau_i - \frac{k}{2K})] = 0. \tag{8}$$

Since $|X_i(f_l)| \ne 0$, then

$$\sin[2\pi(f_l\tau_i - \frac{k}{2K})] = 0 \tag{9}$$

or

$$2\pi(f_l\tau_i - \frac{k}{2K}) = n\pi, \qquad n = 0, \pm1, \pm2, \ldots \tag{10}$$

Therefore

$$\tau_i = \frac{1}{2f_l}\left(n + \frac{k}{K}\right), \qquad -K < k \le K. \tag{11}$$

which is periodic, with period $1/2f_l$. This is a direct result of assuming periodicity when applying the discrete Fourier Transform. Fortunately, in medical ultrasound applications, the delay between adjacent array elements is usually a small fraction of a period of the pulse. Hence, the time delay is simply given by,

$$\tau_i = \frac{k}{2f_l K}. \tag{12}$$

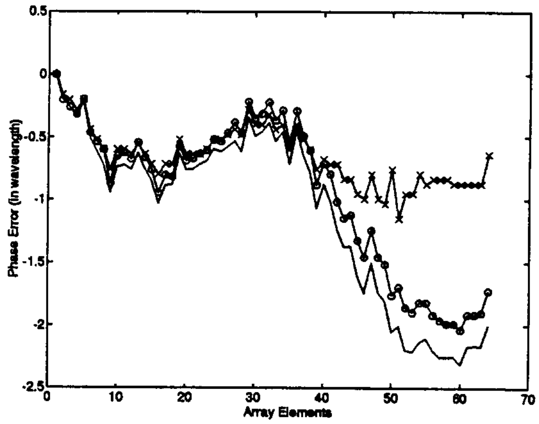

Figure 2: The phase error profiles; solid line is the true phase error, the line with 'o' is the estimates of the frequency method, and the line with 'x' is estimated profile using cross-correlation.

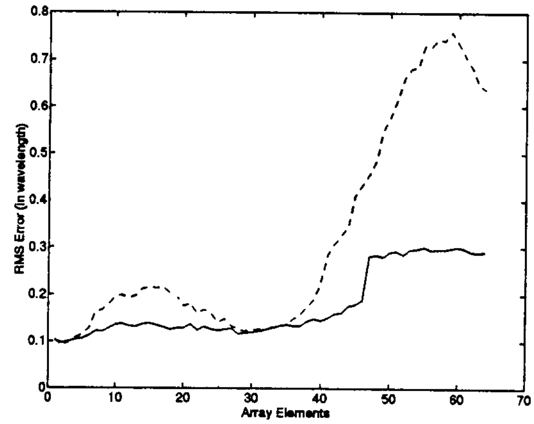

Figure 3: The rms error of the estimates of the frequency method and the cross correlation method as a function of the distance along the array.

To implement this method, the received signals are first windowed and digitized. The 2D FFT is then calculated for each pair of adjacent elements. The search for the peak value at the component corresponding to f_l is then performed. The frequency f_l is selected as that at which the peak of $|X_i(f)|$ occurs. This obviously depends on the characteristics of the transmitted pulse. However, due to wavefront distortion and attenuation as it propagates through tissue, or due to colored non-Gaussian noise, the peak of the 2D spectrum may deviate from its assumed value. In this case, repeating the procedure for different values of f_l in search for the peak provides more accurate results.

This method is equivalent, in concept, to a bandpass filtering operation followed by a periodogram estimator for signals received by each pair of elements. The bandpass filtering action is done by the selection of one or more components of the temporal Fourier transform. The search for the peak in the spatial Fourier domain is equivalent to applying periodogram.

4. SIMULATIONS AND EXPERIMENTAL RESULTS

Simulations were performed to compare the performance of the proposed method with the conventional cross-correlation method for time delay estimation. A wideband pulse with center frequency 2.75 MHz and a 50 % relative bandwidth was used. The array consisted of 64 elements with an inter-element distance of 0.279 mm. The sampling frequency was 48 MHz. The scatterer distribution was assumed to be Gaussian. The phase aberration was simulated by delaying the signals received with a delay profile consisting of three components; sinusoidal, linear, and random. Additive zero mean Gaussian measurement noise was also added to the wideband signals. The data were windowed using a sliding Hamming window. The window size corresponding to a distance of 5 mm in depth. The 2D FFT consisted of 64 points in the spatial direction and 256 points in the temporal direction. The results in Fig. 2 show that both the frequency method and cross-correlation estimates are very close to the actual phase error profile, when the rate of change of the phase profile is relatively low. When the rate of change increases, the frequency method is a more accurate estimate. Figure 3 shows the rms error calculated for 50 independent realizations of the phase profile of Fig. 2.

