Published in final edited form as:

Magn Reson Med. 2019 January ; 81(1): 653–669. doi:10.1002/mrm.27350.

Changes in the specific absorption rate (SAR) of radiofrequency energy in patients with retained cardiac leads during MRI at 1.5 T and 3 T

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Abstract

Purpose—To evaluate the local SAR and heating around retained cardiac leads during MRI at 64 MHz (1.5 T) and 127 MHz (3 T) as a function of RF coil type and imaging landmark.

Materials and Methods—Numerical models of retained cardiac leads were built from CT and X-ray images of six patients with retained cardiac leads. Electromagnetic simulations and bio-heat modeling were performed with MRI RF body and head coils tuned to 64 MHz and 127 MHz and positioned at nine different imaging landmarks covering an area from the head to the lower limbs.

Results—For all patients and at both 1.5 T and 3 T, local transmit head coils produced negligible temperature rise ($\Delta T < 0.1^\circ\text{C}$) for $\|B_1^+\| \leq 3\mu\text{T}$. For body imaging with quadrature-driven coils at 1.5 T, ΔT during a 10-minute scan remained $< 3^\circ\text{C}$ at all imaging landmarks for $\|B_1^+\| \leq 3\mu\text{T}$ and $< 6^\circ\text{C}$ for $\|B_1^+\| \leq 4\mu\text{T}$. For body imaging at 3 T, ΔT during a 10-minute scan remained $< 6^\circ\text{C}$ at all

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imaging landmarks for $\|B_1^+\| \leq 2\mu\text{T}$. For shorter pulse sequences up to 2 minutes, T remained $<6^\circ\text{C}$ for $\|B_1^+\| \leq 3\mu\text{T}$,

Conclusion—For the models based on six patients studied, simulations suggest that MRI could be performed safely using a local head coil at both 1.5 T and 3T, and with a body coil at 1.5 T with pulses that produced $\|B_1^+\| \leq 4\mu\text{T}$. MRI at 3 T could be performed safely in these patients using pulses with $\|B_1^+\| \leq 2\mu\text{T}$.

Keywords

abandoned lead; retained lead; cardiac implanted electronic device; pacemaker; defibrillator; RF heating; safety; computational modeling; anatomical models; finite element method; SAR

Introduction

There is a steady growth in the use of cardiac implantable electronic devices (CIEDs) in the United States. As of 2012, there were more than three million patients with permanent pacemakers in the US [1], with the number growing by 80,000 annually [2]. The trend is likely to continue with an aging population, new indications for use, and technological advancements in device manufacturing [3, 4]. Magnetic resonance imaging (MRI) is considered the modality of choice for a majority of neurologic, cardiac and musculoskeletal disorders because of its excellent soft tissue contrast and non-invasive nature. It is estimated that 50%–75% of patients with CIEDs may need to undergo MRI over their lifetime for both cardiac and non-cardiac indications [5], with many patients requiring repeated MRI examinations [6]. A few studies have assessed the safety of leads connected to working devices, but some patients require additional procedures which result in “lead extraction”. This leaves fractions of the device left in situ, in which case the original safety studies of the device/lead combination may not be applicable.

Until recently, MRI was contraindicated in patients with implanted CIEDs due to concerns for harmful interactions between MRI fields and the implanted device. Since the first report of such adverse incidents [7], significant improvements have been made in CIED design and manufacturing. Such improvements include, for example, reduction of ferromagnetic material to reduce the risk of device dislodgement due to static magnetic fields [8], enhanced device programming to reduce the chance of cardiac stimulation from gradient-field induced currents [2], and improved lead structure to reduce the risk of radiofrequency (RF) heating [9, 10]. These efforts resulted in the development of new generations of MR Conditional devices, the first of which approved by the FDA in February 2011. The term MR Conditional, however, should be interpreted with caution. Adopted from the American Society for Testing and Materials (ASTM) [11], MR Conditional labeling is meant to infer that there are no known hazards to patients with the implanted device under specific conditions of use. As such, MR Conditional labeling of CIEDs only applies to devices with intact leads in their originally intended configuration. In a sizeable fraction of patients however, endocardial leads fail for technical reasons, infection, stenosis of the vein by clot or scar tissue, or lead interference with blood flow back to the heart [12, 13]. In these

situations the patient is often referred for lead extraction and new lead implantation [14], or lead abandonment and implantation of additional leads [15]. Incomplete lead extraction also often happens when the patient undergoes heart transplant but the physician is unable to completely extract the old lead due to adhesion of surrounding fibrotic tissue [16, 17]. In such patients, the leads are often cut in and retained above the level of the superior vena cava (SVC) anastomosis.

As the number of patients with CIEDs increases, there are an increasing number of patients who have retained leads. It has been suggested that leads in place for more than 8 to 10 years are less likely to be successfully removed [16]. A recent study reported that close to half of heart transplant patients had a retained CIED lead [18]. As the number of patients who undergo heart transplant is growing [19] and given the prevalence of retained CIED leads and the high likelihood of the need for MR imaging in patients' lifetime, the related safety risks associated with scanning of these patients need to be addressed. This is particularly important considering that currently, with no official guideline available, the decision to perform an MRI exam in such patients is entirely based on a risk/benefit assessment by the care team.

