Clinical Study

Stereotactic and Functional Neurosurgery

Stereotact Funct Neurosurg 2008;86:92–100
DOI: 10.1159/000112430

Published online: December 12, 2007

Magnetic Resonance Imaging of Implanted Deep Brain Stimulators: Experience in a Large Series

Paul S. Larson[a] R. Mark Richardson[a] Philip A. Starr[a] Alastair J. Martin[b]

Departments of [a]Neurological Surgery and [b]Radiology, University of California, San Francisco, Calif., USA

Key Words
Deep brain stimulation · Specific absorption rate · Magnetic resonance imaging safety

Abstract
Magnetic resonance imaging (MRI) is a commonly used and important imaging modality to evaluate lead location and rule out complications after deep brain stimulation (DBS) surgery. Recent safety concerns have prompted new safety recommendations for the use of MRI in these patients, including a new recommendation to limit the specific absorption rate (SAR) of the MRI sequences used to less than 0.1 W/kg. Following SAR recommendations in real-world situations is problematic for a variety of reasons. We review our experience scanning patients with implanted DBS systems over a 7-year period using a variety of scanning techniques and four scanning platforms. 405 patients with 746 implanted DBS systems were imaged using 1.5-tesla MRI with an SAR of up to 3 W/kg. Many of the DBS systems were imaged multiple times, for a total of 1,071 MRI events in this group of patients with no adverse events. This series strongly suggests that the 0.1 W/kg recommendation for SAR may be unnecessarily low for the prevention of MRI-related adverse events.

Introduction

Postoperative magnetic resonance imaging (MRI) following deep brain stimulation (DBS) surgery is used by many centers to determine lead location relative to the desired brain target and rule out surgical complications such as intracranial hemorrhage [1–9]. Knowledge of appropriate lead location is important in predicting patient outcome and facilitates DBS programming. The immediate feedback that postoperative imaging provides is also critically important for quality assurance, as it allows members of the implanting team involved in microelectrode recording, macrostimulation or other methods of physiological mapping to interpret their intraoperative findings and correlate them with the final electrode placement. Recent safety concerns following two reported cases of patient injury, presumably due to heating of implanted DBS components during MRI, have prompted the company who manufactures the only FDA-approved DBS system to release updated guidelines for MRI of their devices [10–12]. These include a new recommendation to use sequences that limit the applied head specific absorption rate (SAR) to 0.1 W/kg. The new recommendation for SAR represents a reduction from the previously recommended 0.4 W/kg, which was already well below the permitted whole-body SAR limit of 4 W/kg for patients without these devices [12].

KARGER

Fax +41 61 306 12 34
E-Mail karger@karger.ch
www.karger.com

1011–6125/08/0862–0092$24.50/0

Accessible online at:
www.karger.com/sfn

Paul S. Larson, MD
Department of Neurological Surgery, University of California, San Francisco
505 Parnassus Avenue, Box 0112
San Francisco, CA 94143-0112 (USA)
Tel. +1 415 353 3489, Fax +1 415 353 3459, E-Mail larsonp@neurosurg.ucsf.edu

Following SAR recommendations in real-world situations is problematic for a variety of reasons, including varying definitions of SAR across different MRI platforms and the fact that MRI protocols that optimally image DBS leads and subcortical structures tend to have much higher SARs than current recommendations allow. We present our group's experience with imaging over 700 implanted DBS leads using multiple 1.5-tesla MRI platforms and varying techniques over a 7-year period.

Materials and Methods

A retrospective review of a secure database containing a portion of patients implanted in our program from 1999 to March 2006 was performed. Implantations were performed by two surgeons (P.S.L., P.A.S.) at the University of California San Francisco Medical Center and the San Francisco VA Medical Center. Data concerning surgical target, number and sequence of lead implantation (unilateral, staged bilateral, simultaneous bilateral) and pulse generator (IPG) type were obtained. Imaging was performed on 4 1.5-tesla MRI scanners at two hospitals: Siemens Magnetom Vision, Siemens Magnetom Symphony, Philips Intera and General Electric Horizon (Siemens AG, Munich, Germany; Philips Medical Systems, Best, The Netherlands; General Electric Medical Systems, Milwaukee, Wisc., USA). All scanners underwent numerous system upgrades during the time period of this review. MRI was used exclusively for both preoperative targeting and postoperative scanning in all patients. Postoperative imaging was generally obtained immediately after surgery, or in some cases on the following day. Image sequences specified by our group were performed in all patients, with some postoperative scans including additional sequences if deemed necessary by the implanting surgeon or radiology staff.

Because many DBS components were scanned multiple times, it was necessary to define an 'MRI event'. An MRI event was defined as a complete, fully internalized DBS system (one lead connected to a pulse generator) undergoing a unique MRI session. Each MRI event consisted of at least two sequence acquisitions. Sequences were repeated if the images were degraded by patient motion. Our practice is to implant the brain electrode and pulse generator on the same day. Accordingly, for staged bilateral implantations, the first complete DBS system implanted was counted as undergoing multiple MRI events: postoperatively after first-side implantation, preoperatively for targeting of the second side, and postoperatively after second-side implantation (a total of 3 MRI events for that lead and pulse generator). Two leads connected to a Medtronic Kinetra dual-channel pulse generator (Medtronic Inc., Minneapolis, Minn., USA) were counted as two separate systems undergoing two separate MRI events.

