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Heating Induced near Deep Brain Stimulation Lead Electrodes during Magnetic Resonance Imaging with a 3T Transceive Volume Head Coil

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Abstract

Heating induced near deep brain stimulation (DBS) lead electrodes during MRI with a 3T transceive head coil was measured, modeled, and imaged in three cadaveric porcine heads (mean body weight = 85.47 ± 3.19 kg, mean head weight = 5.78 ± 0.32 kg). The effect of the placement of the extra-cranial portion of the DBS lead on the heating was investigated by looping the extra-cranial lead on the top, side, and back of the head; and placing it parallel to the coil's longitudinal axial direction. The heating was induced using a 641 s long turbo spin echo sequence with the mean whole head average SAR of 3.16 W/kg. Temperatures were measured using fluoroptic probes at the scalp, first and second electrodes from the distal lead tip, and 6 mm distal from electrode 1 ($T_{6\text{mm}}$). The heating was modeled using the maximum $T_{6\text{mm}}$ and imaged using a proton resonance frequency shift based MR thermometry method. Results showed that the heating was significantly reduced when the extra-cranial lead was placed in the longitudinal direction

compared to the other placements (peak temperature change = 1.5–3.2 °C vs 5.1–24.7 °C). Thermal modeling and MR thermometry may be used together to determine the heating and improve patient safety online.

Keywords

MRI; Safety; DBS; Heating; 3T

INTRODUCTION

Clinically harmful heating induced near deep brain stimulation (DBS) lead electrode – tissue contacts during magnetic resonance imaging (MRI) is a safety concern. Timevarying electromagnetic (EM) field produced during MRI induces current in a conductive lead. This current dissipates in the adjacent tissue through the lead electrode-tissue contacts producing localized heating. The heating depends on the thermodynamic and EM properties of a lead and adjacent tissue, MRI coil, the lead placement with respect to the coil, and blood flow (Baker et al., 2006, Georgi et al., 2004, Rezai et al., 2002, Finelli et al., 2002).

Developing DBS lead implantation and MRI protocols to reduce the MRI induced heating near electrode-tissue contacts is urgently needed to improve safety of patients implanted with DBS devices. Developing such ‘induced-heating-safe’ lead implantation and MRI protocols requires experimentation in suitable animal models and the development of validated, non-invasive engineering tools (i.e., a bioheat transfer model together with appropriately implemented Maxwell equations and an MR thermometry method) to determine in vivo temperatures near conductive medical devices during MRI. Experimentations are needed to develop and validate accurate engineering tools and to better understand the effect of local temperature change on the mammalian brain and body. A first principles based bioheat transfer model together with appropriately implemented Maxwell equations is necessary to design and develop devices and clinically feasible protocols to keep the heating below safe thresholds. Once such devices and protocols are developed, a suitable MR thermometry method is required to image temperatures and verify safety of a patient after an MRI session. Imaging temperatures using endogenous contrast based MR thermometry techniques within the susceptibility artifact region around a conductive device is challenging. Using MR thermometry outside the susceptibility artifact range and thermal modeling within the artifact range with MR temperatures as boundary conditions may be an attractive non-invasive option to improve online safety.

This preliminary work, first, measures heating near DBS lead electrodes in cadaveric porcine heads during imaging with a 3T, transceive volume head coil due to various placements of the extra-cranial portion of the lead. A clinically feasible protocol was identified to place the extra-cranial DBS lead such that to minimize heating near DBS lead electrodes during imaging at 3T at the maximum allowable whole head average specific absorption rate (SAR). Second, the maximum temperature measured outside the susceptibility artifact range of the DBS lead was used in a simple thermal model to simulate time-varying temperature distribution around DBS electrodes to check the feasibility of such models in predicting heating when accurate temperatures outside the susceptibility range are available. Third, thermal resolution constraints for an MR thermometry method were estimated using the thermal model to develop appropriate MR thermometry online safety protocols. Fourth and last, the predictions of the thermal model were compared qualitatively to the temperatures imaged using endogenous proton resonance frequency (PRF) shift based MR thermometry technique to investigate the feasibility of using the two methods together to determine heating near DBS electrodes.