One of the major safety hazards during MRI of patients with abandoned or retained leads is RF heating due to the "antenna effect", where the electric field of the MRI transmit coil couples with conductive leads and amplifies the energy deposition in the tissue surrounding the lead [20–22]. To date, there has been no systematic evaluation of the heating profile induced by retained leads as a function of RF frequency and imaging landmark. Few studies have attempted to measure RF heating in abandoned cardiac leads in phantoms. A major challenge in phantom studies is that the geometry does not accurately reflect the clinical scenario. Another difficulty is that precise measurement of MRI induced heating is time consuming and difficult to achieve [23]. As the problem of RF heating has a large parameter space with many interacting variables [24], the implementation of an exhaustive experimental approach would be practically infeasible. On the other hand, numerical modeling and simulations provide an excellent tool for an exhaustive exploration of thousands of variable combinations and allow for the analysis of parameter values outside the bounds of normal clinical practice [25]. If correctly implemented, computational models can provide a reliable tool for the assessment of safety margins and the evaluation of potential risks [26].

When it comes to the RF heating of elongated implants, the local E-field tangential to the implant is shown to play a critical role [27]. Furthermore, it has been demonstrated that induced E-fields inside the body shift when changing body positioning along the z-direction [28]. Thus, the MRI RF safety on patients with linear implants needs to be investigated separately for each body position (aka imaging landmark) in the scanner. Another important point is to differentiate between "abandoned" leads and "retained" leads. Although these words have been sometimes used interchangeably, they refer to two importantly distinct groups of leads with different safety concerns. This issue is explained in more detail in the Discussion.

This study aims to evaluate changes in specific absorption rate (SAR) and temperature during MRI of a representative group of models based on data from patients with fractions of retained endovascular leads remaining in their torso. We aimed to perform electromagnetic simulations with RF head and body coils tuned to 64 MHz (1.5 T) and 127 MHz (3 T) and positioned at nine different imaging landmarks covering an area from the head to the lower limbs. We also calculated temperature maps for the cases that exhibited the maximum local SAR amplification around the leads. To our knowledge, this is the first study that attempts to quantify SAR and temperature changes as a function of RF frequency, coil type, and imaging landmark in patients with retained endovascular leads. The results of the study may provide useful information for a benefit/risk evaluation for caregivers who try to decide whether to use MRI on a patient with a retained lead who may need MRI for cardiac or non-cardiac indications.

Methods

Patient identification and lead model construction

Secondary use of de-identified patient data for modeling and simulation was approved by the Northwestern University Hospital Internal Review Board. Patients with a history of implanted cardiac devices with retained leads who also had Chest CT and X-ray images were retrospectively identified through a query of the enterprise data warehouse at Northwestern Memorial Hospital. Images of six representative patients with fractions of a retained lead remaining in their torso were used for modeling and simulation (Figure 1). Five patients (Age: 50 ± 12 years old, 4 males) had implanted ICD/CRT followed by heart transplant surgery. During the transplant surgery, the attempt for complete extraction of some of the leads was unsuccessful. Therefore, patients were left with some parts of the lead (including cardiovascular shock coil on right ventricle (RV)) retained in their endovascular system in contact with venous wall. One patient (male, 22 years old) had both right atrium pacemaker lead and right ventricular ICD lead in close proximity of each other (~ 1 mm apart). This case was particularly interesting as it represented a scenario where potential RF coupling could happen between the leads. The mean time interval between the date of lead implantation and date of attempt to remove the lead was 64 months (range: 9–126 months).

To extract the lead trajectory, a semi-automatic thresholding algorithm based on an intensity histogram analysis was applied to generate a preliminary mask of the hyperdense lead from the CT images. The lead centerlines were manually extracted and exported to a CAD tool for 3D surface reconstruction. Endocardial leads typically include individual wires wound as a coil that extend from the proximal to the distal end of the lead [9, 29, 30]. The original topology of these thin coiled wires may change during the extraction procedure due to stretching. X-ray images were used to reconstruct the details of the lead structure, including the number of micro-loops and variations in their pitch (Figure 1). The reconstructed lead model was incorporated into the ANSYS human body model (ANSYS Inc., Canonsburg, PA) [31] and was assigned the electrical properties of platinum ($\sigma = 9.3 \times 10^6$ S/m). To position the lead inside the body model, a triangulated surface of the patient's ribcage was generated from CT images and used for alignment of the lead into the multi-compartment

body model (Figure 2). Different tissue classes were then combined to form a homogenous body with relative permittivity of $\epsilon_r = 80$ and bulk conductivity of $\sigma = 0.47 \text{ s/m}$ [32].

Table I gives patient information including lead implantation and extraction dates, patient's age at the time of extraction surgery, lead model and manufacturer, as well as the effective length of the lead calculated from its 3D trajectory. For comparison, the actual length of the looped wire that constructed the lead was reported as well.

MRI RF Coils

Electromagnetic simulations were implemented in ANSYS Electronics Desktop (HFSS 16.2, Designer, ANSYS Inc., Canonsburg, PA). Numerical models of two high-pass birdcage body coils (620mm length, 607mm diameter) and two low-pass head-only coils (292mm length, 356mm diameter) were implemented and tuned to their respective Larmor frequencies - 64 MHz (1.5 T proton imaging) and 127 MHz (3 T). Body coils were composed of 16 rungs connected at each end to two end rings and shielded by an open cylinder (1220mm length, 660mm diameter). The coils and shields were made of copper and their dimensions were chosen based on a typical clinical body coil [33]. Head coils were unshielded and composed of 16 legs with dimensions based on typical prototypes built in our lab. A quadrature excitation was implemented by feeding the coils at two ports on the bottom end-ring that were 90° apart in position and phase. Body coils were tuned by lumped capacitors distributed at the end-ring gaps (88 pF at 64 MHz and 17 pF for 127MHz) and matched to 50 Ω using a single capacitor at each port in series with an ideal voltage source with a 50 Ω internal resistance. Head coils were tuned using lumped capacitors distributed at both ends of each leg (10 pF at 64 MHz and 1.5 pF for 127MHz) and a single matching capacitor at each port.

