INTERACTION OF ULTRASOUND WITH BIOLOGICAL MATERIALS

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Abstract. *This paper deals with medical applications of focused ultrasound in minimally invasive treatment of a variety of disorders. The pressure distributions generated by plane as well as focused ultrasound beams are presented. Ultrasound imaging uses the transmission and reflection of high-frequency longitudinal, mechanical waves in tissues. Image forming information is provided by the differing degree to which the energy of such waves is reflected from the surfaces between different tissues; the reflection is due to differences in the mechanical properties of the tissues. Using the wave propagation speed in tissues, time of reflection information can be converted into distance of reflection information. Two interesting features of the interaction including the shape and size of the focal region as well as the spatial distribution of the pressure field in the material have been presented.*

Keywords: *plane waves, focused ultrasound beams, tissues.*

1. INTRODUCTION

The use of ultrasonic imaging as a diagnostic tool has increased considerably. If the transmitting-receiving wave transducer has suitable directional sensitivity, the direction of the reflected signal can be measured and a two-dimensional reflection-signal image formed. Acoustic output of machines has also increased. In some cases it can affect some potential hazards. There has been considerable prior work for study the interaction between ultrasound and biological matters [1-8]. However, the widespread use of therapeutic ultrasound in clinical environments has so far been limited, in part, due to incomplete understanding of the interaction process. In order to treat a narrow target area with minimal damage to the surrounding tissue, it is necessary to predict the path of the ultrasound beam from the transducer to the target. Since experimental studies using living tissue is difficult and costly, theoretical simulation of the problem can be extremely useful in providing a firm scientific basis for future clinical applications of focused ultrasound. Furthermore, the complex geometry and acoustic properties of human tissue often preclude their treatment using analytical methods and numerical methods (e.g., finite element or FEM) are needed in order to obtain quantitative information on the interaction between ultrasound and biological materials. In this study the ultrasound wave propagation simulation provided by BioSono Inc., is used to obtain the wave propagates in free 2D space.

2. PLANE WAVES

In most applications of ultrasound, the objective is to treat a small area in the path of the ultrasonic beam. Thus, the theoretical simulations must be able to isolate the local effects associated with the interaction process from boundary and other effects. Ultrasound wave propagation simulation is based on wave equation, demonstrating how the wave propagates in free 2D space. The incident wave is unfocused plane ultrasound wave. The simulation is numerical with the parameters at each grid point of the field being refreshed at certain time interval. The object shape does not affect calculation time. Two conditions, namely, symmetry and absorption are used in simulating propagation in unbounded media. In order to verify the accuracy of these approximations, the problem of plane waves propagating across a solid layer between two different fluid media is considered.

The three-layered materials system is shown in Fig. 1. It is infinite in the vertical, y-direction and semi-infinite along the positive x direction. The thickness of the first layer is h_1 and of the second layer is h_2. A uniformly distributed pressure provided by an ultrasound transducer, *f(t)*, is applied at $y=0$. The ray-tracing approximation has been extended to refraction and reflection at multiple interfaces, considering propagation through multiple media [9].

When the ultrasound pulse hits a boundary between two tissue structures the pulse will be partially reflected and partially transmitted. Reflection depends on the difference between the impedances of the two materials. In soft tissue, reflection is small. We place the phantoms on top of the steel-block and obtain the pulse-echoes from the fat-muscle interface as well as the muscle-steel block interface. The purpose is to create a highly reflective interface under two layers (fat and muscle). The values of the media parameters of the phantom are given in Table 1.

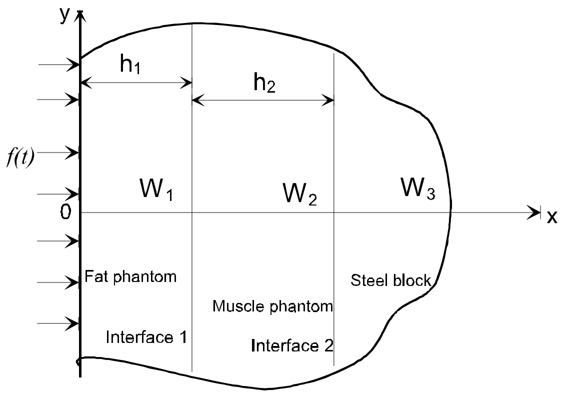

Fig. 1. A three-layered medium.

