The Physics of Magnetic Resonance Imaging Safety

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KEYWORDS

• Magnetic resonance safety • Static magnetic field • Spatial field gradient
• Pulsed gradient magnetic field • Radiofrequency field • Heating • Acoustic noise
• Peripheral stimulation

KEY POINTS

- The primary safety risks in MRI arise from the 3 unique magnetic fields used: the static magnetic field (B0), the time-varying radiofrequency magnetic field (B1) and the time-varying gradient magnetic field (dB/dt). Understanding and controlling the spatiotemporal distribution of these fields helps ensure MR safety for patients.
- Static magnetic field risk is dominated by the potential for ferromagnetic objects brought into the room to become projectiles as well as potential displacement, disruption, or damage to external or implanted devices.
- Diffuse heating from the time-varying radiofrequency magnetic field is characterized by the specific absorption rate to help manage core body temperature and avoid undue thermal stress on patients.
- Focal heating from the time-varying radiofrequency magnetic field can arise from interaction with conducting materials in the bore and is one of the most often reported injuries. Minimizing conductors in the bore, avoiding close proximity of conductors to each other, and insulation between patient and conducting surfaces can help mitigate risks.
- The time-varying gradient magnetic field can induce peripheral nerve stimulation in patients and is also the source of acoustic noise requiring hearing protection in MRI. This field can also interact with external or implanted medical devices as well, possibly resulting in unintended stimulation, as well as disruption or damage of the implant.

INTRODUCTION

Compared with other advanced tomographic imaging modalities, magnetic resonance (MR) imaging is an extremely versatile modality capable of exquisite soft tissue contrast as well as functional and metabolic information. As technology has advanced, costs for high-performance systems have come down, and wider availability and clinical acceptance of this modality have grown, so have use and demand. MR imaging is now ubiquitous in radiology departments worldwide and is rapidly expanding to areas outside traditional diagnostic radiology departments, such as interventional, intraoperative, or radiation oncology hybrid suites.

MR imaging is a nonionizing radiation modality. The safety considerations and risks are unique and radically different from routine imaging sources using ionizing radiation. In the hands of, or under the direct supervision of, properly trained personnel, MR imaging is one of the safest imaging modalities. However, because of the propensity for serious or life-threatening injury to untrained personnel even entering the MR environment, a highly structured set of safety guidelines are used to help minimize the risk associated with MR imaging.[1,2]

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The Earth's magnetic field is on the order of 0.5 G (0.05 mT, where 1 T = 10,000 G). The unique safety concerns in MR imaging are caused by the generation and/or presence of 3 independent magnetic fields used for imaging by the MR scanner:

- Static magnetic field (B_0): a very strong magnetic field (0.5–7 T) is used to generate a potential energy difference in tissue proton spin population. This field, and the large spatial field gradients (T/m) associated with sharp transitions near the scanner, are the source for potentially large magnetic forces on objects entering the MR environment and can also potentially affect devices and personnel outside the MR suite. For most scanners in use today, this field is usually generated by large currents circulating in cryogen-cooled superconducting coils and so are generally always on, making the safety concerns caused by this field omnipresent and requiring substantial access and supervisory control and vigilance over personnel and items entering the MR environment.
- Time-varying radiofrequency (RF) magnetic field (B_1): a much smaller magnetic field (μT) oscillating at or near the MR frequency (megahertz) of protons is generated orthogonal to the static field by another set of current-carrying coils close to the patient but inside the bore of the MR scanner during imaging to excite and/or manipulate the polarized spins for signal and contrast. This field, characterized by its amplitude, frequency, and duty cycle, is responsible for risks caused by heating in the bore of the scanner. This field is only on during imaging.
- Time-varying magnetic field gradient (G): 3 orthogonal linear gradient magnetic fields (mT/m) are generated by another set of coils in the bore and are pulsed for image encoding and contrast manipulation during image acquisition. This rapid switching of the large currents through these coils is the generating source of the loud acoustic noise generated during MR imaging, whereas the rapid switching of the resulting magnetic field (dB/dt) at points in the gradient field can cause muscle stimulation.

Magnetic Resonance Safety Operating Modes

The output of these fields on the scanner is tightly controlled. In the United States, the Food and Drug Administration (FDA) provides guidelines for marketing of clinical MR scanners. Most safety limits and nomenclature used by the FDA come from the International Electrotechnical Commission (IEC) publication 60601-2-33, "Requirements for the Safety of MR Equipment for Medical Diagnosis." These agencies have established different operating modes for MR imaging based on the exposure of the patient to each of the 3 magnetic fields and the likelihood that this exposure may cause stress to the patient. Normal operating mode is one in which there is no expectation that scanning will result in physiologic stress and can be used safely for all patients. First-level controlled operating mode is one in which 1 or more outputs may reach a value that may cause physiologic stress to patients. It is expected that this mode is used to achieve a clinical benefit to the patient that outweighs these risks and that appropriate medical supervision is provided. There is a second-level controlled mode that allows higher outputs that can be used for research purposes after review by an independent review board for ethical and safety consideration.

Note that MR scanner operating modes govern outputs designed to manage physiologic stress to the patient and do not relate directly to prevention of safety events associated with improper patient screening or positioning as well as devices in the MR environment. These events are prevented by learning and understanding the appropriate use of the equipment. Similarly, although operating modes and associated outputs are often cited as part of the technical conditions for safely scanning medical implants, these modes and output measures were not created with those objectives in mind, but they are often the only way of specifying system output in a way that the user has control over.

There are many excellent reviews on the MR physics underlying MR safety.[3–7] This article reviews some of the practical aspects of physics needed to understand the system operation, outputs, and basic interactions with tissue and other materials with a view to underscoring how this information may be translated into the clinical management of patients or operation of the scanner.

STATIC MAGNETIC FIELD

The primary safety concerns to consider with a strong static magnetic field include potential direct interaction with patient (biological effects); ferromagnetic objects becoming projectiles (missile effect); interference or damage of ancillary medical equipment; disruption, damage, or displacement of implanted medical devices; and, for superconducting systems, cryogen safety considerations.

A strong, static magnetic field is used to polarize the spins. The field acts as a reservoir of potential energy for the spins. The stronger the static magnetic field, the larger population of spins aligned with the field and hence the larger potential signal that can be produced by the system. Large magnetic field strengths are typically cited in Tesla (T), whereas smaller fields, such as the so-called fringe fields further from the magnet, may be cited in Gauss (1 G = 0.1 mT). Current clinical imaging field strength ranges from 0.2 T to 7.0 T, with most current systems being 1.5-T or 3.0-T fields generated by cylindrical superconducting magnet technology. These field strengths are on the order of 50,000 times the Earth's magnetic field. Of course, objects exposed to these fields can be subject to varying degrees of magnetic force.

For siting purposes, magnets used for MR imaging tend to have a very high, homogeneous static magnetic field over a spherical volume for imaging (ie, 50 × 50 × 50 cm), and then have a very sharp spatial field gradient moving away from the imaging volume to minimize the penetration of the fringe magnetic field into the surrounding area and help keep within a controlled area, such as the MR suite.

Magnetic fields are generated by charges in motion (ie, currents) and it is often convenient to represent this source current density in terms of a magnetic dipole moment, which is a measure of the potential interaction between this current density and an externally applied magnetic field.

Magnetic Materials

Magnetic materials are those in which an applied external magnetic field can induce a net magnetization (M) from the current density induced within the material.[8] The ratio of magnetization induced per unit magnetic field strength is called the magnetic susceptibility (χ) and is a property of the material. Although the relationship can be complicated for three-dimensional materials, generally the induced magnetization is proportional to the local magnetic flux density (B), with the proportionality constant being a function of the susceptibility.

Most materials encountered in routine clinical imaging lie in the range of approximately $-10^{-5}<\chi<10^{-5}$. Diamagnetic materials (ie, water, soft tissue, deoxygenated blood, copper) generally resist the applied field and hence have a small induced magnetism against the field ($-10^{-5}<\chi<0$). Paramagnetic materials (ie, gadolinium, calcium, titanium) have a weak induced magnetism that aligns with the applied field ($0<\chi<10^{-5}$). Water susceptibility is approximately -9×10^{-6} and susceptibility differences greater than 10^{-5} from this value are likely to result in substantial artifacts and distortion on MR images.

In the paramagnetic and diamagnetic materials described earlier, generally no appreciable forces are observed on these objects at current field strengths used in MR imaging. However, ferromagnetic materials (ie, iron, stainless steels, cobalt, nickel) have unique magnetic domains that result in a much stronger induced magnetization from susceptibility ($\chi>10^{-2}$). Therefore, in addition to substantial artifacts and distortion that may negate the diagnostic quality of the acquired images, much stronger magnetic forces exist on these materials when placed in an external magnetic field. The forces on these objects are a paramount safety concern.

