Vol 18 No 11, November 2009
1674-1056/2009/18(11)/4932-06

Chinese Physics B

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Influence of the abdominal wall on the nonlinear propagation of focused therapeutic ultrasound*

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(Received 15 April 2009; revised manuscript received 7 June 2009)

This article theoretically studies the influence of inhomogeneous abdominal walls on focused therapeutic ultrasound based on the phase screen model. An inhomogeneous tissue is considered as a combination of a homogeneous medium and a phase aberration screen. Variations of acoustic parameters such as peak positive pressure, peak negative pressure, and acoustic intensity are discussed with respect to the phase screen statistics of human abdominal walls. Results indicate that the abdominal wall can result in energy loss of the sound in the focal plane. For a typical human abdominal wall with correlation length of 7.9 mm and variance of 0.36, the peak acoustic intensity radiated from a 1 MHz transmitter with a radius of 30 mm can be reduced by about 14% at the focal plane.

Keywords: phase aberration, focused therapeutic ultrasound, abdominal wall
PACC: 4335, 8750C

1. Introduction

High intensity focused ultrasound (HIFU) is a noninvasive treatment that focuses high intensity ultrasound energy to the targeted area of tissue within the body without damage to intervening tissue.[1,2] The principle of this technique is to raise the tissue temperature to relatively high values and cause thermal coagulation and ablation of cells by using focused ultrasound. HIFU devices are currently under investigation as an effective treatment modality for ablating solid tumors of the prostate, liver, breast, kidney, and brain.[3−6] For the purposes of accurate targeting and effective heating, it is essential to understand the propagation of HIFU beam in tissues. Most numerical simulations of nonlinear wave propagation and heating within tissues have used limited sets of parameters for specific tissue models.[7,8] Few studies have considered the variability in propagation path or physical properties of the tissue. However, scattering and phase aberration brought about by the inhomogeneity of tissues have significant effects on the accurate prediction of heat deposition at the target tissue.

The body wall plays the most important role among all the phase fluctuations introduced by the inhomogeneity inside the human body.[9] For example, the abdomen layer, consisting of skin, entangled fat, muscle, connective tissues as well as septa within subcutaneous fat, deteriorates the focusing of an ultrasound beam in the treatment of liver tumors owing to phase fluctuation. Additionally, large sound speed variation within the body wall can affect the energy deposition while the inhomogeneity within viscera only affects local deposition.[10] Hence, phase aberration is only considered to exist in the body walls, while the internal organs are considered to be homogeneous. Previous studies have been performed to determine the influence of abdominal walls on phase fluctuations in diagnostic ultrasound. Krämmer and Hassler[11] made a one-dimensional measurement of the time delay fluctuations introduced by the human abdominal wall. Shmulewitz *et al*[12] found that muscles significantly contribute to the formation of wave distortion, while atherosclerosis, lower limb paralysis, ribs, intestinal gas, or an aponeurosis contribute to the wave front distortion. Sumino and Waag[13] found that human abdominal walls are statistically related, and the correlation length is on the order of several millimeters. An inhomogeneous medium can be considered as a homogeneous medium and a phase aberration screen that represents the cumulative phase aber-

*Project supported by the Program for New Century Excellent Talents in University of China (Grant No 06-0450), the National Natural Science Foundation of China (Grant No 10774071) and the Natural Science Foundation of Jiangsu Province, China (Grant No BK2007518).
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ration in tissues. Several models based on phase screen are developed for characterizing the fluctuations induced by abdominal walls. Hinkelman *et al*[14] suggested that a significant portion of the time-shift aberration caused by the abdominal walls can be explained by using single or multi phase screen models. Huang and Tsao[15] proposed a multilayer phase screen model in a linear field and Cristopher[16] proposed a nonlinear field for characterizing the fluctuations introduced by the abdominal walls. It has been pointed out that as the thickness of a typical abdominal wall is of the order of the correlation length, the phase fluctuations introduced by the abdominal walls can be simplified to a single phase screen model.

The purpose of this article is to evaluate the influence of the abdominal wall on HIFU therapy based on the phase screen model. The phase screen is constructed to provide phase variations having random distributions over the plane perpendicular to the ultrasound propagation axis. It possesses statistics that are similar to those provided by Hinkleman based on measurements[14] of human abdominal wall. Several ultrasonic parameters including positive pressure, negative pressure, and acoustic intensity, are discussed with respect to correlation length and variance.

