Ultrasound Beam Focusing Considering the Cutaneous Fat Layer Effects

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Received: September 7, 2012 **Accepted: December 28, 2012**

Published: January 8, 2013

Abstract: *Commercial medical ultrasound scanners assume average sound velocity of 1540 m/s while sound speed varies at different tissues. This assumption limits focusing quality and degrades contrast and resolution, particularly for patients with fatty abdominal wall. This paper presents a simple two layer model to demonstrate the effect of ultrasound beam focusing quality in inhomogeneous medium based on Huygens's principle. A time delay function for ultrasonic phase array has been derived using in vivo information of fat layer and considering refraction in the interface of two layers. Simulated beam pattern and corresponding beam profiles at the focal depth using conventional delay time and that for proposed two layer model are compared. An experimental setup was designed to assess the image quality using a commercial ultrasound scanner and a phantom of two layers with different sound velocity. Simulated and experimental results indicate that obtained images using time delays for two layer model show better contrast resolution.*

Keywords: *Ultrasound beam focusing, Fat layer effects, Sound speeds, Beamforming, Ultrasound imaging.*

Introduction

In ultrasound imaging, phased-array transducers are used for beam focusing and steering. Beam focusing is the major factor that determines the lateral spatial resolution in ultrasound imaging. In conventional ultrasound scanner, it is assumed that the ultrasound beam travels in a homogeneous media at a constant speed. However, human body consists of many different organs and tissues in which the sound velocity may change significantly. Non-uniform sound velocity makes aberration in the beam focusing, which may result in degradation of contrast and spatial resolution.

In abdomen imaging, the fat layer underneath the skin may have significant effects on beam focusing performance when the fat layer is very thick. Fat tissue has the average sound speed of 1460 m/s which is much slower than the average sound speed in the muscle, 1540 m/s. The difference in sound speed and the refraction on the fat surface introduce error in phase delay calculation for transmit and receive beam formation using phase array and hence make aberration at the focal point. Aberration leads to degradation of resolution, causes a spreading in the main lobe energy and gives rise to high isolated side lobes in the beam pattern and hence off axis response become significant [1, 2].

There are many methods to minimize the effect of aberration problem [3-13]. Most of the methods analyze the received echo signal in the pair of different elements or elements groups to measure the deviation from expected arrival times. These deviations are used to modify time delays for received beamformer. A cross-correlation based method for estimation of arrival time fluctuations was proposed by S. W. Flax and M. O'Donnel [3]. M. Fink described a time reversal mirror, which carries information about the pulse including arrival time and amplitude distortions. But this method requires a point reflector in the insonified medium [5]. A method based on speckle brightness is proposed by Nock et al. where the time delays are calculated corresponding to brightness maximization over a region of interest [6]. K. W. Rigby et al. presented some estimation and correction of time delays based on cross correlation and summing of echo signals in neighboring elements pairs [9-11]. S.-E. Måsøy et al. proposed aberration estimation algorithm to estimate arrival time and amplitude fluctuations with signals from random scatterers in both frequency domain [12] and time domain [13] and analyzed variance of the estimates. Most of the existing methods consider the propagation medium as homogeneous and neglect refraction of acoustics rays in the interface of different tissue layers. Furthermore there are very few research works on aberration correction using in vivo information. In order to consider refraction, information about fat layer thickness and sound velocity are necessary to calculate the time delay. So to calculate more accurate time delay to obtain better focus quality and hence better resolution in ultrasound imaging in inhomogeneous media using phase array, in vivo information about fat layer thickness and sound velocity in the medium should be incorporated in time delay calculations.

In this work, we evaluate the effect of fat layer on the beam focusing quality using two layer model of sound wave propagation. The time delay function is derived using in vivo information of fat layer thickness and velocity of sound propagation through the inhomogeneous medium. Simulation and experimental results using time delays based on both conventional and proposed method are presented.

Materials and methods

Effects of fat layer on beam focusing quality

We first investigated the effects of a cutaneous fat layer on the beam focusing quality. With the two layer model shown in Fig. 1, we calculated the beam profiles at the assumed focal depth. The sound speeds of the layer 1 (fat) and layer 2 (non-fatty tissues) were assumed to be 1460 m/s and 1540 m/s, respectively. On top of the fat layer, $(2n + 1)$ point sources are linearly placed with the interval of Δ.