Forces on Magnetic Materials

Of the magnetic materials discussed, it is ferromagnetic objects that experience the greatest forces in the MR environment. The external static magnetic field induces a field in the object as characterized by its large susceptibility. However, these materials generally can only be magnetized up to a point before reaching a maximal saturating magnetization (M_s). Typically, ferromagnetic objects saturate between 0.25 and 2.5 T, depending on the material, with iron at the highest end of the scale. In addition to size and shape, this can affect the net induced magnetization. The external field can result in a demagnetization factor such that the magnetic susceptibility becomes a strong function of object shape, which can be characterized by a shape-dependent susceptibility constant, χ_D,[8,9] which accounts for this demagnetization effect.

The net induced magnetization in magnetic objects creates an effective net dipole moment ($\boldsymbol{m}$) in the material. The potential energy (U) between this dipole moment and the external field is given by $U = -\boldsymbol{m}\bullet\boldsymbol{B}$. This dot product shows that the energy of the system depends on both the magnitude of these values and their alignment with each other. Energy is minimized with the dipole moments and field aligned with each other. Forces arise on objects in a potential field when there are spatial gradients in the field. In general, $\boldsymbol{F} = \nabla U$. This treatment shows 2 primary forces of concern for objects placed in a magnetic field: displacement forces arising from spatial variations in the magnetic field itself, and rotational forces (torques) attempting to align the dipole moment with the external magnetic field.

Magnetically Induced Displacement Force

Strong displacement forces on a magnetic material arise from the spatial variations (gradients) in the static magnetic field (B). The relationship between the displacement force (F_d) experienced by a magnetic object with a dipole moment ($\boldsymbol{m}$) object in a static magnetic field ($\boldsymbol{B}$) can be represented by the relationship:[9]

$$F_d \propto |\nabla(\boldsymbol{m} \cdot \boldsymbol{B})| \propto V \cdot M |\nabla B| \qquad \textbf{Equation 1}$$

That is, the force on the object is strongly dependent on the mass of magnetic material, via the volume (V), the induced magnetization (M) in the magnetic material from its susceptibility (χ) and static magnetic field value at a particular location, and the spatial gradient of the static magnetic field ($|\nabla B|$) at that location. So, for objects in an appreciable magnetic field, force increases where rapid changes in the magnetic field (ie, high spatial field gradients) exist. These areas are conspicuous on isofield plots provided by the vendor because these are usually represented by tighter clusters of field lines. Objects are likely to accelerate their greatest in these regions. For a cylindrical-bore, superconducting-magnet design, these gradients are greatest at the magnet face and the edges of the bore (Fig. 1).

An obvious reference force for displacement is the weight of the object from gravity (F_g). As F_d becomes appreciable, the object may move on smooth surfaces on which there is little or no opposing frictional force. However, as the force of gravity is exceeded, unrestrained objects may begin to levitate and accelerate into the magnet bore. In particular, ferromagnetic objects may accelerate with a force many times their weight. To get an understanding of these values, the ratio of these forces can be approximated by:

$$\frac{F_d}{F_g} \cong C \cdot M \cdot |\nabla B| \qquad \textbf{Equation 2}$$

where the constant of proportionality is related to the object density (ρ), permeability of free space (μ_0), and gravitational constant via $C = (\rho\ \mu_0\ g)^{-1}$. Note, as discussed previously, the amount of magnetization that can be induced in a ferromagnetic object is limited ($M \leq M_s$) and so an object placed in the static magnetic field does not necessarily yield an induced magnetism $M = B$ in regions of higher magnetic inductance, but tracks with the local value of B until reaching M_s.

By way of example, for fully saturated iron, $M \sim 2.2$ T and C ~ 10 ($\rho = 7900$ kg/m^3; $\mu_0 = 4\pi \times 10^{-7}$ H/m, and $g = 9.8$ m/s^2). Just inside and near the edge of the bore of a 3-T scanner, the spatial field gradient can exceed 5 T/m, resulting in forces exceeding 120 times the weight of the object. However, the trouble with controlling object displacements starts much further away than that. At 0.4 to 0.5 m from the bore face, where the object is not fully saturated and the field is less than one-tenth of its maximum, the displacement force may still be strong enough to exceed the weight of the object, depending on the size and magnetization properties of the material. Going out to the edge of the patient table, where the field can be in the range of 3 to 5 mT, forces can still be appreciable enough to move unrestrained or unhindered objects.[10]

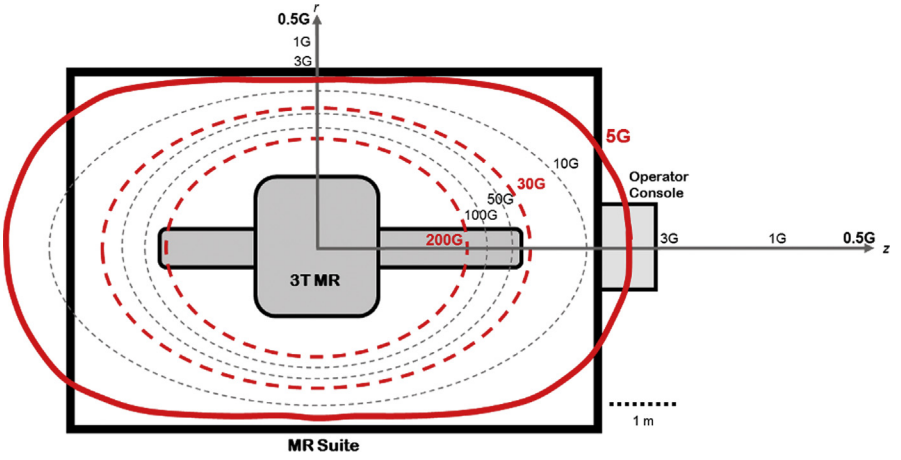

Fig. 1. Fringe field of a typical MR imaging suite showing the 5-G limit. All areas containing the 5-G line, including above and below the unit, must be access controlled. Because this safety limit is attributed to medical implants (eg, pacemakers), screening of personnel is required before entering. The 30-G zone is the limit for significant kinetic forces to be exerted on ferromagnetic objects. Refer to Table 1 for a summary of relevant isofield regions and distances from the magnet. Units: 1 G = 0.1 mT.

Of course, for medical implants that may enter the MR imaging scanner, it is important to assess the magnitude of the displacement force on the object. To this end, a simple deflection test has been devised in which the object is suspended and a protractor used to measure the maximal potential displacement location of the object, which is where the maximal spatial field gradient occurs near the edge of the face of the magnet. The angle of deflection can be compared against the force of gravity on the object, with a 45° deflection being the point at which $F_d/F_g \geq 1$. From this information, maximal static field and spatial field gradient conditions can be elucidated.[11–14]

Displacement forces are therefore the first forces experienced as objects or people enter the MR environment. As shown later, forces that may affect implanted medical devices can extend outside the room containing the scanner. Given the acute nature of the problem, safety interlocks by way of access control and screening of both objects and personnel have been put in place to reduce the risks associated with the MR environment. Note that the location and magnitude of these forces depend on both the value and spatial gradient of the static magnetic field in a given location. These properties vary as a function of magnet design characteristics, such as field strength, active versus passive shielding, and bore dimensions. Knowledge of the areas where the forces are significant and for what objects is necessary for maintaining a safe environment and is needed for each installation. Plots of the magnetic field, spatial field gradient, and product of magnetic field and spatial field gradient is one way of obtaining this information, and some form of this information is supplied by the vendor operator manuals for safety considerations.[10,15]

Magnetically Induced Torque

In asymmetric magnetic objects (ie, ellipsoidal or elongated), a strong net dipole moment not aligned with the static magnetic field may be present. In order to achieve the lowest energy state, a rotational force (torque) attempts to align the dipole with the external field. The torque (***L***) generated by this force is proportional to the cross-product between the dipole moment (***m***) and magnetic field (***B***).[3,4,9]

$$\boldsymbol{L} \propto \boldsymbol{m} \times \mathbf{B} \Rightarrow L \propto V \cdot M \cdot B \cdot \sin\theta \qquad \textbf{Equation 3}$$

Similar to the displacement force, the total torque depends on the amount of the magnetized material (V) and induced magnetization (M); however, it also depends directly on field strength (B) and not the spatial gradient, as well as the angle between ***m*** and ***B*** (θ). This equation indicates that, if the dipole of the object manages to align itself with the field, the torque goes away. The maximum torque experienced by the object can be represented by the following relation:

$$L_{max} \propto V \cdot M_s^2 \cdot \chi_D^2 \qquad \textbf{Equation 4}$$

For ferromagnetic objects subject to demagnetization factors, it can be seen that, as the field increased, the dependence of the torque becomes a function of the object size, the saturated magnetization of the object, and a strongly shape-dependent modified susceptibility coefficient that accounts for the demagnetization effects.[16] In general, elongated objects experience the strongest torques, whereas more isotropic objects experience less.

