A Study of Effects of Tissue Inhomogeneity on HIFU Beam

Viren R. Amin  
_Iowa State University_

Ronald A. Roberts  
_Iowa State University, rroberts@iastate.edu_

Tao Long  
_Iowa State University_

R. Bruce Thompson  
_Iowa State University_

Timothy Ryken  
_University of Iowa_

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Abstract
The potential of high-intensity focused ultrasound (HIFU) will not be realized unless the effects of overlaying tissues are understood in such a way that allows for estimation of HIFU dose distribution at a target tissue. We employ computational models to examine the impact of phase aberration on tissue ablation. Thompson and Roberts have recently studied the effects of phase aberration on ultrasound focusing in aerospace engine materials such as titanium alloy, and have developed a computational model to examine these effects. The ultrasound beam observed after transmission through the fused quartz (homogeneous) and that observed after transmission through the titanium (inhomogeneous) demonstrate the severe beam wavefield amplitude distortion introduced by the velocity inhomogeneity-induced phase aberration. We study applicability of this approach to model phase aberration in inhomogeneous tissues and its effect on HIFU dose distribution around the focus. It is hypothesized that the ill-effects of phase aberration accumulate during propagation through intervening tissue in which field intensities are substantially lower than that in the focal zone, and it is therefore appropriate to use a linear acoustic model to describe the transport of energy from the transducer to the volume targeted for ablation. We present initial results of the simulation and experiments of beam measurements under water without and with different tissue layers. © 2006 American Institute of Physics

Keywords
ultrasonic effects, biological tissues, computer aided analysis, medical diagnostic computing, Electrical and Computer Engineering

Disciplines
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Comments
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A Study of Effects of Tissue Inhomogeneity on HIFU Beam

Viren Amin$^{1,2}$, Ron Roberts$^1$, Tao Long$^{1,2}$, R. B. Thompson$^1$ and Timothy Ryken$^3$

$^1$Center for Nondestructive Evaluation and $^2$Department of Electrical and Computer Engineering, Iowa State University, Ames, Iowa, USA
$^3$Departments of Neurosurgery and Radiation Oncology, University of Iowa, Iowa City, Iowa, USA

Abstract. The potential of high-intensity focused ultrasound (HIFU) will not be realized unless the effects of overlaying tissues are understood in such a way that allows for estimation of HIFU dose distribution at a target tissue. We employ computational models to examine the impact of phase aberration on tissue ablation. Thompson and Roberts have recently studied the effects of phase aberration on ultrasound focusing in aerospace engine materials such as titanium alloy, and have developed a computational model to examine these effects. The ultrasound beam observed after transmission through the fused quartz (homogeneous) and that observed after transmission through the titanium (inhomogeneous) demonstrate the severe beam wavefield amplitude distortion introduced by the velocity inhomogeneity-induced phase aberration. We study applicability of this approach to model phase aberration in inhomogeneous tissues and its effect on HIFU dose distribution around the focus. It is hypothesized that the ill-effects of phase aberration accumulate during propagation through intervening tissue in which field intensities are substantially lower than that in the focal zone, and it is therefore appropriate to use a linear acoustic model to describe the transport of energy from the transducer to the volume targeted for ablation. We present initial results of the simulation and experiments of beam measurements under water without and with different tissue layers.

Keywords: HIFU, ultrasound beam, simulation, tissue inhomogeneity, velocity.

INTRODUCTION

High-intensity focused ultrasound (HIFU) can potentially produce a noninvasive, trackless (no destruction of overlaying tissues), painless and bloodless lesion to a specific target. However, in real life, tissue inhomogeneity, organ motion, and target identification pose significant challenges and are being addressed by researchers to realize the full potential of HIFU. The HIFU application for destroying complex targets such as brain tumors and pediatric heart requires high precision and tools for dosimetry therapy planning in addition to real-time feedback of HIFU effects. The effects of overlaying tissues need to be understood in a way that allows for estimation of HIFU dose distribution at a target tissue. We present initial results of experimental and computational studies to examine the impact of phase aberration through overlaying tissues on ultrasound beam intensity distribution.
BACKGROUND

Some HIFU dosimetry work has been done using empirical experimental data analysis on different tissues. For example, Wang et al. [1] demonstrated, with in vitro and in vivo experiments on pig liver, that the HIFU lesion occurs at so-called “biological focal region” as a function of acoustic focal region (as measured in water), acoustic intensity, exposure time, irradiation depth within the tissue or overlaying tissues, tissue structure and its functional status. The effect of inhomogeneous overlaying tissues on the heating patterns and steady-state temperature distributions have been studied analytically using a random-phase screen model [2]. These model findings suggested the presence of many hot and cold spots throughout the medium.

Researchers including Thompson and Roberts (co-authors) recently studied the effects of phase aberration on ultrasound focusing in aerospace engine materials such as titanium alloy, and have developed a computational model to examine these effects [3,4]. The ultrasound beam observed after transmission through the fused quartz (homogeneous) and that observed after transmission through the titanium (inhomogeneous) demonstrate the severe beam wavefield amplitude distortion introduced by the velocity inhomogeneity-induced phase aberration (Fig. 1). We are studying applicability of this approach to model phase aberration in inhomogeneous tissues and its effect on ultrasound field intensity distribution around the focus.

