Calorimetric Calibration of SAR Estimates Displayed on GE MR Scanners

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Introduction: Specific absorption rate (SAR) as estimated by an MR scanner is a useful predictor of RF heating (1). Those calculations, however, are highly model-dependent. Manufacturers of implanted devices (e.g., DBS stimulators) sometimes specify the maximal allowable SAR values for safe MRI scanning of their devices (2). The purpose of this work is to use calorimetry (3) to refine the model-dependent SAR estimates provided by the scanner. This is important because being overly restrictive in SAR can preclude the use of clinically useful imaging protocols, while not being restrictive enough can lead to injury or death of the patient (4).

Theory: Phantom temperature T (in ^{o}C) as a function of time t (in seconds) can be modeled by the differential equation

$$\frac{dT}{dt} = -k(T - T_A) + \frac{e\,S}{c}, \qquad [1]$$

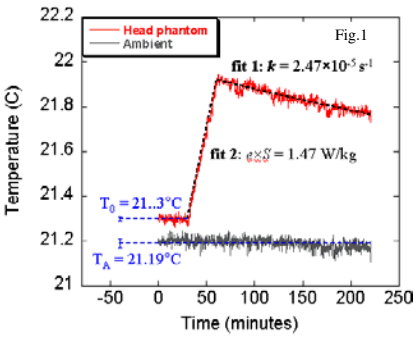
where k is Newton's cooling constant in s^{-1}, T_A is the ambient temperature in ^{o}C, S is the average head SAR calculated by the scanner in W/kg, e is the dimensionless coil efficiency, and c is the heat capacity (i.e., 4186 J kg^{-1} $^{o}C^{-1}$ for water). The product $e \times S$ represents the actual, average head SAR that is applied. Assuming that the application of RF power begins at $t = 0$, Eq. [1] has the solution

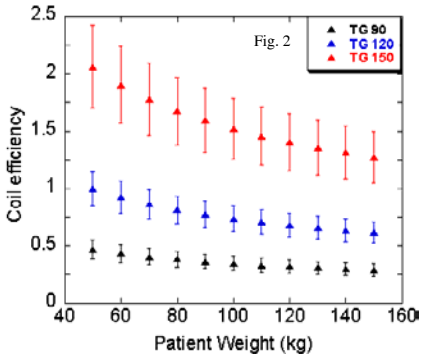
$$T(t) = T_0 \exp\left(-kt\right) + \left(T_A + \frac{e\,S}{k\,c}\right)\left(1 - \exp\left(-kt\right)\right), \qquad [2]$$

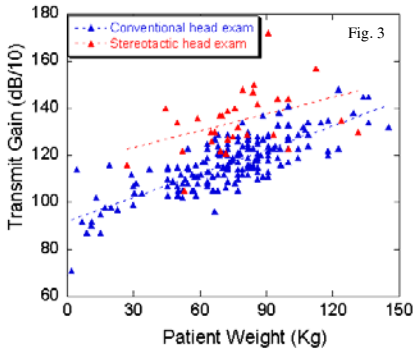
where T_0 is the phantom temperature at $t = 0$.

Methods: Calorimetric tests were performed to measure the temperature increase due to the deposition of RF energy during MRI scanning. The measurements were performed on three equivalent clinical 1.5T General Electric MR scanners running 11.0 M4 software, using standard transmit-receive head coils and a thermally insulated 4-liter cylindrical phantom (20cm diameter, 50cm length) filled with 0.45% NaCl solution. The temperatures were recorded simultaneously at four probe locations using an MR-compatible Model 750 Fluoroptic thermometry system (Luxtron, Santa Clara, CA). Three of the probes were inserted into the head phantom. The fourth probe, measuring the ambient temperature, was placed in an open container with water outside the magnet. The relevant imaging parameters included: RF spin echo with 4-echoes, TR/TE1/TE2/TE3/TE4 100/20/40/60/80ms, 256×256 matrix, a single 3mm-slice, and 70 signal averages to prolong the scan time to 30min. The fan inside the bore was turned off, and the gradients were disabled to avoid possible damage caused by their continual activation during a 30 minute-long sequence. The measurements were performed for a range of three RF output power levels (i.e., transmit gain - TG): 90, 120, and 150 (in units of 0.1dB) that were matched to values typically observed in clinical head scanning. The scanner's SAR estimate S was recorded for patient weights 50-150kg. Before scanning, the system was allowed to thermally stabilize for at least 30min. The temperature measurement continued approximately 3 hours after scanning to allow determination of the cooling constant k by fitting to Eq. [2] with $S = 0$. After k is determined, the coil efficiency e was determined by fitting the measured temperature history during application of RF to Eq. [2].

Results: Fig. 1 shows a typical experimental run for TG = 150. Fig. 2 shows the fitted coil efficiencies, where the error bars indicate the standard deviation of the measurements from three scanners. The measured coil efficiency depends on patient weight because S in Eq [1] does as well. Fig. 3 shows a plot of TG (as optimized by







the scanner's standard autoprescan measurement) versus patient weight for 212 consecutive patients. Figure 3 generally indicates that TG increases with patient weight, which is consistent with the results of Fig. 2. Note that the presence of a stereotactic head frame used in our Practice tends to increase the TG value, suggesting that the frame detunes the head coil, i.e., it absorbs some RF power itself or causes the reflected power to increase.

Discussion: A very wide range of efficiencies is (0.3-2) is observed in Fig.2. As indicated on Fig. 3, however, this full range of efficiencies would not be used, because TG generally increases with patient weight.

Suppose the patient has an implanted device that (per the manufacturers recommendation) requires the actual deposited average head SAR be kept below 0.1 W/kg with the transmit/receive head coil. The patient weight is correctly entered into the scanner, and autoprescan is run as usual on the patient. Based on the measured TG value, the coil efficiency is estimated from Fig 2. For example, if the patient weighs 80kg and TG = 120, then from Fig. 2 we estimate that the efficiency is approximately 0.8. Therefore we will allow the use of pulse sequences with calculated SAR up to approximately 0.1/0.8 ≈ 0.12 W/kg on this particular patient.

Although we have confidence in this methodology, ***we strongly caution other sites that they should not use the results of Fig. 2 without first replicating this experiment on their own MRI system.*** The validity of the results depends on 1) the assumption that the TG is accurately calibrated on the scanner. This calibration is a standard service procedure, but it can drift over time or become unstable if the RF amplifier is defective. 2) The RF amplifier, head coil, and SAR model all match the ones that we used. In our experience, an equipment vendor can modify the SAR model between software releases, and SAR models can vary widely among vendors.

Conclusion: The efficiency of a T/R head coil was calibrated as a function of RF transmit gain and patient weight. A wide range efficiencies was observed, indicating that this methodology is a useful refinement compared to simply using the model-dependent SAR value that is displayed by the MRI system.

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