Table 1. The media parameters of the phantom

Parameters	Phantom 1 (fat)	Phantom 2 (muscle)
Relative density (ref water)	1.010	1.041
Velocity (m/s)	1508	1582
Attenuation *(dB/cm/ MHz)*	0.067	0.188
Reflectivity of normally incident waves at interface fat - muscle	0.10	

We used “fat" and “muscle" mimicking phantoms. These physical objects (designed from water-salt-gel mixtures) imitate some ultrasound characteristics (such as propagation velocity) in human fat and muscle. We also used a steel-block about 25 mm thick.

The pressure field within the medium can be calculated by using a standard theoretical approach. We first solve the propagation problem in the frequency domain and invert the resulting spectral response to obtain the time domain result. In the frequency domain, the displacement in water can be expressed in the form

$$W_1(x,\ \omega) = C\left[e^{-ik_1(x-h_1)} + R\ e^{ik_1(x-h_1)}\right],\quad 0 < x < h_1 \tag{1}$$

where the time dependence, $e^{i\omega t}$ has been suppressed.

In (1), $k_1 = \omega / \alpha_1$ is the wave number, ω is the angular frequency, α_1 is the velocity of the acoustic waves in fat and *R* is the reflection coefficient. The displacements in the 2nd and 3rd layers are of the form

$$W_2(x,\omega) = Ae^{-ik_2(x-h_1)} + B\,e^{ik_2(x-h_1)}, \quad 0 < x < h_1 \tag{2}$$

$$W_3(x,\omega) = Te^{-ik_3(x-h)}, \quad h_1 < x < h \tag{3}$$

where $h = h_1 + h_2$, *A, B* are constants, *T* is the transmission coefficient, and k_1, k_2 are the wave numbers in the 2nd and 3rd layers, related to the wave velocities, α_1, α_2, through $k_j = \omega / \alpha_j$.

The pressure is related to the displacement through the relation

$$P(x,\omega) = (\lambda + 2\mu)\frac{\partial w}{\partial x} = \rho\alpha^2 \frac{\partial w}{\partial x} \tag{4}$$

Applying the conditions for the pressure at $x = 0$ and the continuity conditions for the pressure and displacement at the interfaces, $x = h_1$, and h, the following system of equations is obtained for the unknown coefficients, *C, R, A, B,* and *T*:

$$\begin{cases} i\omega Z_1 C e^{ik_1h_1} = F(\omega) \\ \dfrac{F(\omega)}{i\omega Z_1}(1+R)e^{-ik_1h_1} = A + B \\ F(\omega)(-1+R)e^{-ik_1h_1} = i\omega Z_2(-A+B) \\ Ae^{-ik_2h_2} + Be^{ik_2h_2} = T \\ -Ae^{-ik_2h_2} + Be^{ik_2h_2} = -Z_{32}T \end{cases}$$

where:

$$Z_i = \rho_i\alpha_i, \; Z_{ij} = \frac{\rho_i\alpha_i}{\rho_j\alpha_j}$$

and *F(ω)* is the Fourier time transform of the applies pressure, *f(t)*. In the present calculations *f(t)* is assumed to be a single cycle sine pulse of period τ given by:

$$f(t) = \sin\left(\frac{2\pi t}{\tau}\right)H(t-\tau)$$

Thus

$$F(\omega) = \int_0^\infty f(t)\,e^{-i\omega t}\,dt = 2\frac{\pi\tau(e^{-i\tau\omega}-1)}{\omega^2\tau^2 - 4\pi^2}$$

Solving the system of equations, we obtain,

$$C = \frac{F(\omega)e^{-ik_1h_1}}{i\omega Z_1}$$

$$R = -\frac{(k_2\alpha_2 Z_3 - k_1\alpha_1 Z_1)(e^{2ik_2h_2}+1) - (k_1\alpha_1 Z_1 Z_{32} - k_2\alpha_2 Z_2)(e^{2ik_2h_2}-1)}{(k_2\alpha_2 Z_3 + k_1\alpha_1 Z_1)(e^{2ik_2h_2}+1) + (k_1\alpha_1 Z_1 Z_{32} + k_2\alpha_2 Z_2)(e^{2ik_2h_2}-1)}$$

$$P(x,\omega) = (\lambda + 2\mu)\frac{\partial w_1}{\partial x} = \rho_1\alpha_1^2\frac{\partial w_1}{\partial x} = F(\omega)(-e^{-ik_1x} + R\,e^{ik_1(x-2h_1)})$$

3. RESULTS AND DISCUSSION

The plot in Fig. 2a to the left shows the echo from the first interface. This can be used for rudimentary corrections to the velocity scaling for some simple parallel-interface cases. The Fig. 2b detailed the second echo. However for general media interface geometries, this may not be possible. For example in the $\theta = 30^\circ$ case, the first echo disappears. For general curved interfaces, more sophisticated estimation of the interfaces is necessary. Prior knowledge of the number of layers is needed.