A search was made through our Philips scanner image archives to reconstruct the scanning conditions and SAR for a random sampling of patients from our DBS database. SAR is a value that is displayed on the MRI console at the time of scanning and is not recorded in routine clinical practice. The exact calculation of SAR can depend on the software version the scanner is running at the time and the radiofrequency (RF) coils being used [13]. On GE and Siemens platforms, its calculation is also dependent on the patient's weight as entered by the technologist performing the scan, and therefore is not readily available for retrospective analysis. However, the Philips platforms predict SAR independent of patient weight, so for patients scanned on this machine retrospective data regarding coil type, sequences acquired, scan time and SAR were available. Retrospective data were also available for 1 patient scanned on the Siemens platform.

Results

MRI events were confirmed in 405 patients with 746 implanted DBS systems. Implanted targets included the thalamus, subthalamic nucleus, globus pallidus and the hypothalamus. 100 leads were placed unilaterally, 265 were placed as staged bilateral implants, and 381 were placed as simultaneous bilateral implants. Some patients had more than 2 brain leads implanted. 739 of the leads placed were Medtronic Model 3387, and 7 were ANS Quattrode leads (Advanced Neuromodulation Systems, Plano, Tex., USA). Lead extension lengths for the Medtronic devices included 40-, 51- and 66-cm extensions. Pulse generators implanted included Medtronic Itrel II, Soletra and Kinetra models as well as 7 ANS Libra models. There were 4 pulse generators placed in the abdomen; the remainder were placed in the chest just below the clavicle. Preoperative imaging sequences consisted of a minimum of whole-brain 3-dimensional gradient echo acquisition as well as either T_2-weighted fast spin echo (FSE) or inversion recovery FSE imaging as a slab through the target region. Postoperative imaging consisted of whole-brain 3-dimensional gradient echo and axial T_2-weighted slab acquisitions. In addition, some patients received other postoperative imaging sequences, including diffusion-weighted imaging, fluid-attenuated inversion recovery and MR angiography. Early in our series, our practice was to optimize MRI sequences to obtain adequate visualization of the DBS lead and the intended target with no regard to SAR whatsoever. Likewise, we initially did not specify the use of specific RF coil types. A summary of the older imaging protocols for each of our scanning platforms at the time is shown in table 1. Our more recent imaging protocols with relatively lower SAR sequences are outlined in table 2.

A total of 1,071 MRI events were identified in this group of patients with no adverse events. All patients were scanned with sequences having a SAR well in excess of 0.1 W/kg, in most cases many times greater. The specific scan parameters and sequences obtained for a ran-

Table 1. Initial MRI protocols used for preoperative and postoperative imaging at our center

MRI protocol	Parameter	1.5-tesla MRI scanner (model)		
		General Electric (Horizon)	Siemens (Vision)	Philips (Intera)
T_2-weighted FSE	Field of view	260 mm	260 mm	260 mm
	Matrix	384 × 256	256 × 256	512 × 268
	Slice thickness	2 mm	2 mm	2 mm
	Number of slices	24	26	21
	Interleaved	yes	yes	yes
	TR/TE/flip angle	3,000 ms/87 ms/90°	2,500 ms/108 ms/90°	3,000 ms/90 ms/90°
	Echo train length	10	9	16
	Bandwidth	15.6 kHz	195 Hz/pixel	183 Hz/pixel
	Signal averages	4	4	6
	Scan time	10:48	9:08	8:42
	Reported SAR	varies by patient weight	varies by patient weight	3.0 W/kg
3-Dimensional GRE	Field of view	260 mm	MPRAGE 260 mm	260 mm
	Matrix	256 × 192	256 × 256	256 × 192
	Slice thickness	1.5 mm	1.5 mm	1.5 mm
	Number of slices	114	128	120
	TR/TE/flip angle	36 ms/8 ms/35°	15 ms/4.4 ms/15°	20 ms/2.9 ms/30°
	Bandwidth	15.6 kHz	130 Hz/pixel	259 Hz/pixel
	Signal averages	0.75	1	1
	Scan time	11:00	8:10	8:46
	Reported SAR	varies by patient weight	varies by patient weight	0.6 W/kg
IR FSE	Field of view	NA	260 mm	260 mm
	Matrix		256 × 230	512 × 256
	Slice thickness		2 mm	2 mm
	Number of slices		27	28
	Interleaved		yes	yes
	TR/TE/flip angle		3,000 ms/40 ms/90°	3,000 ms/40 ms/90°
	Inversion time		200 ms	200 ms
	Echo train length		9	5
	Bandwidth		208 Hz/pixel	120 Hz/pixel
	Signal averages		3	3
	Scan time		11:53	11:48
	Reported SAR		varies by patient weight	1.4 W/kg

GRE = Gradient echo; IR = inversion recovery; TR = repetition time; TE = echo time; MPRAGE = magnetization-prepared rapid gradient echo.

dom sample of 13 MRI events is shown in table 3. A combination of body transmit/head receive and head transmit/receive (T/R) coils were used, with SAR ranging from 0.1 to 3.0 W/kg. A representative postoperative scan using our current imaging protocols is shown in figure 1. This patient had cervical and truncal dystonia as well as spasmodic dysphonia. The patient received simultaneous bilateral subthalamic nucleus and globus pallidus DBS implantations with 4 Medtronic 3387 leads and 2 Kinetra pulse generators placed in the same setting. Pre- and postoperative imaging was obtained with a Philips Intera 1.5T, running the Achieva release 1.5 software package and using a head T/R coil. The 3-dimensional T_1-weighted sequence head SAR was 0.3 W/kg. The T_2-weighted FSE sequence head SAR was 1.4 W/kg.