Obviously, a torque on an implanted device in a patient can cause harm if strong enough to break free. Measurement of magnetically induced torque on a ferromagnetic device or implant can be performed by affixing the object in a rigid holder attached to a torsion spring and measuring the maximum deflection.[16] Similar to the deflection comparison with the force from the object weight, if the measured torque is less than the length of the object times the force of gravity, the object is unlikely to carry significant additional risk. These thresholds are both conservative and may not prohibit entry into the MR environment depending on how the device is secured in place. This test is often coupled with the deflection test for displacement force[11] to assess risk of an object in the bore of the magnet. For larger objects for which these tests are either inappropriate or the device is not designed to enter the bore, customized tests may need to be performed to establish some reasonable set of conditions for safe use of the equipment in the MR environment.[17]

Note that a weaker torque may be experienced by conducting objects moving within the magnetic field. As mentioned earlier, when magnetic flux through a conducting object, such as from motion, is changing, Faraday's law indicates that eddy currents arise in the object in proportion to the rate of change of flux. Lenz's law further states that this force works to oppose the motion generating force. This force on conductors can result in some unexpected dampening or resistance of large conductors to motion in the magnet. For nonferrous objects, such as metal heart valves, moving rapidly in the magnetic field, it raises concerns that this force may be substantial enough to hinder normal valve function and require additional monitoring of the patient during MR imaging,

particularly as higher magnetic fields are used for imaging.[18,19]

Cryogen Concerns

An additional safety concern with superconducting magnets is the substantial amount of liquid helium used to supercool the coils. If a leak develops, the helium can enter the room as a colorless and odorless gas that displaces oxygen and may result in asphyxiation. The Occupational Safety and Health Administration (OSHA) standard states that oxygen depletion in an enclosed space resulting in less than 19.5% oxygen is hazardous to unprotected personnel (29 CFR 1910.146). In addition, MR systems are at risk of a quench. A quench is a sudden loss in the superconductivity of the magnet when the cryogen cooling is no longer sufficient to maintain this state. The time for a typical 1.5-T to 3.0-T magnet to go from field to 20 to 50 mT is on the order of 20 to 120 seconds (depending on the magnet design and field strength). Loss of superconductivity means a high resistance is suddenly present in the coils and the several hundred Amperes of current quickly generates substantial heating. Heating of the liquid helium causes it to boil off as an extremely cold gas. The liquid to gas transition results in a tremendous volume expansion. Gas is expelled from the room to the outside via a quench pipe. If the quench pipe fails, wholly or in part, the cold helium gas will enter the examination room and may result in hypothermia or asphyxiation. Emergency procedures for addressing a quench incident should be developed for all MR imaging facilities, but, importantly, should also be initially addressed during facility siting and design.

Static Magnetic Field Safety Limits

The magnetic forces in MR imaging do interact with human physiology.[20] Possible side effects of moving or working in higher fields (ie, $\geq$2 T) include transient nausea, vertigo, metallic taste, and/or phosphenes.[20,21] Effects can be minimized by moving slower (ie, $\leq$3 T/s)[10] when near the front of the bore and have not proved to be of significant concern at field strengths up to 3 T.[22] Magnetohydrodynamic effects on flowing blood can result in increased T waves on electrocardiogram (ECG), which can lead to issues with both proper monitoring of patients and triggering acquisitions.[23] Although systems with higher field strengths are entering the market, the FDA has noted that patient exposure to static magnetic fields less than or equal to 8 T for adults, children, and infants more than 1 month of age is not considered a significant risk.[24]

Electronically powered or magnetically programmed active implanted medical devices (AIMDs), include device families such as cardiac implanted electronic devices (CIEDs), brain/nerve/spine/bladder stimulators, and drug infusion pumps. AIMDs may not only suffer from issues of displacement forces or torques as passive devices do but may also be susceptible to temporary or permanent B_0-field–induced device malfunction.[25] One prominent example of this type of disruption is that of the impact of the magnetic field on the reed switch of CIED. The reed switch enables an external magnet to be used to program the pacing or therapy delivery mode of the device. Exposure to the MR environment without disabling this switch in older non-MR conditional CIEDs, and appropriately monitoring the patient, has resulted in fatality.[26] Because of the sensitivity of such devices, any area containing fringe static magnetic fields of 0.5 mT (5 G) or higher must be controlled and clearly marked,[15,25] with personnel being screened for these and other implanted devices before entry.[1,2] Note that this controlled zone can extend above or below the MR suite as well as around.

Static Magnetic Field Safety Summary

Magnetic field bioeffects do not present a high-risk safety issue at routine clinical field strengths ($\leq$3 T). However, the potential for ferromagnetic objects to be pulled violently into the magnet (missile effect) is the primary safety risk from the static magnetic field and is second to RF-induced thermal injury in terms of reported safety events related to MR system magnetic fields.[27] The forces on these large, heavy objects are high enough to result in severe injury or death to anyone caught between the object and its path to the magnet, and the difference between a moderate pull on the object and substantial forces is just a few meters (see **Fig. 1**). Similarly, the static magnetic field may present a hazard to patients with ferromagnetic implants because of strong displacement forces, which are maximal as the patient passes the mouth of the bore because of the high spatial field gradients (**Fig. 2**), as well as strong torques, which are maximal in the region around magnet isocenter (**Table 1**).

These strong fields can also damage and disrupt normal operation of AIMDs, such as pacemakers, at a distance far from the magnet isocenter in the fringe fields. For this reason, access to areas containing this region must be identified and tightly access controlled. Note that, during initial siting or site safety reviews, these fields extend above and below the magnet as well as all around the

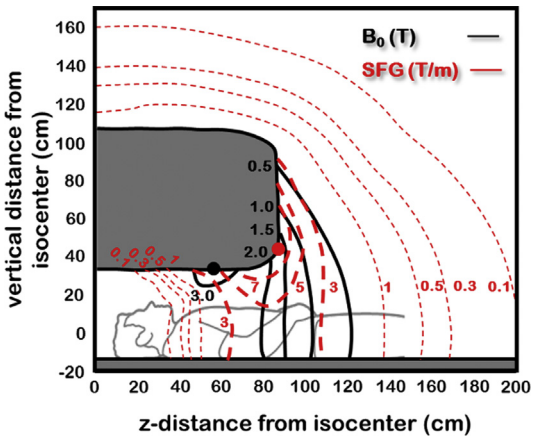

Fig. 2. The approximate location and magnitude of B_0 isofield lines (*black*) in units of Tesla and spatial field gradient (SFG) lines (*red*) in units of T/m as a patient enters a 3 T scanner. The maximum field (*black circle*) is within the bore and is approximately 3.5 T, whereas the maximum SFG (*red circle*) is located at the bore opening and is approximately 10 T/m. The region of extremely high spatial gradient–generated displacement force that may be a hazard to patients with ferromagnetic medical implants is emphasized using bold SFG lines ($\geq$3 T/m) and begins at the bore opening. Further inside the bore, magnetic torque forces become stronger and are the dominant consideration as the spatial gradients weaken near magnet isocenter. Vendors often quote the maximum SFG exposure within a cylindrical region so that users can understand the maximum SFG that might be experienced by a particular device or implant on the patient. Units: 1 T/m = 100 G/cm.

system and may penetrate into adjacent spaces and present a safety hazard or affect nearby sensitive equipment (see **Table 1**). Note that a magnet located on an upper floor with fringe field lines going out the window still may present a hazard to the window washers or the equipment they use. It is important to fully assess the location and potential impacts of the magnet fringe fields. Modern scanners actively shield their systems to minimize the range of this field. In some cases, permeable metal magnetic shielding may be needed to confine the fringe fields to the MR suite.

In addition, the cryogens associated with cooling the superconducting coils needed to generate the static magnetic field are a safety concern as well. Helium leaks can displace the oxygen in the room, resulting in risk of asphyxiation. Accidental or controlled magnet quenches run a risk of a violent release of helium gas into the MR room, or exposure of personnel at the output of the quench pipe. Quench procedures tailored to the specific facility and routine checking of the quench pipe are needed.