MATERIALS AND METHODS

DBS Lead

The DBS lead used in this work (model number 3389, Medtronic Inc., Minneapolis, overall length = ~46 cm, width = 1.1 mm) was a coiled wire insulated using polyurethane with four bare electrodes near the tip. Each electrode was measured as ~1.5 mm long. Inter-electrode spacing was measured as ~1.0 mm (Figure 1). First and the distal most DBS electrode was located 1.0 mm away from the DBS lead tip.

Extra-cranial DBS Lead Placement

The current work studies heating induced near DBS lead electrode-tissue contacts during imaging with a 3T head coil with the extracranial lead looped on the top, side, and back of the porcine head; and placed parallel to the coil's longitudinal axial direction. Typically, in a clinical situation, part of the DBS lead with the electrodes near the tip is implanted in the brain (intra-cranial DBS lead) by drilling a burr hole through the skull. The rest of the DBS lead (the extra-cranial part) after exiting the skull is looped under the scalp and near the burr hole. The extra-cranial DBS lead can be looped in three perpendicular directions with respect to an imaging coil and can be run parallel to the coil (Rezai et al., 2002, Bhidayasiri et al., 2005, Baker et al., 2005). The extra-cranial lead was looped only once (i.e., number of turns = 1) on the porcine head to maximize the loop area and thus, induced currents and heating.

During imaging with a transceive head coil, the placement of the extra-cranial portion of the DBS lead is expected to affect heating near DBS lead electrodes the most due to its relative closeness to the coil compared to the rest of the device (i.e., a connecting lead and a generator). This is so because the extra-cranial portion of the DBS lead can be placed such that the rest of the DBS stays relatively away from the coil and thus, does not affect the heating as much.

Porcine Model

Porcine brains were used instead of tissue mimicking gels to measure heating near DBS electrodes. Use of the porcine brain inherently simulated appropriate electromagnetic properties of the porcine and human brain at 3T (Larmor water proton frequency = ~123 MHz) (instead of preparing gel such that to mimic appropriate electromagnetic properties of the brain at ~123 MHz, a method which is challenging and only approximate) (Gabriel et al., 1996). Thermal conductivity of the porcine head (e.g., epidermis = 0.21 W(mK)^{-1} , fat = $0.16\text{--}0.40 \text{ W(mK)}^{-1}$, brain = $\sim 0.5 \text{ W(mK)}^{-1}$, muscle = $0.46\text{--}0.62 \text{ W(mK)}^{-1}$) is comparable to the thermal conductivity of the human head (e.g., epidermis = 0.21 W(mK)^{-1} , fat = $0.20\text{--}0.22 \text{ W(mK)}^{-1}$, brain = $0.50\text{--}0.56 \text{ W(mK)}^{-1}$, muscle = $0.49\text{--}0.59 \text{ W(mK)}^{-1}$) (Holmes, 1998). Further, use of the un-perfused porcine head provided the worst case heating data near DBS electrodes.

MRI Protocol to Induce Heating

Experiments were conducted on a Siemens' 3T trio scanner. The heating was induced with a 641 s long turbo spin echo (TSE) sequence. Flip angles of the sequence were modified to obtain the scanner reported whole head average SAR of ~3.2 W/kg. The scan time of 641 s was comparable to a typical clinical scan time of 10 min (Larson et al., 2008). The whole head average SAR of 3.2 W/kg was the maximum allowable SAR (IEC, 2010).

Thermal Modeling

A transient, two-dimensional thermal model was used to estimate the temperature change distribution near electrode-brain contacts (Shrivastava et al., 2011, Shrivastava and Vaughan, 2009). The following parameters were used: density $\rho = 1000 \text{ kg/m}^3$, specific heat $C_p = 3600 \text{ J/(kg.K)}$, thermal conductivity $k = 0.5 \text{ W/(m.K)}$, domain dimensions = (120mm) (120mm), boundary temperature change on the domain sides = 0 (Holmes, 1998). Assuming the boundary temperature change as 0 was reasonable since heating near the electrode-brain contacts was localized (details are presented in the results section). The heating source term $Q (\text{W/m}^3)$ was assumed to be located at the electrode-brain contacts. The four source terms were assumed as equal. The magnitude of the source term was determined such that the thermal model simulated the measured temperature change outside the susceptibility artifact range at the end of the TSE sequence (i.e., temperature 6 mm distal from electrode 1, $dT_{6\text{mm}}$) accurately. The effect of the absence and presence of the blood flow was simulated by assuming blood-tissue heat transfer coefficient as 0 and $9000 \text{ W/(m}^3\text{K)}$, respectively. The blood volume fraction and temperature were assumed as 8% and 37°C , respectively (Shrivastava et al., 2011). The two-dimensional thermal model was implemented instead of a three-dimensional model because the two-dimensional model provided sufficient accuracy when compared to the experimental results and was faster to execute.