Fig. 1 Two layer model with linear array of point sources

A wavefront propagates away from an exciting point source at the speed of sound in the propagating media and it does so equally in all directions. Therefore after a given time it has formed a spherical wavefront with a radius proportional to the time interval. Considering adiabatic condition the acoustics wave equation in the homogeneous medium with sound velocity c can be written as:

$$\nabla^2 p = \frac{1}{c^2}\frac{\partial^2 p}{\partial t^2} \tag{1}$$

The solution of wave equation for a point source in unbound homogeneous medium can be expressed as:

$$p = p_0 \frac{e^{i(\omega t - kr)}}{r} \tag{2}$$

where $\omega = 2\pi f$ and $k = \frac{\omega}{c} = \frac{2\pi}{\lambda}$. In the above equation, p_0 is the constant amplitude for all sources, r is the radial distance, f is the frequency, ω is the angular frequency, λ is the wavelength and k is the wave number. Amplitude decaying factor $\frac{1}{r}$ is applied in the above equation in order to satisfy the law of conservation of energy.

When a number of point sources excited simultaneously or in different instant of time, the spherical wavefronts generated from the sources may add together or cancelled out each other due to phase differences. Hence constructive or destructive interference occur as a result of superposition of wavefronts from different point sources. The resultant wavefront will be the algebraic sum of the individual waves which can be written as:

$$p = \sum_{j=1}^{N} p_0 \frac{e^{i(\omega(t-\tau_j) - kr_j)}}{r_j} \tag{3}$$

where t is the reference time, τ_j is delay time for j-th source, r_j is radial distance from j-th source and N is total number of point sources.

If the point sources are excited with delay τ_j starting from $t = 0$ then,

$$p = \sum_{j=1}^{N} p_0 \frac{e^{i(-\omega\tau_j - kr_j)}}{r_j} \tag{4}$$

Now according to Huygens's Principle, each point on that wavefront acts as a source of successive wavefronts, which propagate in the same fashion. Therefore in the interface of two layers propagating wavefront gives rise to secondary sources, each of which radiates wavefronts. These sources appear continuously at all positions in the interface and combined together to produce new wavefronts in accordance to their individual phase. The wavefronts in the second layer can be calculated as:

$$p = \sum_{q=1}^{M} |p_q| \frac{e^{i(\angle p_q - k_2 r_q)}}{r_q} \qquad (5)$$

where $|p_q|$ and $\angle p_q$ is the magnitude and angle of q-th secondary source pressure, r_q is radial distance from q-th source and M is total number of point sources and k_2 is the wave number for second layer.

The beam pattern in the imaging plane can be shown from these pressure distributions. To keep the continuity in pressure distribution throughout two layers a multiplication factor (m_f) equal to the ratio of energy in interface (secondary sources) to primary sources is introduced with r_q in the denominator.

$$p = \sum_{q=1}^{M} |p_q| \frac{e^{i(\angle p_q - k_2 r_q)}}{m_f r_q} \qquad (6)$$

where, normalizing factor $m_f = \frac{E_2}{E_1}$, energy on the transmitter surface: $E_1 = N \times |p_0|^2$, energy on the interface (secondary source): $E_2 = \sum_{q=1}^{M} |p_q|^2$.

Derivation of phase delay function to compensate the fat layer effects

Ultrasound beam focusing is alignment of propagating waves from point sources to a specific point at the same time. Focusing can be accomplished by delayed excitation of point sources. The delay time can be calculated from the geometrical relationship of acoustics ray path through medium of different velocity. The calculated delay times can be used for both transmit and receive beamforming to investigate the fat layer effects. Fig. 2 shows the geometrical relationship of acoustics ray path to a specific focal point.