2. Principles and methods

The phase screen method is often used for modeling random media.[17] The principle of this method is that an extended random medium can be separated into a series of slabs with a thickness Δz. The phase variations ϕ inside the slab is projected onto a screen of zero thickness, situated in the center of the slab. The phase fluctuation introduced by one such screen is $\phi(X, Y)$. The extended medium is thus approximated by a series of such screens. When the slab thickness is of the order of the correlation length of inhomogeneity, the single phase screen can be employed for modeling the inhomogeneous media.

For simplicity, the phase variation $\phi(X, Y)$ introduced by the screen possesses a zero mean, variance σ, and a Gaussian correlation function,

$$\mathrm{Cor}_\phi(X, Y) = \exp\left(-\frac{X^2 + Y^2}{l^2}\right), \tag{1}$$

where l is the correlation length. A two-dimensional (2D) Fourier transform is applied to the Gaussian distribution function

$$H(f_X, f_Y) = \sqrt{F\{\mathrm{Cor}_\phi(X, Y)\}}. \tag{2}$$

A 2D random noise signal $\eta(X, Y)$ with zero mean and variance σ is input into the filter $H(f_X, f_Y)$. The output is

$$\Phi(f_X, f_Y) = H(f_X, f_Y) \cdot F\{\eta(X, Y)\}. \tag{3}$$

The expected phase aberration screen is obtained by using an inverse Fourier transform

$$\phi(X, Y) = F^{-1}\{\Phi(f_X, f_Y)\}. \tag{4}$$

Figure 1 depicts three typical phase screens, where Fig.1(a) shows a phase screen for $l = a/5$ (a is the radius of the transmitter) and $\sigma = 0.36$; Figure 1(b) is for $l = a/10$ and $\sigma = 0.36$; Figure 1(c) is for $l = a/5$ and $\sigma = 0.76$. The circles in the figure indicate the relative size of the transducer.

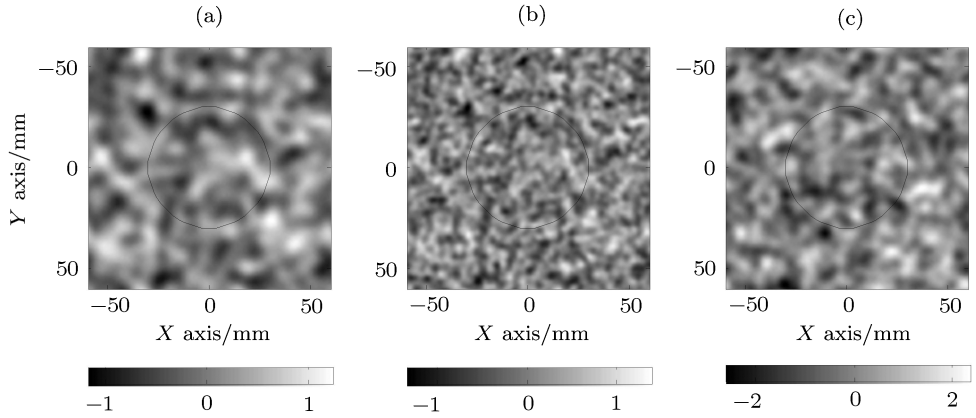


Fig.1. Three typical phase screens: (a) $l = a/5$ and $\sigma = 0.36$; (b) $l = a/10$ and $\sigma = 0.36$; (c) $\sigma = 0.76$ and $l = a/5$ ($a = 30$ mm).

The widely used Khokhlov–Zabolotskaya–Kuznetsov (KZK) equation is employed to model the nonlinear sound propagation in the homogeneous medium,[18−20] which is written in the following non-dimensional form

$$\frac{\partial}{\partial T}\left(\frac{\partial P}{\partial Z}-NP\frac{\partial P}{\partial T}-A\frac{\partial^2 P}{\partial T^2}\right)=\frac{1}{4G}\Delta'_{\perp}P, \qquad (5)$$

where $P=p/p_0$ is the normalized pressure, in which p is the sound pressure and p_0 is the sound pressure at the surface of the transmitter. $\Delta'_{\perp}=\partial^2/\partial X^2+\partial^2/\partial Y^2$ is the non-dimensional transverse Laplace operator with respect to X and Y. $X=x/a$, $Y=y/a$ and $Z=z/D$ are normalized coordinates, in which a is the radius of the transmitter, D is the geometrical focus distance. $T=\omega\tau$, here $\tau=t-z/c_0$ is the retarded time, ω is the angular frequency. $N=D/Z_n$, in which $Z_n=\rho_0 c_0^3/\omega_0\beta p_0$ is the shock wave formation distance of a plane wave, c_0 is the sound speed; ρ_0 is the density; β is the acoustic nonlinear coefficient. $A=\alpha D$, in which $\alpha=\omega_0^2 b/2\rho_0 c_0^3$ is the absorption coefficient, b is the dissipative parameter of sound. $G=z_d/D$, in which $z_d=\omega a^2/2c_0$ is the Rayleigh distance.