FIGURE 1. Experimental results on ultrasound beam thru 3in fused quartz (FQ, homogeneous) and titanium (Ti, inhomogeneous) material. Left panel shows experimental setup for thru transmission measurement of beam intensity distribution. Second panel shows cross-section of the beam thru FQ and the right two panels show two incidences thru Ti. Color scale black=low amplitude, and red=high amplitude.

As seen in Fig. 1, the ultrasound energy is localized in “hot spots.” For flaw detection, this introduces a signal variance and for high-precision thermal therapy, this could introduce uneven tissue damage. The accuracy and repeatability of intended focus and heat distribution are crucial for the success of the HIFU therapy. However, the significance of the effects of tissue inhomogeneity for HIFU effect needs to be determined.
METHODS

Simulation. We used approach described in [3, 4] for simulation of the ultrasound wave field in inhomogeneous material. The wavefield in inhomogeneous media is governed by the volume integral equation

$$\phi(x') + \int \phi(x) v(x) G(|x-x'|) \, dx = \phi^0 (x')$$

where

- $\phi(x)$ = time harmonic wavefield,
- $v(x)$ = scattering potential = $1 - c_0^2 / c^2(x)$,
- $c(x)$ = wave velocity (spatially varying),
- $c_0$ = mean wave velocity (constant),
- $\phi^0 (x)$ = wavefield in homogeneous medium $c_0$,
- $G(x|x)$ = time harmonic point source (Green function) for $c_0$,

For small $c(x)$ inhomogeneity (< 5% variation), this can be solved as Neumann series and can be efficiently computed using FFT. For simulation of tissue layers (muscle and liver), each pixel $c_{ij}$ has random velocity value of magnitude $\varepsilon$ where $c_{ij} = c_0 (1 + \varepsilon r)$, $-1 < r < 1$, $\varepsilon = 0, .01, .03$; $r$ is random number with uniform probability. The small scale structure ($\lambda/4$, $\lambda$ = wavelength) was assumed. The Medium is further divided into rectangular cells with about two wavelengths (9 pixels) in length (propagation direction) and about one wavelength (5 pixels) wide. The Velocity within cell is raised or lowered by a random number $r$, i.e., $c(x) = c_{ij} (1 + \varepsilon r)$, $-1 < r < 1$, $x$ within cell for larger scale structure (about $2 \lambda$, $\lambda$ = wavelength).

Experiments. Experimental setup for measuring ultrasound beam intensity variations through tissue slices is shown in Fig. 2. We used precisely cut slices of liver and bovine muscle to measure the effects of tissue inhomogeneity on ultrasound beam intensity distribution. We measured the ultrasound beam pattern around the focus for a given transducer using different overlying tissues in bubble-free water path, at room temp. For water-only scan, an axial scan size was 200 (W) by 500 (L) steps for 127mm (5in) and cross-sectional scan size was 50.8mm (2in) by 50.8mm (2in). The scans with intervening tissues were done using 10mm thick bovine skeletal muscle layer and 20mm thick bovine liver layer. The intensity distribution at beam cross-sections was measured at the focus and at 12.7mm (.5”) pre- and post-focus. The scan step size was 0.2mm (0.08”) and scan area was 20.3mm (.8”) by 20.3mm (0.8”) providing 100 by 100 pixels. The measurements were made using a HIFU transducer (center frequency of 465.0 kHz, focal depth of 51.74mm and diameter of 81.8mm) driven using conventional pulser/receiver (Penametrics 5058), measured for third harmonic around 1.5 MHz (using 1.0-3.0 MHz filtering) using a receiving transducer sharply focused at 1.0in (FWHM=0.83mm at focus). The cross-sectional measurements were done at the maximum intensity along the axis of the beam beyond the intervening tissue layer.
RESULTS AND DISCUSSION

Fig. 3 shows simulation results for a two-layered tissue model with velocity variations of 0% (homogeneous), 1% and 3% for effects on beam intensity distributions for two transducers. It also shows (right panel) the results of the cross-sectional measured beam profile at the focus through different tissue layers.

For the experimental results, the water scan at focus revealed FWHM axial=10.4mm and lateral=1.5mm. The beam cross-sections measured in water at the focus and 12.7mm (.5”)) pre- and post-focus are shown in Fig. 4 for three configurations: without tissue, with 1cm bovine muscle layer, and with 2cm bovine liver layer.
CONCLUSION AND FURTHER WORK

We present initial results of experimental and computational studies to examine the impact of phase aberration through overlaying inhomogeneous tissues on ultrasound beam intensity distribution. We observed significant effects of tissue inhomogeneity on ultrasound beam. Future work includes: map out the velocity variation in thin layers (wavelength scale) of tissues to validate input values to simulation; validate computational models with experimental data initially on layered tissue structures and later with more complex shapes; extend the beam simulation model for 3D; and incorporate BHTE for HIFU heat dose calculations.

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