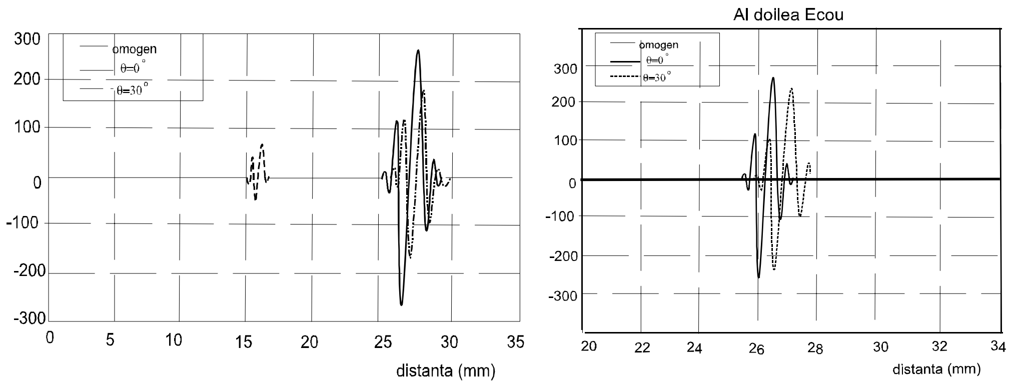

Fig. 2. The echo from the two reflecting interfaces.

The differences of the shape (especially towards the ends of the echoes) are mainly due to errors in estimating the transducer pulse. The other factor is the frequency-dependence of the attenuation coefficient. In this study we took the attenuation at the center-frequency. For the first echo, this was expected to be not much of a problem because the attenuation coefficient itself was small for the first medium. The second medium is more attenuative. For the second echo, we do observe a small frequency-downshift of the experimental pulse. This effect can be attributed the fact that the higher frequencies in the real signal is more attenuated than the attenuation considered at the center-frequency. And the lower ones are less attenuated than at the center frequency considered. Another consideration is that ideally we should also have estimated the attenuation coefficients as well as the velocities, apart from the thicknesses.

As an example of the ultrasound wave propagation simulation in free 2D space, the full-field pressure is shown in Figs. 3 and 4. The interaction of the waves with the interface can be easily identified. Use used the following assumptions:

The surface must be big compared to the ultrasound wavelength when reflection happens.

The acoustic impedance on both side of the surface must be different.

For plane wave, the reflected wave has the same angle to the surface as that of incident wave.

Fig. 3 presents a plane wave and the reflected wave has the same angle to the surface as that of incident wave.

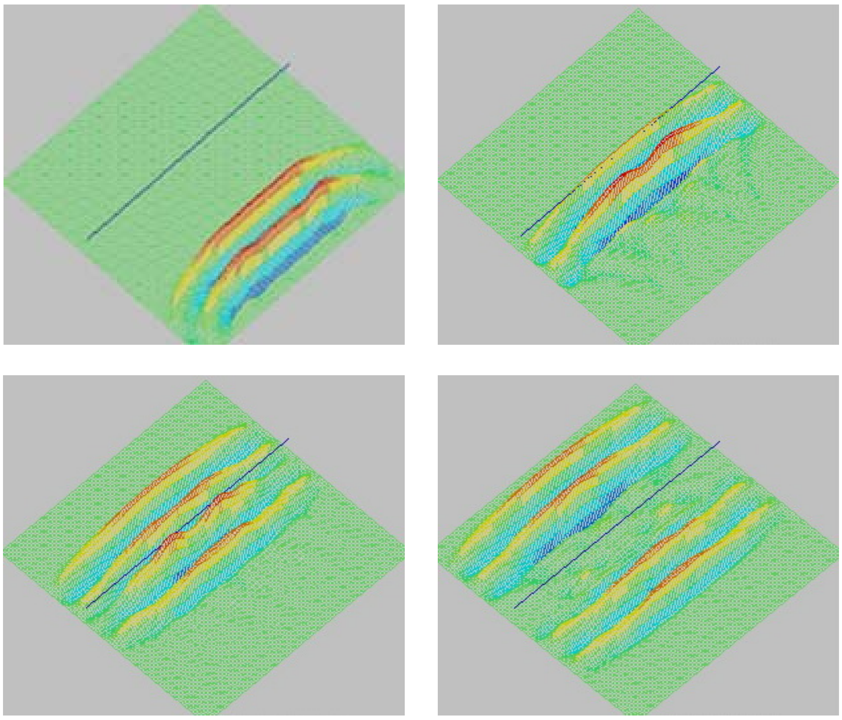

Fig. 3 Reflected and transmitted plane wave by a plane surface parallel with transducer area (by courtesy BioSono Inc.)