Discussion

In this study, we reviewed our experience with MRI of over 700 DBS electrodes, using a variety of MRI scanners and protocols, over a 7-year period. DBS has become a widely accepted and adopted treatment modality for a variety of disorders [14–23]. Many advocate postopera-

Table 2. Current MRI protocols with relatively lower SAR used for preoperative and postoperative imaging at our center

MRI protocol	Parameter	1.5-tesla MRI scanner (model)		
		General Electric (Signa)	Siemens (Symphony)	Philips (Achieva)
T_2-weighted FSE	Field of view	260 mm	260 mm	260 mm
	Matrix	384 × 224	256 × 256	384 × 262
	Slice thickness	2 mm	2 mm	2 mm
	Number of slices	24	26	21
	Interleaved	yes	yes	yes
	TR/TE/flip angle	2,500 ms/90 ms/90°	2,500 ms/108 ms/90°	3,000 ms/90 ms/90°
	Echo train length	10	9	16
	Bandwidth	15.63 kHz	195 Hz/pixel	147 Hz/pixel
	Signal averages	4	4	6
	Scan time	8:10	9:47	8:42
	Reported SAR	1.67 W/kg (for a patient weight of 74.8 kg)	1.4 W/kg (for a patient weight of 74.8 kg)	0.5 W/kg
3-Dimensional GRE	Field of view	260 mm	MPRAGE 260 mm	260 mm
	Matrix	256 × 160	256 × 256	192 × 192
	Slice thickness	1.5 mm	1 mm	1.5 mm
	Number of slices	114	128	120
	TR/TE/flip angle	33 ms/3 ms/35°	1910 ms/10 ms/15°	20 ms/2.9 ms/35°
	Bandwidth	15.63 kHz	130 Hz/pixel	259 Hz/pixel
	Signal averages	0.75	1	1
	Scan time	7:55	8:10	8:50
	Reported SAR	0.50 W/kg (for a patient weight of 74.8 kg)	0.1 W/kg (for a patient weight of 74.8 kg)	0.3 W/kg
IR FSE	Field of view	NA	260 mm	260 mm
	Matrix		256 × 230	304 × 228
	Slice thickness		2 mm	2 mm
	Number of slices		27	28
	Interleaved		yes	yes
	TR/TE/flip angle		3,000 ms/48 ms/90°	3,000 ms/40 ms/90°
	Inversion time		200 ms	200 ms
	Echo train length		9	5
	Bandwidth		208 Hz/pixel	120 Hz/pixel
	Signal averages		3	3
	Scan time		11:53	11:54
	Reported SAR		0.8 W/kg (for a patient weight of 74.8 kg)	0.8 W/kg

GRE = Gradient echo; IR = inversion recovery; TR = repetition time; TE = echo time; MPRAGE = magnetization-prepared rapid gradient echo. All protocols use head T/R coils. SAR for General Electric and Siemens platforms based on a patient weight of 74.8 kg.

tive imaging with either computerized tomography (CT) or MRI to confirm lead location and to rule out complications such as hemorrhage. CT has the advantage of having no potential interactions with the metallic components of a DBS system, but offers no direct visualization of the brain target due to poor tissue discrimination. Postoperative CT must be fused with preoperative MRI to provide any meaningful data regarding lead location, a technique that is time consuming and can introduce error to this process [24–26].

Implanted stimulation devices such as deep brain stimulators, dorsal column stimulators and cardiac pacemakers are potentially problematic when interacting with RF energy in a magnetic field [27–30]. These devices typically consist of a long electrode that runs some distance through the body from the pulse generator to the end organ, often with random turns or loops along its length. The RF pulses and magnetic gradient fields used during MRI produce currents that can be conducted and focused by these devices, with resultant heating at the stimulating

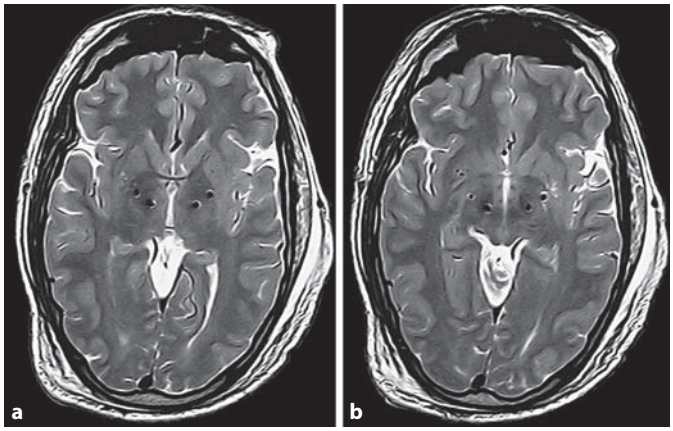

Fig. 1. T_2-weighted images from a patient with bilateral globus pallidus interna and subthalamic nucleus electrodes. Images shown are directly from the MRI scanner and not reformatted to be parallel to the intercommissural (AC-PC) plane; there is some asymmetry with the left side of the brain slightly higher than the right. The slice close to the AC-PC plane shows lateral leads relative to the globus pallidus (**a**), and the slice approximately 4 mm below the AC-PC plane shows medial leads relative to the subthalamic nucleus (**b**).