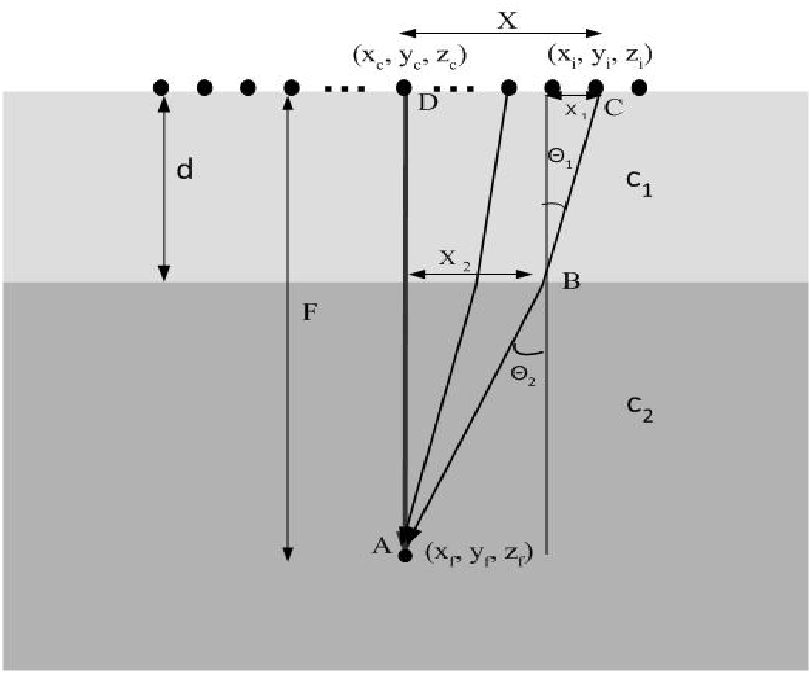

Fig. 2 Derivations of time delays for two layer model

The delay time for *i*-th point source compared to center point source is derived. The propagation times are:

$$t_{AD} = \frac{d}{c_1} + \frac{F-d}{c_2} \quad \text{and} \quad t_{ABC} = \frac{CB}{c_1} + \frac{BA}{c_2} \tag{7}$$

Therefore propagation delay between *i*-th point source and center point source to reach at focal point is the difference between above two propagation times. Referring to Fig. 3, path length *CB* and *BA* can be calculated as:

$$CB = \frac{d}{\cos\theta_1} \text{ and } BA = \frac{F-d}{\cos\theta_2} \tag{8}$$

So the problem is reduced to calculation of θ_1 and θ_2. From the geometry, we can write:

$$d\tan\theta_1 + (F-d)\tan\theta_2 = X \tag{9}$$

From Snell's law for refraction of non-perpendicular sound waves through medium of different velocity, we have,

$$d\tan(\sin^{-1}(\frac{c_1}{c_2}\sin\theta_2) + (F-d)\tan\theta_2 = X \tag{10}$$

Numerical methods are applied for solving above equation to obtain the values of θ_1 and θ_2 which are used to calculate the time delay between point sources. Fig. 3 shows the difference in propagation times as well as difference in excitation times for the point sources considering different thickness of fat layer in comparison with single layer.

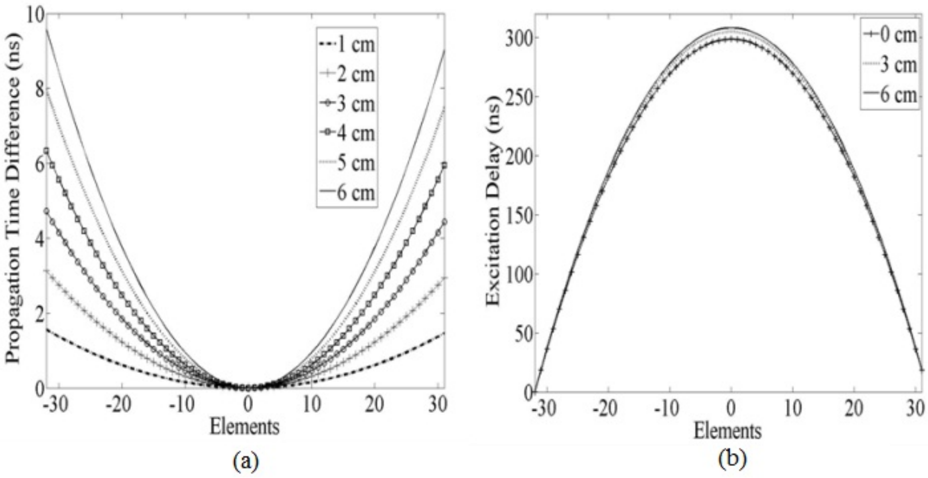

Fig. 3 (a) Elements-wise difference in propagation time from that of single layer case for different fat layer thickness, (b) Excitation delay for 64 elements with center to center spacing of 0.3 mm and focal depth of 10 cm along center line for different thickness of fat layer. Excitation starts from most distant element with respect to center element.