TIME-VARYING RADIOFREQUENCY MAGNETIC FIELD

Polarized spins aligned with the static magnetic field are modulated using RF magnetic field pulses tuned to or near the resonant (Larmor) frequency for both signal generation and contrast manipulation purposes. The resonance frequency is the product of the gyromagnetic ratio (γ) of the imaged nuclei and the static field strength (B_0). For protons, γ = 42.58 MHz/T, yielding resonance frequencies (f) of approximately 64 MHz and 128 MHz for 1.5 T and 3.0 T respectively. At these frequencies, the primary safety concerns are whole-body and localized heating from the absorption of the applied RF energy.[28] Understanding and controlling this heating is paramount in MR imaging because thermal injury is the most prevalent reported injury.[27]

A fraction of the energy in the applied RF magnetic field (B_1) is absorbed, resulting in heating. From Maxwell's equations, a time-varying B_1 field is the source of an induced electric field (E_1), from which a current density (J_C) is generated via tissue conductivity (σ). Simultaneously, the time rate of change of this induced E_1 field gives rise to an opposing displacement current (J_D) via the tissue permittivity (ε). The total induced current (J) would then be given by the expression in Equation 5 as:

$$J = J_C + J_D = \sigma E_1 + \varepsilon \frac{\partial E_1}{\partial t} \qquad \textbf{Equation 5}$$

Thus, the dielectric properties of the medium determine conversion of the applied RF field to currents that are ultimately absorbed locally, resulting in tissue heating. Because this is the key means of heating, at the wavelength used in MR imaging, a large fraction of the energy is deposited superficially, with less heating observed at depth.

The amplitude (B_{1p}) of the applied RF magnetic field used in MR imaging is much smaller (μT) than either the static or gradient magnetic fields and is applied on the order of milliseconds using high-power (kilowatt) amplifiers to drive whole-body or head/extremity-sized multirung, circularly polarized transmit coils. Remembering that the spins are polarized to have a net magnetization aligned with the direction of B_0, the applied B_1 field is often designated by a superscript (B_1^+), which signifies the excitation pulse is circularly polarized along this direction. Polarization of the excitation pulse has an efficiency advantage in that the applied power is primarily used to excite the spins polarized with the field magnetic field, as opposed to linear polarization, which wastefully excites both B_1^+ and B_1^-

Table 1
Summary of general static field regions, limits, and potential hazards in a magnetic resonance imaging suite with sample approximate distances for an actively shielded, wide-bore 3-T scanner

Field Strength	Location	Significance and Potential Disruptive Impact
0.05 mT (0.5 G)	r = 3.5 m z = 8.5 m	Earth's magnetic field (average)
0.05–3 mT (0.5–30 G)	r = 3.5–2.0 m z = 8.5–3.0 m	Medical equipment: photomultiplier tubes, image intensifiers, gamma cameras, PET, cyclotrons, electron microscopes, linear accelerators, ultrasonography, x-ray tubes, computed tomography units, color/monochrome monitors AIMDs: CIED, stimulators, insulin pumps, hearing implants Devices: watches, small motors, cameras, credit cards, magnetic data carriers, processors, oscilloscopes
0.5 mT (5 G)	r = 2.5 m z = 4.5 m	**Controlled access to MR environment limit** **Mandatory posting of potential safety hazards** **Mandatory personnel screening**
3 mT (30 G)	r = 2.0 m z = 3.0 m	Threshold for onset of kinetic energy hazards from small ferrous objects
20 mT (200 G)	r = 1.75 m z = 2.3 m	**Often used as limit for movement of ferrous objects within room (ie, for service personnel)**
20–500 mT (200–5000 G)	$r \leq$1.75 m z = 2.3–1.3 m	Range of MR conditional anesthesia, patient monitoring, injectors location/tethering limits
High SFG (≥3 T/m)	Bore entry z = 1.1–0.7 m	**Region of strongest magnetic displacement forces** **Largest risk for ferromagnetic implants**
1.5–3.0 T	Isocenter	Region of strongest magnetic torque forces Field strengths for most commercial MR imaging
>3.0 T	Isocenter	First level controlled operating mode for B_0
7 T	Isocenter	Current highest commercial field strength
>8.0 T	Isocenter	Second level controlled operating mode

See Figs. 1 and 2 for visualization of regions.

precessing spins. The $B_{1\rho}$ for a rectangular pulse is approximated by $\alpha/\gamma\tau_{rf}$, where α is the flip angle and τ_{rf} is the time the pulse is played out (ie, inverse of bandwidth). Note the lack of dependence on the value of B_0. This calculation yields 11.7 μT for a 1-millisecond 180° pulse. Many RF pulses are optimized (ie, timing, spatial-spectral selectivity, power, and so forth) and so do not have such a simple dependence on these parameters, but the case discussed earlier is illustrative nonetheless.

The B_1^+ field has wavelength $\lambda = c/f$, where c is the speed of light (3×10^8 m/s), which is about 4.7 m and 2.3 m for 1.5 T and 3.0 T respectively. However, of interested here is propagation through the body, which has different dielectric properties than air. Assuming a relative permittivity $\varepsilon_r = 60$ (muscle), a new, much smaller wavelength of $\lambda_{tissue} = \lambda_{air}/\sqrt{\varepsilon_r}$ would be expected, which is about 50 cm and 25 cm for 1.5 T and 3.0 T respectively. So, the effective wavelengths get smaller with increasing field strength and with tissue permittivity. The smaller wavelengths affect patients more strongly, affecting both propagation through the patient as well as absorption of energy.

The power applied to tissue is generally a function of field strength, pulse sequence, and patient size. A fraction is absorbed in patient, implants, and/or conductors as heat. The primary safety concerns are heat stress from sustained whole-body temperature increases and potential for tissue damage from localized high-temperature exposures. Although temperature control is the aim, temperature cannot be easily measured internally during routine clinical MR imaging. So, temperature control in MR imaging focuses on controlling system power output in conjunction with theoretic and/or empirical thresholds of damage.

Specific Absorption Rate

The rate of energy deposition in tissue is the specific absorption rate (SAR), which is often expressed in units of watts per kilogram. Absent any losses, the initial change in temperature (ΔT) in time (Δt) is proportional to the SAR via the heat capacity (C) of the material so that $\Delta T = C \cdot SAR \cdot \Delta t$. Therefore, for an insulated slab with $C = 3.5$ kJ/kg/°C (similar to tissue), this simple estimate with a SAR of 1 W/kg results in a 1°C temperature increase in 1 hour. In 15 minutes, 2 W/kg results in a 0.5°C increase and 4 W/kg results in a 1°C increase.

Ignoring the impact of permittivity in Equation 5, a relationship between the applied B_1^+ peak amplitude (B_p), induced electric field (E_p), and SAR for a homogeneous spherical object of radius (R) and density (ρ) can be approximated by[3,4]

$$SAR = \frac{\sigma |E_p|^2}{2\rho} = \frac{\sigma}{2\rho} \cdot (\pi \cdot R \cdot f \cdot B_p)^2 \cdot D$$

Equation 6

where the tissue conductivity (σ) generally ranges from about 0.4 to 1.0 S/m for soft tissue for the Larmor frequencies ($f = \gamma B_0$) used in clinical MR imaging, with progressively lower values for tissue such as lung, bone, and adipose that are in the range of 0.1 to 0.3 S/m.[29]

The duty cycle (D) is a factor that reduces the SAR based on the ratio of time the RF pulse is on versus the time period over which it is averaged and is determined by the pulse sequence parameters used during imaging. For many two-dimensional acquisitions, the duty cycle can be estimated as $D = (\tau_{rf} \cdot N_{echoes} \cdot N_{slices})/TR$. Here TR is the pulse repetition time (milliseconds), τ_{rf} is the RF pulse duration (milliseconds), with N_{echoes} and N_{slices} being the number of interleaved echoes and/or slices per TR period. Noting that B_p depends on the RF pulse flip angle (α) used, $SAR \sim B_0^2 \cdot \alpha^2 \cdot D$ (patient size). Control over the flip angle and duty cycle are 2 powerful means for controlling SAR from an acquisition standpoint. Some suggestions for reducing SAR in MR scanning are given in **Table 2**.

To illustrate the magnitude of the effect, a simple calculation is instructional using a spin-echo acquisition applied to a spherical object with R = 10 cm (ie, human head), σ = 0.5 S/m, $\rho = 1000$ kg/m^3, and $B_p = 10$ μT for 0.5 millisecond (180° pulse). One 90° to 180° sequence per slice and TR = 400 millisecond, with 15 slices per TR period, gives $D = 0.03$, resulting in an estimated SAR = 0.3 W/kg.

