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69 COMPARISON OF DERATING METHODS FOR NONLINEAR ULTRASOUND FIELDS OF DIAGNOSTIC-TYPE TRANSDUCERS

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OBJECTIVES

There are therapeutic and diagnostic uses of imaging probes, which benefit from exceeding the Mechanical Index limits of diagnostic ultrasound and support that these benefits occur without negative bioeffects. Without imbedded software restrictions, the *in situ* pressure levels of these devices can exceed the typical diagnostic limits on Mechanical Index and spatial peak pulse average intensity (I_{SPPA}). When calibrating imaging probes at these levels in water, nonlinear propagation effects are present, which complicates the derating process for estimating *in situ* fields. Different derating approaches have been proposed to predict pressures in tissue from measurements in water. One conventional derating method is to scale the focal pressures and another method is to scale the source amplitude to compensate for linear losses on the way to focus. This second method is described as nonlinear derating and has been shown to provide accurate results for strongly focused therapeutic transducers. However, applicability of these derating methods to diagnostic probes operating at therapeutic intensities is still in question. Here, the derating methods were tested for a diagnostic probe used in kidney stone propulsion technology.

METHODS

A standard diagnostic ultrasound curved array probe operating at 2.3 MHz (C5-2, Philips Ultrasound, Andover, MA, USA) was considered. The array comprises 128 elements; however, the results presented hereafter were obtained by considering 64 central elements of the array to be active (Figure 1). In the azimuthal plane the focus can be steered electronically, while a cylindrical acoustic lens focuses the field at a fixed depth in the elevation plane.
A combined measurement and modelling approach was used to establish an equivalent source boundary condition for nonlinear simulations of the array field in water based on the 3D Westervelt equation [1]. Simulations were performed for propagation entirely in water and in the presence of a tissue mimicking phantom placed at a distance of 1 cm in front of the probe surface. The acoustic properties of the phantom were set the same as in water, except for the frequency dependent absorption of α_0 = 0.5 dB/cm/MHz with power exponent n = 1.2 and corresponding dispersion that were set according to the properties of the phantom.
Two derating methods were tested to estimate the *in situ* (z = 50 mm) pressure field in tissue from the waveforms simulated in water. Derated waveforms were then compared with direct simulation results in tissue. In the conventional derating method, the pressure field calculated in water at the focus was multiplied by the absorption exponent accounting for the propagation distance of 40 mm in tissue to the focus. In the nonlinear derating method, the pressure amplitude at the focus in tissue was assumed to be the same as in water for the lower source voltage scaled with the same absorption exponent value.

RESULTS

Axial distributions of the peak positive and peak negative pressures are shown in Figure2 for several output voltages in the free field in water (a) and in the presence of the tissue phantom (b). The focal lobe of the probe (20 mm long) is relatively large in comparison with the focal length of 50 mm because the transducer has a relatively low linear focusing gain (9.3). Therefore, at high power outputs nonlinear propagation effects accumulate over the long propagation distance and are not localized near the focus as is the case for strongly focused therapeutic sources.
Peak positive and peak negative pressures at the focus, z = 50 mm, as functions of source voltage are shown in Figure3. The nonlinear saturation of the peak positive pressure is clearly seen for propagation in water (black curve) and in tissue (blue curve), though at higher voltage levels. The conventional derating process of scaling focal pressures is illustrated by the green curve in Figure3. As denoted by the vertical dashed arrows, this method overestimates peak positive pressure at moderate voltages (up to 50 V) and underestimates it at higher voltages. Nonlinear derating is illustrated by the red curve. As shown by the horizontal arrows, peak positive pressures are significantly overestimated (by up to 50%) for source voltages higher than 20 V. For lower voltages, the nonlinear derating matches results in tissue within 10% of accuracy. Peak negative pressure magnitudes estimated by derating methods are smaller than those obtained in direct numerical simulations. Peak negative pressures predicted by the conventional derating method can be 50% smaller than in simulations, while the discrepancy remains less than 20% for the nonlinear derating method.
Focal waveforms obtained in simulations in tissue (blue curve) and using the derating methods (red and green curves) are compared in Figure4 for 55 V (a) and 90 V (b) outputs. At 55 V the nonlinear derating method predicts 40% higher peak positive pressure than simulations, while peak negative pressures are in closer agreement. The waveform resulting from conventional derating is fortuitously close to the simulated waveform. At 90 V all waveforms are significantly different; the peak positive pressure obtained in simulations is approximately in the middle between the results of the two derating processes.

