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Local SAR near deep brain stimulation (DBS) electrodes at 64 MHz and 127 MHz: A simulation study of the effect of extracranial loops

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Abstract

Purpose—Magnetic resonance imaging (MRI) may cause brain tissue around deep brain stimulation (DBS) electrodes to become excessively hot, causing lesions. The presence of extracranial loops in the DBS lead trajectory has been shown to affect the specific absorption rate (SAR) of the radiofrequency energy at the electrode tip but experimental studies have reported controversial results. The goal of this study was to perform a systematic numerical study to provide a better understanding of the effects of extracranial loops in DBS leads on the local SAR during MRI at 64 MHz and 127 MHz.

Methods—A total of 160 numerical simulations were performed on patient-derived data, where relevant factors including lead length and trajectory, loop location and topology, as well as frequency of MRI RF transmitter were assessed.

Results—Overall, the presence of extracranial loops reduced the local SAR in the tissue around the DBS tip compared to straight trajectories with the same length. SAR reduction was significantly larger at 127 MHz compared to 64 MHz. SAR reduction was significantly more sensitive to variable loop parameters (e.g., topology and location) at 127 MHz compared to 64 MHz.

Conclusion—There might exist lead management strategies that significantly reduce risks of 3.0 T MRI for DBS patients.

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Keywords

atlas-based segmentation; medical implant; RF heating; Finite Element Method (FEM); safety

Introduction

Many patients with deep brain stimulation (DBS) implants will benefit from regular MRI examinations throughout the course of their lives, as MRI is often the standard diagnostic tool of choice for monitoring structural changes in the brain. Moreover, multi-modal studies that utilize functional MRI in combination with DBS have proven to be advantageous in clinical settings to help better understanding the therapeutic effects of electrical stimulation on DBS target structures (1-3). Today however, high-field MRI (>1.5 T) is largely inaccessible to patients with DBS due to safety concerns (4), and postoperative MRI at 1.5 T is practiced in only few centers and under strict safety guidelines (B_1^+ rms fields <2.0 μ T or whole-head averaged SAR <0.1 W/kg) that limit image quality (5,6).

As DBS patients become more likely to need MRI over the course of their lives (7), an important safety concern is the so-called “antenna effect” of the DBS leads. Here, the RF fields of the MRI transmitter couple with the long conductive leads and cause the specific absorption rate (SAR) of the radiofrequency (RF) energy to significantly amplify near the implant's tip (8,9). Injuries reported when FDA-approved guidelines were not followed have underscored these safety concerns (10,11).

A major hurdle in devising a systematic methodology that consistently and reliably estimates the implant-induced heating is that the problem has a large parameter space with many interacting factors. The coupling between MR-generated RF fields and DBS leads depends on the frequency, geometry, and electrical implementation of the transmit/receive RF coils, the patient's position inside the magnet (i.e., imaging landmark), the location of implant within the head (12), the lead trajectory (13-16), as well as implant materials (17) and design (18). Accordingly, complex computational models are used by device manufacturers and researchers to evaluate the safety of patients with DBS undergoing MRI (19). Such models may need to include detailed description of RF coils, head model, and implant geometry, therefore requiring considerable computational time and resources (20). One particular difficulty is to extract realistic DBS lead trajectories from post-operative computed tomography (CT) images that are routinely administered for electrode localization (21,22). In many practical cases surgeons choose to keep the DBS extension connector at the level of cranium to avoid positioning at the neck. As a result the lead is looped several times, making segmentation of the trajectory very challenging because of overlapping sections that cannot be easily distinguished on CT images (see Figure 1).