Of course, a more detailed spatiotemporal estimation of SAR in human patients is more complicated and requires realistic modeling of the transmitted RF field propagation, absorption, and bioheat transfer in tissue. A great deal of research has been done in this area and very accurate simulation is possible to aid in characterizing heating, such as for investigating safety issues associated with higher field strengths or medical implants in the MR environment.[30–34]

MR scanners provide a conservative estimate of SAR during prescription of pulse sequences using a process that requires both system-specific and patient-specific information as input. The B_1^+ power needed to excite the spins is estimated via a prescan calibration process and can be used to make SAR predictions. This information is combined with information about the pulses and timing used in the prescribed pulse sequence to estimate $B_1^+{}_{rms}$ for the sequence over an appropriate averaging time, such as the TR period. During the scan, RF power delivered to the patient is estimated in real time using the pulse-energy method[10,35] to update predictions. In either case, the estimated power delivered to the patient is normalized by the patient weight to estimate whole-body SAR. Peak SAR can be roughly estimated to be approximately 2.5 times higher.[36] Partial-body SAR is useful when the amount of tissue exposed is reduced (ie, head transmit coil) and is calculated by estimating the mass for normalization from the fraction of exposed tissue in the coil. Whole-body, partial-body, and head SAR are calculated for volume excitation coils. When local excitation coils are used, whole-body and local SAR, power averaged over any 10 g of the patient's body, are used to control the system output.[37]

Radiofrequency Field Safety Limits

SAR is a measure of RF power absorbed in tissue, and the estimate of this system output operates as a surrogate for managing temperature effects in MR imaging because patient temperature is not easily measured in the area of heating during imaging. Whole-body and localized heating are the primary concerns with absorption of RF power during imaging. The thermoregulatory system helps the body counteract this thermal stress and manages through a combination of heat radiation and evaporation from the surface of the body as well as convection and conduction. For the exposures expected in MR imaging, this means the patient may feel heat sensations in the skin, increased perspiration, and increased pulse rate.

So, with respect to temperature effects, whole-body RF field exposures resulting in less than or

Table 2
Summary of common specific absorption rate reduction techniques for magnetic resonance acquisitions

Acquisition Modification	Potential Tradeoff
K-space View Reduction	
Reduced phase encodes	Resolution loss
Rectangular field of view	Not amenable to all anatomy
Parallel/compressed acquisition	SNR loss and potential artifacts
RF Pulses	
Reduced flip angle excite and/or refocus	SNR loss and contrast changes
Pulse amplitude/width modulation	SNR loss; sequence timing issues
Saturation/suppression pulse reduction	Contrast changes; artifacts
Time Efficiency	
Increase concatenations	Longer acquisition times
Reduce ETL; increase ESP and/or TR	Longer acquisition times
Reduce anatomic coverage	Need for multiple acquisitions
Increase slice thickness/spacing	Slice resolution loss
Pulse Sequence	
Gradient vs spin echo or bSSFP	Contrast and SNR considerations
RF Coil Selection or Patient Positioning	
Smaller volume transmit coils	Coverage, uniformity, availability

Note that increases in acquisition times add to motion problems and increase overall active scan times. Increasing RF pulse excitation may be more susceptible to both motion and relaxation effects.

Abbreviations: bSSFP, balanced steady-state free precession; ESP, echo spacing; ETL, echo train length; SNR, signal/noise ratio.

equal to 1°C increase in body core temperature are not expected to result in adverse health effects.[38] In persons at risk for thermoregulatory impairment (eg, infants, pregnant women, persons with cardiocirculatory impairment), the temperature increase should be less than or equal to 0.5°C. Maximum temperatures in localized regions should be managed as well for head (≤38°C), trunk (≤39°C), and limbs (≤40°C).

To prevent undue heat stress and tissue damage, the SAR output of the MR imaging is governed with the goal of limiting increases in local and whole-body temperature. Normal operating mode limits total body core or local temperature to less than or equal to 39°C and temperature changes to less than or equal to 0.5°C. First-level controlled operating mode limits total body core or local temperature to less than or equal to 40°C and temperature changes to less than or equal to 1°C.

In terms of how this translates to SAR limits, for volume transmit coils (ie, body, head, extremity), the whole-body SAR normal operating mode limit is 2 W/kg and for first-level controlled operating mode first-level controlled operating mode it is 4 W/kg.[10] Partial-body SAR ranges from those whole-body limits up to 10 W/kg as the fraction of exposed tissue decreases, whereas SAR in the head specifically is limited to 3.2 W/kg for either operator mode. Local transmit coils have limits of 10 W/kg and 20 W/kg for normal and first-level controlled modes, respectively in the head and trunk and double these values for extremities. All limits are 6-minute averages, with the average in any 10-second window being less than twice the stated limit. In addition, with the higher SAR of local transmit coils, care needs to be taken with sensitive tissues in the field, such as the orbits of the eyes, to keep temperature increases less than 1°C. Exceeding the first-level controlled operating mode first-level controlled operating mode limits is only done within the confines of human-subject research studies.

Note that first-level controlled operating mode first-level controlled operating mode whole-body limits assumes patients have uncompromised thermoregulatory capability. Thermoregulatory capabilities can be compromised by the ambient room environment limiting heat exchange with the environment. The SAR limits assumed an ambient temperature of less than or equal to 25°C and less than or equal to 60% humidity. Whole-body SAR limits

may be reduced by 0.25 W/kg per degree Celsius exceeding 25°C until SAR is returned to the normal operating mode limit.

MR examinations that expose more of the body over longer periods of time (eg, whole-body examinations, cervical-thoracic-lumbar spine examinations, combined abdomen-pelvic examinations, PET/MR imaging) have increased in frequency. To help balance the thermoregulatory stress in light of sustained power deposition, a simple active scan time metric for the total energy delivered to the patient has been developed for aiding in promoting best practices during these examinations by limiting, or giving patients time to recover between, long examination times.[10] The specific absorbed energy (SAE) is an estimate of the total energy delivered into the patient during the active scan time. The current recommended maximum SAE is 14.4 kJ/kg (or 240 W*min/kg). If followed, this SAE recommendation limits active scanning at the normal operating mode SAR limit of 2 W/kg to less than 120 minutes and for first-level controlled operating mode SAR limit of 4 W/kg to less than 60 minutes. If SAE limits are reached during an examination, different vendors may have different safety interlocks in place that can warn, or stop, the user. In any event, if the SAE is reached, it is recommended to check on the patient's status and potentially allow a period of time to cool down if needed.

Irreversible thermal tissue damage is a function of both temperature and exposure. Unless temperatures are very high (>57°C), damage takes seconds to minutes to develop. The primary mechanism to irreversible damage of tissue is denaturation of key proteins needed to maintain cellular hemostasis or membrane activity so that the rate of transition from the normal to damaged state (Ω) can be modeled as a function of temperature (T) via a first order Arrhenius rate[39]

$$\Omega = A\int_0^t e^{-\frac{E_a}{RT(\tau)}}d\tau \qquad \textbf{Equation 7}$$

where the estimated fraction of tissue in the damaged state (F_D) would be given by $F_D = 1\text{-}e^{-\Omega}$, and R is the universal gas constant (8.3145 J/mol/K). The original work on high-temperature skin burns established a frequency factor $A = 3.1 \times 10^{98}\ s^{-1}$, and $E_a = 6.28 \times 10^5\ J\ mol^{-1}$is the activation energy for the process. Although these original parameters are still often used for predicting tissue damage at high temperatures, a tremendous amount of work has been performed not only to refine the original values and identify temperature dependent breakpoints but to successfully adapt to different tissues and processes.[40,41] Pioneering work by Sapareto and Dewey[42] resulted in a simplified version of this model that is useful for hyperthermia dosimetry,[42,43] where many different isoeffects could be characterized simply by the characterizing the cumulative equivalent minutes spent at 43°C (CEM_{43}).

$$CEM_{43} = \sum_{t=0}^{n\cdot\Delta t} R^{(43-T_n)}\cdot\Delta t,\ with$$
$$R = \begin{cases} 0.25 & T_n < 43^\circ C \\ 0.50 & T_n \geq 43^\circ C \end{cases}$$

Equation 8

Here R is not the universal gas constant, but represents the rate of damage accumulation with time. Although values for R can vary, the most important result from a safety aspect is that at more than 43°C, damage begins to accumulate exponentially faster with dose and approximately doubles for each degree Celsius increase in temperature. Damage that takes 60 minutes to accumulate at 43°C, takes 15 minutes at 45°C or 4 hours at 42°C. Because of the exponential nature of this curve and the high uncertainty this introduces at low temperatures, application to assessing risk at higher temperatures, such as those associated with sustained focal heating, make thermal dosimetry a powerful compliment to modeling and measurement techniques, but not immediately useful with respect to controlling whole-body heating.[44,45]

It cannot be stressed enough that neither SAR limits nor SAE guidance have been developed with the prevention of RF burns in mind. These limits are designed to diminish discomfort from thermal stress or potential damage that can accumulate over long periods of exposure time with slow heating. Heating rate is proportional to SAR, and so the high SAR associated with focal heating can result in much faster increases in temperature. Unlike diffuse heating of the patient that leads to discomfort, focal heating may heat locations where the patient is less sensitive to the pain, or at very fast rates, such that patient monitoring based on verbal feedback or the bulb to stop the scan may be ineffective. Therefore, prevention of focal heating in the patient is an important separate consideration. The potential for focal heating in routine MR imaging has a strong dependence on proper patient screening, preparation, and positioning, as well as appropriate management of what materials or devices accompany the patient into the scanner.

