Proceedings of the ASME 2007 Summer Bioengineering Conference (SBC2007)
June 20-24, Keystone Resort & Conference Center, Keystone, Colorado, USA

SBC2007-175317

ENTROPY BASED PHASE ABERRATION CORRECTION TECHNIQUE IN ULTRASOUND IMAGING

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INTRODUCTION

Ultrasound technology has been widely used in medical imaging. Techniques using phased array transducers use an array of transducer elements to transmit a focused beam into the body, and each element then becomes a receiver to collect the echoes. The received echoes from each element are dynamically focused to form an image. These systems assume a constant acoustic velocity in the tissue of 1540 m/s while steering and focusing the beam. However, soft tissues have a range of acoustic velocities that vary from 1470 m/s for fat to 1665 m/s for collagen [1]. The acoustic wavefront propagation through a region with locally different acoustic velocities will be phase shifted relative to the rest of the wavefront. This effect is known as phase aberration. The effects of phase aberration include broadening of the system point spread function that deteriorate the image resolution, and increasing the off-axis response leading to multiple images for the target [2].

The aberator can be modeled as a near field thin phase screen, consistent with the assumption that the layer of subcutaneous fat immediately at the face of the transducer causes the greatest degradation of the beamforming stage [3]. Different techniques have been proposed to correct the phase aberration problem. These techniques are based on measuring and correcting the phase error whether in the time domain or in the frequency domain [2, 4].

This paper describes a simple non-invasive method to correct the phase aberration resulted from the thin layer of subcutaneous fat immediately at the face of the transducer. The proposed technique searches for the best thickness of the fat layer, which achieves the maximum image quality. This algorithm assumes that the acoustic wave is passing through a uniform distributed fat layer with an acoustic velocity of 1470 m/s [5].

METHODS

A. Digital Beamforming

The beamforming technique used for image reconstruction is based on dividing the field of view into different point targets (raster points). These points are considered focal points which are separated laterally and axially by a small distance. Receive beamforming only is considered without loss of generality. That is, the beamforming is applied only on the received signals, so the focusing and steering are performed in the receiving mode using *N*-channel beamformer. The number of channels *N* is changed proportional to the depth of focus at each point for a fixed *f-number* equals two. The received signals are up-sampled to decrease the quantization to 10 ns. A time gain followed by Hilbert transform are applied to the radio frequency RF signals to decrease the attenuation effect and to calculate the signal envelop. The samples corresponding to each focal point are synchronized and added. Then a logarithmic compression is applied to decrease the dynamic range of the reconstructed image. The technique is used to reconstruct different images from different datasets collected by the SAFT technique [6].

B. Aberration Correction technique

In this work, we propose a simple method that based on modeling the subcutaneous fat layer as a uniform layer at the face of the transducer with a specified thickness F_L such as in figure 1. Then by determining the thickness of this layer the delays can be determined accurately. The time delay is divided into two components; the first component is due to the propagation of the ultrasonic waves in the fat layer, so that the delay is calculated based on the path length L_f and the velocity of the ultrasonic wave c_f in the fat layer. The second delay component is due to the propagation of the ultrasonic wave through the entire tissue, which is assumed to have the same velocity

c, and the delay is calculated based on the path length L_n and the velocity in this tissue, which many systems assumed it to be 1540 m/s. The total delay T_t is calculated using the equation below:

$$T_t = \frac{L_f}{c_f} + \frac{L_n}{c}. \quad (1)$$

The lengths of L_f and L_n can be determined from the geometry as shown in figure 1, and the length L_f can be determined as follow:

$$L_f = \frac{F_L * R_n(i,j)}{l_f}. \quad (2)$$

The determination of the thickness of the subcutaneous fat layer can be made qualitatively by the user or quantitatively by using a simple automatic algorithm. In the first method the user can change the thickness of the fat layer manually from 0 mm to 50 mm by a specific increment such as 1 mm, and the user can determine qualitatively the thickness that achieves the maximum image quality. The second automatically determines the best thickness of the fat layer that enhances the image quality and decreases the bad effects of the fat layer on the image such as the image blurring. In this research the value of entropy was used as the cost function. Classical entropy-based criteria describe information-related properties for an accurate representation of a given signal. The observations reveal that the value of the entropy decreases as the energy distribution within a region of interest (ROI) in the image becomes more precise and has sharper edges, *i.e.* as the effect of image blurring decreases as the value of the entropy decreases within a specified ROI within the image, or simply the ROI may be the point spread function (PSF). So this algorithm searches for the minimum value of the entropy for different thicknesses.

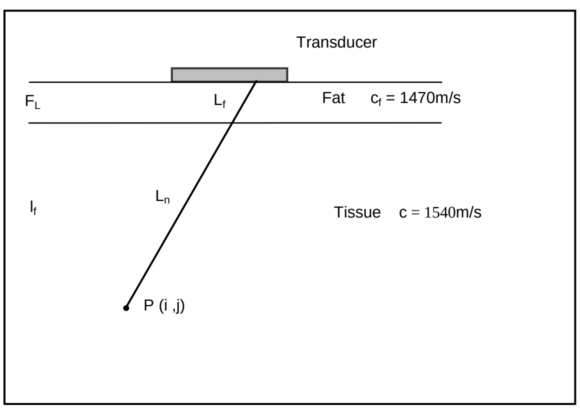

Figure 1. The path of the ultrasonic beam through the fat and tissue layers.

The best thickness for the fat layer will be at the minimum value of entropy. The type of entropy used in this algorithm was the (non-normalized) Shannon entropy, which can be calculated using the following equation:

$$E(s) = -\sum_i s_i^2 \log\left(s_i^2\right), \quad (3)$$

where s is the signal and $(s_i)_i$ the coefficients of s in an orthonormal basis.

RESULTS

The proposed technique for the aberration correction was tested at different fat layer thickness from 1 mm to 50 mm. Figure 2 shows the aberrated and the corrected PSFs at 3 cm simulated fat layer thickness. The corrected PSF has higher amplitude main loop, and it becomes less broaden than in the aberrated PSF, which improves the lateral resolution.

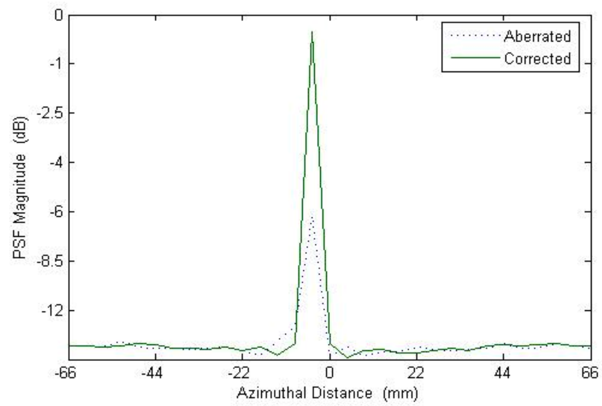

Figure 2. Aberrated and corrected 2D PSFs.

CONCLUSION

A fast and accurate correction scheme for the problem of phase aberration in ultrasound images that results from the subcutaneous fat layer is described. The new method relies on calculating the time delay taking into account the difference in the ultrasonic velocity between the fat and the tissue. This method assumes a planar model for the subcutaneous fat layer. The thickness of the fat layer is determined non-invasively using both manual method, and Entropy based automatic method. Experimental results from both simulations and real data acquired from two different ultrasound systems are used to verify the solution using images reconstructed from a beamforming algorithm applied on receiving mode based on raster points. The results indicate that a significant improvement using the proposed methods and suggest their practicality and clinical utility.

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