The effect of extracranial loops on the MRI-generated heating of the tissue around DBS electrode contacts is not well understood and reports are controversial. Baker et al (16) observed a significant reduction in temperature rise around the tip of an implanted DBS device in a gel phantom during MRI at 1.5 T and 3.0 T, when the lead formed small concentric loops at the level of phantom cranium. In contrary, Shrivastava et al (13,14) predicted a significant increase in the temperature rise at electrode contacts of a DBS lead

implanted in a porcine head during MRI at 3.0 T and 7.0 T when extracranial loops were present. Part of the discrepancy is due to the heterogeneity of experimental settings. Indeed, performing experimental studies that control a plurality of factors that affect the heating of implanted leads is substantially cumbersome as underscored in literature (12,23). Simulation studies, on the other hand, provide a great opportunity to study heating mechanism as they allow for a systematic and controlled way of changing each variable to better study its effect differentially, or in conjunction with other variables.

The goal of the present work was to apply a controlled simulation study and assess the presence of extracranial loops on the local SAR amplification at and around DBS electrode contacts during MRI at 64 MHz and 127 MHz. A total of 160 numerical simulations were performed on patient-derived data, where fine features of DBS lead including electrode contacts, insulation, core, and interconnections were represented. To explore the effect of transmit field frequency, lead trajectories with and without extracranial loops were simulated inside 64 MHz and 127 MHz RF coils. Each simulation scenario was repeated at 15 different RF coil-head orientations to account for variations across different coil systems. Both figure-8 and concentric loop topologies were studied at each frequency. Two different loop locations were investigated by positioning the loops at the side and top of the head. Finally, for lead trajectories without loops, additional simulations were performed with extended leads to control for the effect of lead length.

Methods

Secondary use of patient data for modeling and simulations was approved by Massachusetts General Hospital Internal Review Board (IRB). Finite Element Method (FEM) simulations were performed on a homogenous head model which was built by combining different tissue classes of a multimodal imaging-based detailed anatomical (MIDA) model of human head and neck, reported elsewhere (24,25). Post-operative computer tomography (CT) images of a patient operated at Massachusetts General Hospital were used to extract the DBS lead trajectory. The lead trajectory was manually segmented from CT images and co-registered to the MIDA head and neck model. Co-registration of patient's CT data and MIDA's MRI data was performed using a non-linear approach based on free-form deformations (FFDs) and normalized mutual information (NMI). The segmentation of the lead was then projected onto the MIDA model using the deformation vector field resulting from the co-registration.

DBS electrodes composed of four cylindrical contacts (outer diameter=1.27 mm, thickness=150 μm , height=1.5 mm) were constructed at the tip of the lead and were connected through a solid straight central core (diameter=260 μm). Electrode contacts were 0.5 mm apart, similar to those reported in (26). The electrodes and the core were made of 90%:10% (Pt:Ir) platinum-iridium ($\sigma = 4 \times 10^6 \text{S/m}$) and were separated by an insulation made of polyurethane ($\sigma = 10^{-10} \text{S/m}$, $\epsilon_r = 3.5$). Details of electrode contacts and their interconnections are given in Figure 2.

RF coil models

MRI head coils were modeled as low-pass 16-rung birdcages with 35 cm diameter and 26 cm length. Commercially available software ANSYS Electronics Desktop 16.2 (ANSYS,

Canonsburg, PA, USA) was used to perform a combined FEM-circuit analysis as described in previous studies (27,28). In the first step, the multi-port scattering matrix of the RF coil in the presence of the head model and implant was calculated in ANSYS HFSS with tuning and matching circuit components substituted with equivalent lumped ports. The multiport S-parameter matrix was then exported to ANSYS Designer to be tuned and matched at 127 MHz (3.0 T) and 64 MHz (1.5 T). The updated port voltage and current values were then pushed back to ANSYS HFSS to calculate electric and magnetic field distributions. A detailed description of this procedure can be found in (27). This method has shown to give an accurate estimation of magnitude of field values in phantom experiments (29).