The ensemble of coil and body model was enclosed in an air-box with 50% padding in transverse direction and 20% padding in z direction. For each simulation, the final mesh on the lead was visually inspected to assure that fine features of the loops were properly represented, specifically to assure that closely wound loops were not bridged during the meshing process. Each simulation took approximately one hour and a half on a Dell PowerEdge R730 system with 16×32GB = 512GB of RAM, an NVIDIA K80 GPU and 28 cores (2×Intel Xeon CPU with each 14 cores) running 64-bit Windows Server 2012 system.

The amount of global and local energy absorption in the human body during MRI is affected by the patient's position inside the coil [34]. By the same token, tissue heating near implant will vary depending on imaging landmark [28]. To investigate the effect of body-coil position on SAR changes around the lead, simulations were performed for RF coils at nine different landmarks covering the area from the head to the lower limbs (L1–L9 in Figure 3).

SAR and B_1^+ calculations

For all simulations, the input power of the coil was adjusted to produce a mean B_1^+ of 3 μT inside the body on an axial plane passing through the center of the transmit coil. 1g-averaged SAR (1gSAR) was calculated according to IEEE STD P1528.4 recommendation [35], using ANSYS built-in SAR calculation module which calculates the local SAR at each mesh point as:

$$1gSAR = \frac{1}{2\rho}\sigma\|E\|^2 \quad [1]$$

where $\|E\|$ is the complex magnitude of the electric field, ρ is the density of the material, and σ is the material conductivity. The $1gSAR$ was calculated over a volume that surrounded each mesh point determined by the material's mass density to contain 1g mass. To quantify implant-induced SAR changes inside the tissue, the maximum of $1gSAR$ (referred to as $Max1gSAR_{Lead}$) was reported inside a conformal cylindrical volume (diameter 20 mm) that surrounded the lead. The maximum of the $1gSAR$ was also calculated throughout the body outside of this cylindrical volume and its value was reported for comparison (referred to as $Max1gSAR_{Body}$).

Thermal simulations

Thermal simulations were performed using the transient thermal analyses solver of ANSYS Mechanical (ANSYS Inc., Canonsburg, PA) which solves the Pennine bio heat equation:

$$c\rho\frac{\partial T}{\partial t} - \nabla k \nabla T = \rho(SAR) \quad [2]$$

where c is the specific heat capacity of the tissue (4150 Jkg⁻¹C⁻¹ W/m/C [36]), k is the isotropic thermal conductivity (0.42 W/m⁻¹C⁻¹ [36]), ρ is the density (1000 kg/m³), and T is the temperature. Thermal simulations were performed for all patients using local head coils and body coils at different imaging landmarks with input powers adjusted to produce a mean of $\|B_1^+\| = 3\mu T$. For the case that produced the maximum heating, simulations were reformed for a range of $\|B_1^+\|$ from 2 μT to 6 μT .

Results

SAR changes: 3T vs. 1.5 T

Figure 4 gives means and standard deviations of $Max1gSAR_{Lead}$ (averaged over all patients) at imaging landmarks L1–L5 for both 1.5 T and 3T coils, as well as the P-values of the paired 2 sample t-test assessing the null hypothesis that the difference between the means was zero. The results are given for the specific case of $B_1^+=3\mu T$ but can be scaled to other field values. Although a trend for higher SAR values at 3T was observed, the difference was not statistically significant (P-value> 0.06 at all imaging landmarks).

SAR changes: effect of imaging landmark

Figure 5 shows $Max1gSAR_{Lead}$ calculated for all patients at imaging landmarks L1 to L9, for both 1.5 T and 3 T head and body coils. Tables 2 and 3 give values of $Max1gSAR_{Lead}$ and $Max1gSAR_{Body}$ as well as their locations for the adjusted input power that produced a mean $\|B_1^+\| = 3\mu T$ on an axial plane passing through coil's iso-center.

For all patients and at both 1.5 T and 3 T, local head coils produced a negligible 1g-averaged SAR around the lead. The value of $Max1gSAR_{Lead}$ was observed to be below 1 W/kg at 1.5 T and below 4 W/kg at 3 T in all patients. These values are roughly 3 times less than the maximum reference SAR of 3 W/kg at 1.5 T and 12 W/kg at 3 T observed at the periphery of the head.

When using body coils, for all patients and at both resonant frequencies, the $Max1gSAR_{Lead}$ was substantially lower than $Max1gSAR_{Body}$ when the imaging landmark was at the umbilicus region (L5) or lower. $Max1gSAR_{Lead}$ values peaked when the landmark was at the level of mid-chest (L3), followed by landmark at mid-torso (L4), and neck levels (L2).