surfaces of the electrode [31–34]. Heating of the stimulating surfaces in a cardiac pacemaker can cause local degradation of tissue contact that can reduce the effectiveness of the device, but this phenomenon is particularly worrisome for devices placed in the brain, as excessive tissue heating is associated with permanent damage [35, 36]. Temperature changes in the range of 5–7°C are associated with reversible lesions, while temperature elevations of 8°C or more are believed to cause irreversible thermal injury [31, 37]. In addition, the pulse generators of stimulating devices can be switched on and off repeatedly during MRI, causing unwanted, target-specific side effects such as paresthesias in the case of DBS systems implanted in the thalamus or subthalamic nucleus.

Manufacturer Guidelines

In recent years, Medtronic has evolved specific guidelines for the use of MRI in patients with implanted DBS components. Those guidelines were most recently updated in November 2005 to include the following: (1) turn the device off, with the amplitude set to zero and the stimulator mode in bipolar, (2) use only horizontal bore MRI systems operating at a static magnetic field of 1.5 T, (3) use only a T/R head coil that does not extend over the chest (IPG) area, (4) enter the correct patient weight into the MRI console, (5) limit the gradient field to 20 T/s or less, and (6) use exam parameters that limit the displayed average head SAR (or applied SAR if known) to 0.1 W/kg or less for all RF pulse sequences. The last requirement represents a decrease in SAR from the prior recommendation of 0.4 W/kg [12, 38].

The concept of SAR was developed to provide a measure for the rate of RF energy absorption by the body and was first proposed in 1979 by the National Council on Radiation Protection and Measurements. At the time, the agency was primarily concerned with exposure to radio waves, which is the frequency range in which MRI systems operate. The aim of measuring SAR in MRI systems is to limit the rise in tissue temperatures due to RF energy deposition, and mirrors restrictions that are placed on other RF sources such as cell phones and radio towers. It can be difficult to measure, however, so MRI manufacturers typically use proprietary 'predictive models' to assure that the industry standards for allowable SAR are not broken; they further employ variable safety margins and frequently report their SAR values as 'less than' rather than providing a specific value. The net effect is that SAR is a rather poor index for predicting the potential for localized heating near DBS leads, but it is the only measure currently available.

From a practical standpoint, following a recommendation based on SAR is problematic for a variety of reasons. SAR is a calculated estimate, not a measured value, and therefore the SAR that is presented to the MRI technologist at the time of scanning is dependent on many factors. The method of SAR calculation itself is not standardized and varies from scanner to scanner. Two identical model scanners from the same manufacturer can differ in their method of SAR calculation if the software version they are running is different. In some scanners, such as our GE and Siemens platforms, the SAR calculation is dependent on the MRI technologist entering the patient's body weight. This is often inaccurately recorded in the chart, underestimated by the patient and/or poorly estimated by the technologist, as was the case with patient 13 in table 3 (weight entered 104 kg, actual weight 132 kg). In some instances, the scanner will not proceed with a particular sequence if the SAR exceeds a certain level set by the scanner software. Some technologists will intentionally enter an inaccurate body weight in such circumstances in order to allow scanning to proceed [Larson, pers. commun.].

Reported Adverse Events

The issue of safety in patients undergoing MRI with implanted DBS systems came to the forefront several