A numerical solution for Eq.(5) is obtained by employing a finite difference method in the frequency domain. The sound pressure takes the form of a complex Fourier series expansion and is expressed as

$$P(X,Y,Z,T)=\sum_{n=-\infty}^{\infty}C_n(X,Y,Z)\mathrm{e}^{-\mathrm{i}\,nT}, \qquad (6)$$

where $C_n(X,Y,Z)$ is the complex amplitude of the n^{th} harmonics ($-\infty<n<\infty$), $C_{-n}=C_n^*$, and C_n^* denotes the complex conjugate of C_n. Substituting Eq.(6) into Eq.(5), we gain the following set of coupled partial differential equations,

$$\frac{\partial C_n}{\partial Z}=-\frac{\mathrm{i}}{2}nN\sum_{k=-\infty}^{\infty}C_kC_{n-k}-An^2C_n+\frac{\mathrm{i}}{4Gn}\Delta_{\perp}C_n. \qquad (7)$$

In the case of a uniform circular focused piston source, the boundary condition at $Z=0$ is

$$\begin{cases} C_n(X,Y,0)=0.5\exp[-\mathrm{i}\,G(X^2+Y^2)], & X^2+Y^2\le 1,\ \ n=1;\\ C_n(X,Y,0)=0, & X^2+Y^2>1\ \ \text{or}\ \ n\ne 1. \end{cases} \qquad (8)$$

An inhomogeneous medium can be regarded as a combination of the homogeneous medium and a phase screen. When ultrasonic waves encounter a phase screen, the phase change can be expressed as

$$C_n(X,Y,Z)=C_n(X,Y,Z)\cdot n\cdot\phi(X,Y), \qquad (9)$$

where n is the harmonic order number, $\phi(X,Y)$ is a random phase fluctuation in the 2D plane.

3. Results and discussion

In this study, a 1.0 MHz transducer (radius = 30 mm, focal length = 180 mm) with an acoustic pressure of 0.1 MPa, is used as the transmitter. Ten groups of random phase screens are created and positioned at the source plane, with σ ranging from 0.16 to 0.76 and l varying from 1.9 mm to 11.2 mm based on previous studies.[9,14] In order to study the influence of the body wall on the nonlinear acoustic field of HIFU, we discuss the dependences of the peak positive pressure (p^+), peak negative pressure (p^-), and acoustic intensity patterns on the correlation length and variance. The above-mentioned parameters are selected from the focal plane that is located at 0.92–0.95d (d is the geometrical focal distance of the transducer) and is parallel to the transducer surface.

3.1. Dependence on correlation length

The correlation length (inhomogeneity length scale) l determines the spatial distribution of the random phase screen. Figure 2 gives an example of beam patterns of acoustic intensity at the focal plane in the presence

of a phase screen of $\sigma = 0.36$, where Figs.2(b) and 2(c) are lateral patterns at $l = a/5$ and $a/10$ respectively. For comparison, the lateral pattern without a phase aberration screen is also given in Fig.2(a). The lateral beam pattern without the phase screen (Fig.2(a)) is undistorted, where the side lobe ring structure is clearly defined. Owing to the effect of the phase screen, the main lobes are slightly distorted, and the structure of the first few side lobes is destroyed as shown in Figs.2(b) and 2(c). As the correlation length decreases, more side lobes are destroyed, but there is almost no change in the main lobe. Since the distribution of the scattered field is determined by the correlation length, the scattered field is concentrated near the propagation axis in the case of large correlation length. As a result, most of the beam distortion occurs near the main lobe.

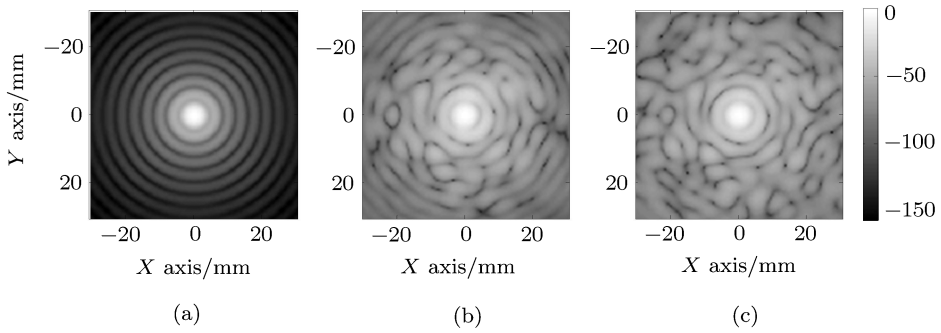


Fig.2. Intensity distributions in focal plane with different correlation lengths: (a) without phase aberration screen; (b) phase screen with$l = a/5$; (c) $l = a/10$ (in dB scale).