At 1.5 T, $Max1gSAR_{Lead}$ dropped on average by 72% compared to its peak value when the coil was at the umbilicus level and by more than 98% when the coil was at the pelvic level (L6) or lower. At 3 T, $Max1gSAR_{Lead}$ dropped on average by about 93% compared to its peak value when the coil was at the umbilicus level and by more than 99% when the coil was at the pelvic level (L6) or lower.

Temperature maps

Table 4 gives the maximum temperature rise T around the leads for each patient and at each imaging landmark during 10 minutes of continuous RF exposure (100% duty cycle). For all patients, the imaging landmark that generated the maximum local SAR also produced the maximum temperature rise in the tissue around the lead, which was landmark L3 at both 1.5 T and 3 T. Figure 6 gives time profiles of the maximum T (observed in patient 5 at L3) for different B_1^+ values at 1.5 T and 3 T, respectively.

At both 1.5 T and 3T, T remained below 0.1°C for all patients when a local transmit head coils was used. For body coils at 1.5 T, T remained <3°C at all imaging landmarks during a 10-minute scan at 100% duty cycle when $\|B_1^+\|$ was equal to 3 μ T and <3°C at all imaging landmarks for $\|B_1^+\| = 4\mu$ T for scan durations below 2.5 minutes.

For body coils at 3 T, T remained <3°C at all imaging landmarks for $\|B_1^+\| = 2\mu$ T when the scan duration was below 2.5 minutes and <5°C for scan durations up to 8 minutes. For $\|B_1^+\| = 3\mu$ T, T remained <7°C at all imaging landmarks for short scan durations below 2.5 minutes and <12°C for scan times up to 8 minutes.

Discussion

This study simulates, for the first time, the local SAR and temperature changes in patients with fractions of retained cardiac leads remaining in their torso during MRI at 1.5 T and 3 T. Simulations were performed on representative models based on real patient data. The effect of coil type (local transmit head coils vs body coils) and patients' position in the scanner was investigated by simulating RF coils at 9 different positions, ranging from head-centered to centered on the lower limbs.

Few studies have attempted to address the problem of tissue heating around abandoned cardiac leads, where they have relied on phantom experiments with limited lead configurations [37, 38]. Langman et al. [37] measured the temperature rise close to the tips of five custom length cardiac leads (20cm–60cm) positioned circularly along the circumference of a cylindrical phantom and placed at the iso-center of a 1.5 T MRI scanner. The authors concluded that based on their length, abandoned leads could show resonance behavior and exhibit greater heating than pacemaker attached leads. Mattei et al. [38] performed similar experiments on four different leads from MR Conditional cardiac pacing devices (lengths 52cm–65cm) positioned inside an ASTM body phantom where the leads were manually routed to mimic a typical implantation path. Similar to the former study, the authors concluded that the presence of abandoned leads could pose an additional risk for the patient. Both studies only investigated 1.5 T systems and considered a single imaging landmark, namely the phantom placed at the center of the RF coil. Although this central position is suggested to generate the maximum heating [39, 40], a quantitative measure of power deposition around the implant when the RF coil is positioned at other imaging landmarks provides useful comparative information. Moreover, with an increasing prevalence of clinical 3 T scanners and the wealth of information on safety of 1.5 T systems, a comparative study between 1.5 T and 3 T systems is also useful.

The importance of adopting a full-wave simulation scheme to understand and predict RF heating effects in the tissue has been established by the pioneering works of Ibrahim, Collins, and Jin [22, 40–43]. Recently simulations have been also extensively used to guide MRI hardware design to improve the safety of imaging patients with implants [44–47]. In the past 20 years, several models of RF birdcage coils have been introduced for this purpose following different levels of complexity. This work adopted a *specific* model which replicates the number and position of excitation ports in a physical coil and includes lumped elements for tuning and matching (i.e., resistors, capacitors). The technique has been shown to accurately predict the distribution of MRI RF fields inside both unloaded body coils and body coils loaded with standard phantoms [48], as well as head coils loaded with anthropomorphic head phantoms [47]. We compared the maximum of 1g-averaged SAR around leads ($Max1gSAR_{Lead}$) with the maximum of 1g-averaged SAR throughout the body ($Max1gSAR_{Body}$) in a group of six patients with retained endovascular leads. Our results showed that for both 1.5 T (64MHz) and 3 T (127 MHz) systems, local transmit head coils generated $Max1gSAR_{Lead}$ values substantially lower than the reference SAR observed at the periphery of the body. The maximum temperature rise around the lead during 10 minutes of continuous RF exposure was also negligible in all cases (<0.1 °C). Considering that the temperature rise in the tissue is approximately proportional to the magnitude of the dissipated power (see Eq 2), T remains $<3^{\circ}\text{C}$ for B_1^+ field values up to 15 μT . Most clinical neuroimaging pulse sequences produce B_1^+ values well below this limit (2 μT –4 μT) and have short durations (<5 minutes), thus such sequences can be safely applied using a local transmit head coil. This applies for example, to patients with retained cardiac leads who need head scans for a neurologic indication.

It is important to note however, that there is not a well-established universal consensus on the maximum local temperature rise that can be considered safe for implant imaging. Some guidelines for example, have considered $T = 3^{\circ}\text{C}$ to be the safety limit. More recent reports

that studied threshold of thermal damage in different body tissues have suggested a higher limit of 6°C for 26 minutes to induce any measurable effect in muscles [49]. As the majority of retained lead cases include the conductor being in touch with the vein wall muscles authors believe that the 6°C is a more reasonable threshold, especially considering that these are highly conservative calculations (no perfusion effect, 100 duty cycle).