DBS lead trajectories

Five different DBS lead trajectories were studied with details given in Figure 1. First, a realistic patient-derived ~50 cm lead was constructed from CT images, consisting of a figure-8 loop on the side of the head (model A). To study the effect of loop topology, the model was morphed to form a concentric loop, with the rest of the trajectory kept unchanged and the overall length preserved (model B). To investigate the effect of loop's position, another model was built where the loop was moved to the top of the head close to the surgical bur hole. The loop's surface and topology were kept unchanged, but its orientation was adjusted to follow the skull's curvature (model C). Finally, two models were constructed in which loops were eliminated. In one, the loop was cut out with the rest of the path kept the same, resulting in a shorter overall length of ~39 cm (model D). In the other, the lead was extended at the neck level to compensate for the shortening of the path (model E). A summary of model characteristics is given below.

- Model A: figure-8 loop on side of the head, lead ~50 cm
- Model B: concentric loop on side of the head, lead ~50 cm
- Model C: concentric loop on top of the head, lead ~50 cm
- Model D: no loop, lead ~39 cm
- Model E: no loop, lead ~50 cm

Each head and lead model was simulated in both 64 MHz and 127 MHz quadrature head coils. Coils were excited at two rungs that were 90° apart in position and signal phase (P_I and P_Q in Figure. 2). Excitation ports may have arbitrary position from one MRI site to another depending on the installation process. To account for the variability in the relative orientation of the head and RF coils, coils were rotated around each head model with 22.5° increments (a total of 15 coil positions). A reference head model with no lead was also simulated for the purpose of field normalization (not shown in the figure).

Numerical convergence

At the start of each simulation ANSYS HFSS was set to follow an adaptive mesh scheme. The algorithm started with a user-controlled initial tetrahedral mesh which forced a fine resolution on the DBS lead (maximum tetrahedron edge < 0.5 mm). The adaptive algorithm started by refining the mesh by 30% between each two iterative simulations. At each step, the maximum change in the magnitude of S-parameters, S , was defined as $S = \text{Max}_{ij} |S_{Nij}|$

S_{N-1ij} , where i and j covers all matrix entries and N represents the iteration number. The adaptive simulation continued until the threshold of $S < 0.01$ was reached. All simulations converged after $N=4$ adaptive passes. Once convergence was reached, the mesh was visually inspected to assure that fine details of DBS leads were properly captured. Details of mesh statistics for a typical simulation (model A, feed at $\theta = 0^\circ$) is given in Figure 2.

SAR and B_1^+ field calculations

The total power absorbed in the head was calculated by computing the volume integral of the loss density in the head. The whole-head-averaged SAR (whSAR) was then calculated by dividing this total absorbed power by the head mass:

$$whSAR = \frac{\frac{1}{2} \int \int \int_{Head} \sigma |E|^2}{head \ mass} \quad [1]$$

where $|E|$ is the magnitude of electric field phasor and σ is the conductivity of the tissue. The overall mass of the head used in the calculation was ~ 5.1 kg. 1g-averaged SAR was calculated based on IEEE STD P1528.4 recommendation (30) using the built-in SAR calculator module in ANSYS HFSS 16.2. The maximum 1g-averaged SAR (referred to as $MaxSAR_{1g}$) was then calculated as the maximum of the 1g-averaged SAR in a $30 \times 30 \times 30$ mm³ cubic volume that encompassed all four electrode contacts.

The counter-clockwise rotating component of the RF magnetic field was computed as $B_1^+ = 0.5 (B_{1x} + jB_{1y})$ (31). For comparison, the spatial root mean square (rms) of the B_1^+ field was recorded from an axial reference plane passing through the head, ~ 7 cm from top of the head. The magnitude of the B_1^+ field phasor was sampled with a $1 \text{ mm} \times 1 \text{ mm}$ resolution from this axial plane and samples were used to calculate the B_1^+ rms as:

$$B_1^+ rms = \sqrt{\frac{\sum_N |B_{1N}^+|^2}{N}} \quad [2]$$

where $|B_{1N}^+|$ is the magnitude of the N^{th} sample and the summation is over all samples. $MaxSAR_{1g}$ and whSAR of models A-E are given in Table S1 in the Supporting Material for both 64 MHz and 127 MHz coils and at all 15 feed positions.