Fig. 4 presents the reflected and transmitted plane waves by a tilt plane surface.

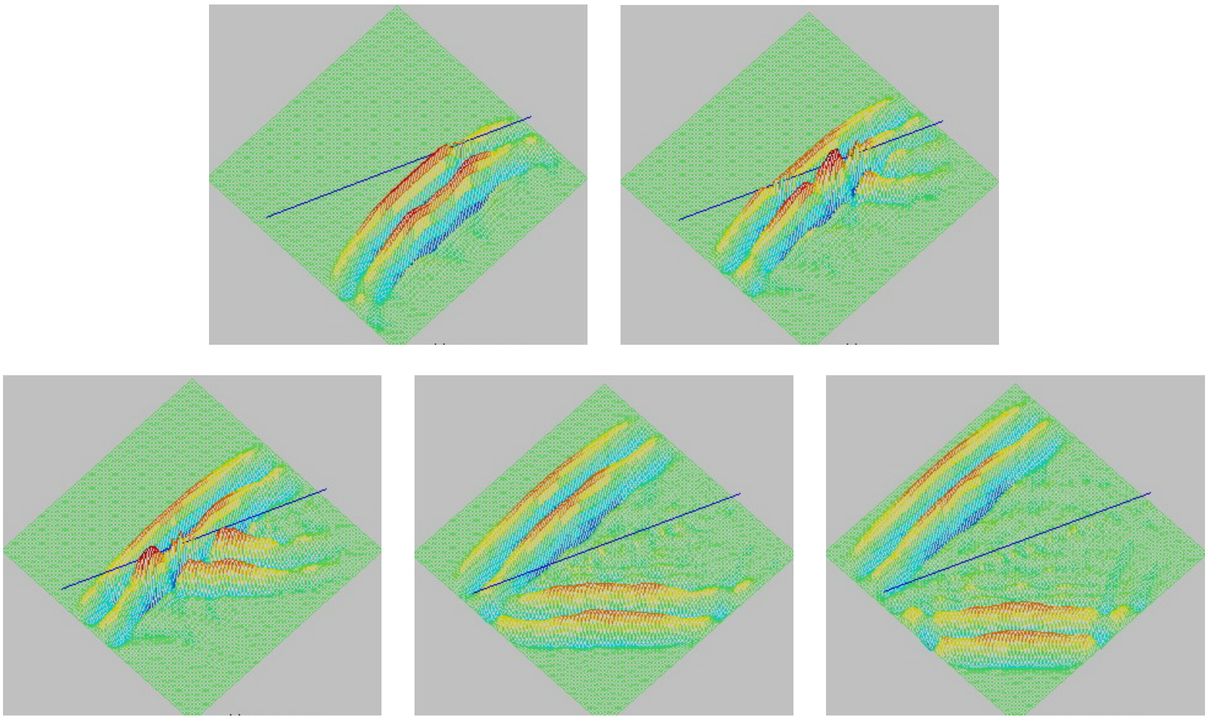

Fig. 4 Reflected and transmitted plane wave by a tilt plane surface (by courtesy BioSono Inc.)

The results, presented in Figs. 3 and 4, show that after the ultrasound passes through the layer and a plane surface, it is refocused at a point below the layer, as can be predicted on the basis of ray theory.

4. CONCLUSIONS

Locating the focal region of therapeutic ultrasound in complex biological materials systems is an important issue that needs to be properly addressed before wider clinical applications are possible. Lack of adequate knowledge on the temperature distribution in the biomaterial can lead to thermal exposure to a larger area for longer period of time and use of higher intensity ultrasound than may be needed. The knowledge gathered not only can reduce the needed intensity of the input ultrasound, but also reduce possible damage to the surrounding tissues. The characteristics of the interaction between focused ultrasound and a simple model of the biological material have been simulated in this paper. The spatial distribution of the pressure field in the material has been presented.

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