For body imaging at 1.5 T, the maximum T remained <3°C at all imaging landmarks for $\|B_1^+\| \leq 3\mu\text{T}$ and <6°C for $\|B_1^+\| \leq 4\mu\text{T}$ for scan times with durations up to 10 minutes. For higher fields with $\|B_1^+\| \leq 6\mu\text{T}$, T remained <6°C for short scans up to 2 minutes. Our results indicate that body imaging these six patients, including imaging of the thoracic region, could be safely performed at 1.5 T for $\|B_1^+\|$ up to 4 μT . For body imaging at 3 T, the maximum T remained <6°C at all imaging landmarks for $\|B_1^+\| \leq 2\mu\text{T}$ during a 10-minute scan. For pulse sequences with $\|B_1^+\| \leq 3\mu\text{T}$, T remained <6°C for short scans up to 2 minutes. Typical cardiac imaging pulse sequences (e.g., 2 chamber cine true FISP) produce B_1^+ values in the range of 5 μT and have short durations typically <2 minutes, thus, can be safely applied on these patients at 1.5 T. At 3 T however, caution must be used and the use of low-power pulses with $\|B_1^+\| \leq 2\mu\text{T}$ is advised.

We observed a non-significant difference between the means of local SAR values around the lead at 1.5 T and 3 T. It should be noted however, that the N=6 is a relatively small sample size. From the results of this study, one can calculate the sample size required to obtain the study power of 0.8 (80% confidence that the means are actually not significantly different), by calculating the standardized difference between the two samples as [50]:

$$\text{standardized difference} = \frac{\text{difference in the mean}}{\text{sample standard deviation}}$$

and then find the corresponding sample size from the lookup tables. From Figure 4, the maximum difference between means of SAR values at 1.5 T and 3T was 65 W/kg which occurred at imaging landmark L3 (mid-chest). Based on this value and the standard deviation of 79 W/kg, the standardized difference between the two samples was 0.8, which gives the required sample size of N=26 [50].

The location of the $Max1gSAR_{Lead}$ changed at different imaging landmarks and was not always at the distal tip. Patient 5, for example, had two fractions of the lead separated by a small gap, and the maximum SAR was observed in the gap between the two pieces at some imaging landmarks (see Tables 2 and 3). As abrupt changes occur in the electric field at locations of breaks and discontinuities along a wire conductor, such locations should be always monitored for potential SAR amplification.

This work has some limitations that should be considered. First, temperature maps were calculated for only one value of tissue thermal conductivity at 0.42 W/m⁻¹C⁻¹. Tissue thermal conductivities have been measured to vary from 0.2 W/m⁻¹C⁻¹ to 0.5 W/m⁻¹C⁻¹ depending on the fat content, and the effect on the temperature increase has been shown to

be significant in ablation studies [51]. The extent of the sensitivity of MRI induced temperature rise around metallic implants to thermal conductivity of adjacent tissues remains to be evaluated and considered in uncertainty analysis. Moreover, the leads studied in this work were limited to endovascular leads with RV shock coil which are directly in contact with blood. The temperature rise around these leads can be mitigated by blood flow's cooling effect which was not included in modeling. In general, this effect can be beneficial as it can decrease the temperature rise to an amount that can be tolerated by the surrounding tissues. Furthermore, the current study used a homogenous body model and a generic geometry for the RF coils which should be noted as limitations.

The current study is also limited in that it presents the results of a relatively small representative group of patients which should not be taken as a surrogate for the larger population. Although cases presented here were selected after careful review of more than thirty patient charts, our database only consisted of retained leads in the left thoracic region. It has been shown however, that the position of the lead has a non-negligible effect on the heating at the tip [23]. Therefore, future studies should include scenarios where retained leads in both left and right thoracic regions are investigated. Our results however, strongly warrant the need for such studies as these patients are currently prevented from all types of MRI exams.

Finally, we emphasize the difference between “abandoned” leads and “retained” leads. Although these words are sometimes used interchangeably, they refer to two distinct categories of leads which should be treated separately from the MRI safety point of view. Abandoned leads are those that are disconnected from the CIED generator and left intact in the endovascular system. Their trajectory and geometry remains as originally implanted and thus, their tip is in contact with the cardiac tissue. This could cause an additional safety concern due to the transiently induced voltages from the gradient coils which might cause unintended heart pacing. Retained leads on the other hand, are dislodged and partially extracted from the body and in most cases (including all patients in our database) are no longer in contact with the cardiac tissue. Therefore, gradient-induced voltages do not pose a safety risk for retained leads. It is advisable however, to consult CT images of patients individually to ensure that this is in fact the case. If in a patient part of the retained lead was found to be in contact with the cardiac tissue, it is prudent to scan the patient while monitoring their electrocardiogram and stop the scanning if any anomalies in the heart rate were detected.

Acknowledgments

This work was supported by the NIH grants K99EB021320, R01EB006847, and R01MH111875.

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Figure 1. 3D rendering of CT images and frontal views of the chest X-rays of six representative patients with fractions of bare retained cardiac leads remaining in their torso. Details of leads' structure and dimension such as diameter and variations in the pitch were extracted from the X-ray.

Figure 2.