Table 3. Scan conditions and SAR for a sample of MRI events

Patient No.	Year	Implanted lead(s), IPG(s)	Scanner	Coils used	Sequences	Scan time	SAR W/kg
1	2001	Medtronic 3387, Soletra	Philips	body transmit, birdcage head receive	3D GRE IR TSE	8:50 11:54	0.6 1.6
2	2002	Medtronic 3387, Itrel II	Philips	body transmit, birdcage head receive	3D GRE IR TSE	8:50 11:54	0.6 1.6
3	2002	Medtronic 3387 ×2, Soletra ×2	Philips	body transmit, birdcage head receive	3D GRE T_2-weighted axial diffusion FLAIR	8:50 8:42 1:06 3:53	0.6 3.0 1.0 1.6
4	2003	Medtronic 3387 ×2, Soletra ×2	Philips	body transmit, birdcage head receive	3D GRE 3D GRE[1] T_2-weighted axial T_2-weighted axial[1]	8:50 8:50 8:42 8:42	0.6 0.6 3.0 3.0
5	2004	Medtronic 3387 ×2, Soletra ×2	Philips	body transmit, birdcage head receive	T_1-weighted GRE 3D GRE T_2-weighted axial T_2-weighted axial[1]	3:10 8:50 8:42 8:42	1.8 0.6 3.0 3.0
6	2004	Medtronic 3387 ×4, Soletra ×4	Philips	body transmit, SENSE head receive	3D GRE T_2-weighted axial	8:50 8:42	0.6 3.0
7	2004	Medtronic 3387, Soletra	Philips	body transmit, SENSE head receive	T_1-weighted GRE 3D GRE T_2-weighted axial	3:10 8:50 8:42	1.8 0.6 3.0
8	2004	Medtronic 3387 ×2, Soletra ×2	Philips	body transmit, birdcage head receive	T_1-weighted sagittal 3D GRE T_2-weighted axial	2:30 8:50 8:42	1.0 0.6 3.0
9	2005	Medtronic 3387 ×2, Kinetra	Philips	body transmit, SENSE head receive	T_1-weighted sagittal 3D GRE T_2-weighted axial	2:30 8:50 8:42	1.0 0.6 3.0
10	2005	Medtronic 3387 ×2, Soletra ×2	Philips	body transmit, birdcage head receive	3D GRE T_2-weighted axial	8:50 8:42	0.6 3.0
11	2005	Medtronic 3387 ×2, Kinetra	Philips	body transmit, birdcage head receive	3D GRE T_2-weighted axial T_2-weighted axial[1]	8:50 8:42 8:42	0.6 3.0 3.0
12	2006	Medtronic 3387 ×2, Kinetra	Philips	head T/R	3D GRE T_2-weighted axial	8:50 8:42	0.3 1.4
13	2006	ANS Quattrode ×2, Libra ×2	Siemens	head T/R	T_1-weighted MPRAGE T_2-weighted axial	8:10 9:47	0.1[2] 1.5[2]

Philips scanner (patients 1–12) predicts SAR independent of patient weight, so SAR for a given sequence and coil arrangement will be the same from patient to patient. Siemens scanner (patient 13) calculates SAR based on patient weight as entered by the MRI technologist. GRE = Gradient echo; IR = inversion recovery; TSE = turbo spin echo; FLAIR = fluid-attenuated inversion recovery; MPRAGE = magnetization-prepared rapid gradient echo.

[1] Sequence repeated due to motion.

[2] Siemens scanner, patient weight entered by the technologist 104 kg, actual patient weight 132 kg.

years ago after two reports of serious injury during MRI. The first case involved a patient with bilateral DBS leads and Soletra pulse generators, one in the typical subclavicular location on the right and the other in the abdominal wall with a longer lead extension on the left. The patient had a lumbar MRI scan using a 1.0-tesla scanner and a whole-body RF coil. The patient emerged with altered mental status, and subsequent CT and MRI of the brain showed a 2- to 3-cm presumed thermocoagulation lesion around the left brain lead with a small amount of associated hemorrhage. There was no injury associated with the right-sided lead, which was connected to the IPG in the

chest. The patient was left with permanent disability [10, 39]. The second case involved a patient with bilaterally implanted DBS leads that were externalized. The patient underwent cranial MRI using a head T/R coil, also in a 1.0-tesla scanner, with both leads fixed outside the coil in a straight manner. The patient emerged with dystonic and ballistic movements in the left leg, and although CT imaging was inconclusive and the patient did not consent to further MRI, an acute thermal injury of the right subthalamic nucleus was presumed. The dystonic and ballistic left leg movements resolved by 17 months after the incident [11].

Both of these cases represent a unique set of circumstances that likely contributed to or were the sole cause of these adverse events. Both incidents occurred in 1.0-tesla scanners which are uncommon, use a different RF wavelength and whose field interactions with DBS systems are not well characterized, although there is a recent report of 1.0-tesla MRI being used for postoperative imaging in DBS patients [40]. In our series, all patients were scanned on 1.5-tesla platforms. The first case involved the use of a whole-body RF coil for spine imaging and a patient with an abdominally placed IPG. Earlier in our series, before the potential dangers of using body coils were described, we scanned many patients using body transmit/head receive coils without adverse events. Four patients in our series were scanned with abdominally placed IPGs, although we cannot retrospectively confirm what transmit coils were used in those cases. In addition, at least 19 spine MRI scans (cervical and lumbar) were performed for spine issues in our implanted DBS patients, also without incident. The second case of patient injury involved bilaterally externalized DBS leads; in our series, no patients were scanned with externalized leads.

It is unclear if having the pulse generator amplitude set to zero and having the stimulating mode set in a bipolar configuration has any role in the generation of heating or other adverse events. While it is our current practice to turn the IPG off and set the amplitude to zero, we have not routinely placed the stimulating mode to bipolar prior to performing MRI. We have occasionally scanned patients with the IPG turned off but the amplitude still set to therapeutic levels of stimulation; some of these patients have reported transient paresthesias that would be consistent with the IPG being switched on and off during imaging. No long-term adverse effects have occurred in these patients, and no damage or alteration in IPG function has been observed. All patients with implanted IPGs should undergo interrogation of their devices following MRI to confirm proper stimulation parameters.

