# Temperature and perfusion monitoring during RF heating of the human calf muscle

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**Introduction** One of the main concerns for patient safety in MR imaging is radiofrequency (RF) induced tissue heating. This heating arises from the intense RF exposure involved in MR imaging and the complex, spatially heterogeneous RF absorption in the human body. Recent simulation studies indicate that even within global SAR limitations in certain regions the local SAR levels and the resulting local temperature can exceed safety guidelines [1,2]. Murbach et al. concluded that local thermoregulation is crucial in this process and should be considered in safety determination [2]. In addition to being topic of investigation, MRI also provides non-invasive measurement techniques to study this local behavior in human volunteers. With Proton Resonance Frequency Shift (PRFS) MR thermometry one can monitor rises in tissue temperatures. Perfusion measurements can be performed without the need for contrast agents, using Arterial Spin Labeling (ASL). This method quantifies the increase of blood flow and its regulatory effect. By performing an MR experiment in which both temperature and perfusion are monitored, we hope to obtain good insight in the local thermoregulatory system of the body.

*Figure 1: The setup inside the scanner*

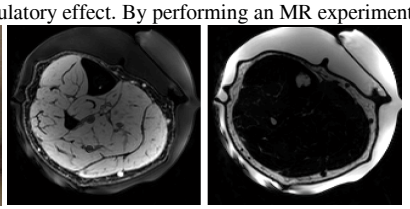

*Figure 2: Dixon water (left) and fat (right) images of the reference bag wrapped around a calf muscle*

**Materials and methods** All measurements were performed on a Philips 3T Achieva MRI system with a dedicated knee coil (Philips SENSE T/R Knee coil 16-elements). A sphygmomanometer bag for the upper leg was filled with sunflower oil and wrapped around the calf muscle of the volunteers (n=2) to use as a non temperature-dependent reference (see Figure 1 and 2). This was used for estimation and compensation of the spatially varying magnet drift over the scan. A multi slice, spoiled gradient echo was used for the PRFS method (flip angle = 40°, TE= 15 ms, TR= 400 ms, 25 dynamics, acq. resolution 2.5x2.5x5 $mm^3$). The phase difference images were compensated for background changes using the method of Salomir et al. [3]. The closed border, which is the input for this method was manually selected within the sunflower oil. The calf was first scanned for 10 minutes at low RF settings (<2% of maximum local SAR in extremities ($SAR_{ext}$)) and then a similar scan followed in which the excitation was preceded by a high amplitude, rectangular off-resonance pulse in order to induce RF heating. In this heating sequence, maximum RF power was applied as allowed by the scanner (≈100% of $SAR_{ext}$). In one subject this scan was followed by exercise of the calf muscle until fatigue. A pulsed ASL scan was used for perfusion measurements. This was done at rest, directly after RF heating and directly after exercise (TR= 4000 ms, TE= 13 ms, acq. resolution 4x4x10 $mm^3$, post label delay 1500 ms). Perfusion was quantified on the ΔM images of the ASL scan [4]. As a control three optical temperature probes (Opsens OTP-M) were placed around the leg on the skin. In addition to the measurements, temperature simulations were performed with SEMCAD X based on the Pennes bioheat equation. For this purpose, a generic 16 elements birdcage coil was modeled with similar dimensions as the knee coil used for the experiments. The coil was placed around the calf of the male model of the virtual family (Duke). Dielectric and thermal tissue properties were assigned according to Hasgall [5], constant core heating was neglected, and starting body temperature was 37°C with a surrounding of 21°C. The model was simulated for 30 min without heating to obtain equilibrium conditions. The applied RF power in the simulations was normalized according to the $B_1^+$ amplitude and the duty cycle used in the heating sequence (20 μT and 6.3% respectively). Next to the perfusion values of muscle from the database (3.9 ml/100g/min), the mean perfusion value of the calf muscle found by ASL was also tested as input for muscle perfusion. The reference bag was not included in the simulations.