Radiofrequency-Induced Focal Heating

Higher-caliber currents that may result in focal resistive heating in tissue can also be induced by the applied RF field and are a function of material conductivity, geometry, and location in the excitation coil. The primary concern along with these distributions of current are focal areas of high resistance that can lead to resistive heating of tissue. Materials with high conductivity, such as metals, tend to have current density highest at the surface. For smaller conducting materials (<2 cm) there is not a high probability of significant heating unless there are adjacent conductors within about 3 cm that may couple for enhanced heating.[46] Larger conductors, such as a hip or spine prosthetic, can generate a significant amount of current. If the object is large and smooth, these currents tend to distribute uniformly across the material, spreading the energy over a large volume. Medical implants, such as a drug infusion pump, result in heating distributed across their volume, which can be managed. However, at sharp corners or disconnects, or when in close proximity to another conductor, there is potential for high electric fields and resistive heating in the adjacent tissue.

Conducting loops

One specific geometry that presents risks in MR imaging is conductors forming loops in the RF field that are nearly perpendicular to the applied field. As discussed regarding Faraday's law, a time-varying magnetic field induces an electromotive force in these effective conducting loops generating a current proportional to the area of the loop and magnetic flux. Larger loops result in larger induced currents. This current, and hence the heating, will be distributed all along the loop if there are no areas of high resistance. However, areas of high resistance, such as breaks in the loop, will generate hotspots where this large current can be turned into heat, which can be substantial. In some cases, the electrical properties of the loop have a resonance frequency close to the Larmor frequency, in which a very large amount of current can be generated. Although it is unlikely for a random conducting loop to be near resonance in the RF field, the receive array coils are designed to be at resonance and are actively blanked during transmit. Because of the potential of surface coils to heat during imaging, vendors design and test coils in both connected and unconnected modes. In general, temperatures should not exceed 41°C.[10,47]

Of course, metal is not the only conductor in the magnet capable of high-caliber current loops. Human skin and/or damp clothing can also be highly conductive. The clasping of hands can form a large-diameter conducting loop from the arms, chest, and shoulders. The point of contact (ie, fingers or hands) is a potential region of high resistance where rapid, high heating can occur.[48] Similarly, crossing of legs (ankle to ankle or calf to calf), hands to outer thighs, or inner thigh to thigh contact have been associated with RF-induced thermal injury. Next to direct contact with external conductors, skin-to-skin contact is one of the most identified root causes for reported burns in MR scanners.[27]

Antenna effect

Long, cylindrical conductors, such as needles, wires, or leads, may be implanted, or partially implanted, in patients, and are additional sources of potential heating. For these objects, the induced tangential electric field drives the current density along the length of the conductor. As with loops, in general, the longer the length, the higher the potential current density and potential to deposit energy at a location of high resistance, such as the lead tip. Also as with a current loop, there is a resonance phenomenon in which the transfer of power from the RF field to the conductor is dramatically enhanced. In particular, when the object length is close to the half-wavelength of the RF field in tissue, the potential for very rapid and high heating exists (antenna effect).[49,50] As discussed earlier, the wavelength in tissue is reduced by the relative permittivity. However, there is an additional small loss for conductance that should be accounted for as well.[50,51] Therefore, the effects are highly tissue dependent. Using dielectric properties of muscle, which is relevant to implant locations, the half-wavelengths are approximately 20 cm (1.5 T) and 12 cm (3.0 T). The degree of heating can easily reach higher than 10°C in seconds, resulting in potential localized tissue damage early in an acquisition. In addition to this critical length effect and associated tissue dependence, for leads, the local temperature increase is a complex function of configuration, amount of material within tissue versus air, amount of insulation, as well as location in the RF coil. Implants closer to the edge of the coil tend to experience higher electric fields.

$B_1^+{}_{rms}$ and focal heating

Note that SAR operates as a dosimetric unit for diffuse heating over a large volume. However, because of the complexities that go into SAR prediction for these purposes, it does not track well with the extent of focal heating from a conductor in the RF field. For focal heating, estimated maximum $B_1^+{}_{rms}$, which might better relate to

the maximum induced E_1 field and hence temperature, is a better output control variable. This output should be displayed on all MR scanners for helping to manage metal devices or implants that may be in the field.[10]

With this in mind, whether SAR or $B_1{}^+_{rms}$ is used for RF output control in the presence of conducting passive or active implants or external devices, primary guidance comes from standardized testing of these objects, which may include both theoretic and empirical results.[17,25,52,53] Just as with static field force effects being dependent on the magnet design, it is important to understand the limitations of the testing conditions for RF heating. Higher field (>3 T) scanners now routinely use 2-channel or higher transmit body coils in order to better tailor the RF transmission field to the size and shape of the anatomy of interest. The changes in amplitude and phase associated with these techniques have not been applied to the device being tested to assess the MR conditions in the MR scanner. In addition, lower field vertical magnets also orient the RF field in a different direction than the assumed direction of the implant and can have a very different (favorable or unfavorable) effect on the heating.[54] For these reasons, considerations of the conditions under which an assessment of safe conditions for MR scanning has been made should be considered when a conducting device is being considered for scanning.

Summary of Radiofrequency Safety in Magnetic Resonance Imaging

The primary risk associated with the time-varying RF magnetic field in MR imaging is tissue heating from absorption of currents induced during excitation. For the RF energy delivered to patients in routine MR imaging, these currents are distributed widely over the patient volume, resulting in diffuse heating that may lead to uncomfortable thermal stress to the patient. The increase in body core temperature is primarily controlled by regulating the dosimetry output of SAR, which approximates the power delivered to the patient, and monitoring the patient during the exposure. Long exposures may require breaks and time to cool off, and patients should be visually and audibly monitored. In addition, the temperature, humidity, and air flow in the MR scanner can influence the ability of patients to dissipate heat. Some patients are at risk because of conditions that may compromise their thermoregulatory capability, or ability to feel or report issues. A summary of these considerations is given in **Table 3**A. These patients may require additional medical supervision and appropriate monitoring to manage risks from these stresses, such as ECG monitoring for cardiovascular stress.

The induced current density can also become concentrated in regions of higher resistance, resulting in higher temperature focal heating that can lead to irreversible tissue damage (ie, burns) when certain configurations of conductors are present. This focal heating is the leading cause of reported injuries in the MR environment. Neither the SAR output control nor exposure considerations are designed to prevent these thermal events. In addition, damage may happen so quickly that the patient is not able to report a problem until it is too late. Avoiding these thermal events is primarily a matter of proper patient screening for conducting objects and implants; mindful patient and conducting device positioning to avoid and insulate contact between any conducting surfaces, including the MR equipment and patient themselves; as well as avoiding exposures of these conductors to high B_1 hotspots near the transmitting coil (ie, bore wall) (**Fig. 3**). A summary of the consideration for focal heating is also provided in **Table 3**B.

Conducting passive implanted medical devices or AIMDs are of particular concern for heating. In addition, the RF field may induce unintended stimulation as well as temporary or permanent device malfunction for certain AIMDs.[25] Methodologies for empirical and model-based testing to help characterize conditions for safe scanning of these devices have evolved substantially over time and many specific devices, or device groups, have some form of guidelines with conditions under the control of the user by which both patient and device can be appropriately managed for RF heating in the MR environment.

As MR technology evolves, such as the move toward higher fields and multitransmit technologies, as well as a greater variety of devices being implanted in patients in order to manage conditions associated with, among other things, chronic disease and extended life span, understanding and keeping current with the physics and technology associated with RF heating by practitioners of MR imaging is paramount to maintaining and steadily improving the overall safety profile that continues to provide increasing patient access to this critical technology for managing their care.

Time-Varying Gradient Magnetic Field

During pulse sequencing, additional coils within the bore are used to create time-varying gradient magnetic fields for image encoding purposes.