Next, the thermal model was used to simulate the maximum temperature change outside the artifact range for the maximum temperature change of 1°C at the DBS electrodes as a function of scan time. Current regulatory guidelines recommend maximum allowable in vivo brain temperature change of 1°C (CDRH-FDA, 2003, IEC, 2010, ICNIRP, 2004). Thus, the simulated temperatures are the maximum temperature change that must be imaged accurately and precisely by MR thermometry to comply with regulatory guidelines. The thermal model was also used to simulate the maximum temperature change at the DBS electrodes for 1°C maximum temperature change outside the artifact range as a function of scan time. 1°C is the typical accuracy of the current endogenous contrast based MR thermometry techniques in vivo (Kickhefel et al., 2010, Rieke et al., 2007, Lepetit-Coiffe et al., 2006). The simulated temperatures present the ‘grave’ potential of tissue damage near DBS electrodes during MRI in the absence of more accurate and precise MR thermometry.

MR Thermometry

MR thermometry was performed by measuring phases using a gradient recalled echo (GRE) sequence with the following parameters: TR = 70 ms, TE = 10 ms, Flip angle = 25° , FOV = $256 \times 256 \text{ mm}^2$, Resolution = 128×128 , Slice thickness = 2 mm, Averages = 3, Bandwidth = 320 Hz/px, Scan time = 28 s, Scanner reported whole head average SAR = 0.1 W/kg . The phases were measured before and after a TSE sequence with the extra-cranial DBS lead looped on the top, side, and back of the porcine head; and placed parallel to the coil’s longitudinal axial direction. The PRF shift coefficient of $-0.015 \text{ ppm/}^\circ\text{C}$ was used to convert phase differences into temperature changes (Weis et al., 2009). The scan time was kept within 30 s to keep the change in the temperatures outside the susceptibility artifact range within the scan time less than 0.2°C (i.e., standard deviation of the ‘gold standard’ fluoroptic probes that were used to make direct temperature measurements).

Measuring heating at DBS electrode-brain contacts using endogenous proton resonance frequency (PRF) shift method is challenging due to susceptibility artifacts of the lead. Temperature changes can be measured more reliably outside the artifact range using the PRF shift method. The PRF method is the most accurate, non-invasive MR thermometry imaging method (Wlodarczyk et al., 1999). The average temperature change adjacent to the artifact range may be used as the average temperature change at the artifact boundary in a thermal model to estimate heating at the electrode-brain contacts.

Experiment Protocol

Temperatures were measured in three cadaveric porcine heads (mean body weight = 85.47 ± 3.19 kg, mean head weight = 5.78 ± 0.32 kg) implanted with DBS leads. Three porcine heads were chosen since a minimum number of $N=2.7$ animals was required to yield >95% power with $\alpha=0.05$ (two-sided) to detect a minimum temperature change of 0.25°C . The heating was measured in the cadaveric porcine heads for afore mentioned electromagnetic and thermodynamic tissue property similarities between the porcine and human brains. The head coil was used since a head coil had been shown to produce less heating compared to a body coil (Rezai et al., 2002) and has been recommended for imaging DBS patients at 1.5 T by the device manufacturer.

Temperatures were measured at the following four locations in each pig using fluoroptic temperature probes (Luxtron Corporation, model m3300): scalp, near the first and second electrode-brain contacts from the distal DBS lead tip, and 6 mm distal from electrode 1 in the brain. Temperature at the epidermis of the scalp was monitored to measure direct heating of the scalp due to imaging. Temperatures in the brain were measured to measure heating near DBS electrodes. The fluoroptic probes were taped to the lead to keep them at the electrodes and 6 mm away from electrode 1 (Figure 2).