Phantom Model Imaging

Experimental testing has been performed by various groups to examine the heating of DBS components in the MRI environment. The results are quite variable and dependent on numerous factors, such as the scanner configuration (including field strength and RF wavelength), placement of the DBS system with respect to the bore of the MRI scanner and/or coils, coil selection, method and location of temperature measurements, and pulse sequences used [13, 34, 41, 42]. Finelli et al. [31] performed phantom studies of DBS electrodes examining the potential for heating at the electrode tip with a variety of clinically used pulse sequences at 1.5 T. They found a linear relationship between SAR and electrode heating, although temperature elevations with local (head) SAR <2.4 W/kg were less than 2°C, i.e. in the clinically tolerable range [31]. This group also demonstrated that the method of testing is important and can have a profound impact on the heating seen at the electrode tip. The type of RF coil used, for example, had a profound impact on electrode heating. 'Worst case' scenarios using body T/R coils and pulse sequences with SAR up to 3.9 W/kg lasting up to 15 min produced temperature changes at the electrode tip of up to 23.5°C. By contrast, switching to the head only T/R coil produced a maximum temperature change of 7.1°C [27]. A more recent study examined the use of a head only T/R coil in two different generation scanners from the same manufacturer. Temperature changes were measured and normalized to the head SAR values displayed by the scanner console. A statistically significant 3.5- to 5.5-fold difference was seen in the slope of normalized temperature change between the two scanners, indicating that console-reported SAR itself is not a reliable index of heating for DBS leads [13]. Others have stated that using SAR as a safety recommendation in the setting of an implanted device is misleading and potentially dangerous [43].

Clinical Imaging

It is unfortunate that the imaging sequences that provide the best visualization of the target nuclei, T_2-weighted FSE and inversion recovery FSE, typically have inherently higher SAR values. Reducing the SAR of these sequences to 0.1 W/kg or less may degrade image quality such that the scan is no longer anatomically informative or create a scan that is prohibitively long. Furthermore, there are presently no reliable data reflecting the relative risk of MRI at specific SAR levels against which the clinical benefits of the procedure may be weighed. Some scanners do provide the option of using a 'low SAR' T_2-weighted protocol, however, these T_2-weighted sequences

do not reduce the SAR to the 0.1 W/kg recommendation. Custom sequences can be developed to provide the lowest SAR possible, but in most cases this requires the expertise of a dedicated MRI physicist with access to and control over details of the RF pulses used in these acquisitions. Many centers unfortunately do not have access to such resources; moreover, the differences in the way various manufacturers' platforms work are significant enough that one person may not be able to customize sequences for multiple platforms.

At our center, we have experimented with ultralow SAR T_2-weighted sequences on our Philips scanner to minimize image degradation and achieve an SAR of 0.2 W/kg. This is accomplished by reducing the number and amplitude of RF pulses and altering pulse shape. We found an effective strategy of reducing the number of RF refocusing pulses by half, which would normally double the scan time. We compensated for this by performing half the number of signal averages. Signal-to-noise ratio can be maintained by taking advantage of the wider RF pulse spacing to reduce the acquisition bandwidth by half. RF refocusing pulses that are substantially less than 180° may also be employed to reduce SAR in FSE images. Likewise, reducing the peak amplitude of RF pulse reduces SAR but lengthens the pulse duration, which can cause timing problems in the pulse train. When this occurs, the shape of the pulse can be changed, typically sacrificing the integrity of the slice profile for shorter RF pulse durations but maintaining the flip angle. Despite this ability to perform acceptable T_2-weighted imaging with ultralow SAR on our Philips scanner, we still use a T_2-weighted sequence that has an SAR of 0.5 W/kg due to concern regarding some of the compromises necessary for further SAR reduction.

It is important to recognize that the series of patients reported here were scanned under varying circumstances that would be encountered at any center. A wide variety of scanning parameters were used in a nonprospective manner on multiple hardware and software environments over a long time period, with MRI technologists of varying skill levels. We are of the impression that scanning at an SAR of 0.1 W/kg results in zero risk of thermal damage due to RF-induced heating under typical conditions. Unfortunately, it is not clear what the risks of imaging at an SAR of 0.4 W/kg or higher really are. We have carefully modified our scanning methods over the years to remain compliant with most manufacturer guidelines, but have never scanned a patient with an SAR of less than 0.1 W/kg in all pulse sequences, nor experienced any MRI-related adverse events.

Conclusions

Postoperative 1.5-tesla MRI of implanted DBS systems have been performed at our center in a large series of patients under a wide variety of circumstances without adverse events. This is only one center's experience, and these data do not imply that there are no safety concerns when performing MRI in patients with implanted DBS systems. However, this series strongly suggests that the 0.1 W/kg recommended safety margin for SAR, which is impractical for high-quality imaging, may be unnecessarily low for prevention of MRI-related adverse events. Particularly in light of the fact that following such a recommendation may ultimately decrease the quality of DBS therapy by discouraging or even preventing adequate postoperative imaging, experienced DBS centers with a preexisting track record of successfully scanning implanted patients with no adverse events should question the logic of stopping their postoperative MRI practices based on SAR alone.

References

1 Tisch S, Zrinzo L, Limousin P, Bhatia KP, Quinn N, Ashkan K, Hariz M: The effect of electrode contact location on clinical efficacy of pallidal deep brain stimulation in primary generalised dystonia. J Neurol Neurosurg Psychiatry, in press.

2 Pinto S, Le Bas JF, Castana L, Krack P, Pollak P, Benabid AL: Comparison of two techniques to postoperatively localize the electrode contacts used for subthalamic nucleus stimulation. Neurosurgery 2007;60:285–292.