CONCLUSIONS

Nonlinear acoustic fields generated by a diagnostic ultrasound probe are simulated in water and in a tissue phantom using the 3D Westervelt equation. Two derating approaches were applied to estimate the pressure field in tissue using the results obtained in water. It was shown that the conventional derating method can either overestimate (up to 50%) or underestimate (up to 25%) peak positive pressure depending on the source voltage, while it underestimates peak negative pressures by up to 50%. The nonlinear derating method provides accurate results at low intensities (here up to 20 V); however, it overestimates peak positive pressures by up to 50% at higher intensity levels. These simple derating procedures therefore cannot substitute direct numerical modelling to provide reasonable accuracy for nonlinear in situ pressures for diagnostic probes. *The study was supported by the grants RSF 14-12-00974, NIH EB007643 and DK043881.*

REFERENCES

[1] Karzova *et al.*, AIP Conf. Proc. (1685) 040002-1, 2015

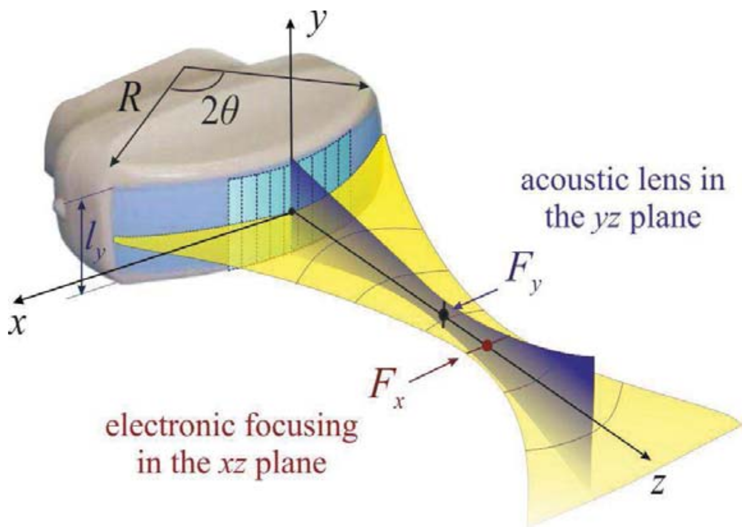

Figure 1: Geometry of focused ultrasound beam produced by C5-2 array probe with 64 active elements.

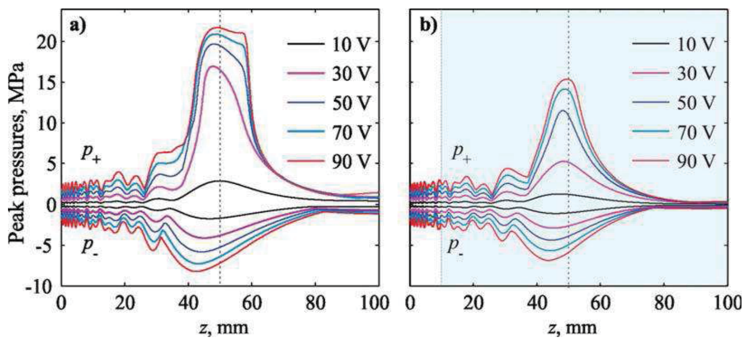

Figure 2: Axial distributions of the peak positive and peak negative pressures for (a) free field propagation in water and (b) propagation in the presence of a tissue phantom (shown in blue) for outputs of 10, 30, 50, 70 and 90 V. The water/phantom interface is positioned 1 cm from the array centre.

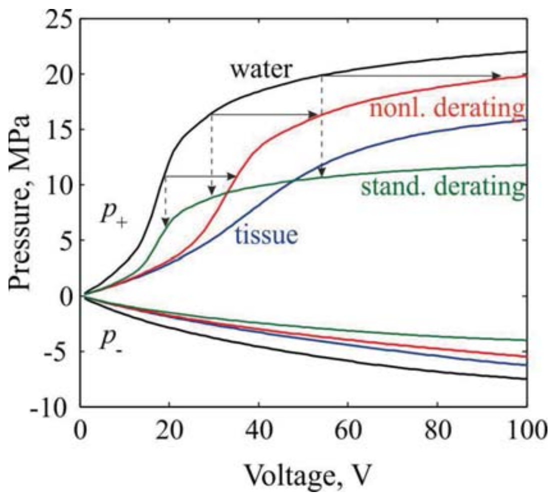

Figure 3: Peak positive and negative pressures at the focus (z = 50 mm) versus transducer output voltage. Propagation is simulated in water (black curve) and in a tissue phantom (blue curve). Horizontal and vertical arrows illustrate nonlinear derating by scaling the source amplitude (red curve) and conventional derating by scaling the focal pressures (green curve).

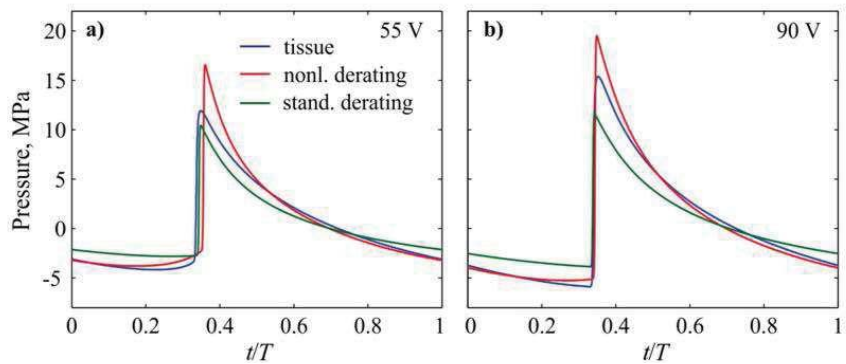

Figure 4: Focal waveforms (z = 50 mm) obtained by direct modelling of propagation in the tissue phantom (blue curve) as well as by derating from free field water simulations (red curve for nonlinear derating, green curve for conventional derating). Results are shown for output levels corresponding to (a) 55 V and (b) 90 V.