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NONLINEAR DERATING OF HIGH-INTENSITY FOCUSED ULTRASOUND USING HYDROPHONE MEASUREMENTS IN WATER

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INTRODUCTION

In the design and testing of high-intensity focused ultrasound (HIFU) devices, it is important to know the HIFU-induced temperature rise in the tissue of interest. In order to reduce the required number of animal and human experiments for assessing the influence of HIFU, semi-empirical mathematical models can be used for predicting the thermal effects. However, these models require values of the pressure amplitude in the tissue, which can be difficult to obtain experimentally. Pressure measurements can be much more readily performed in water. Therefore, techniques for predicting the pressure field in tissue using measured pressures in water, i.e., derating techniques, can be valuable for characterizing HIFU systems. At low ultrasound intensities, pressure amplitudes in tissue can be estimated by measuring pressure in water and reducing it by a factor based upon typical tissue attenuation [1] (linear derating). At higher ultrasound intensities, harmonics of the fundamental frequency are generated due to nonlinear propagation effects. Higher harmonics are attenuated differently in water and tissue [2]. Energy is also converted into heat at rates that depends on the frequency of the harmonic modes. Techniques for nonlinearly transforming measured pressures in water to values appropriate for tissue are defined as *nonlinear derating*. These techniques are desirable when bioeffects of higher intensity exposure are being studied.

In this research we present a nonlinear derating method [3] that uses pressure measurements obtained in water to model the nonlinear interaction between higher harmonics in tissue. To account for the differences in modal amplitudes between tissue and water, in the modeling the water harmonics are scaled down with the aid of contours derived from numerical simulations. The results obtained from the derating procedure are compared with direct measurement of the focal pressure in the tissue phantom using hydrophone.

METHODS

Propagation of HIFU beams are often modeled using the nonlinear Khokhlov-Zabolotskaya-Kuznetsov (KZK) equation [2]. Modeling the time-harmonic pressure field as a sum of modes with a Gaussian transverse profile, the KZK equation may be reduced to a simpler system of ordinary differential equations for the mode amplitudes. In the context of the nonlinear derating technique discussed here, the system takes the following form [3]:

$$\frac{d\tilde{a}_n}{d\zeta}+\left[\gamma_n-\frac{G+i}{G-(G+i)\zeta}\right]\tilde{a}_n=\frac{inN}{4}\sum_{m=1}^{\infty}a_m\left(a_{n-m}+2^{*}_{m-n}\right) \tag{1}$$

$$\text{where: } \gamma(f)=d\alpha(f)-\frac{2id}{\pi}\alpha(f)\log\left(\frac{f}{f_0}\right) \tag{2}$$

Here $\tilde{a}_n$ and a_n are the scaled complex pressure amplitudes at $\rho = 0$ (where ρ is the radial coordinate scaled by the transducer radius) corresponding to the n^{th} harmonic in tissue and water, respectively. The '*' denotes complex conjugation, $n = 1, 2, \ldots$ is the harmonic number, d is the transducer focal length, $\zeta = z/d$ is the scaled axial coordinate, G is the linear pressure gain, f is the transducer frequency, and f_0 is the reference frequency usually 1 MHz. $N = 2\pi p_0\beta df/(\rho_0 c_0^3)$ is the coefficient of nonlinearity [3], where β is the nonlinear parameter for the propagation medium, p_0 is the pressure at the transducer surface, ρ_0 is the mass density and c_0 the small-signal sound speed. N in eq. (1) is corresponding to the coefficient of nonlinearity in tissue. The relationship between absorption and frequency is given by $\alpha(f) = \alpha_0 (f/f_0)^{\eta}$, where η = 1for tissue, η = 2 for water and α_0 is the acoustic absorption at the reference frequency f_0.

A HIFU transducer with focal length of 6.26 cm and diameter of 6.4 cm was used to produce the pressure field in water. The operating frequency and the linear pressure gain were 1.05 MHz and 34.17, respectively. Material parameters included c_0 = 1482 m/s, ρ_0 = 1000 kg/m^3, β = 3.5 in water, and c_0 = 1579 m/s, ρ_0 = 1030 kg/m^3, β = 4.5 in tissue phantom. In order to account for the difference in attenuation between water and tissue, two types of pressure amplitude reductions were performed. The first, labeled "source scaling", involves performing the water measurements at a lower source pressure in water than the value for tissue. This results in a lower coefficient of nonlinearity in water than tissue (N_w/N_t < 1). The second, labeled "endpoint scaling", reduces the measured amplitudes of the water harmonics by the factor exp($-\varepsilon n \alpha_t z$), where $0 \leq \varepsilon \leq 1$ and α_t is the acoustic absorption for tissue phantom. The factor ε is required because amplitude reduction using the full attenuation leads to excessive attenuation when combined with source scaling. In our previous study [3], a set of pressure error contour plots was provided as a function of N_w/N_t and ε. The factors (N_w/N_t) = 0.82 and ε = 0.01 used in this study were chosen from those error contour plots.