(A) Example of a reconstructed lead model for finite element simulations. (B) The mesh was visually inspected to ensure detailed features of the lead were well represented. (C) Lead model was then co-registered with the ANSYS human body model and inserted into the MRI RF coils model for simulations.

Figure 3.

(A): Relative locations of retained leads in the body model. (B): A closer image of leads, showing their detailed structure and locations of the maximum 1gSAR as give in Tables 2 and 3. (C): Locations of different imaging landmarks. Bottom row middle: Electromagnetic simulations calculating B_1^+ and 1gSAR. (D): Temperature maps showing the maximum temperature after 600 s of exposure.

Figure 4. Mean and standard deviation of $Max1gSAR_{Lead}$ (averaged over all patients) at different imaging landmarks for both 1.5 T and 3 T simulations for the case of $B_1^+=3\mu T$.

Figure 5. Max1gSARLead at (A) 1.5 T and (B) 3 T for the head coil and body coil at imaging landmarks L1–L9. The input power of the coils was adjusted to produce a mean $B_1^+ = 3\mu\text{T}$ on a transverse plane passing through coils' iso-center.

Figure 6.

Temperature profiles for the case of maximum temperature rise (patient 5, landmark L3) calculated for different values of $\|B_1^+\|$ at (A) 64 MHz RF exposure (1.5T) and (B) 127 MHz (3 T).

Table 1

Patient information for cases used in this study

Patient ID	Age	Lead type	Manufacturer and model	Date of implantation (m/y)	Date of extraction (m/y)	Lead Length [mm]	Wire Length [mm]
1	57	RV shock coil	Medtronic Lead 6949	8/2012	05/2013	45	469
2	29	RV shock coil	Medtronic Lead 6949	2/2005	08/2015	85	960
3	57	RV shock coil	Medtronic Lead 6947	4/2003	12/2009	155	543
4	56	RV shock coil	Medtronic Lead 6944	5/2002	04/2007	62	904
5	52	RV shock coil	Boston Scientific Lead 0185	1/2007	08/2011	89 & 44	133 & 65
6	22	RV lead, RA lead	Medtronic 5076 & 6931	9/2007	9/2012	134 & 105	679 & 105

Table 2

The values and locations of $MaxIgSAR_{Lead}$ and $MaxIgSAR_{Body}$ at different imaging landmarks for 1.5 T system.

1gSAR [W/kg] @ 1.5 T (64 MHz), mean $B_1^+ = 3\mu T$				
Landmark	Patient -model 1			
	$MaxIgSAR_{Lead}$ [W/kg]	$MaxIgSAR_{Lead}$ Location	$MaxIgSAR_{Body}$ [W/kg]	$MaxIgSAR_{Body}$ Location
Head Coil	<0.1	upper tip (A)	3	eyes and nose
L1	1	upper tip (A), lower tip (B)	11	lateral neck
L2	4	lower tip (B), upper tip (A)	12	latera neck, under arms
L3	7	lower tip (B), upper tip (A)	11	lateral neck, under arms, lateral waist
L4	6	lower tip (B)	12	lateral waist, under arms, hand fingers
L5	4	lower tip (B)	20	perineum, lateral waist, arms
L6	<0.5	NA	21	perineum, lateral waist, arms, under arms
L7	<0.1	NA	18	perineum, lateral buttocks, wrists
L8	<0.1	NA	12	legs, foot toes
L9	<0.1	NA	7	legs, foot toes
Landmark	Patient -model 2			
	$MaxIgSAR_{Lead}$ [W/kg]	$MaxIgSAR_{Lead}$ Location	$MaxIgSAR_{Body}$ [W/kg]	$MaxIgSAR_{Body}$ Location
Head Coil	<0.1	upper tip (A)	3	between the eyes
L1	3	upper tip (A)	10	lateral neck
L2	6	upper tip (A)	12	latera neck, under arms
L3	8	upper tip (A)	12	lateral neck, under arms, lateral waist
L4	6	upper tip (A)	13	lateral waist, under arms, hand fingers
L5	1	lower tip (B)	17	perineum, lateral waist, arms
L6	<0.1	NA	23	perineum, lateral waist, arms, under arms
L7	<0.1	NA	19	perineum, lateral buttocks, wrists
L8	<0.1	NA	10	legs, foot toes
L9	<0.1	NA	7	legs, foot toes
Landmark	Patient -model			
	$MaxIgSAR_{Lead}$ [W/kg]	$MaxIgSAR_{Lead}$ Location	$MaxIgSAR_{Body}$ [W/kg]	$MaxIgSAR_{Body}$ Location
Head Coil	1	upper tip (A)	4	between the eyes
L1	15	lower tip (B)	11	lateral neck
L2	32	lower tip (B)	11	latera neck, under arms
L3	45	lower tip (B)	12	lateral neck, under arms, lateral waist
L4	35	lower tip (B)	12	lateral waist, under arms, hand fingers
L5	10	lower tip (B)	20	perineum, lateral waist, arms
L6	<0.5	NA	24	perineum, lateral waist, arms, under arms
L7	<0.5	NA	21	perineum, lateral buttocks, wrists