A porcine head was harvested after appropriate euthanasia. The head was cut off the body at the level of the first cervical vertebra (i.e., C1 level). Ears were removed since they obstructed appropriate placement of the porcine head in the head coil. The weight of the head was measured. An $\sim 18\text{G}$ hole was drilled through the porcine cranium perpendicular to the coil plane (i.e., in the coil radial direction) to place the DBS lead and temperature probes deep within the porcine brain. The hole was drilled 45 mm away from the back of the skull and 5 mm left to the line that divided the head in two equal halves. The hole was drilled such that the dura was not punctured. The distance from the scalp to the dura was measured. The head was placed in the volume head coil. The marked DBS lead and three taped fluoroptic probes were inserted through the dura within the brain to pre-determined depths. It was assumed that placing the DBS lead and the taped temperature probes through an 18 G wide and ~ 3 cm deep cranial 'guide' hole (skin to dura distance = 33.8 mm, $\text{SD} = 1.9$ mm) assured appropriate placement of the temperature probes near the DBS lead electrodes.

The DBS lead was placed such that the distal lead tip was 60 mm away from the top of the skin (Figure 3). Placing the distal DBS lead tip 60 mm deep in the porcine brain was appropriate for the following three reasons. One, DBS electrodes were placed deep in the brain close to the clinically relevant deep brain structures like thalamus. Two, the scalp to the lead tip distance of 60 mm was close to a clinical situation of a DBS lead placement (Rezai et al., 2002). Three, the electrodes were placed away from the surface to minimize the effects of surface cooling. No specific brain structure was targeted since the goal of this work was to study the effect of the extra-cranial DBS lead orientation on the heating. Next, the extra-cranial portion of the DBS lead was appropriately placed 11 either as a loop on the top, side, or back of the head; or parallel to the axial direction of the coil. A pair of GRE sequences was run to acquire baseline phases and study the error in temperature mapping in the absence of any significant heating for each animal. A GRE-TSE-GRE sequence train was run for each extra-cranial DBS lead placement to acquire base-line phases, produce heating, and acquire phases after the heating.

Statistical Analyses

Measured temperature changes at a given location were compared to one another for all DBS lead placements in each pig to study the significance of the extra-cranial DBS lead placement on the heating. Measured temperature changes at the four locations were

compared to one another for a given extra-cranial DBS lead placement in each pig to study the nature of the heating. Each experimental fluoroptic temperature curve was estimated using local linear regression (Fan, 1992), a technique commonly employed in functional data analyses (Ramsay and Silverman, 2005) because of its appealing optimality properties. In local linear regression, the degree of smoothness exhibited by the estimate is controlled by a parameter, λ , with smaller values of λ producing a rougher curve, larger values a smoother curve. λ was chosen automatically using a well-known technique called repeated learning-testing (RLT) (Burman, 1989). Our implementation of RLT used test sets of size 20 and $\lfloor n/20 \rfloor$ iterations, where n is the sample size and $\lfloor . \rfloor$ denotes the floor function. This produced estimated temperature curves that appear to capture the true features of the data while being nearly free of artifacts. Additionally, approximate simultaneous 95% confidence bands were obtained. Although a simultaneous confidence band is considerably more conservative than a pointwise band, the simultaneous bands for these data were quite narrow because the 12 sampling frequency was high and the signal-to-noise ratio was large. Even the more conservative simultaneous bands indicated that we can be quite certain about the shape of the true temperature curves.

RESULTS

Clinically harmful heating was produced near DBS electrode-brain contacts when the extra-cranial portion of the DBS lead was looped on the porcine head (Table 1, Figure 2). The measured maximum temperature change at the electrodes (i.e., peak temperature change) varied between 5.1 °C and 24.7 °C during the 641 s long TSE imaging with the 3T transceiver head coil. The scanner reported whole head average SAR was ~3.2 W/kg. The heating was affected by the placement of the extra-cranial DBS lead loop inside the volume head coil (Table 1, Figure 4).

The heating was significantly reduced when the extra-cranial lead was placed parallel to the longitudinal axial direction of the head coil compared to when the lead was looped. The peak temperature change varied between 1.5 °C and 3.2 °C when the lead was placed in the parallel configuration (Figure 4, Table 1).

Strong temporal and spatial temperature gradients may exist near DBS electrodebrain contact points. The peak temperature change varied from 0.95 °C to 18.76 °C within 1 minute of the heating. The peak temperature change was significantly higher than the corresponding maximum temperature change measured 6 mm distal from electrode 1 (Figures 4 and 5, Table 1).