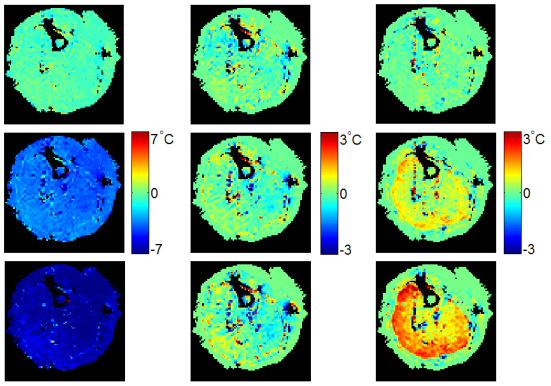


*Figure 3:Temperature maps of the calf muscle. Columns show the temperature change over 10 minutes: left is without compensation and <2% of $SAR_{ext}$, the middle column shows the same data compensated for magnet drift, the right column portrays the compensated scan with ≈100% of $SAR_{ext}$.*

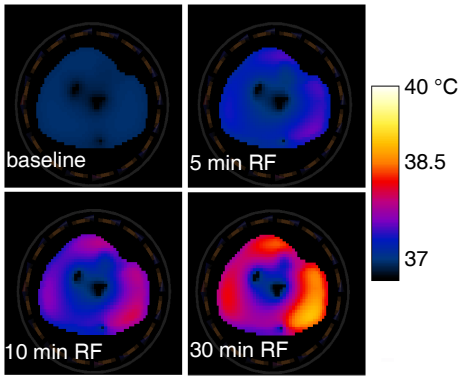


*Figure 4: Heating simulations of the calf muscle in SEMCAD X, perfusion is set to 3.9 ml/100g/min*

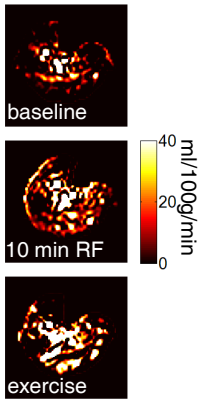


*Figure 5: Perfusion maps measured with ASL*

| | *baseline T* | *10 min RF max ΔT* | *30 min RF max ΔT* |
|---|---|---|---|
| **volunteer 1** | - | +2.5 °C | - |
| **optical probes 1** | 31.7 °C | +1.1 °C | - |
| **volunteer 2** | - | +2.1 °C | - |
| **optical probes 2** | 31.5 °C | +1.3 °C | - |
| **simulation (1.7 ml/100g/min)** | - | +1.2 °C | +2.4 °C |
| **simulation (3.9 ml/100g/min)** | - | +1.1 °C | +1.9 °C |

**Results and discussion** During the scan without RF, no heating is observed (-0.05°C with a standard deviation of 0.8) and skin temperature stays around 31-32°C. During the heating sequence the temperature clearly increases; mainly in the outer regions (Figure 3). This is due to the quadrature drive of the coil which results in destructive interference of the electric field in the center. This heating pattern is also observed in the simulations (Figure 4). Measured and simulated temperatures did not exceed IEC safety guidelines for extremities (40°C). The probes showed a smaller maximum temperature increase than the MR thermometry (see the table above) which might be due to the cooling of the skin by the bag filled with sunflower oil. The average perfusion level of one subject at baseline was measured to be 1.7 ml/100g/min over a single slice, which was also used as input for the simulations. After RF heating this had increased to 3.4 ml/100g/min with the largest differences occurring predominantly in the region of maximum heating (Figure 5). The perfusion increase induced by calf muscle exercise until fatigue was in the same order (mean: 3.8 ml/100g/min). We did not observe clear perfusion changes in the other volunteer. We believe this is related to the low perfusion of the calf muscle, which can cause the ASL scans to have a low SNR. The simulations with database values and measured perfusion values showed a minimal difference after 10 minutes of RF exposure. However, after 30 minutes the difference was more pronounced, proving the importance of perfusion values in simulations. The simulated maximum temperature rise after 10 minutes was approximately 1°C lower than MR thermometry values. To match simulations better with experimental conditions, we plan to make subject specific models using Dixon water/fat images and ASL perfusion maps.

**Conclusions** During a high SAR scan at clinical settings RF heating can lead to significant heating of the calf muscle. The MRI techniques presented offer the opportunity to measure and study the effect of RF absorption on local tissue temperature and perfusion. This is a first step in understanding the thermoregulatory system that comes into play during RF heating, and can help in making simulations more accurate.

*Refs. [1] van Lier, A.L.H.M.W. et al., jMRI 2012; 35:795–803 [2] Murbach, M. et al., Proc. Intl. Soc. Mag. Reson. Med. 2012; 20:313 [3] Salomir, R. et al., IEEE Trans Med Imaging 2012; 31:287-301. [4] Luh, W.M. et al. MRM 1999; 41:1246–1254 [5] Hasgall, P.A. et al. 2012;www.itis.ethz.ch/database.* *Acknowledgments: ZonMw program Electromagnetic Fields and Health-Basic Research*