Corrections and Calibration of MR Thermography for Hyperthermia Monitoring in the Hyperthermia/MR Hybrid System

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Introduction:
Deep-seated and largely infiltrating tumors still leave a considerable gap in the oncology, which could be preferably filled by the RF deep-body hyperthermia (HT), mostly applied in conjunction with other treatment modalities [eg. 1]. However, as with other therapy modalities also in HT, verification of the applied dose is the highest imperative. Under many non-invasive approaches proposed in the last years, methods of the MR thermography (MRTh) proved to be most promising (for review see eg. [2]). Thereupon a simultaneous operation of both HT and MR systems was attempted and finally achieved in form of the clinical HT/MR hybrid system [3]. Given the meanwhile evident applicability of the MRTh methods for ablation monitoring and their sufficient accuracy (<1°C) in diversely heated small objects [eg. 4], their use for monitoring slow and non-ablative deep heating in large objects in the HT/MR hybrid system is hampered by many artifacts. Their main sources, when applying the MRTh methods based on the proton resonance frequency (PRF) [5] and before considering any in vivo situation, are B0 drifts [6], susceptibility [7] and conductivity changes [8]. In this contribution, we present an approach for correction of errors caused by B0 drifts and conductivity changes.

Methods:
Investigations were performed in a large 3-D pelvic phantom (**Fig. 1a**) and in patients undergoing the HT therapy in the clinical HT/MR hybrid system consisting of two simultaneously operating units: the RF HT unit (BSD-2000-3D, BSD Medical Corp., Salt Lake City, UT, USA) and the MR unit (1.5T Symphony, Siemens Medical Systems, Erlangen, Germany) [3]. The pelvic phantom was built using the plastic presentation skeleton and the outer shell made from acryl glass. Before filling catheters for guiding E-field and temperature sensors were placed. The filling by agar solution was doped to match the electrical and magnetic properties of muscle tissue. HT was applied throughout 30 min in phantom (**Fig. 1b**) and 60 to 80 min in patients. In addition to the heating, MRTh monitoring was performed up to 30 min after turn-off of HT power. The PRF-based MRTh was performed by the volume covering (multi-slice) double echo spoiled gradient-echo sequence (FLASH2D, TR/TE1/TE2/FA=600/4/20/50). Both, phase and magnitude information from the two echoes was used. The B0 drifts were corrected by linear spatial interpolation using the water bolus as a reference volume at known temperature (spatially distributed sensor-based measurements in the water bolus). The segmentation and labeling of the water bolus was performed automatically utilizing the strong T1 contrast of the magnitude images. Spatially distributed errors (offsets of the thermal phase coefficient) caused by temperature-dependent conductivity changes were corrected using the phase difference from the two echoes. After achieving a high linearity and convergence of the correlation plots the final calibration was performed.

Results:
The PRF-measured temperature without any corrections is underestimated and nonlinear due to B0 drifts (**Fig. 1c,d**). After drift correction (**Fig. 2a**) the non-linearity disappears, but an overestimation increasing medially occurs and a divergence of the correlation plots remains, both due to radially dependent B1 retardations (**Fig. 2b**). After this correction the correlation plots do not diverge more and can be finally calibrated (**Fig. 2c,d**) resulting in calibration constants of –0.011 and –0.009 ppm/°C for bolus water and the phantom filling, respectively. The corresponding values from the patient measurements, which are of comparable regression accuracy, were measured only in the poorly perfused necrotic tumor regions and are between –0.008 and –0.009 ppm/°C.

Conclusions:
After correction of the drift and conductivity errors, the accuracy better than 0.5°C was achieved for the PRF-monitoring of HT in the HT/MR hybrid system in a large phantom and poorly perfused tissue.

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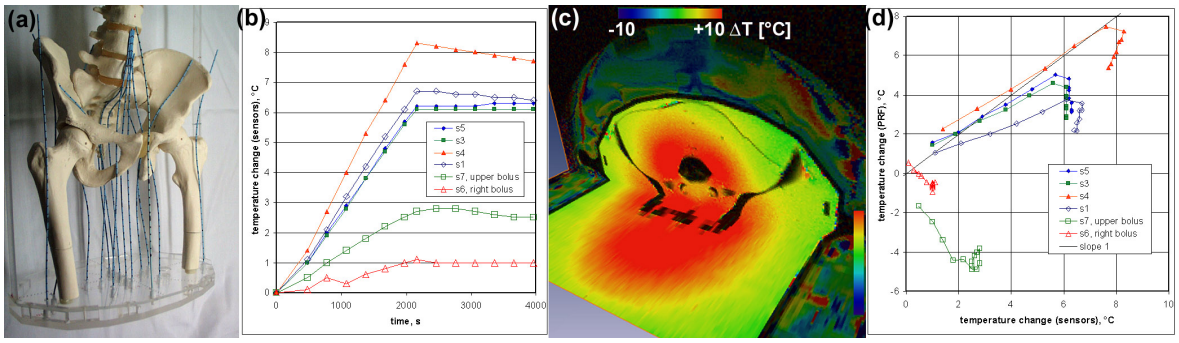

Figure 1. Pelvic phantom before filling by the muscle-equivalent agar solution (a). Evolution of temperature changes in the pelvic phantom and water bolus measured by sensors (b). Example of distribution of the temperature changes measured by the PRF-based MRTh without any corrections (c) and corresponding correlation plots (d).

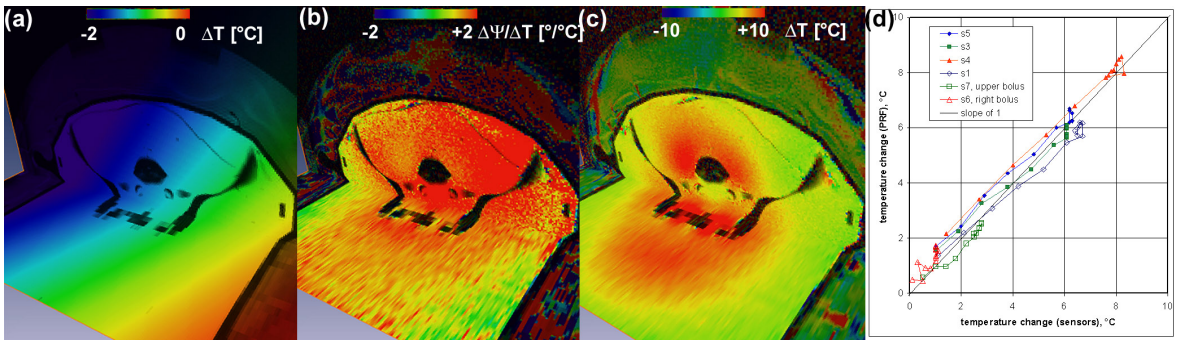

Figure 2. Examples of distributions of the B0 drift as erroneous temperature changes (a), of the offset of the thermal phase coefficient (b), of the temperature changes measured by the PRF-based MRTh after drift and conductivity corrections as well as final calibration (c) and corresponding correlation plots (d).