For steering case, the perpendicularly distant point source on the surface from the focal point is considered as reference point for time delay calculation.

Results and discussion

Simulation results

To analyze the effect of fat layer on the focus quality we simulated ultrasound beam pattern based on point source model of sound wave propagation. The beam patterns has been obtained by applying the conventional time delay and calculated time delay considering various fat layer thicknesses to excite point sources. In this simulation we considered an array of 256 point sources with center to center spacing of 0.3 mm and a focal depth of 10 cm. The beam profiles in the focal depth obtained using time delays considering single layer for different fat layer thickness are compared. Fig. 4 shows the simulated beam pattern in both single layer and double layer and beam profiles with fat layer thickness of 4 cm at the focal depth of 10 cm. It is observed that the beam profile obtained from the homogeneous model has a wider and weaker main lobe than that obtained from the two layer model. This is due to the shifting of focal point from desired location. As the fat thickness increases, the aberration also increases.

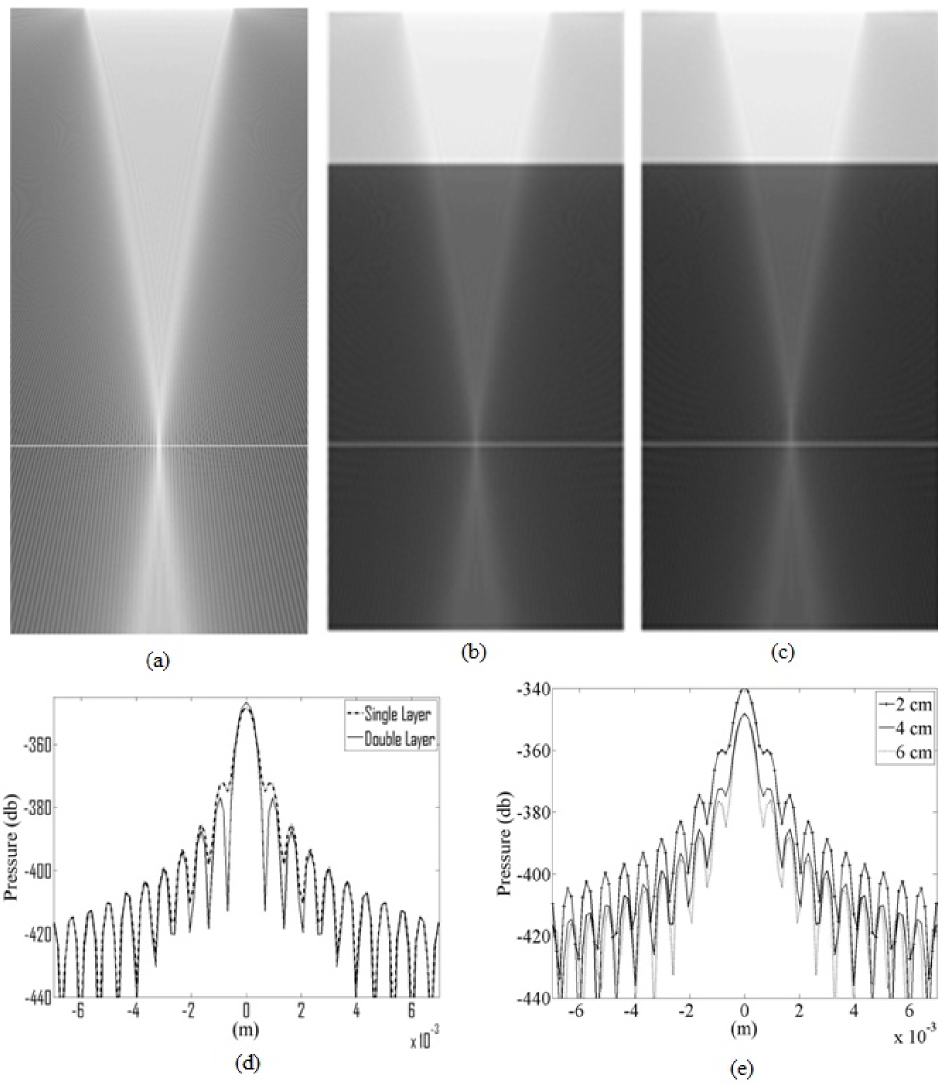

Fig. 4 (a) Simulated beam pattern for homogeneous medium, (b) Simulated beam pattern using time delays calculated from two layer model, (c) Simulated beam pattern using time delays calculated from single layer model, (d) Comparison of beam profiles along focal depth for single and two layer, and (e) Comparison of beam profiles along focal depth using single layer time delay with fat layer thickness of 2 cm, 4 cm, and 6 cm

Experimental results

We made a phantom using butter fat, agar, gelatin, and water to implement two layer model of medium with different sound speeds. The 1.7 cm thick upper layer was made of butter fat, agar, gelatin, and water to mimic fat layers and the 8.3 cm thick lower layer was made of agar, gelatin, and water to mimic non-fatty layers. To calculate the phase delays based on the exact sound speed of each layer, we measured the sound speeds of the layers at separate A-mode experiments. It has been found that the upper and lower layers have the sound speeds of 1645 m/s and 1695 m/s, respectively. To evaluate