The simple thermal model simulated temperature changes comparable to the measured temperature changes near DBS lead electrodes. Figure 5 presents simulated and 13 measured temperature changes for a given extra-cranial DBS lead placement in a cadaveric porcine head. The simulated and measured transient temperatures near the scalp, electrode 2, and 6 mm distal from electrode 1 were comparable. Relatively large deviation was found in the simulated and measured temperatures near electrode 1. The deviation was attributed to the strong temporal and spatial temperature gradients and the placement of the fluoroptic probe near electrode 1.

The simple thermal model was used to simulate thermal resolution constraints for a suitable MR thermometry approach for determining heating near DBS electrodes during MRI (Figure 6 and 7) to verify safety. Figure 6 presents the simulated maximum and minimum temperature change 6 mm away from the DBS lead electrodes per unit peak temperature change as a function of the scan time in un-perfused and perfused porcine heads. The maximum temperature change was found lateral to electrode 2. The minimum temperature

change was found distal to electrode 1. The minimum temperature change must be imaged accurately and precisely by an MR thermometry sequence to determine peak temperature using the thermal model. The temperatures were simulated 6 mm away from the DBS lead electrodes since MR thermometry can be performed more reliably outside the lead's susceptibility artifact range. The lead's susceptibility artifact range was ~5 mm since 1 mm wide DBS lead was imaged as 1 cm wide for the PRF shift MR thermometry sequence (Figure 3). The maximum temperature change outside the susceptibility artifact range occurred lateral to electrode 2 due to the superposition of heating from nearby electrodes. The minimum temperature change outside the susceptibility artifact range occurred distal to electrode 1 due to its relatively larger distance from other heated electrodes.

Figure 7 presents the simulated peak temperature change per unit maximum temperature change 6 mm away from the DBS lead electrodes as a function of the scan time in unperfused and perfused porcine heads. The peak temperature increases as the scan time decreases for a given maximum temperature change away from the lead electrodes.

Imaged (Figure 8A) and simulated (Figure 8B) temperature changes match qualitatively outside the susceptibility artifact range of the DBS lead. The temperature changes were imaged using the MR thermometry method and simulated using the thermal model at the end of the TSE sequence.

DISCUSSION

Several important observations were made. First, clinically harmful heating near DBS electrodes may be induced during MRI with a 3T volume head coil when the extracranial portion of the DBS lead is looped (Table 1, Figure 4). The heating was dependent on the placement of the extra-cranial lead (Table 1, Figure 4). The observations were explained since the heating was dependent on the currents induced in the DBS lead by the time-varying electromagnetic fields during MRI. The induced currents in the extracranial portion of the DBS lead depended on the placement of the extra-cranial DBS lead with respect to the transmit RF coil. Strong dependence of the heating near DBS electrodes on the placement of the extra-cranial DBS lead with respect to an RF coil suggested that the heating may be reduced significantly by suitably placing the extracranial DBS lead with respect to a coil. Such a placement protocol may be developed for a given RF coil and DBS lead.

Heating in a conductive device is proportional to the induced currents and thus voltage due to changing electromagnetic fields during MRI. The induced voltage along the length of a device is given by integral Maxwell equation as follows: $\int -E \cdot dl = \int \frac{dB}{dt} \cdot ds$, where E = electric field, l = length, B = magnetic flux density, s = surface area, and t = time. Thus, heating in a conductive device can be produced by increasing rate of change of B , surface area s , or their product. This is evident since for elongated structures the coupling to the RF electric fields produced by the MR coil can cause significant heating, without loops in the wire.

Second, the heating near DBS electrodes in the brain was reduced when the extracranial portion of the DBS lead was placed parallel to the longitudinal axial direction of the coil (Figure 4, Table 1). The observation was explained since the magnetic flux crossing the DBS coiled wires approached zero in the coil when the lead was placed parallel inducing relatively less electromotive force and thus, currents and heating. The observation suggested that less heating may be induced during human imaging with the 3T head coil if the extra-cranial portion of the DBS lead was placed behind the ears and parallel to the coil's longitudinal axis with no loops.

Third, strong temporal and spatial temperature gradients may exist near DBS electrodes (Figures 4 and 5, Table 1). Strong temporal and spatial temperature gradients near DBS electrodes suggested that accurate computational modeling of electromagnetic and resultant temperature fields required high temporal and spatial resolution and/or higher order elements. Convergence of a solution must be checked by increasing resolution to have confidence in the simulated electromagnetic and temperature fields.