3 Vergani F, Landi A, Antonini A, Parolin M, Cilia R, Grimaldi M, Ferrarese C, Gaini SM, Sganzerla EP: Anatomical identification of active contacts in subthalamic deep brain stimulation. Surg Neurol 2007;67:140–147.

4 Pollo C, Vingerhoets F, Pralong E, Ghika J, Maeder P, Meuli R, Thiran JP, Villemure JG: Localization of electrodes in the subthalamic nucleus on magnetic resonance imaging. J Neurosurg 2007;106:36–44.

5 Arantes PR, Cardoso EF, Barreiros MA, Teixeira MJ, Goncalves MR, Barbosa ER, Sukwinder SS, Leite CC, Amaro E Jr: Performing functional magnetic resonance imaging in patients with Parkinson's disease treated with deep brain stimulation. Mov Disord 2006;21:1154–1162.

6 Ondo WG, Bronte-Stewart H: The North American survey of placement and adjustment strategies for deep brain stimulation. Stereotact Funct Neurosurg 2005;83:142–147.

7 Hamid NA, Mitchell RD, Mocroft P, Westby GW, Milner J, Pall H: Targeting the subthalamic nucleus for deep brain stimulation: technical approach and fusion of pre- and postoperative MR images to define accuracy of lead placement. J Neurol Neurosurg Psychiatry 2005;76:409–414.
8 Starr PA, Christine CW, Theodosopoulos PV, Lindsey N, Byrd D, Mosley A, Marks WJ Jr: Implantation of deep brain stimulators into the subthalamic nucleus: technical approach and magnetic resonance imaging-verified lead locations. J Neurosurg 2002;97:370–387.
9 Ryu SI, Romanelli P, Heit G: Asymptomatic transient MRI signal changes after unilateral deep brain stimulation electrode implantation for movement disorder. Stereotact Funct Neurosurg 2004;82:65–69.
10 Henderson JM, Tkach J, Phillips M, Baker K, Shellock FG, Rezai AR: Permanent neurological deficit related to magnetic resonance imaging in a patient with implanted deep brain stimulation electrodes for Parkinson's disease: case report. Neurosurgery 2005;57: E1063.
11 Spiegel J, Fuss G, Backens M, Reith W, Magnus T, Becker G, Moringlane JR, Dillmann U: Transient dystonia following magnetic resonance imaging in a patient with deep brain stimulation electrodes for the treatment of Parkinson disease. Case report. J Neurosurg 2003;99:772–774.
12 Tremmel J (ed): Urgent Device Correction – Change of Safe Limits for MRI Procedures Used with Medtronic Activa Deep Brain Stimulation Systems. Medtronic, 2005.
13 Baker KB, Tkach JA, Phillips MD, Rezai AR: Variability in RF-induced heating of a deep brain stimulation implant across MR systems. J Magn Reson Imaging 2006;24:1236–1242.
14 Deep-brain stimulation of the subthalamic nucleus or the pars interna of the globus pallidus in Parkinson's disease. N Engl J Med 2001;345:956–963.
15 Burchiel KJ, Anderson VC, Favre J, Hammerstad JP: Comparison of pallidal and subthalamic nucleus deep brain stimulation for advanced Parkinson's disease: results of a randomized, blinded pilot study. Neurosurgery 1999;45:1375–1382.
16 Rodriguez-Oroz MC, Obeso JA, Lang AE, Houeto JL, Pollak P, Rehncrona S, Kulisevsky J, Albanese A, Volkmann J, Hariz MI, Quinn NP, Speelman JD, Guridi J, Zamarbide I, Gironell A, Molet J, Pascual-Sedano B, Pidoux B, Bonnet AM, Agid Y, Xie J, Benabid AL, Lozano AM, Saint-Cyr J, Romito L, Contarino MF, Scerrati M, Fraix V, Van Blercom N: Bilateral deep brain stimulation in Parkinson's disease: a multicentre study with 4 years follow-up. Brain 2005;128:2240–2249.
17 Koller WC, Lyons KE, Wilkinson SB, Troster AI, Pahwa R: Long-term safety and efficacy of unilateral deep brain stimulation of the thalamus in essential tremor. Mov Disord 2001;16: 464–468.
18 Lee JY, Kondziolka D: Thalamic deep brain stimulation for management of essential tremor. J Neurosurg 2005;103:400–403.
19 Papavassiliou E, Rau G, Heath S, Abosch A, Barbaro NM, Larson PS, Lamborn K, Starr PA: Thalamic deep brain stimulation for essential tremor: relation of lead location to outcome. Neurosurgery 2004;54:1120–1130.
20 Putzke JD, Wharen RE Jr, Obwegeser AA, Wszolek ZK, Lucas JA, Turk MF, Uitti RJ: Thalamic deep brain stimulation for essential tremor: recommendations for long-term outcome analysis. Can J Neurol Sci 2004;31:333–342.
21 Kupsch A, Benecke R, Muller J, Trottenberg T, Schneider GH, Poewe W, Eisner W, Wolters A, Muller JU, Deuschl G, Pinsker MO, Skogseid IM, Roeste GK, Vollmer-Haase J, Brentrup A, Krause M, Tronnier V, Schnitzler A, Voges J, Nikkhah G, Vesper J, Naumann M, Volkmann J: Pallidal deep-brain stimulation in primary generalized or segmental dystonia. N Engl J Med 2006;355:1978–1990.