Results

Table S1 in Supporting Material gives the values of $MaxSAR_{1g}$ and whSAR for models A-E at 64 MHz and 127 MHz for different coil orientations. At each field strengths, the reference B_1^+ rms (B_{1ref}^+ rms) was calculated as the spatial rms of B_1^+ field recorded from the reference plane, when the coil operated at its default position ($\theta = 0^\circ$), and generated a whSAR of 3.2 W/kg inside a head model with no implant. In all other simulations, the coil input power was adjusted to generate the same B_1^+ rms on the reference plane as B_{1ref}^+ rms which equaled to 6.7 μT at 64 MHz and 3.0 μT at 127 MHz.

Effect of extracranial loops on the $MaxSAR_{1g}$ at 127 MHz

Presence of extracranial loops, either concentric (Models B and C) or figure-8 (Model A), significantly reduced the SAR amplification at DBS electrode contacts at 127 MHz (75%-99%). Results were consistent across different coil-head orientations and when compared against both leads with controlled length (Model E) and un-controlled length (Model D).

The extent of SAR reduction however, varied with loop topology and location. Maximum SAR reduction was observed in Model B (2.1 ± 0.4 W/kg), followed by Model A (18.7 ± 4.9 W/kg) and Model C (46.2 ± 16.8 W/kg) with respect to both Models D (207.3 ± 68.7 W/kg) and E (317.0 ± 100.5 W/kg). Interestingly, $MaxSAR_{1g}$ values of Model B were even smaller than SAR values at the periphery of the head far from the electrode (see Figure 5).

Effect of extracranial loops on the $MaxSAR_{1g}$ at 64 MHz

When controlled for the total length, the presence of extracranial loops consistently reduced the $MaxSAR_{1g}$ at 64 MHz. Specifically, a 3 fold reduction in $MaxSAR_{1g}$ was observed at all 15 feed orientations for both figure-8 loop (Model A) and concentric loops (Models B and C) with respect to the lead trajectory with the same length but no loop (Model E). However, compared to the lead trajectory with no loop and a shorter length (Model D), concentric loops only introduced a modest reduction in the $MaxSAR_{1g}$ (5% to 15%) and figure-8 loop did not consistently reduced the SAR (-6% to 2%).

Comparative effect of the lead trajectory at 64 and 127 MHz

At both frequencies and for all 15 feed positions, the 50 cm lead with no loop (Model E) generated a higher $MaxSAR_{1g}$ than the 39 cm lead with no loop (Model D). The difference however, was significantly larger at 64 MHz (1586.8 ± 254.2 W/kg vs. 589.4 ± 98.6 W/kg) than at 127 MHz (317.0 ± 100.5 W/kg vs. 207.3 ± 65.7 W/kg). Figures 3 and 4 give means and standard deviations of $MaxSAR_{1g}$ values (calculated over all coil orientations) for 64 MHz and 127 MHz coils, when the input power was adjusted to generate the same B_1^+ rms as the B_1^{+ref} rms or B_1^+ rms = $1 \mu T$, respectively. As it can be observed from Figures 3 and 4, presence of loops had a significantly larger SAR reduction effect at 127 MHz compared to 64 MHz. For example, concentric loop on the side (Model B) introduced a ~144 fold SAR reduction factor compared to Model E at 127 MHz, whereas it only reduced the SAR by a factor of 3 at 64 MHz.

Interestingly, the differential effect of loop's topology and location was also more significant at 127 MHz than at 64 MHz. Two-tail t-test comparing means of Models A and B (effect of loop's topology) had a p-value = 0.04 at 64 MHz and p-value = $1.3e-9$ at 127 MHz. Two-tail t-test comparing means of Models B and C (effect of loop's location) had a non-significant p-value at 64 MHz (p-value = 0.2) but a p-value = $3.9e-8$ at 127 MHz.

