

ORAL PRESENTATION

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Quantifying perfusion-related energy losses during magnetic resonance-guided focused ultrasound

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Background/introduction

The focused ultrasound power required for successful ablation of uterine fibroid tissue varies substantially between patients and within single treatments.[1,2] Fibroids with high signal intensity in pretreatment T2-weighted MR images have been shown to require increased power to achieve adequate temperature elevation for ablation:[2,4] thus, T2-weighted signal intensity has been suggested as a predictor of MRgFUS treatment response.[2,3] Physiologically, the high intensity of T2-weighted MR images of uterine fibroids may represent vascularization, fluid-rich tissues, or degeneration.[4,6] By quantifying perfusion-related energy losses (Q_b) during MRgFUS treatments, this study is the first step in linking perfusion-related energy losses with MR perfusion imaging. This knowledge could be used to improve biothermal modeling of MRgFUS fibroid treatments and as a potential independent predictor of treatment response and outcome.

Methods

Experiments were performed in *ex vivo* porcine kidneys perfused with a heparin- H_2O solution in variable flow (0, 20, 40 mL/min) situations and embedded in a gelatin phantom (Figure 1). Heating was achieved by electronically steering a phased-array ultrasound transducer (256 elements, $f=1$ MHz) in an 8 mm-radius circle for 120 s (Figure 2). MR temperature data (Figure 3) were acquired with a 3T Siemens Trio MRI (3D segmented-EPI, TR/TE=30/11 ms, FA=15°, EPI factor=9, 2x2x3 mm³, 3.3 s acquisition, ZFI to 0.5-mm isotropic spacing). Based on conservation of energy principles, deviation of a thermal model that excludes perfusion effects from the experimental temperatures was used to quantify Q_b . Estimates of Q_b were obtained at the time of each MR acquisition during

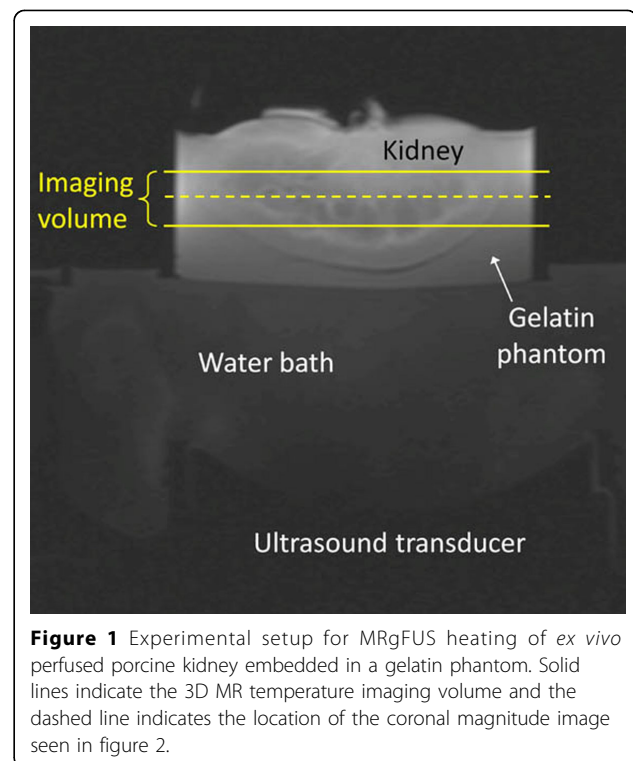


Figure 1 Experimental setup for MRgFUS heating of *ex vivo* perfused porcine kidney embedded in a gelatin phantom. Solid lines indicate the 3D MR temperature imaging volume and the dashed line indicates the location of the coronal magnitude image seen in figure 2.

cooling, transformed into perfusion values via the Pennes bioheat transfer equation,[7] and averaged to mitigate the effects of noise.

Results and conclusions

High perfusion values (Figure 4) correspond to regions of increased cooling (Figure 3) and likely indicate locations of discrete vasculature. Constant, uniform perfusion values ranged from -0.7–0.1, 1.6–3.9, and 3.4–4.4 kg/m³/s for 0, 20, and 40 mL/min flow rates, respectively, following anticipated trends with perfusion approximately zero for the no flow case and increasing