Figure 3 depicts the dependence of the normalized p^+, p^- and acoustic intensity in the focal plane along the propagating axis, which is normalized to their corresponding values without phase screen. The results are obtained from a mean of 10 random phase screens, and the standard deviation is indicated by error bars. Owing to the existence of the phase screen, p^+, p^- and intensity are decreased by about 6.9%–7.9%, 6.0%–6.6%, 12.5%–13.8% respectively in the focal plane. In addition, as the correlation length decreases, both the values of decrement and standard deviation induced by the phase screen are decreased, for example, the standard deviation for acoustic intensity is decreased from 7.3% to 3% corresponding to the change of a/l from 2.8 to 6.8. The large value of standard deviation indicates a more random distribution of the acoustic field.

Figure 4 shows the averaged –3 dB focal region of acoustic intensity in the focal plane as a function of the correlation length, where the values are normalized by the corresponding value without the phase screen. In the case of $a/l < 5.8$, the –3 dB focal region in the focal plane rapidly decreases as a/l increases. When $a/l > 5.8$, the –3 dB focal region reaches

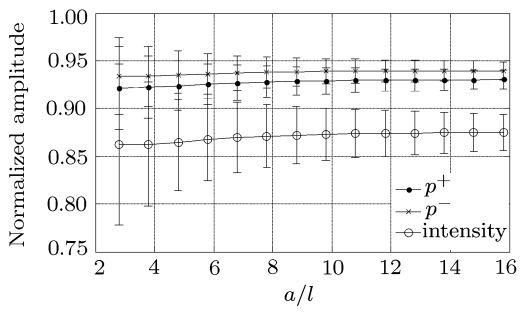


Fig.3. Correlation length dependence of p^+, p^- and acoustic intensity in the focal plane.

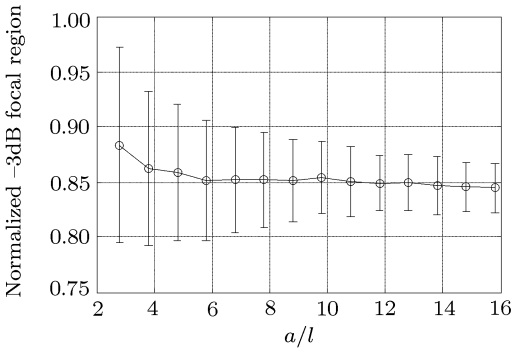


Fig.4. Correlation length dependence of the –3 dB focal region of acoustic intensity in the focal plane.

almost a constant, which is about 85% of that without phase screen. Previous studies[9,21] indicate that a human abdominal wall with an averaged correlation length of 7.9 mm can lead to a decrease of acoustic intensity of more than 10%. Therefore, the radius of the therapeutic transducer is required to be larger than 45.8 mm ($a/l > 5.8$). In this case, the acoustic intensity and –3 dB focal region are decreased to about 85% of those without phase screen.

3.2. Dependence on phase variance

The variance determines the phase variation of the random phase screen. Figure 5 shows the distribution of acoustic intensity in the focal plane in the case of the correlation length $l = 5$ mm, and varied variance, where Fig.5(a) is for $\sigma = 0.16$ and Fig.5(b) is for $\sigma = 0.76$. It is shown that the phase screen with $\sigma = 0.76$ leads to a larger phase variation (−2.3–−2.1 radian) compared with the screen with $\sigma = 0.16$ (−0.5–0.5 radian).