Discussion and Conclusion

This study provides novel results of a systematic evaluation of the effect of extracranial loops on the SAR amplification around DBS electrode contacts during MRI at 64 MHz and

127 MHz. To date, reports of temperature rise at and around DBS electrodes have drawn controversial conclusions as to the effect of extracranial loops (13,14,16). The discrepancy is partially due to the heterogeneity of the experimental settings and the methodology applied to measure or predict the temperature. For example Baker (16) used fluoroptic probes to directly measure the temperature rise at the distal DBS electrode contact immersed in a phantom gel. Shrivastava (14) on the other hand, applied *in-vivo* MR thermometry to estimate the temperature rise a few millimeters away from the tip, and used the estimated temperatures to calibrate a numerical thermal equation solver to predict the maximum heating. Apart from obvious differences in the experimental setup (gel phantom vs. animal model) and the methodology (direct measurement vs. numerical interpolation based on MR thermometry) these studies did not control details of the lead trajectory or position and orientation of loops with respect to the RF coil. Such details however, could significantly affect the heating as highlighted in previous reports (12,32).

Performing experimental studies that control a plurality of interacting factors that affect the heating of implants is substantially cumbersome. Simulation studies on the other hand, allow a more affordable systematic evaluation of such factors including lead length and trajectory, loops position and topology, and RF coil's geometry and position. This work was a first attempt to provide better insight into the effect of extracranial loops on the local SAR at the tip of a DBS lead, by carefully crafting models that were identical in all aspects except the one under study.

Overall, we observed that for the specific length of the lead under study (~50 cm) and specific locations and orientations of loops (side and top of the head), the presence of extracranial loops reduced the local SAR at the DBS tip compared to straight trajectories with the same length. SAR reduction however, was significantly larger at 127 MHz compared to 64 MHz. Specifically, at 127 MHz there existed lead configurations (concentric loops on the side of head) that reduced the $MaxSAR_{lg}$ down to background levels observed at the periphery of the head. We also observed that SAR reduction was significantly more sensitive to differential loop parameters (e.g., topology and location) at 127 MHz compared to 64 MHz.

The mechanism by which extracranial loops affect the SAR is not completely understood. Baker (16) suggested two possible scenarios. First, the loops may add a reactive inductance that produces a source of impedance near the end of the electrode, reducing the induced current in the wire. This mechanism is similar to the reduction of RF-induced temperature rise observed for example by Ladd and Quick in catheter by addition of chokes (33) or by coiling the wires (34,35). Another possibility is that the voltage induced in the loops because of coupling with the magnetic field (from Faraday's law) cancels out the voltage generated by coupling of the straight part of the lead with the RF electric field. As the Faraday coupling effect scales up with the rate of change of magnetic flux, we expected that the former mechanism (inductive effect) be more dominant at 64 MHz, while the latter (Faraday coupling) be dominant at 127 MHz. This was in line with our observation of relative insensitivity of SAR reduction to the location and topology of the loop at 64 MHz vs. its high dependency on these parameters at 127 MHz. Specifically at 127 MHz, the loop on the side of the head with a surface exposed to a larger normal magnetic flux reduced the SAR

much more effectively than the loop on top of the head where the B_1^+ field was more tangential to its surface.