Fourth, the simple, two-dimensional thermal model simulated the heating near DBS electrode-brain contacts accurately (Figure 5). This thermal model together with the MR thermometry can be used in industry standard ASTM gel phantoms and tissue phantoms (including animal and humans cadavers) to determine worst case temperature field near conductive medical implants (CMDs) for given implant placements during MRI. The thermal model and MR thermometry together will help perform worst case thermal dosimetry and identify upper limits of the whole-head average specific absorption rate (SAR) vs scan time to minimize tissue heating/damage.

Regarding using the thermal model for in vivo predictions, strong spatial and temporal temperature gradients near the electrodes require new development and evaluations of the parameters (e.g., diffusion coefficients, blood-tissue heat transfer coefficients, etc.) of the bioheat model for simulating a non-linear temperature field in vivo (Shrivastava and Vaughan, 2009). The new thermal model must be validated by placing ‘sufficient’ number of fluoroptic temperature probes near the electrodes in animal models.

Fifth, an MR thermometry technique with sub-degree C accuracy and precision in vivo is needed to keep peak temperature change below safe thresholds to assure patient safety (Figures 6–8). Temperature changes outside the susceptibility artifact range were significantly lower than the peak temperature change. Shorter scans produced much lower temperature changes outside the artifact range for the same peak temperature change due to finite thermal diffusion time stressing the need for sub-degree C accuracy and precision of an MR thermometry method to reduce peak heating.

Improved MR thermometry with sub-degree C accuracy and precision is a key to determine accurate peak temperature change as well as temperature field near a conductive medical device (CMD) using thermal modeling. The temperatures simulated in Figure 6–7 are expected to be reasonably accurate since a) the thermal model has recently been validated to predict in vivo radiofrequency heating in swine due to MRI; and b) the thermal model predicts accurate heating near DBS devices in cadaveric porcine heads in the current study.

Sixth, qualitatively comparable imaged (Figure 8A) and simulated (Figure 8B) temperature change maps were obtained in the porcine head outside the susceptibility artifact range of the intra-cranial DBS lead. This suggested the feasibility of using MR thermometry and thermal modeling together to determine peak temperature change. Figure 5 showed that the temperature change outside the susceptibility artifact range can be used in thermal models to simulate heating near electrodes. However, accurate quantification of the PRF shift coefficient in the imaged tissue is needed for developing PRF shift based MR thermometry with sub-degree C accuracy and precision. PRF shift coefficients ranging from 0.01–0.02 ppm/°C have been reported in the porcine brain (Weis et al., 2009, Peters et al., 1998). Alternatively, an exogenous contrast based minimally-invasive MR thermometry may need to be developed to determine temperatures near DBS electrodes (James et al., 2009, Pakin et al., 2006, Hekmatyar et al., 2005b, Hekmatyar et al., 2005a, Hekmatyar et al., 2002). Exogenous contrast based MR thermometry techniques have been shown to be significantly more sensitive than the endogenous PRF based techniques. Use of such agents in animals will help accurately estimate heating near CMDs during MRI and validate to-be-developed

improved 18 endogenous PRF based MR thermometry techniques and thermal modeling. Use of such agents in humans, if proven safe, will help improve patient safety online.