the lateral resolution, we inserted two nylon monofilaments with the diameter of 0.156 mm in parallel and 2 mm apart from each other in the lower layer as shown in Fig. 5. We used a commercial ultrasound scanner (GE Healthcare, LOGIQ P5) to take linear scan images. The ultrasound scanner has a 128-element linear phase array probe with the element-to-element interval of 0.3 mm, where 64 elements are active for beamforming.

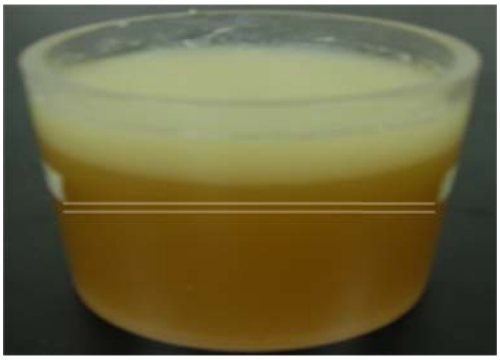

Fig. 5 Phantom of two layer model

We have formed three kinds of images from the RF echo data to compare the image quality. Among them, one directly from the ultrasound scanner which makes the image with the assumed sound speed of 1540 m/s, and two from the receive beam formation using the time delays calculated for the single layer with the sound speed of 1695 m/s and two layer models with the sound speed of 1645 m/s and 1695 m/s, are shown in Fig. 6. Fig. 6b shows the ROI images of nylon monofilament inserts for better comparison. In all cases, the scanner average sound speed of 1540 m/s is used for transmit beamforming. The image obtained after received beamforming with the sound speed of 1645 m/s and 1695 m/s in the two layer model, shows a little better contrast as well as lateral resolution. With the two-layer phantom which has small sound speed difference between the layers, the effect of taking account of two layers in calculating phase delays seems to be marginal. However, in human imaging where the sound speed of fatty tissue is much slower than the phantom, the effect is expected to be bigger than in the case of phantom imaging with conventional transmit beamforming using average sound speed. We are now trying to develop new phantoms which better mimics fatty tissues in terms of sound velocity.

The proposed model considered refraction of sound waves due to inhomogeneity and corresponding propagation paths for time delay calculation rather than consideration of straight beam path in most of the existing methods. Focusing has been improved for more accurate time delays calculated from two layer model but with extra computational burden. Additional studies are necessary to automatically estimate fat layer thickness and sound speeds from echo signals for clinical application of the proposed method.