Table 3
Summary of radiofrequency heating risk factors for patients

RF Heating Risk	Risk Factors
Whole-body or Partial-body Heating[a]	
Environment	• Bore temperature (ideally ≤22°C; not to exceed 25°C) • Humidity should be maintained ≤60% • Unobstructed air flow in bore (fan)
Patient	• Compromised thermoregulatory capability • Fever, cardiac decompensation, inability to perspire, hypertension, diabetes, obesity, elderly, certain patients with cancer • Pregnant patients, neonates, and low-birth-weight infants • Medications (eg, diuretics, β-blockers, calcium blockers, amphetamines, sedatives,) • Avoid thermal insulation (eg, heavy clothing, blankets) • Unconscious, sedated, or loss of feeling in any body part • Appropriate patient monitoring plan
RF exposure	• High SAR for sustained period • Single acquisition >15 min • Active scan time examination >60 min
Focal Heating	
Patient preparation	• Electrical insulation between patient and any conductors • Patient isolated from bore, RF coils and cables, and skin-to-skin contact via vendor-recommended padding • RF coils, cables, and leads isolated from bore wall and each other • All coils/leads properly engaged (ie, plugged in) • Use only properly maintained, operated, and undamaged MR conditional equipment • Uncommunicative and/or anesthetized patients • Appropriate patient monitoring plan
Conductors	• Avoid unnecessary conductors and/or metallic objects • Damp clothing, diapers, hair or skin • Jewelry, tattoos, makeup, hair products, clothing • Medicinal transdermal patches • Active, passive, or on-body implanted medical devices • Read and follow MR conditions on all devices or implants • Metallic objects as far from bore wall as possible • Two or more conductors in close proximity (≤3 cm)
Current loops	• Avoid circular, U-shaped, S-shaped conductor configurations • Skin-to-skin contact • Coil cables and conducting leads • Tattoos • Jewelry
Antenna effect	• Elongated implants, including fully or partially implanted leads, interventional needles, and guidewires • Greatest risks near resonant length or longer
RF exposure	• High SAR for short or sustained periods • Use lowest possible SAR

[a] Consider restricting to normal operator mode and/or use appropriate medical supervision and patient monitoring if risk factors cannot be mitigated.

The coils seek to modulate the value of the axial (B_z) by an amount small with respect to the magnetic field (mT) linearly around isocenter of each logical axis (x, y, z). These gradient magnetic fields (G_x, G_y, G_z) range from their lowest value at the isocenter to their largest positive or negative values near the edge of the usable field of view in each direction. Gradients in each direction can be run simultaneously, providing for a rapid change in the local magnetic field (dB/dt) at points away from the isocenter. The rapid switching of these gradients during acquisitions and associated dB/

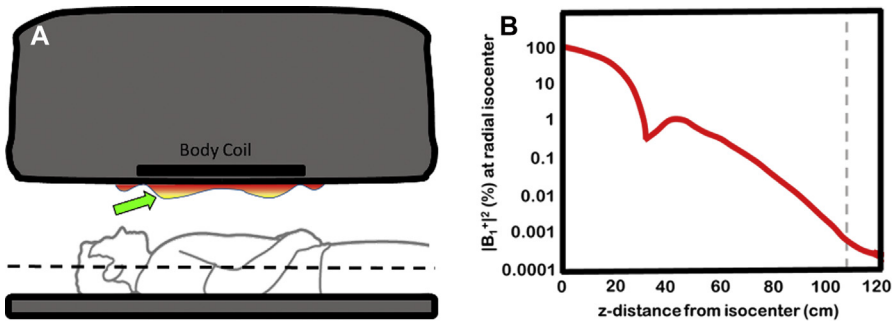

Fig. 3. Spatial distribution of RF power in MR imaging. The internal whole-body volume transmit-receive coil is the primary source for RF excitation in modern MR imaging. The coil is usually a multirung birdcage quadrature coil. Near the edge of the bore, the RF field intensity can be large (*A*), with the highest intensity occurring near the edge of the coil in the z-direction away from isocenter. Because the rungs are configured in a circle around the bore, these hotspots occur periodically around the bore as well, which can be a safety concern for patient anatomy (especially upper extremities or shoulders), devices, and/or medical implants abutted up against the edge of the bore and not protected by proper padding as recommended by the vendor. Vendor-supplied documents on the spatial distribution of the RF field (*B*) show the transmitted RF power along the magnet isocenter and do not include these hot spots. However, such plots are still useful for estimating the exposure to a particular region of anatomy, device, or implant in the field. RF power decreases to less than 1% at the edge of the coil (approximately z = 30 cm for this system) and is many decades attenuated by the edge of the bore (*gray dashed line*). Therefore, although the highest RF power experienced by most tissue lies near isocenter, hotspots at the radial edge of the bore wall must be considered for patients and devices in these areas.

dt away from the isocenter is responsible for the primary safety concerns from this field[55] because this generates the loud acoustic noise associated with MR imaging; can induce potentially uncomfortable nerve stimulation in the patient; or, in the case of AIMDs in particular, may result in unintended stimulation to the patient, device vibration and heating, as well as temporary disruption or permanent damage to the device.[25]

Powerful gradients are a hallmark of modern MR scanners because they facilitate rapid, high-resolution volume imaging and shorter echo times and echo spacing, among other performance enhancements. Large currents (kiloamps) are switched on the gradient coils in the bore to generate the linear magnetic field in each direction. Generating this amount of current results in significant equipment heating, which requires active cooling of the system. Imaging when the cooling system is not fully functional can result in overheating and permanent damage. Gradient performance is generally characterized by the maximum amplitude per axis (40–80 mT), fastest increase time (100–400 microseconds), and maximum slew rate (130–200 T/m/s). These values are usually part of standard gradient performance specifications provided by the vendor in describing the system and can be useful for characterizing the system, although the stated specifications may exceed what the system can actually apply during routine imaging because of safety or system gradient heating limitations.

As the gradients switch on and off during an acquisition, the associated dB/dt is largest away from the isocenter (**Fig. 4**), and it is near these locations that safety concerns with respect to interaction with patients and devices are greatest. In particular, the largest potential value when all gradients are simultaneously at maximum slew happens at a location away from the z-isocenter, near the edge of the bore. The spatial distribution of these values (T/s) for making safety decisions can be found in the vendor system manual.

Peripheral Nerve Stimulation

As previously discussed, time-varying magnetic fields result in an electromotive force and resulting electric field in conducting materials via Faradays' law of induction. Time-varying gradients differ from RF in that dB/dt has a higher amplitude but a much lower frequency (kilohertz). However, even the systems with the weakest gradients on the market are capable of exceeding the stimulation threshold for the tissue. The system operating mode safety limits have been developed to limit this peripheral nerve stimulation (PNS), which may result in patient discomfort. Beyond comfort, it is important to limit PNS because examination efficacy can be compromised by movement associated with this patient discomfort or agitation. In addition, PNS thresholds in the current region of operation have also been shown to be below cardiac stimulation thresholds, which could result in

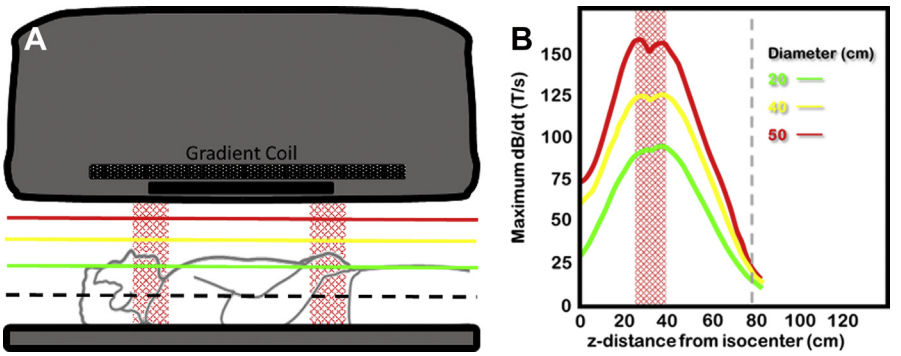

Fig. 4. The distribution of maximum gradient magnetic field dB/dt in MR imaging. A patient lying in an MR scanner (*A*) experiences increasing dB/dt as a function of radial distance from isocenter. Here, radii for diameters around isocenter are shown for 20 cm (*green*), 40 cm (*yellow*), and 50 cm (*green*). When combined with the increasing dB/dt in the z-direction moving away from isocenter, the maximum dB/dt the patient can experience happens in the cross-hatched regions near the edge of the gradient coil. These large dB/dt may be the source of painful stimulation or could damage/disrupt AIMDs in the patient. A plot of these regions for a 60-cm diameter bore magnet with powerful gradients (80 mT/m amplitude and 200 T/m/s slew rate) is also shown (*B*).

additional patient safety events, such as induced arrhythmia.