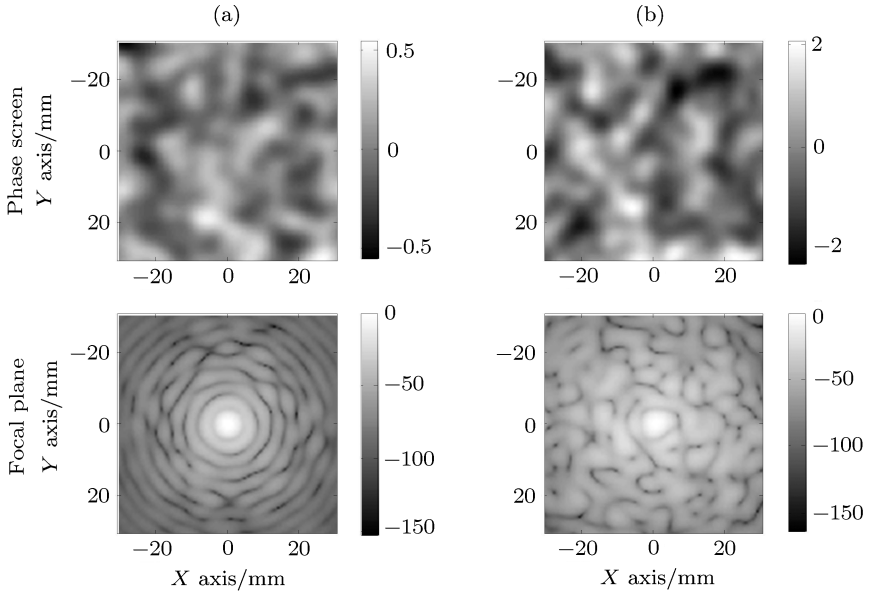


Fig.5. Acoustic intensity distribution in the focal plane with different variances: (a) σ =0.16, (b) σ =0.76.

Figure 6 shows the variance dependences of p^+, p^- and acoustic intensity in the focal plane. It is found that p^+, p^- and acoustic intensity are strongly dependent on the variance of the phase aberration screen. With the increase of variance, the phase screen becomes more irregular, leading to larger phase variations. As a result, the loss of acoustic energy is increased. In the case of $\sigma = 0.16$, the amplitudes of p^+, p^- and intensity are decreased by less than 3%, while in the case of $\sigma = 0.76$, their amplitudes are decreased by more than 45%. Similarly, the –3 dB focal region of the acoustic intensity in the focal plane decreases with the increase of the variance as shown in Fig.7. In the case of $\sigma = 0.76$, it decreases by about 70% as compared with that without phase screen.

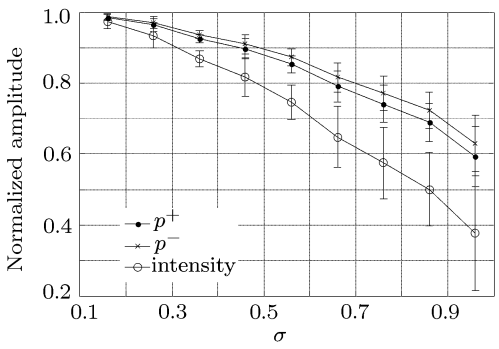


Fig.6. Dependence of the maximum peak p^+, p^- and acoustic intensity on the variance in the focal plane.

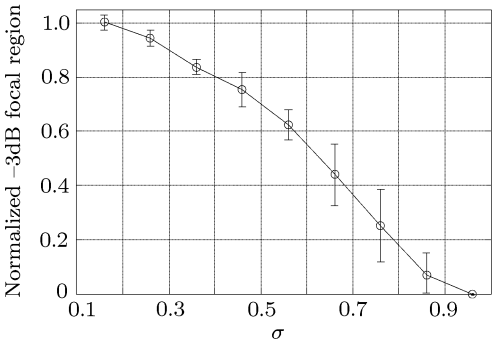


Fig.7. Dependence of –3 dB focal region of acoustic intensity in the focal plane on the variance.

In addition, one can find from Figs.6 and 7 that as σ increases, the standard deviations of acoustic intensity and –3 dB focal region in the focal plane increase from 1.9% to 5.9%, and from 2.6% to 13.3%, respectively. This suggests that a phase screen with a larger variance will introduce larger phase variations, leading to a larger loss of acoustic energy. A previous study indicates that muscle of the abdominal wall possesses a larger variance than fat.[14] As a result, the attenuation of muscle (1.4±0.6 dB/cm) is larger than that of fat (0.7±0.3 dB/cm) at a frequency of 1 MHz at 23 °C.[21]

4. Conclusion

In this article, we theoretically study the influence of inhomogeneous abdominal wall on focused therapeutic ultrasound. The phase aberration screen model and nonlinear KZK equation are employed to describe nonlinear sound propagation in human abdominal wall. Results demonstrate that inhomogeneity of the abdominal wall can lead to a decrease in amplitudes of positive pressure, negative pressure and acoustic intensity. For a typical human abdominal wall with correlation length of 7.9 mm ($a/l = 3.8$) and variance of 0.36, the peak acoustic intensity can be reduced by inhomogeneity by about 14% at the focal plane. In particular, a small different correlation length and large variance may lead to significant loss of ultrasound energy. Further experimental validation is in progress.

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