Next, in comparing the present study with previous studies temperature changes other than the noise could not be detected due to a DBS lead (model 3387), Medtronic Inc., Minneapolis) at 1.5T and 2.35T in a NaCl solution-filled phantom with a body coil (Georgi et al., 2004). Rezai et al. measured the peak temperature change ranging from 2.3 – 7.1 °C in a gel phantom after 15 minutes of MRI with a head coil at 1.5 T. The scanner reported whole body average SAR ranged from 0.07–0.24 W/kg (Rezai et al., 2002). Finelli et al. measured the peak temperature change ranging from 0.2–6.7 °C in a gel phantom after 15 minutes of MRI with a head coil at 1.5 T. Several pulse sequences were run with the scanner reported whole body average SAR ranging from 0.00–0.24 W/kg (Finelli et al., 2002). Shrivastava et al. measured the peak temperature changes of 0–5.24 °C and 16.8–26.8 °C due to 15 minutes of 400 MHz continuous wave (CW) RF power deposition at the whole head average SAR of ~3.0 W/kg during the axial and azimuthal placement of the extra-cranial DBS lead, respectively (Shrivastava et al., 2010). These temperature changes were comparable to the peak temperatures changes measured in the present study of 1.5–3.2 °C and 5.1–24.7 °C due to the axial and loop placement of the extra-cranial DBS lead, respectively. Baker et al. measured the peak temperature change ranging from 0.8–7.3 °C in a gel phantom after 1 minute of a spin echo sequence with a head coil at 3T. Time averaged RF power ranged from 13.2–14.7 W and the scanner reported whole body average SAR ranged from 0.2–0.3 W/kg (Baker et al., 2005). The temperature changes were comparable to the peak temperature changes measured herein of 0.95–18.76 °C. Baker et al. measured the peak temperature change 19 ranging from 0.7–9.3 °C in a gel phantom after 2 minutes of a spin echo sequence with a head coil at 1.5T. The scanner reported whole head average SAR ranged from 0.6–6.5 W/kg (Baker et al., 2006). Carmichael et al. measured the peak temperature change of 1.4 °C (at 1.5 T, RF power deposition time ~3.5 minutes) and 2.2 °C (at 3.0T, RF power deposition time ~2.0 minutes) with the whole head average SAR of 1.45 W/kg and 2.34 W/kg, respectively (Carmichael et al., 2007). The results were comparable to the peak temperature changes of 1.06–21.15 °C measured in the present study in two minutes after the TSE sequence was started. Heating near a DBS device depends on the manufacturer, software, and construction of the head coil for a given placement of the device. A standard guideline is required in manufacturing head coils, exciting tissues, and computing whole head average specific absorption rates (SARs) to improve safety with implanted medical devices.

Conclusions

MRI induced heating near DBS lead electrode-brain contacts is affected by the placement of the extra-cranial portion of the DBS lead with respect to a head coil. The heating may be significantly reduced by placing the extra-cranial DBS lead ‘favorably’ for a given lead and coil. The use of thermal modeling and MR thermometry together may help determine the heating and improve patient safety online.

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Figure 1.
A DBS lead with four electrodes.

Figure 2.
A DBS lead and four electrodes and taped fluoroptic probes. The fluoroptic probes are taped to electrode 1, electrode 2, and 6 mm distal to electrode 1.

Figure 3.
A DBS lead in the porcine head. Approximately 1 mm wide DBS lead was imaged as ~10 mm wide DBS lead by the GRE sequence used for MR thermometry.

Figure 4.
A typical heating near DBS lead electrodes during imaging with a 3T head coil in the cadaveric porcine head.

Figure 5.

A typical comparison between measured and simulated temperatures in the porcine head.

Figure 6.
Simulated temperature change 6 mm away from the DBS lead electrodes per unit peak temperature change as a function of scan time.

Figure 7.

Simulated peak temperature change per unit maximum temperature change 6 mm away from the DBS lead electrodes as a function of scan time.

Figure 8. Imaged (Figure 8A) and simulated (Figure 8B) temperature changes near DBS lead electrodes in the porcine brain.

Maximum temperature change measured at the deep brain stimulation (DBS) electrode 2 (i.e., peak temperature change, dT_{max}) and 6 mm distal from electrode 1 (dT_{6mm}) in the porcine brain due to imaging with a 3T transceive volume head coil. The corresponding scanner reported whole head average specific absorption rate (SAR) is also presented.

Table 1

N	Animal Weight (kg)	Top Loop			Side Loop			Back Loop			Parallel		
		SAR (W/kg)	dT_{max} (°C)	dT_{6mm} (°C)	SAR (W/kg)	dT_{max} (°C)	dT_{6mm} (°C)	SAR (W/kg)	dT_{max} (°C)	dT_{6mm} (°C)	SAR (W/kg)	dT_{max} (°C)	dT_{6mm} (°C)
1	82.8	3.17	5.11	1.11	3.17	16.13	4.17	3.17	8.87	2.05	3.17	1.45	0.22
2	84.6	3.20	16.71	2.22	3.20	24.74	3.53	3.17	23.05	3.16	3.10	3.21	0.06
3	89.0	3.17	8.26	1.75	3.17	18.48	4.81	3.17	10.40	2.41	3.10	2.18	0.61