The experimental set-up was similar to that used in the simulations except that the number of elements was 48. The received rf signals were per channel data recorded from an ATL-Ultramark®8 scanner. In Fig 4-(a) and 4-(d) real ultrasound rf data from a female breast and the corresponding A-line (obtain by envelope detection of the coherent summation of the rf data) are shown. The corrected rf data and the resulting A-line after correction using the proposed method are shown in Fig. 4-(b) and 4-(e). Finally the results after correction using cross correlation are seen in Fig. 4-(c) and 4-(f). Figure 5 shows a partial image of a human abdomen before (Fig. 5-(a)), and after correction (Fig. 5-(b)); improvements in resolution and contrast can be clearly observed.

5. CONCLUSION

A phase aberration correction method is proposed for ultrasound imaging systems. The method has the advantages of being simple to implement and providing reliable estimates for the phase errors resulting from tissue inhomogeneities. Compared to the cross-correlation method used for time delay estimation, the proposed method is less sensitive to noise and sudden variations in the phase error profiles.

6. ACKNOWLEDGMENTS

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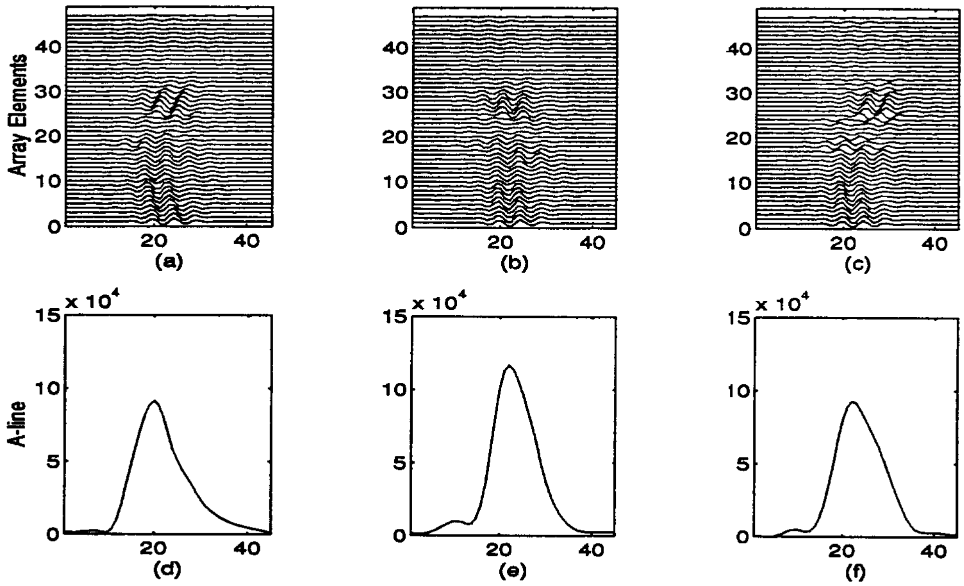

Figure 4: A comparison of real ultrasound rf data and A-lines obtained without correction (a) and (d), after correction using proposed method (b) and (e), and after correction using cross-correlation (c) and (f).

7. REFERENCES

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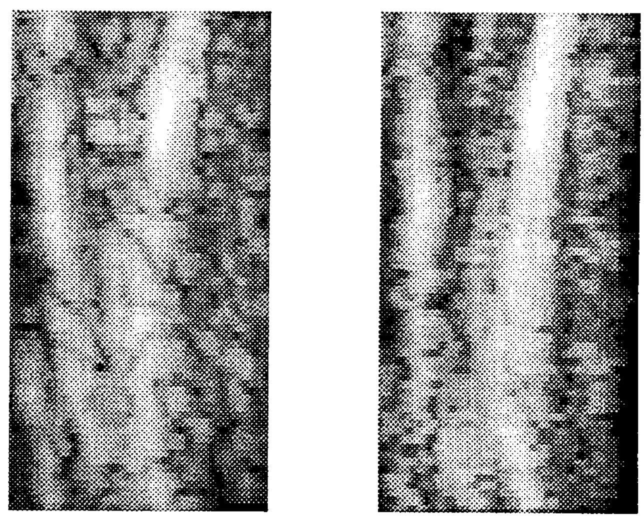

(a) (b)

Figure 5: Partial image of a human abdomen (a) before correction and (b) after correction.