Dynamical model parameter adjustments in model predictive filtering MR thermometry

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INTRODUCTION

In magnetic resonance guided focused ultrasound (MRgFUS) brain applications the fully insonified field-of-view (FOV) is ideally monitored. This can be achieved in a 3D GRE pulse sequence by combining k-space subsampling and a model-based reconstruction method, such as the previously described model predictive filtering (MPF) [1]. MPF uses the Pennes bioheat transfer equation (PBTE) [2] and tissue thermal and acoustic properties determined from a low-power pre-treatment heating (which ideally does not deliver any thermal dose, i.e. ΔT < 6 °C). The accuracy of the determined tissue parameters, and hence of the MPF reconstruction, depends on the low power heating. In this work we investigate dynamical adjustment of model parameters during heating to compensate for errors in model parameters for improved MPF temperature measurement accuracy.

Table 1. *MR and US parameters used for the 3 Low Power heatings (to estimate Q and k), for the fully sampled "truth", and for the subsampled MPF heatings.*
For all scans: TR/TE=22/11 ms, Resolution = 1.15x1.15x2.5 mm, ETL=7, BW = 752 Hz/Px, Flip angle = 15°.

	FOV [mm]	Tacq [s]	FUS
Initial Low Power	147x96x45	4.8	3.6 W 28.18s
Fully sampled "Truth"	147x96x45	4.8	3.6 W 28.18s
MPF	147x110x135	2.4	16 W 28.18s

METHODS

<u>*MPF:*</u> 5 steps are involved in reconstructing temperature maps for time frame (*n+1*): 1) MPF forward predicts the temperature map from time frame (*n*) using the PBTE. 2) The temperature is converted to phase using the standard proton resonance frequency shift (PRFS) equation. 3) The phase is combined with the image magnitude from time frame (*n*) to create a complex image, which is Fourier-transformed into k-space. 4) The acquired k-space lines for time frame (*n+1*) replace the modeled lines. 5) The PRFS equation is used to calculate updated temperature maps.

<u>*Implementation:*</u> The acoustic (power density, *Q*, in W/m^3) and thermal (conductivity, *k*, in W/m°K) properties in the PBTE were determined with recently published methods [3, 4] utilizing an initial low power heating. These methods yield a 3D estimate of *Q* and a scalar estimate of *k*. Tabular values for density ρ (1000 kg/m^3) and specific heat C (3635 J/kg/°C) were assumed. To minimize thermal dose accumulation the temperature rise of the initial low power heating was kept below 6 °C, and was repeated three times for improved SNR.

<u>*MR imaging:*</u> All imaging was performed on a Siemens Tim Trio 3T MRI scanner (Erlangen, Germany) with a segmented echo planar imaging pulse sequence with gradient recalled echo (GRE) readouts (**Table 1**). k-space was subsampled with a recently described variable density subsampling scheme where the center of k-space is sampled more often than the higher frequencies [5]. A subsampling factor of *R*=7 was used for all imaging and all image data was zero-filled interpolated to 0.5 mm voxel spacing to mitigate partial volume effects.

<u>*FUS heating:*</u> Single point FUS heatings were performed in a gelatin phantom with an MR-compatible phased-array US transducer (256 elements, 1 MHz, 13cm radius of curvature, 2x2x8 mm focal spot FWHM, Imasonic, Besançon, France, and Image Guided Therapy, Pessac, France), **Table 1**.