Our study also underscores the fact that 3.0 T MRI should not be considered, per se, more dangerous for patients with conductive leads than 1.5 T MRI. Indeed, our results showed that for a B_1^+ rms=1 μ T the majority of lead configurations introduced less SAR amplification at DBS electrodes at 127 MHz compared to 64 MHz. It should be noted however, that the sensitivity and variation of the results with frequency, lead length, as well as loop location, orientation, and topology warrant the need for a multivariate analysis of all relevant factors before any definitive conclusion on the safety-enhancing effect of extracranial loops can be drawn. There are specifically two important limitations of the current study. First, in all our simulations the central core of the DBS lead was modelled as a single straight solid wire connecting all four electrode contacts. In reality, DBS electrode contacts are connected through tightly-coiled interleaved spiral wires, each connected to one electrode contact, and isolated from each other (36). Similar technique is applied in manufacturing of cardiac pacemakers and defibrillator leads (37,38). It is conceivable that the presence of these small coils, adding to the overall electric length of the lead, interferes with the hypothesized SAR-reduction mechanism of large extracranial loops. A more detailed model of DBS lead interconnection wires is needed to investigate this issue. Second, we report results of numerical simulation of SAR distribution and have not performed measurements to calculate the associated temperature rises. Although SAR has been routinely used and an indirect quantitative measurement for clinical MRI procedures when conductive implants are present (8,39), local SAR values do not linearly translate to temperature rise as factors such as thermal conduction and perfusion should be taken into account (40). Additionally, even if the SAR near the lead tip is reduced, there may still be high values in other regions of the head. Consequently, temperature calculations/measurements over the entire head should be considered instead of monitoring SAR alone near the tip.

Finally, this study highlights the prospect of obtaining a significant reduction in the SAR at 127 MHz by administering lead management strategies. More numerical simulations validated with experimental measurements and inclusive of uncertainty analysis are necessary to verify and optimize such strategies, which could provide a useful surgical guideline to the clinical community.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure.1.

Top row: Five head models with different DBS lead trajectories. Model A was directly constructed from intra-operative CT images shown in the bottom row. The end of the lead was extended behind the ear, similar to what happens in conventional DBS surgical approach, to model the connection of the lead to the pulse generator extension. Model B is similar to model A in all aspects, except for the loop which was morphed to a concentric shape. Model C is derived from model B by cutting the loop and moving it on top of the head close to the surgical bur hole. The loop's surface and topology were kept unchanged, but its orientation was adjusted to follow the skull's curvature. Model D was constructed by cutting out the loop and leaving the lead with the shorter overall length of ~39 cm. Model E was based on model D, except that the lead was further extended to reach the overall length of ~50 cm same as in models A-C. Bottom row: Steps of DBS lead segmentation. 3D view of intra-operative CT images of a patient with the implant is shown. Image contrast threshold and manual refinements were applied to recover overlapping loop segments.

Figure .2. Top left: FEM setup showing the head model and implant inside the RF coil. Coil rotation angle θ and feed positions are illustrated. Top right: Details of DBS lead model including electrode contacts, core and insulation and a sample of FEM adaptive mesh after convergence. Mesh was inspected visually for all simulations to assure fine details of DBS lead were properly captured. Bottom: Mesh statistics for the DBS lead, insulation and head model in a representative simulation (Model A, $\theta = 0^\circ$). Mesh statistics were similar across simulations.

Figure.3.
MaxSAR values for models A-E inside 64 MHz and 127 MHz coils. Input power adjusted to produce a $B_1^{+rms} = B_{1^{+ref}}^{rms}$. At each transmit frequency, $B_{1^{+ref}}^{rms}$ was calculated as the rms of the B_1^{+} field recorded from an axial plane at ~7 cm from top of the head, when the coil was operated at its default feed position ($\theta = 0^\circ$) and generated a whSAR of 3.2 W/kg inside a head model with no implant. $B_{1^{+ref}}^{rms}$ was ~7.65 μ T at 64 MHz and ~3.0 μ T at 127 MHz.

Figure. 4.
MaxSAR values for models A-E inside 64 MHz and 127 MHz coils. Input power adjusted to produce $B_1^{+rms} = 1\mu\text{T}$.

Figure. 5.
Maps of 1g-averaged SAR for Models B and D at 64 MHz and 127 MHz. Coils were operated at their default feed position ($\theta = 0^\circ$) and the input power was normalized to produce $B_1^{+rms} = B_1^{+refrms}$.