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HIGH POWER FOCUSED ULTRASOUND FIELDS IN THERAPEUTIC MEDICAL APPLICATIONS: MODELING AND MEASUREMENTS WITH A FIBER OPTIC HYDROPHONE

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Michael R. Bailey[1], Michael S. Canney[1], Vera A. Khokhlova[1,2], Oleg A. Sapozhnikov[1,2], and Lawrence A. Crum[1]
[1] Center for Industrial and Medical Ultrasound, Applied Physics Laboratory, University of Washington, Seattle, WA, USA; bailey@apl.washington.edu
[2] Dept. of Acoustics, Faculty of Physics, Moscow State University, Moscow, Russia

ABSTRACT

The goal of this work was to determine the acoustic waveform and beam width at the focus of a therapeutic ultrasound source both in water and in a tissue phantom. The source was a 2 MHz transducer of 45 mm focal length, 42 mm diameter, operating at 50 - 300 W acoustic power. Focal waveforms and beam widths calculated with a KZK-type model were in excellent agreement with values measured with a 100-µm, 100-MHz bandwidth fiber optic probe hydrophone (FOPH). Super focusing of the peak positive pressure and a proximal shift in the peak negative pressure were observed. Shocked distorted waveforms reached +70 MPa and -15 MPa. Surface waves on the transducer were measured and included in the model but did not significantly affect the results obtained at focus. The change of the FOPH bandwidth to 30-MHz or the diameter of hydrophone to 500-µm resulted in 20% underestimation of the measured peak positive pressure but did not affect the measured negative peak pressure. Initiation of boiling was observed in tissue phantoms in milliseconds as predicted by weak shock theory due to absorption on the shocks. Work was supported by NIH DK43881, NSBRI SMS00402, and RFBR.

INTRODUCTION

Therapeutic ultrasound generally uses focused sources to deliver acoustic energy through the skin deep into the body. It is used, for example, to break kidney stones, cauterize bleeds, or thermally necrotize tumors. Often there is no feedback imaging to monitor temperature changes at the focus. Therefore it is important to know the exact acoustic field in tissue for proper treatment planning. The first step, as required by the U.S. Food and Drug Administrations and other regulatory agencies worldwide is characterization of the acoustic output in water.

In this study, we investigate a source used in a continuous wave mode to heat tissue in what is termed high intensity focused ultrasound (HIFU). Our 2 MHz source is in the range 1-10 MHz generally used in HIFU treatments and is sharply focused to both localize the therapeutic region and to avoid skin burns during treatment. In similar clinical use, focal acoustic intensities over 10 kW/cm^2 are often reported and desired. The fine focusing and high amplitudes used in treatments make acoustic characterization difficult. Yet, accurate characterization of acoustic sources is important because nonlinear acoustic wave propagation leads to a nonlinear enhancement of the heating rate at high source intensities. In fact, shock waves can form in the tissue and absorption of acoustic energy at the shocks may become the dominant mechanism of tissue heating.

THEORY AND METHODS

The source was a focused, single-element, air-backed piezoceramic transducer. Short 30-cycle bursts were used to reduce cavitation and heating that might result at the hydrophone tip. The water was degassed and filtered.

Measurements of focal waveforms were performed using two different fiber optic probe hydrophones (FOPH 500 and FOPH 2000, RP Acoustics, Germany). The two hydrophones differed in the bandwidth of the electronics used (30 MHz for the FOPH 500 and 100 MHz for the FOPH 2000 as reported by the manufacturer), but both hydrophones had an active diameter of 100 µm. The output of the hydrophone was recorded using a Lecroy LT344 digital oscilloscope sampling at 500 MS/s. Averaging was used because of the high noise level of the hydrophone and typically 200-1000 burst cycles were averaged. The hydrophone was oriented along the acoustic axis of the transducer to minimize averaging of the waveform over the tip of the hydrophone and thus to better capture the shock fronts. Measurements were performed with the tip placed in water, near the polyacrylamide tissue phantom, and behind the phantom [1].

Modeling was performed using a KZK-type equation that accounted for the combined effects of nonlinearity, diffraction, and absorption [2]. The material properties and model parameters are described in a recent paper [3]. The inputs for the numerical model were established using acoustic radiation force balance and field mapping of the transducer at low power levels as discussed next.

RESULTS

As one of the research goals of this work was to define a method to determine the acoustic fields of therapeutic ultrasound devices, this portion of the methods is described as results.
The inputs for the model were determined as follows. Force balance experiments yielded a transducer efficiency of 83% acoustic power versus electric power. The source aperture of 42 mm and focal length of 44.4 mm were determined by matching the axial and transverse distributions of acoustic pressure obtained in modeling assuming uniform source vibration with the experimental data of free field mapping using an SEA-150 needle hydrophone (150 µm active diameter, Specialty Engineering Associates, Soquel, CA). The pressure amplitude at the source was obtained by introducing an additional 70% correction to the acoustic power measured using the force balance. This adjustment was made by matching the focal pressure amplitudes modeled and measured with calibrated PVDF hydrophone at low intensities to exclude unfocused acoustic radiation that occurs due to non-uniform vibration of the source [4].
It was critical to remove the impulse response of the FOPH. The measured signals were de-convolved with the impulse response supplied by the FOPH manufacturer. De-convolution of the raw signal is required because the FOPH has a non-uniform frequency response due to diffraction effects at the tip of the hydrophone [5].