<u>*Experiment:*</u> The model parameters in the PBTE were dynamically updated during heating (*Q*) and cooling (*k*) based on comparison between the modeled temperatures (step 2) above) to the MPF temperatures (step 5) above) under the assumption that the MPF temperatures will be closer to "truth" since a fraction of k-space is updated every time frame. During heating the magnitude of *Q* was adjusted based on the difference in model and MPF maximum temperature, and the width of *Q* was adjusted based on the difference in FWHM of the model and MPF temperatures. During cooling *k* was adjusted based on the difference in FWHM of the model and MPF temperatures. As "truth", a smaller FOV fully sampled volume was acquired using identical sonication parameters. Three different reconstruction cases were compared: **a)** <u>**No adjustment**</u>, i.e. using fixed *Q* and *k* values for all time frames. **b)** <u>**Adjusting next time frame**</u>, i.e. the comparison of model and MPF temperatures made in time frame *n* were used to adjust *Q* and *k* for reconstructing time frame *n+1* (no added reconstruction time). **c)** <u>**Adjusting present time frame**</u>, i.e. the comparison of model and MPF temperatures made in time frame *n* were used to re-reconstruct this time frame once with adjusted values for *Q* and *k* (effectively doubles the reconstruction time). The reconstruction time in non-optimized Matlab implementation was approximately 1.2 s / time frame. Temperature measurement accuracy was evaluated by investigating local focal spot (hottest voxel and voxels with ΔT>20°C) and more global (voxels with ΔT>6°C) temperature root-mean-square-errors (RMSE).

RESULTS

Mean and STD of the temperature rise for the three low power heatings used for tissue property estimations were 5.46±0.05 °C, resulting in a thermal dose of 0.26 CEM43 for the hottest voxel. *k* was estimated as 0.569 W/m°K, compared to 0.549 W/m°K using an invasive method (KD2 Pro, Decagon Devices, Pullman, Wa). Maximum and FWHM of *Q* was estimated as 0.65 mW/m^3 and 1.35 mm, compared to 0.39 mW/m^3 and 1.79 mm as estimated using higher power (16 W). Table 2 shows the RMSEs for the three reconstruction cases and it can be seen that the *adjust present* case performs best, and that the difference in error is larger for the local focal spot errors. Figure 1a shows the temperature rise as a function of time for the hottest voxel, and the error compared to fully sampled "truth" is shown in Figure 1b.

Table 2. *RMSE of the hottest voxel, and all voxels that experienced temperature rises greater than 6 and 20 °C, for the three reconstruction cases.*

	RMSE Hottest voxel	RMSE ΔT>6□C	RMSE ΔT>20□C
a) No adjustment	0.71±0.12	0.55±0.02	1.25±0.06
b) Adjust next	0.62±0.14	0.54±0.02	1.08±0.10
c) Adjust present	0.47±0.10	0.52±0.03	1.04±0.14

DISCUSSION AND CONCLUSION

It is challenging to accurately estimate *k* and *Q* from low power heatings, especially when attempting to deliver minimal thermal dose. With the 3 x 3.6 W heatings in this study *k* was accurately estimated (within 4%), but *Q* magnitude and width was over- and under-estimated, respectively (67% and 25%). This work has shown that increased temperature measurement accuracy can be achieved by adjusting model parameters during heating. Greatest improvements are seen if each time frame is re-reconstructed once with updated parameters, which in the present study could be done during the acquisition time. Improvements are greatest for voxels close to the focal spot center, with improvements up to 34% and 17% for the hottest voxel and voxels with ΔT>20 °C, The overestimated temperatures in Figure 1a suggest that *Q* (magnitude and width) is not fully corrected, whereas the small error during cooling in Figure 1b, less than 0.25 °C for all adjusted time frames, suggests that the dynamic adjustment of *k* works well.

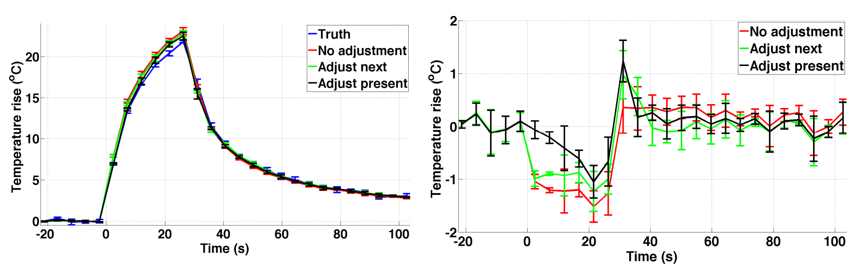

Figure 1. *a) Mean and STD (n=3) of temperature rise for the hottest voxel for fully sampled "truth" and the three variations of MPF reconstruction. b) Mean and STD of temperature error for the hottest voxel for the three variations of MPF reconstruction.*

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