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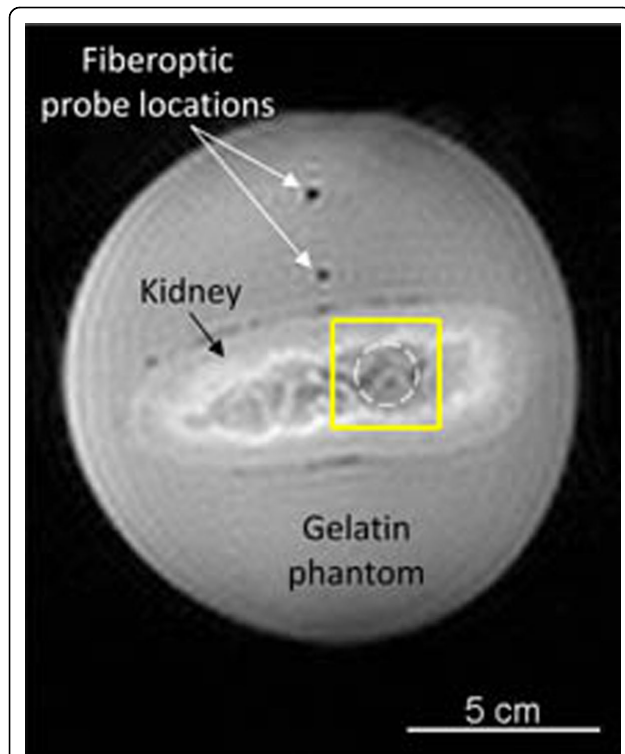


Figure 2 Coronal magnitude image obtained during MRgFUS heating. Fiberoptic probes measured the background temperature. The dashed line indicates the circular heating region and the solid line identifies the region of interest for data presented in figures 3 and 4.

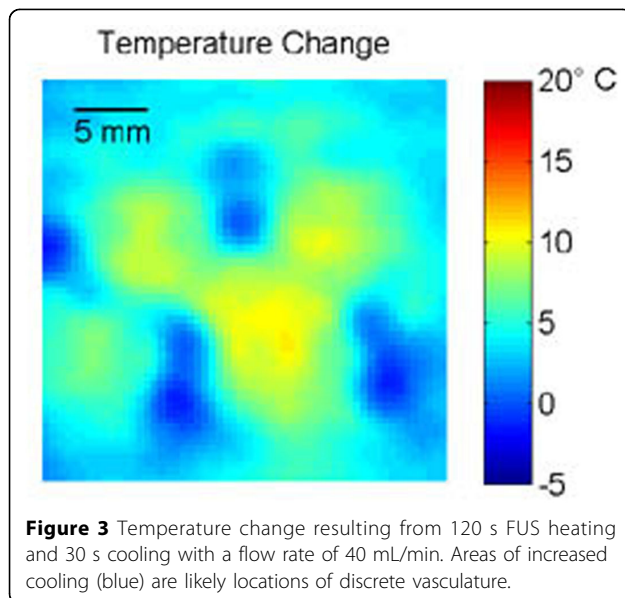


Figure 3 Temperature change resulting from 120 s FUS heating and 30 s cooling with a flow rate of 40 mL/min. Areas of increased cooling (blue) are likely locations of discrete vasculature.

with flow rate. Future work will relate MR perfusion imaging to Q_b , which should eliminate the need for tissue heating for improved biothermal modeling. This study demonstrates that obtaining perfusion estimates

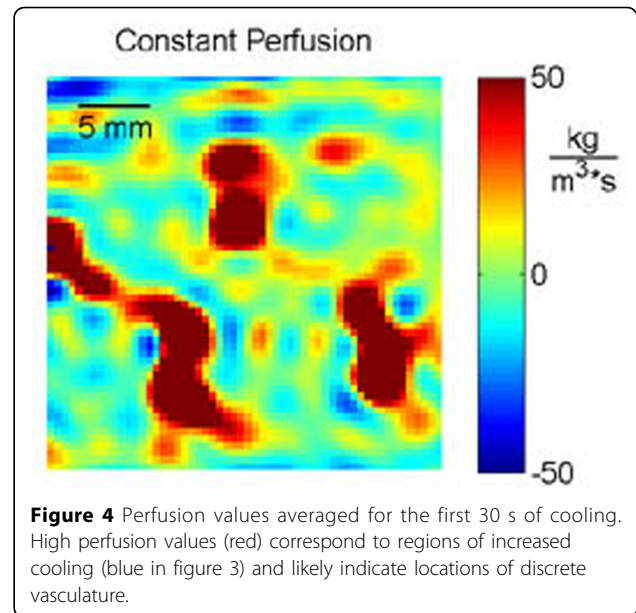


Figure 4 Perfusion values averaged for the first 30 s of cooling. High perfusion values (red) correspond to regions of increased cooling (blue in figure 3) and likely indicate locations of discrete vasculature.

from 3D MR temperature data during MRgFUS is feasible and has the potential to improve biothermal models of MRgFUS fibroid treatments.

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