To use the FOPH in tissue phantom, two calibration experiments were performed prior to casting the hydrophone in the gel. The relation between the output voltage of the hydrophone in gel and the actual measured pressure was determined using two methods. First, the optical index of gel was measured by laser deflection angle and used in the calibration routine provided by the manufacturer. Second, the pressure waveform was measured in water at the distal surface of a 2-cm thick piece of phantom placed between the transducer and the focus. The FOPH tip was then cast in this piece so that the propagation path in gel to the FOPH tip was the same 2 cm. For each experiment, the transducer was positioned so that the FOPH tip was at the focus of the HIFU source. Since the pressure field should be the same in both conditions, a linear calibration factor was obtained by comparing the waveforms measured in water and in the gel. It was found that this calibration factor agreed well with the change in the measured refraction indices.

Once the initial characterization was completed at low power, the focal waveforms were measured and calculated at increasing input powers. At 7 W acoustic input, nearly sinusoidal waves of 2 MPa amplitude were recorded by all 3 hydrophones and agreed very well with the modeled waveforms. However, when the power was increased, disagreement in the value of peak positive pressure was observed (Fig 1). The peak pressures measured in water with the lower and higher bandwidth FOPHs were 20% and 10% lower than the calculated peak pressure. Correspondingly, the spectra of the waveforms agreed at lower frequencies and disagreed at higher frequencies. Agreement in the phantom was better as the most absorptive phantom more readily absorbed high frequency components. This result was good evidence that shocked waveforms of several hundred bars could be measured and that the differences were likely due to the limited bandwidth of the hydrophones. However, the bandwidth determined this way was less for each hydrophone than reported by the manufacturer, so other possible experimental errors were considered.

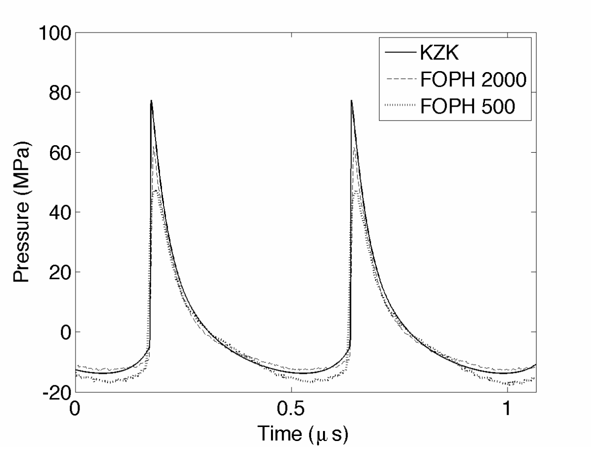

Figure 1. Focal waveforms measured by two FOPH hydrophones and calculated using the KZK-type model at 213 W acoustic power. Agreement is good, but the measured peak positive pressure is reduced due to the limited bandwidths of the hydrophones.

Suppression of the peak amplitude by spatial averaging over the FOPH tip was tested. Measured and calculated beam widths were in excellent agreement indicating little effect of spatial averaging. At high amplitude, the peak positive pressure is narrower than the beam width determined at low amplitude. The opposite is true for the negative pressure. Averaging the calculated pressures over the 100-µm tip reduced the peak positive amplitude only 1-2%. Loss in peak pressure value due to waveform averaging was tested by averaging a differing numbers of times without a measured change.

The effect of spatial distortion of the acoustic field due to non-uniform vibration of the piston source was tested. The true displacements of the source surface were measured using acoustic holography and used as an input to the model [4]. Low amplitude comparisons were used to recalibrate the efficiency of the source. Negligible change in focal waveform was observed at all higher input powers. It was shown therefore that the simpler modeling of a uniform piston was adequate.

Finally, to test the accuracy of the waveforms and elucidate a rapid mechanism of heating in high power therapeutic ultrasound, weak shock theory was used to predict the time to reach 100ºC. A range of measured and calculated focal peak pressures were used; the lower measured pressures yielded a slightly longer time. The times were compared to time to observe a boiling bubble in the phantom using a high peed camera. Initiation of boiling in time durations from 100 to 1 ms were observed. The results of modeling the time to boiling temperature based on calculated shock amplitudes were in near exact agreement with the measurements.

CONCLUSIONS

High power, highly focused therapeutic ultrasound fields can be determined through a combination of measurements and simulations in water and tissue phantoms. Comparison to the time to boiling indicates that calculated waveforms are more accurate, and the measurements underestimate the peak positive pressure. The measurements alone therefore may not be sufficient as even a 100 MHz bandwidth can limit the peak pressure. This pressure, through shock losses, can be the waveform feature most dominant in heating in HIFU fields and can cause boiling in milliseconds.

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