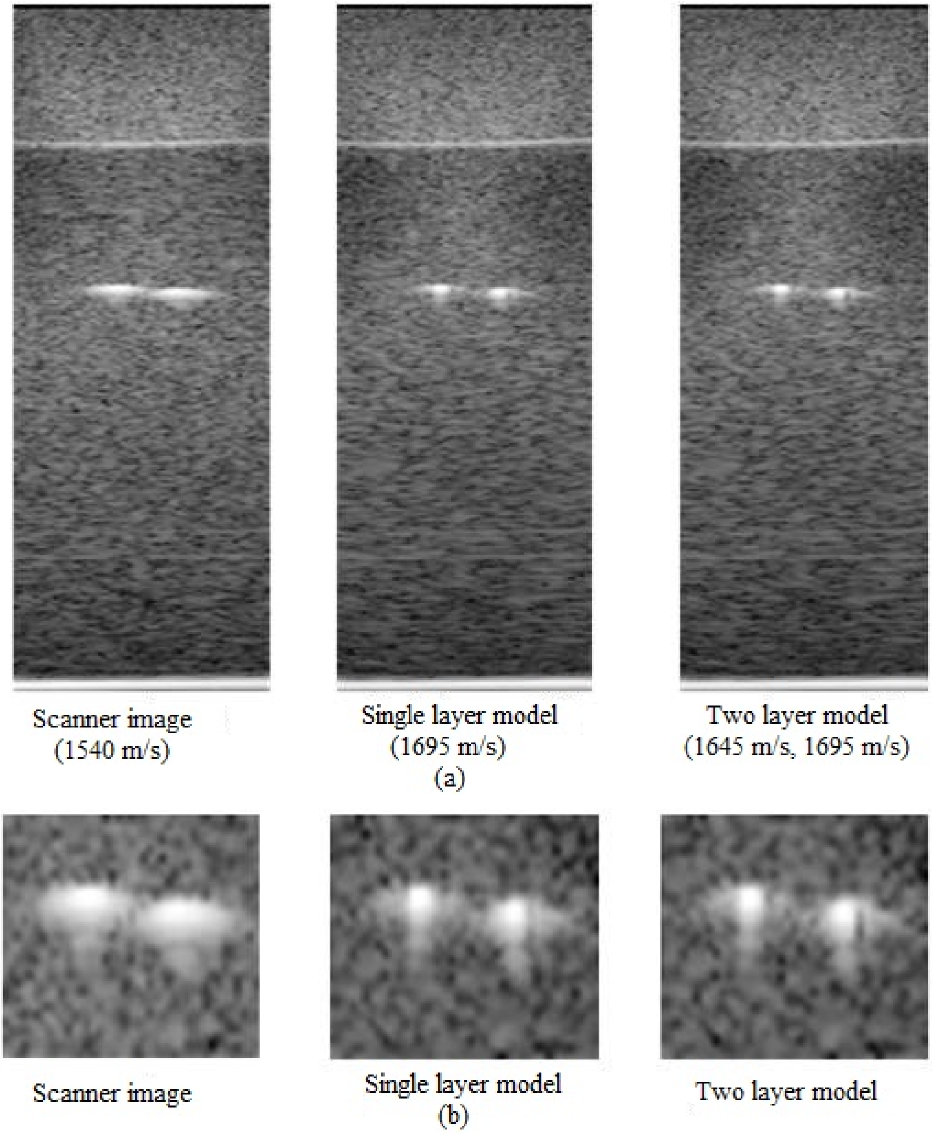

Fig. 6 (a) Images of the phantom for different models and
(b) ROI images of nylon monofilament inserts

Conclusion

Two different models of sound wave propagation through inhomogeneous media are presented to find the time delay function that can be used for beamforming in ultrasound imaging. Degradation in the beam focusing quality has been observed in the single layer model as the effect of fat layer. The time delay functions have been derived in the double layer model considering the refraction of acoustic waves on the interface of two layers as well as the different propagation speeds through medium. Both simulation and experimental results show improvements in image quality with finer beam focusing.

Acknowledgements

This study was supported by the grants from the National Research Foundation of Korea (2009-0078310) and the grant from the Ministry of Knowledge and Economy of Korea (10033627).

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