Real-time Correction of respiratory-induced Field Disturbances for PRFS-based MR-Thermometry in the Human Breast

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Introduction

Real-time MR-thermometry based on the proton resonance frequency (PRF) technique [1] allows monitoring of the local temperature evolution during radio-frequency, laser, cryogenics or focused ultrasound thermal ablation. Commonly, PRF-based MR-thermometry relies on the evaluation of phase differences between sequentially acquired images. However, for the particular case of MR-thermometry of the human breast, magnetic field variations induced by the respiratory cycle lead to phase fluctuations, which require a suitable correction strategy to prevent thermometry artefacts. For this purpose a look-up-table-based multi-baseline correction algorithm [2] is applied to MR-thermometry to correct for the periodic B_0-field changes. The proposed correction method is compatible with a variety of sensors monitoring the current respiratory state, such as respiratory pressure sensors, navigator data derived from images [3] or pencil-beam navigators [4]. The ability to remove phase artefacts during MR-thermometry in real-time in the human breast is demonstrated experimentally in a healthy volunteer during 5min of free-breathing using pencil-beam navigators for respiratory control.

Materials and Methods

MR-thermometry using the PRF method is based on the calculation of a relative temperature change between the current image and a reference image given by $\Delta T=(\varphi-\varphi_{ref})/(B_0\gamma\cdot\alpha TE)$ (1) ,where φ and φ_{ref} denote the phase of the current image and the reference image, B_0 is the main magnetic field strength, γ is the gyromagnetic ratio of protons, TE is the echo time and α gives the temperature dependence of the proton resonance frequency, where α=0.094ppm was used in this work. The applied algorithm improves this approach by correcting the current phase with a pre-registered correction which corresponds to the same respiratory state. The correction procedure can be divided into two parts:

Learning phase: The amplitude of the respiratory motion is measured by a pencil-beam navigator positioned across the diaphragm during several respiratory cycles, prior to MR-thermometry. Simultaneously, the magnitude and phase images are stored in a look-up table according to the navigator signal.

MR-thermometry: For every new image the current respiratory amplitude is compared with those in the look-up table. The corresponding phase of the match is chosen as the reference phase φ_{ref} for the temperature calculation.

All experiments were performed on a 1.5 T Achieva clinical scanner (Philips Medical Systems, Best, the Netherlands). A healthy female volunteer was placed in prone position and a 16-cm-diameter surface coil was used for signal registration. For PRF-thermometry a 2D EPI multi-shot gradient echo sequence (TE=10ms, TR=21ms, flip angle 15°, FOV=350x158mm², slice thickness 5mm, coronal, 7 k-space lines per excitation) was applied. In total 1000 dynamics were acquired with an acquisition time of 335ms per image. The position of the diaphragm was measured before every excitation pulse with the pencil-beam navigator of the Achieva platform. The resulting position together with phase and magnitude images were streamed with the IMF interventional RT-toolkit to an in-house developed real-time reconstructor which performed phase corrections and temperature calculations. For demonstration purposes a non-water-selective excitation pulse was used but the presented method was also verified in glandular tissue using a 121 binomial water-selective pulse (data not shown).

Results and Discussion

The observed periodical magnetic field changes of up to 0.15 ppm correspond well with the values reported by Sprinkhuizen et al [3] and lead to phase changes as shown in Fig. 1. The corresponding artefactual temperature fluctuations show a magnitude of up to 30°C peak to peak.

Using (1) the phase changes can be exploited to calculate maps of the resulting apparent temperature change although no temperature change was present. Figures 2 a and 2 b show an example of the temperature evolution for one pixel within the breast (white arrow). Note that the temperature values have been corrected for offsets and linear temperature drift. Without any correction the temperature oscillates with a standard deviation of 8.2°C following the breathing amplitude. For comparison, the red line shows the theoretical temperature precision limit given by $\sigma_T=\sqrt{2}/(SNR\cdot B_0\gamma\cdot\alpha TE)$ (2) . The proposed correction is able to remove the respiratory induced oscillations within the theoretical limit leading to a reduced standard deviation of 2.8°C as shown in Fig. 2 b. Spatial maps of the temperature standard deviation in the left breast (Fig. 2 c,d) also demonstrate the overall improvement of temperature accuracy if the correction is applied.

Figure 1: The coronal phase images of the left breast acquired at the start of the respiratory cycle, after 2s and after 4s demonstrate the influence of the breathing cycle on the image phase.

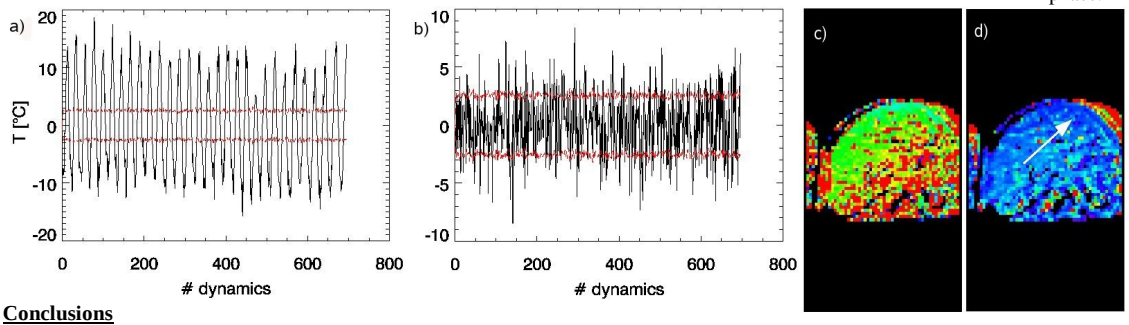

Figure 2: a and b: Evolution of the uncorrected and corrected temperature for one pixel (white arrow in d). The red line shows the theoretical limit for the temperature accuracy calculated from (2). c and d: Map of the standard deviation of the uncorrected (c) and corrected (d) temperature in the left breast.

Conclusions

The applied look-up-table-based multi-baseline approach is able to correct MR-thermometry in the breast in real-time. The resulting accuracy of the temperature maps is within the theoretical limit given by the SNR of the acquired image. Provided that the employed imaging sequence is able to resolve the respiratory cycle, the proposed method is compatible with different motion sensors and sequence types. Hence, this correction procedure provides a simple mean for accurate real-time MR-thermometry as a step towards MR-guided thermal ablations of breast cancer.

References

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