Changes in attenuation coefficient in MRgFUS treatments of in-vivo rabbit thigh estimated using MRTI-derived specific absorption rate patterns

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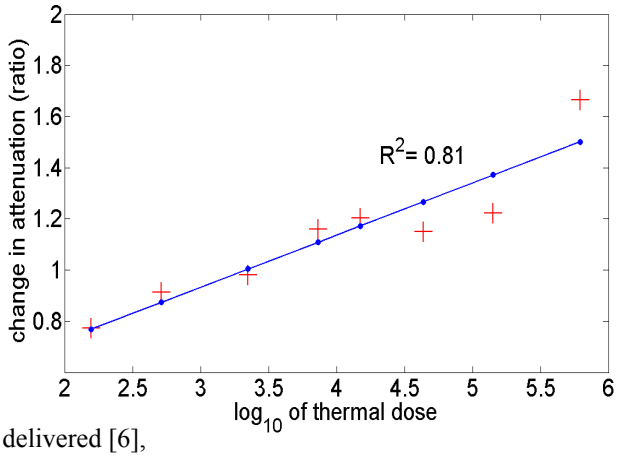
Introduction: MR-guided focused ultrasound surgery (MRgFUS) is a non-invasive means of causing selective tissue necrosis using high-power ultrasound and MR temperature imaging (MRTI). Predicting the power deposition pattern in the inhomogeneous tissue requires a beam propagation technique that has the ability to handle inhomogeneous acoustic properties. Beyond the innate inhomogeneities of tissue, it has been shown [1] [2] that the attenuation coefficient of tissue increases significantly (two-fold or more) and irreversibly at the high temperatures that are common in MRgFUS treatments. The goal of this research is to develop a non-invasive technique to estimate changes in tissue attenuation coefficient during MRgFUS treatments.

Methods: The technique uses an optimization routine to minimize the squared difference between an experimentally obtained power deposition pattern (SAR_{exp}) and a simulated power deposition pattern obtained from a beam simulation technique (SAR_{sim}). The optimization routine adjusts the values of the acoustic properties (speed of sound and attenuation coefficient) that are inputs to the beam simulation technique, and the values that result in the minimization of the difference between the two SAR patterns (over a tissue volume around the focal zone) provide an estimate of the acoustic properties of the media. This inverse parameter estimation technique has been validated previously in homogeneous phantoms [3] and it is used here to estimate HIFU treatment-induced changes in attenuation coefficient in *in-vivo* rabbit thigh muscle.. *Experimental SAR pattern:* It has been shown theoretically and experimentally that SAR information can be obtained from the rate of temperature increase (before thermal conduction or perfusion effects become significant) immediately following a step change in applied power [4]. In order to obtain SAR_{exp}, a short, low-power interrogation pulse is applied and MRTI is used to measure the resulting temperature change. The first few points in the heating curve are used to calculate the SAR_{exp} in the sample as follows: $SAR_{exp,i} = c_{p,i} dT_i/dt$, where $c_{p,i}$ is the specific heat of the voxel i. *Simulation of SAR pattern:* SAR_{sim} patterns are obtained using our previously developed hybrid angular spectrum (HAS) beam propagation technique [5] that accounts for the effects of reflection, refraction and attenuation of the ultrasound beam in complex inhomogeneous media; $SAR_{sim,i} = 2\alpha_i I_i/\rho_i$, where α_i, I_i, and ρ_i denote the attenuation coefficient, intensity (power density), and density at voxel i respectively. The rapid speed of the HAS beam simulation technique (each run in a 201x201x201 model takes 2 s) makes it ideal for use in this optimization technique where multiple forward solutions are required.



Experiment: All experiments were performed in an MRgHIFU system consisting of a Siemens TIM Trio 3T MRI scanner, a 256-element phased-array HIFU transducer (Imasonics, Inc.), and hardware and software for beam steering and data visualization (Image Guided Therapy). The Institutional Animal Care and Use Committee approved these animal experiments. One female white New Zealand rabbit was used for this study. The thighs were shaved and depilatory cream was applied. HIFU heating was monitored with a 2D gradient echo sequence with parameters TR/TE = 45/10 ms, 2x2x3-mm spatial resolution (3-mm slice thickness) and 4.7-s temporal resolution. The following sequence of acoustic power was applied to each thigh: low-power pre-treatment interrogation pulse to estimate the initial attenuation coefficient (12 W acoustic for 30 s), 10-minute cooling period, high-power treatment pulse (31 W acoustic for 35 s) to cause change in attenuation due to thermal dose delivered [6], 15-minute cooling period, and post-treatment low-power interrogation pulse (12 W acoustic for 30 s) to measure the changed attenuation coefficient due to treatment. Pre-treatment and post-treatment attenuation values at a total of twenty-four voxels at and around the focal zone were estimated using the optimization routine. The attenuation change ratio (i.e., post-treatment attenuation/pre-treatment attenuation) was calculated for each voxel.

Results and Conclusion: Since the treatment pulse resulted in a heterogeneous distribution of thermal dose around the focal zone, each voxel had a unique value of thermal dose. In order to calculate the average change in attenuation with thermal dose, the data were divided into eight groups, each group with a range of $\log_{10}$ of thermal dose equal to ± 0.2. Each group included three voxels; the average attenuation change ratio and the average $\log_{10}$ of thermal dose for each group were calculated and are plotted in the figure above. As shown by previous authors *in-vitro* [1], the attenuation coefficient increases approximately linearly with $\log_{10}$ of thermal dose; the slope of the curve found using the non-invasive technique in *in-vivo* rabbit thigh (0.20) corresponds well with the slope found by previous studies in *in-vitro* dog muscle (0.21) [1].

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