L8	<0.5	NA	11	legs, foot toes
L9	<0.1	NA	7	legs, foot toes
Landmark	Patient -model 4			
	$MaxI_{gSAR_{Lead}}$ [W/kg]	$MaxI_{gSAR_{Lead}}$ Location	$MaxI_{gSAR_{Body}}$ [W/kg]	$MaxI_{gSAR_{Body}}$ Location
Head Coil	<0.1	upper tip (A)	3	between the eyes
L1	7	upper tip (A)	11	lateral neck
L2	21	upper tip (A)	12	latera neck, under arms
L3	26	upper tip (A), at the pitch change (C)	13	lateral neck, under arms, lateral waist
L4	24	upper tip (A)	14	lateral waist, under arms, hand fingers
L5	9	upper tip (A)	19	perineum, lateral waist, arms
L6	<0.5	NA	21	perineum, lateral waist, arms, under arms
L7	<0.2	NA	19	perineum, lateral buttocks, wrists
L8	<0.1	NA	10	legs, foot toes
L9	<0.1	NA	7	legs, foot toes
Landmark	Patient -model 5			
	$MaxI_{gSAR_{Lead}}$ [W/kg]	$MaxI_{gSAR_{Lead}}$ Location	$MaxI_{gSAR_{Body}}$ [W/kg]	$MaxI_{gSAR_{Body}}$ Location
Head Coil	1	upper tip (A)	3	between the eyes
L1	10	gap between two pieces (C)	10	lateral neck
L2	30	gap between two pieces (C)	11	latera neck, under arms
L3	48	upper tip (A)	12	lateral neck, under arms, lateral waist
L4	30	upper tip (A)	11	lateral waist, under arms, hand fingers
L5	14	gap between two pieces (C)	20	perineum, lateral waist, arms
L6	1	gap between two pieces (C)	25	perineum, lateral waist, arms, under arms
L7	<0.1	NA	21	perineum, lateral buttocks, wrists
L8	<0.1	NA	10	legs, foot toes
L9	<0.1	NA	7	legs, foot toes
Landmark	Patient -model 6			
	$MaxI_{gSAR_{Lead}}$ [W/kg]	$MaxI_{gSAR_{Lead}}$ Location	$MaxI_{gSAR_{Body}}$ [W/kg]	$MaxI_{gSAR_{Body}}$ Location
Head Coil	1	upper tip of RA lead (A)	3	between the eyes
L1	24	lower tip of the RA lead (C)	11	lateral neck
L2	34	lower tip of the RA lead (C)	12	latera neck, under arms
L3	42	lower tip of the RA lead (C)	11	lateral neck, under arms, lateral waist
L4	38	lower tip of the RA lead (C)	14	lateral waist, under arms, hand fingers
L5	12	lower tip of the RV lead (B)	19	perineum, lateral waist, arms
L6	<0.5	NA	18	perineum, lateral waist, arms, under arms
L7	<0.5	NA	19	perineum, lateral buttocks, wrists
L8	<0.1	NA	10	legs, foot toes
L9	<0.1	NA	6	legs, foot toes

Table 3

The values and locations of $MaxI_{gSAR_{Lead}}$ and $MaxI_{gSAR_{Body}}$ at different imaging landmarks for 3 T system.

1gSAR [W/kg] @ 3 T (127 MHz), mean $B_1^+ = 3\mu T$				
Landmark	Patient -model 1			
	$MaxI_{gSAR_{Lead}}$ [W/kg]	$MaxI_{gSAR_{Lead}}$ Location	$MaxI_{gSAR_{Body}}$ [W/kg]	$MaxI_{gSAR_{Body}}$ Location
Head Coil	<0.1	upper tip (A)	11	between the eyes
L1	3	upper tip (A)	50	frontal neck
L2	6	along the whole lead	42	frontal neck
L3	18	lower tip (B)	78	lateral and frontal neck, under arms, shoulders, wrists
L4	9	lower tip (B)	42	arms, wrists, hand fingers
L5	5	lower tip (B)	63	upper arms
L6	2	lower tip (B)	85	elbow, under arm, upper thigh
L7	<0.1	NA	32	wrists, upper thighs
L8	<0.1	NA	29	upper thighs
L9	<0.1	NA	34	upper thighs
Landmark	Patient -model 2			
	$MaxI_{gSAR_{Lead}}$ [W/kg]	$MaxI_{gSAR_{Lead}}$ Location	$MaxI_{gSAR_{Body}}$ [W/kg]	$MaxI_{gSAR_{Body}}$ Location
Head Coil	<0.2	upper tip (A)	10	between the eyes
L1	10	upper tip (A)	60	frontal neck
L2	12	upper tip (A)	42	frontal neck
L3	25	upper tip (A)	65	lateral and frontal neck, under arms, shoulders, wrists
L4	8	upper tip (A)	37	arms, wrists, hand fingers
L5	2	lower tip (B)	56	upper arms
L6	1	lower tip (B)	70	elbow, under arm, upper thigh
L7	<0.5	NA	29	wrists, upper thighs
L8	<0.1	NA	26	upper thighs
L9	<0.1	NA	36	upper thighs
Landmark	Patient -model 3			
	$MaxI_{gSAR_{Lead}}$ [W/kg]	$MaxI_{gSAR_{Lead}}$ Location	$MaxI_{gSAR_{Body}}$ [W/kg]	$MaxI_{gSAR_{Body}}$ Location
Head Coil	3	upper tip (A)	12	between the eyes
L1	11	lower tip (B)	56	frontal neck
L2	19	lower tip (B)	42	frontal neck
L3	42	lower tip (B)	55	lateral and frontal neck, under arms, shoulders, wrists
L4	23	lower tip (B)	43	arms, wrists, hand fingers
L5	5	upper tip (A)	65	upper arms