Pressures were measured along the beam axis in water, using a hydrophone. The experiments in water were performed at the source-scaled pressure of 0.24 MPa. The acquired pressure trace was Fourier transformed to obtain the modal amplitudes a_n, which were inserted into the right side of eq. (1), after endpoint scaling was performed. The resulting linear, first-order equations were solved to obtain the modal amplitudes in tissue phantom ($\tilde{a}_n$). The estimated modal amplitudes were then used to construct the focal pressure waveform in tissue phantom.

In addition, a gelrite-based tissue mimicking material (TMM) was constructed based on the King et al. [4] protocol. The transducer and the tissue phantom resided in a tank of degassed water. The pressure trace at the transducer focus, right behind the tissue phantom, was recorded using a hydrophone. The source pressure at the transducer surface was p_{0t} – 0.28 MPa.

The pressure waveform at the focus as determined using the derating process was compared with direct measurement in tissue phantom. The error was calculated as:

$$pressure\,error = \frac{\left|p_{+,\,direct} - p_{+,\,derating}\right| + \left|p_{-,\,direct} - p_{-,\,derating}\right|}{p_{+,\,direct} + \left|p_{-,\,direct}\right|} \quad (3)$$

Here p_+ and p_- denote the peak positive and negative pressures, respectively.

RESULTS

Comparison between the derated modal amplitudes, $\tilde{a}_n$, at the focus and direct measurement in the tissue phantom (Fourier transform of the recorded focal pressure in TMM) is provided in Figure 1. The difference between the derated modal amplitudes and direct measurement in TMM for the first, second and third harmonic is 7%, 31% and 4%.

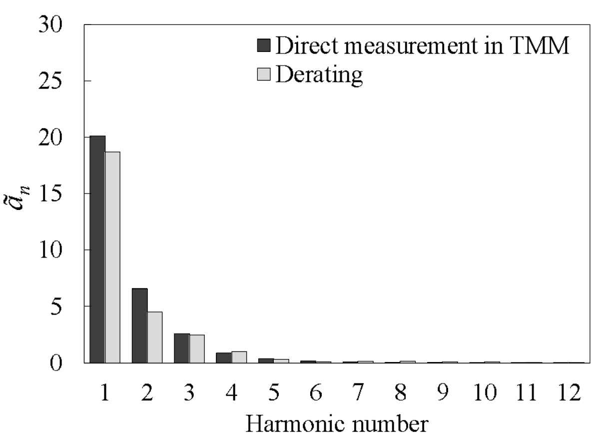

Figure 1. Modal amplitudes in TMM, $\tilde{a}_n$, at focus derived by the derating technique and direct measurement in TMM

Figure 2 shows the focal pressure waveforms obtained by derating and direct measurement in TMM. The peak positive pressure by derating and direct measurement in TMM is 6.59MPa and 8.25 MPa, respectively. The derated peak negative pressure (4.36MPa) is quite similar to the direct measurement in TMM (4.64 MPa). The focal pressure predicted by derating agrees with direct measurement in TMM within 15%.

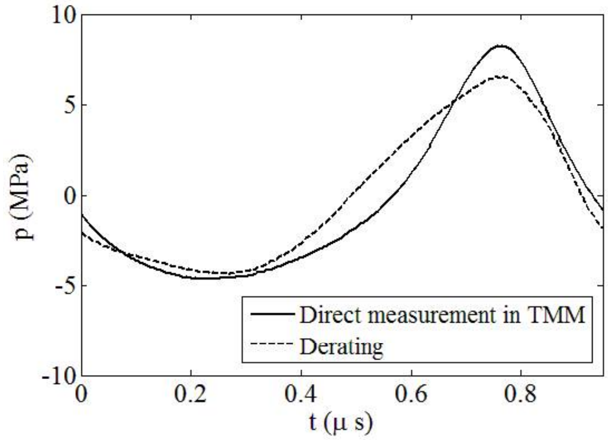

Figure 2. Focal pressure waveform in TMM, derived by the derating technique and direct measurement in TMM

DISCUSSION

The nonlinear derating technique was applied to estimate the focal pressure in TMM, using the measured axial pressures in water by hydrophone. The applied combination of source and endpoint scaling resulted in an estimation of focal pressure in TMM within 15% error. The error may be reduced by increasing the amount of source pressure in water. These experiments are underway. Finally, the technique requires experimental validation for temperature rise. The estimated temperature rise by this nonlinear derating technique will be compared with the direct measurement of the temperature rise inside the TMM using thin wire thermocouples. The technique has the potential to convert pressure measurements in water to pressure and temperature measurements in any tissue of interest, at clinically relevant power levels.

ACKNOWLEDGEMENTS

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