22 Starr PA, Turner RS, Rau G, Lindsey N, Heath S, Volz M, Ostrem JL, Marks WJ Jr: Microelectrode-guided implantation of deep brain stimulators into the globus pallidus internus for dystonia: techniques, electrode locations, and outcomes. Neurosurg Focus 2004;17:E4.
23 Vidailhet M, Vercueil L, Houeto JL, Krystkowiak P, Benabid AL, Cornu P, Lagrange C, Tezenas du Montcel S, Dormont D, Grand S, Blond S, Detante O, Pillon B, Ardouin C, Agid Y, Destee A, Pollak P: Bilateral deep-brain stimulation of the globus pallidus in primary generalized dystonia. N Engl J Med 2005;352: 459–467.
24 Krempien RC, Daeuber S, Hensley FW, Wannenmacher M, Harms W: Image fusion of CT and MRI data enables improved target volume definition in 3D-brachytherapy treatment planning. Brachytherapy 2003;2:164–171.
25 Ferroli P, Franzini A, Marras C, Maccagnano E, D'Incerti L, Broggi G: A simple method to assess accuracy of deep brain stimulation electrode placement: pre-operative stereotactic CT + postoperative MR image fusion. Stereotact Funct Neurosurg 2004;82:14–19.
26 Duffner F, Schiffbauer H, Breit S, Friese S, Freudenstein D: Relevance of image fusion for target point determination in functional neurosurgery. Acta Neurochir (Wien) 2002;144: 445–451.
27 Rezai AR, Finelli D, Nyenhuis JA, Hrdlicka G, Tkach J, Sharan A, Rugieri P, Stypulkowski PH, Shellock FG: Neurostimulation systems for deep brain stimulation: in vitro evaluation of magnetic resonance imaging-related heating at 1.5 Tesla. J Magn Reson Imaging 2002; 15:241–250.
28 Liem LA, van Dongen VC: Magnetic resonance imaging and spinal cord stimulation systems. Pain 1997;70:95–97.
29 Shinbane JS, Colletti PM, Shellock FG: MR in patients with pacemakers and ICDs: defining the issues. J Cardiovasc Magn Reson 2007;9:5–13.
30 Georgi JC, Stippich C, Tronnier VM, Heiland S: Active deep brain stimulation during MRI: a feasibility study. Magn Reson Med 2004;51: 380–388.
31 Finelli DA, Rezai AR, Ruggieri PM, Tkach JA, Nyenhuis JA, Hrdlicka G, Sharan A, Gonzalez-Martinez J, Stypulkowski PH, Shellock FG: MR imaging-related heating of deep brain stimulation electrodes: in vitro study. AJNR Am J Neuroradiol 2002;23:1795–1802.
32 Rezai AR, Finelli D, Rugieri P, Tkach J, Nyenhuis JA, Shellock FG: Neurostimulators: potential for excessive heating of deep brain stimulation electrodes during magnetic resonance imaging. J Magn Reson Imaging 2001; 14:488–489.
33 Shellock FG, Cosendai G, Park SM, Nyenhuis JA: Implantable microstimulator: magnetic resonance safety at 1.5 Tesla. Invest Radiol 2004;39:591–599.
34 Schaefer DJ: Safety aspects of radiofrequency power deposition in magnetic resonance. Magn Reson Imaging Clin N Am 1998;6:775–789.
35 Schueler BA, Parrish TB, Lin JC, Hammer BE, Pangrle BJ, Ritenour ER, Kucharczyk J, Truwit CL: MRI compatibility and visibility assessment of implantable medical devices. J Magn Reson Imaging 1999;9:596–603.
36 Sawyer-Glover AM, Shellock FG: Pre-MRI procedure screening: recommendations and safety considerations for biomedical implants and devices. J Magn Reson Imaging 2000;12: 92–106.
37 Heistad DD, Abboud FM: Factors that influence blood flow in skeletal muscle and skin. Anesthesiology 1974;41:139–156.
38 Medtronic: MRI Guidelines for DBS. Medtronic, 2006.
39 FDA: Adverse Event Report: Med Rel, Inc Soletra IPG. FDA, 2003.
40 Kovacs N, Nagy F, Kover F, Feldmann A, Llumiguano C, Janszky J, Kotek G, Doczi T, Balas I: Implanted deep brain stimulator and 1.0-Tesla magnetic resonance imaging. J Magn Reson Imaging 2006;24:1409–1412.
41 Tronnier VM, Staubert A, Hahnel S, Sarem-Aslani A: Magnetic resonance imaging with implanted neurostimulators: an in vitro and in vivo study. Neurosurgery 1999;44:118–125.
42 Rezai AR, Phillips M, Baker KB, Sharan AD, Nyenhuis J, Tkach J, Henderson J, Shellock FG: Neurostimulation system used for deep brain stimulation (DBS): MR safety issues and implications of failing to follow safety recommendations. Invest Radiol 2004;39:300–303.
43 Nitz WR, Brinker G, Diehl D, Frese G: Specific absorption rate as a poor indicator of magnetic resonance-related implant heating. Invest Radiol 2005;40:773–776.