L6	1	upper tip (A)	89	elbow, under arm, upper thigh
L7	<0.5	NA	37	wrists, upper thighs
L8	<0.1	NA	26	upper thighs
L9	<0.1	NA	36	upper thighs
Landmark	Patient -model 4			
	$Max I g SAR_{Lead}$ [W/kg]	$Max I g SAR_{Lead}$ Location	$Max I g SAR_{Body}$ [W/kg]	$Max I g SAR_{Body}$ Location
Head Coil	1	upper tip (A)	10	between the eyes
L1	31	upper tip (A), lower tip (B)	61	frontal neck
L2	50	upper tip (A), at the pitch change (C)	48	frontal neck
L3	123	upper tip (A), at the pitch change (C)	77	lateral and frontal neck, under arms, shoulders, wrists
L4	33	upper tip (A), at the pitch change (C)	53	arms, wrists, hand fingers
L5	4	upper tip (A), lower tip (B)	66	upper arms
L6	1	lower tip (B)	97	elbow, under arm, upper thigh
L7	<0.1	NA	30	wrists, upper thighs
L8	<0.1	NA	27	upper thighs
L9	<0.1	NA	33	upper thighs
Landmark	Patient -model 5			
	$Max I g SAR_{Lead}$ [W/kg]	$Max I g SAR_{Lead}$ Location	$Max I g SAR_{Body}$ [W/kg]	$Max I g SAR_{Body}$ Location
Head Coil	4	upper tip (A)	10	between the eyes
L1	53	at the gap between two pieces (C)	55	frontal neck
L2	100	lower tip (B), at the gap between two pieces (C)	44	frontal neck
L3	218	upper tip (A)	64	lateral and frontal neck, under arms, shoulders, wrists
L4	84	upper tip (A)	43	arms, wrists, hand fingers
L5	13	at the gap between two pieces (C)	66	upper arms
L6	2	at the gap between two pieces (C)	71	elbow, under arm, upper thigh
L7	1	at the gap between two pieces (C)	29	wrists, upper thighs
L8	<0.5	NA	30	upper thighs
L9	<0.1	NA	40	upper thighs
Landmark	Patient -model 6			
	$Max I g SAR_{Lead}$ [W/kg]	$Max I g SAR_{Lead}$ Location	$Max I g SAR_{Body}$ [W/kg]	$Max I g SAR_{Body}$ Location
Head Coil	4	upper tip of RA lead (A)	10	between the eyes
L1	25	lower tip of the RA lead (C)	55	frontal neck
L2	69	lower tip of the RA lead (C)	53	frontal neck
L3	139	lower tip of the RA lead (C)	59	lateral and frontal neck, under arms, shoulders, wrists
L4	64	lower tip of the RA lead (C)	43	arms, wrists, hand fingers
L5	13	lower tip of RV lead (B)	70	upper arms

L6	1	lower tip of RA lead (C)	85	elbow, under arm, upper thigh
L7	<0.5	NA	26	wrists, upper thighs
L8	<0.1	NA	29	upper thighs
L9	<0.1	NA	32	upper thighs

Table 4

Maximum temperature rise around the lead after 10 minutes continuous RF exposure, $B_1^+ = 3\mu T$, 64 MHz

T_{max} around the lead after 10 minutes continuous RF exposure, $B_1^+ = 3\mu T$, 64 MHz						
	ID1	ID2	ID3	ID4	ID5	ID6
Head Coil	<0.1	<0.1	<0.1	<0.1	<0.1	<0.1
Body Coil						
L1	0.1	0.1	0.6	0.2	0.7	0.2
L2	0.2	0.2	1	0.5	2.3	1.5
L3	0.3	0.5	1.5	0.6	2.9	2
L4	0.3	0.3	1.2	0.5	2.3	1.8
L5	0.2	0.1	0.3	0.2	1	0.8
L6	<0.1	<0.1	<0.1	<0.1	<0.1	<0.1
L7	<0.1	<0.1	<0.1	<0.1	<0.1	<0.1
L8	<0.1	<0.1	<0.1	<0.1	<0.1	<0.1
L9	<0.1	<0.1	<0.1	<0.1	<0.1	<0.1

T_{max} around the lead after 10 minutes continuous RF exposure, $B_1^+ = 3\mu T$, 127 MHz						
	ID1	ID2	ID3	ID4	ID5	ID6
Head Coil	<0.1	<0.1	0.1	<0.1	<0.1	0.1
Body Coil						
L1	0.4	0.6	0.5	1.4	3.9	1
L2	0.3	0.6	0.8	2.4	7.7	2.7
L3	1.4	1	1.8	5.7	12.0	5.6
L4	0.7	0.4	1	1.4	6.8	2.4
L5	0.3	0.2	0.2	0.2	1.1	0.6
L6	0.2	0.1	<0.1	0.1	0.1	<0.1
L7	<0.1	<0.1	<0.1	<0.1	0.1	<0.1

$ 1,0\rangle$	$ 1,0\rangle$	$ 1,0\rangle$	$ 1,0\rangle$	$ 1,0\rangle$	$ 1,0\rangle$	61
$\langle 1,0 $	$\langle 1,0 $	$\langle 1,0 $	$\langle 1,0 $	$\langle 1,0 $	$\langle 1,0 $	81

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