Optimizing Spatial Resolution for MR Thermal Imaging of Transurethral Ultrasound Prostate Ablation

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Introduction

Complete ablation of cancer in the prostate gland under MRI guidance while sparing normal tissues may be facilitated by concurrent transurethral high-intensity ultrasound treatment delivery and magnetic resonance thermal (MRT) imaging, which afford real-time control over the ablation procedure. To date, a systematic determination of the optimal parameters such as pixel size in MRT imaging for prostate therapy has not been studied. This is important because MRT imaging differs from anatomical MRI where accurate depiction of sharp tissue boundaries demands high spatial resolution. In MRT imaging the temperature distribution being imaged is slowly varying in space and does not tend to have the sharp edges that anatomical images do. The goal of this work is to refine MRT imaging during treatment with transurethral high-intensity ultrasound by determining the optimal MRT imaging spatial resolution to accurately represent realistic temperature distributions, while still maintaining high signal to noise ratio (SNR) and short imaging times which allow for high temporal resolution and multiple slice acquisitions.

Methods

The approach of this study was to simulate a temperature distribution, Fourier transform it to k-space, add noise, take the inverse Fourier transform, and calculate the root mean squared difference between the simulated temperature distribution and the inverse Fourier transform. Temperature distributions of a 90° sectored tubular and a planar transurethral applicator[1] are calculated[2], informed by measured temperature distributions from *in vivo* canine prostate studies. The SNR of typical prostate studies with 1.25 mm x 0.6 mm pixels was measured from MRT images acquired with an endorectal coil on a 0.5T interventional MRI scanner (Signa SP, GE Medical Systems, Milwaukee WI). The measured SNR range was extrapolated to other pixel sizes using the linear dependence of SNR on pixel size with other imaging parameters fixed. Simulated signals with SNR levels in this experimentally relevant range were created by adding white noise to the Fourier transform of the calculated temperature distribution. The inverse Fourier transform of the simulated signal was then performed with various lowpass filters to simulate MRT images acquired at different image resolutions (i.e. different pixel sizes). The simulated MRT images with different pixel sizes were compared to the noiseless simulated temperature distributions by calculating the root mean squared (RMS) difference between the two. We defined the overall optimal pixel size to be the one that minimized the average RMS difference over the experimental SNR range.

Results

Figure 1 is a contour plot of the simulated temperature (°C) distribution of the 90° sectored tubular applicator at 8 MHz. For this simulation blood perfusion is 5 kg m^{-3} s^{-1}, the ultrasound absorption coefficient, α, is 8 Np m^{-1} MHz^{-1}, and dynamic changes in blood perfusion and acoustic attenuation during accumulation of thermal dose were incorporated, as well as the effects of transurethral and endorectal cooling with ambient temperature water. In Figure 2 the center profile through the heated portion of the simulated temperature distribution from the tubular applicator is shown. The simulated MRT images with 0.5 mm, 1.7 mm, and 6.4 mm pixel sizes are also shown. Figure 3 is a contour plot of the RMS difference for the tubular applicator over a range of pixel sizes and SNR levels. The measured SNR range is indicated by the straight dotted lines. As shown by the hatched line, high SNR levels allow smaller optimal pixel sizes, while low SNR levels require larger pixel sizes. For the tubular applicator, the optimal pixel sizes within our measured SNR range vary from 1.4 mm to 1.9 mm, and the overall optimal pixel size across the experimental SNR range is 1.7 mm. For the planar applicator optimal pixel sizes range from 0.8 mm to 1.5 mm, and the overall optimal pixel size is 1.2 mm.

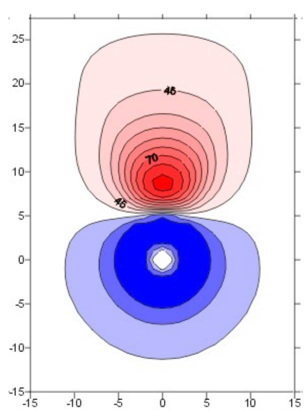

Figure 1: Calculated temperature distribution(°C) for tubular applicator.

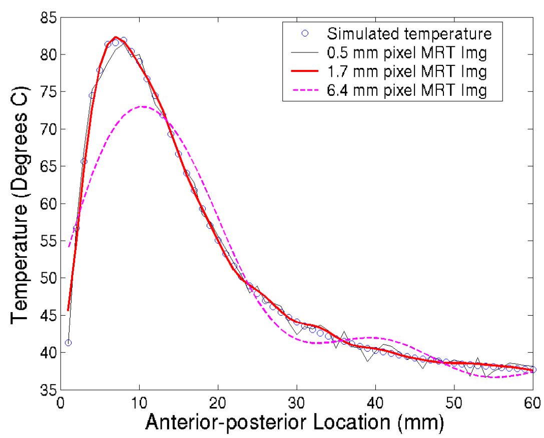

Figure 2: Center profile of temperature distribution.

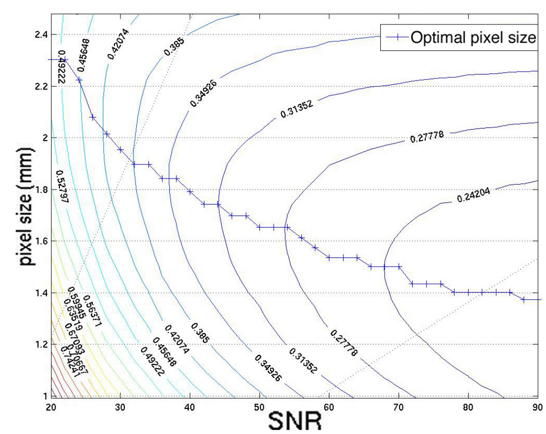

Figure 3: RMS difference (°C)

Discussion

This work provides quantitative justification for the pixel sizes used in MRT imaging. Small pixel sizes are noisy while larger pixel sizes do not adequately represent the underlying heat distribution. Optimal pixel sizes depend on the underlying heat distribution (heat source, anatomical treatment site) and SNR levels (imaging system and parameters). The SNR advantage of larger pixels cannot be fully realized by retrospectively adjusting reconstruction parameters, therefore MRT image acquisition should be adjusted to reflect prospective analysis and optimization.

References

[1] Ross AB, Diederich CJ, Nau WH, Gill H, Bouley D, Daniel B, Rieke V, Butts RK, Sommer FG, Highly directional transurethral ultrasound applicators with rotational control of MRI guided prostatic thermal therapy, Physics in Medicine and Biology, In Press.

[2] Tyréus PD and Diederich CJ, Phys. Med. Biol. 47, 1073–1089, 2002.

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