Novel PRF thermometry method using spatially selective 2DRF excitations and a parametric model

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Introduction: We present a novel method to monitor the temperature of the target tissue during real-time-MRI-guided radiofrequency (RF) ablation. During RF ablation, commonly used to treat cardiac arrhythmias such as atrial fibrillation, scars are intentionally created in the atrial wall to isolate the sources/triggers of abnormal electrical activity. Poor imaging guidance, today provided by X-ray and ultrasound, results in the insufficient deposition of RF energy which leads to the incomplete electrical isolation of the AF triggers and, in turn, the recurrence of symptoms. One benefit of performing catheter ablation under MRI guidance is the potential to visualize real-time lesion formation. In principle, MR thermometry could help determine the quality of lesions (size, transmurality, and extent of damage) created while RF ablation is being performed and thereby improve outcomes. Yet, real-time MRI thermometry in RF atrial ablation is currently unachieved.

The many challenges of performing thermometry in the atrium include the motion of the heart, the low SNR due to the thinness of the atrial wall, the closeness to organ interfaces (which causes rapid anatomical phase variations due to the large susceptibility changes and therefore makes referenceless methods more challenging), and the susceptibility artifact created by the RF ablation catheter tip. On the positive side, unlike in other therapies such as HIFU, heating in RF ablation is limited to a compact region of tissue. Furthermore the exact location of the ablation catheter is known due to the use of active tracking coils embedded in the catheter, and therefore the precise location of the hotspot can be well predicted.

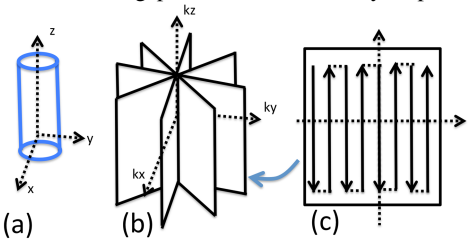

Figure 1: Geometry of 2DRF excitation and EPI readout

A method that is able to estimate temperature without the use of miniature intracardiac coils[1] would be desirable because of the difficulties of constructing an expandable coil, the instability of phase measurements due to coil motion, and the cumbersomeness of the use of such coils.

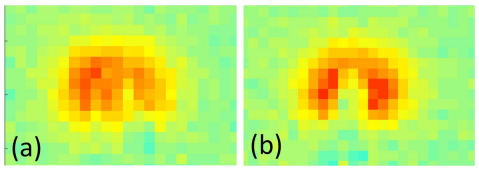

Figure 2: Phase subtraction images of the hotspot from two perpendicular projection views

Methods: Our proposed scheme, exploits this prior knowledge, by performing a spatially local 2D RF excitation of a narrow cylindrical region (diameter < 20mm), as seen in Fig 1a, containing the target tissue. This allows the rapid acquisition of projection images of the excited region using EPI readout as a result of its small phase-encoding field of view (Fig 1c). The temperature map of the heated region is modeled with a small number of parameters and therefore the temperature map is reconstructed from a small number of projection view-angles (Fig 1b). The cylinder is positioned so that its axis aligns with the direction of motion of the target tissue, therefore the central k-space line of the EPI readout can be used to estimate the motion of the target tissue in order to perform baseline subtraction. In RF ablation the operator usually wants to heat the tissue from body temperature (37° C) to above 45° C, and no higher than 65° C. An uncertainty in temperature of a few degrees is therefore acceptable.

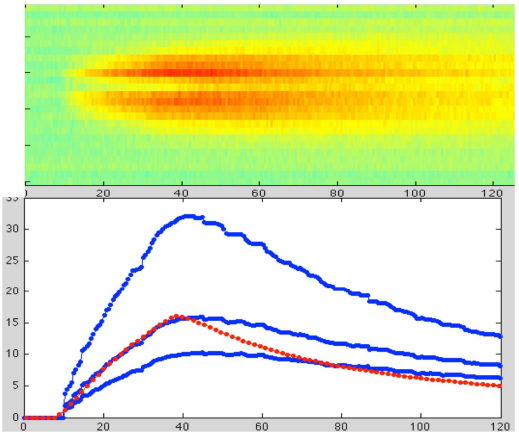

Figure 3: (a) Temporal evolution of single row from Fig 2b (b) Evolution of several pixels (blue) in region of interest compared to optic probe (red). Temperature (Celsius) vs time (seconds)

Results: The method was tested on a custom-made agar phantom, containing sucrose and sodium chloride to simulate a heart, on a 3T Verio MRI scanner (Siemens Healthcare, Erlangen, Germany). An RF generator system (Stockert 70, Stockert GmbH, Freiburg, Germany) was used to deliver power for the RF ablations via a novel, MR-compatible ablation catheter (MRI Interventions, Inc., Irvine, CA). The RF ablation catheter was inserted into the agar phantom, with the tip a few mm away from the air interface. An MR compatible fiber optic probe (Opsens, Quebec City, Quebec, Canada) was placed close to the catheter tip in order to measure temperature during heating.

The imaging sequence was run for a few seconds before the heating was started in order to acquire baseline images. The imaging sequence consisted of a continuous series of 2DRF excitations, each followed by an EPI readout. The nominal diameter of the excitation cylinder was 20 mm, though the short RF pulse duration (12 ms) limits the sharpness of the excitation profile. The cylinder was positioned to contain the catheter tip. For the segmented EPI readout (TE=5ms) an EPI factor of 5 was used with 5 shots comprising a whole image. A PE FOV of 25 mm was used to give a 1mm resolution in the PE direction. The RO resolution was 1.5 mm. TR was 25 ms (i.e. 12ms for 2DRF excitation and 13 ms for EPI readout) and therefore each image was acquired over 125 ms. We believe it is feasible for this acquisition time to be substantially reduced by increasing the EPI factor and reducing the FOV, resolution, and number of shots. The 3D temperature distribution was modeled using a simple Gaussian phase distribution whose amplitude and width were estimated from the phase subtraction projection images using a simple numerical optimization routine. Fig 2 (a) shows two perpendicular projections of the hotspot after 10 seconds of heating. Fig 3(a) shows the temporal evolution of a row from Fig 2 (b). Fig 3 shows the optical probe temperature (in red) compared to the estimated temperature of several pixels in the hot spot region.

Discussion: Because of the simple geometry of our phantom only a single view was used to estimate the 3D temperature map. While the results show the temporal evolution of the temperature estimated by our method to approximately match the evolution of the temperature measured by the optical probe more work is necessary to build a more realistic phase model, using multiple views, and to characterize the error of the estimation method.

References: [1] Volland et al. ISMRM 2012