PNS is characterized by tingling sensations of slight muscle spasms in the ribs, side, abdomen, hip, buttock, or thoracic regions, or along the upper arms or the back muscles in the shoulder region. There is a large variance in stimulation thresholds in patients depending on physiologic conditions. In addition, the specific coil design and gradient pulse shapes are factors as well. As opposed to using derived values, stimulation limits for a specific system can be determined by averaging the individual stimulation thresholds of test subjects. First-level controlled operating mode is such that 50% of all patients experience at least mild stimulations after reaching the stimulation threshold. Normal operating mode limits the scanner to 80% of this threshold.

Mathematically maximum magnetic field switching threshold for PNS can be modeled as a hyperbolic function[55]:

$$\left.\frac{dB}{dt}\right|_{max} = b\left(1+\frac{c}{d}\right) \qquad \textbf{Equation 9}$$

Where b is the mean threshold for stimulation (rheobase) given an infinite pulse duration, and c is the duration at which the stimulation threshold is twice the rheobase. Chronaxie is the smallest time duration required for stimulation for an amplitude twice the rheobase, and d is the pulse duration ($d = 2 \cdot G_{max}/G_{slew}$ for trapezoidal gradients). In general, greater stimulus strengths result in shorter stimulation times. This hyperbolic form for stimulation threshold best fits data from both early numerical simulation work and experimental data.[56–58]

From a fit of the data, threshold output values for PNS on whole-body gradient systems are a rheobase of 20 T/s for first-level operator mode and chronaxie of 0.36 milliseconds. The rheobase is reduced by 80% for normal operator mode. If the vendor has gradients capable of a rheobase higher than 20 T/s, then further testing is needed to establish stimulation thresholds.[15] Hardware is ultimately limited by the 1% cardiac threshold (c = 3 milliseconds and b = 20 T/s) derived from simulation with a safety factor of 3 reducing the cardiac rheobase such that risk of cardiac stimulations at higher ramp times is reduced.[10]

Time-Varying Magnetic Fields and Medical Devices

In the presence of a conducting medical device or implant, rapid switching (dB/dt) of powerful gradients is a consideration.[25] Heating of conducting devices was covered previously for RF pulses. With gradient pulses, the instantaneous power deposited by the eddy currents from the switched gradient field can lead to device heating. Power is estimated similar to Equation 6, but with dB/dt replacing $f \cdot B_1$, making heating proportional to the product of the conductivity, volume, and $|dB/dt|$.[2] Most notably, these effects likely only come into play near the regions of maximum dB/dt for large-volume implants. Vibration can arise from gradient switching because of an eddy current–induced magnetic moment attempting to align with the static field. The induced torque (L) is similar to Equation 3 and is the product of the conductivity, volume, static field, and dB/dt. In addition, for AIMDs, there is the potential for unintended patient stimulation from induced

voltages on leads, and device malfunction, such as device malfunction caused by electrical interference. As with SAR, the calculations of dB/dt provided by the vendor are provided for estimating likelihood of PNS, not for establishing and adhering to thresholds for interactions with implanted or external devices.[25]

Gradient-Induced Acoustic Noise

A current-carrying coil in a strong static magnetic field experiences a force (Lorentz) orthogonal to both the current and the magnetic field. This movement of the coil during gradient slewing is the primary source of acoustic noise in MR scanners.[59,60] Because the noise may be uncomfortable, cause anxiety, or result in temporary hearing loss for patients, vendors work hard to design gradients in which the acoustic noise generated from these mechanical forces is minimized. The physical aspects of acoustic noise are characterized by the frequency spectrum, intensity, and duration of exposure.

Measurements are made versus a reference acoustic pressure value (p_0), such as the threshold for human hearing (20 μPa at 1 kHz). The peak allowable sound pressure level (SPL) when referenced against the lowest sound pressure is given by:

$$\mathrm{SPL(dB)} = 20\ log_{10}\left(\frac{p}{p_0}\right) \qquad \textbf{Equation 10}$$

Peak average SPL exposure for adults is 140 dB (eg, jet engine) and 120 dB for children. Because human hearing is sensitive in a particular part of the audio frequency spectrum, time-limited exposure limits tend to be weighted to emphasize measurement of noise in this part of the audio spectrum (approximately 1–8 kHz) as the human ear might, and hence the SPL is designated as the A-weighted average (dB-A). The IEC limit for patients in MR imaging is less than or equal to 99 dB-A, extrapolated from World Health Organization (WHO) occupational exposures, under the constraining assumption of a single exposure in a day for 1 hour.[10] This value may be increased or decreased by 3 dB-A for halving or doubling of the exposure time, respectively.

In general, noise levels in the bore average around 115 dB-A and can reach peaks of 135 dB-A, with fast sequences, such as echo-planar imaging (EPI), generally being the loudest. At the console, the levels are usually less than 60 dB-A. Therefore, hearing protection should be used to decrease noise to at least 99 dB-A for patients and 85 dB-A for personnel in the examination room[61] via hearing protection (ie, plugs, muffs, or both) with noise reduction ratings greater than 29 dB. Note that, because the noise reduction ratio stated depends on proper placement, it is extremely important to make certain that personnel are properly trained in placing and checking placement of these devices in the various clinical scenarios, such as children, neonates, or anesthetized patients. In addition, an array of quiet or silent sequences has become available from vendors. These sequences often come with a tradeoff between desired timing, contrast, and/or signal/noise ratio. In the end, the established guidelines reflect a continuous exposure, whereas it is unlikely that maximal acoustic noise exposure would be present during the entire time spent on the table.

Safety Concerns of Time-Varying Gradient Magnetic Field Summary

The primary safety concerns arising from the time-varying gradient magnetic field arise from dB/dt, which is responsible for generating the loud acoustic noise associated with MR imaging that mandates hearing protection be used when in the suite during imaging procedures.

Another safety concern is the potentially for uncomfortable PNS when operating in the first-level operating mode. The likelihood of experiencing stimulation is greatly reduced in normal operating mode where gradient performance is derated to 80% of maximum.

In addition, the impact of the gradient magnetic field on implants can result in heating, vibration, unintended patient stimulation, or active device malfunctions. These conditions may also require the system be run in normal operating mode, restriction of positioning of the device within the bore during imaging, or in the future using vendor ability to use the IEC fixed-parameter option for gradients,[10,25] similar to how some vendors have allowed use of a fixed SAR or $B_1{}^+{}_{rms}$ below the normal operating mode limit.

High-performance sequences tend to use large dB/dt gradient pulses, such as EPI, which are used for diffusion or functional MR imaging examinations, as well as balanced steady-state free precession acquisitions, which require rapid, short-TR and short-echo-time (TE) acquisitions for applications such as real-time or CINE cardiac acquisitions. Note that running the MR system in normal operating mode for dB/dt may affect high-performance acquisitions. The speed of acquisitions (including breath-hold times), slice and/or in-plane resolution, and contrast caused by potential increases in echo-spacing or minimum TE (ie, echo-trains, Dixon imaging, in-

phase/out-of-phase imaging) may all be modified and require some attention to ensure the quality and completion of the examination. As gradient performance continues to be pushed, using novel gradient coil or pulse design, and better modeling of patient stimulation or device interactions, it may aid in maintaining both patient safety and access to high-performance acquisitions for a large population of patients.

DISCUSSION

MR imaging has been one of the fastest-proliferating advanced imaging modalities in medical imaging. Despite the rapid growth in both use and technology, MR imaging remains one of the safest advanced imaging services available. In addition, when operated within the established regulated output parameters by knowledgeable teams of personnel who remain diligent and up to date in their MR safety training, MR imaging also remains a nonsignificant medical device according to the FDA.

As shown in this article, the primary safety concerns in the MR environment arise from the physical consequences of the presence of the static magnetic field, which is always on for high-field systems, and the time-varying RF and gradient magnetic fields, which are active during image acquisition. Advances in MR technology, such as higher-field-strength systems with more powerful gradients, remain within the regulated output parameters but may introduce new issues regarding patient safety that require constant vigilance by the MR teams managing the patients. These teams must understand the impact of changes in the technology, including under what circumstances and to what degree patient management may need to change in order to maintain patient safety for a given system.

Knowledge of the basic potential interactions between the patients, devices, and the MR scanner facilitates more informed and timely patient management decisions that help to maintain patient access to MR imaging. This process may increasingly include routing of patients to specific systems with an safety profile amenable to their particular needs and medical conditions, or to manage MR safety conditions on an implanted device. To accomplish this, it is imperative to be able to assess and understand the safety parameters of the MR system and the specific conditions needed to successfully complete an examination while managing the risks.

In the presence of increasing numbers of implantable medical devices, wearable technology, and other devices being tested to establish highly specific conditions for managing the risks associated with scanning in MR imaging, knowledge of field strength and operating mode restrictions is no longer enough. Users must understand the output and output distribution of which their system is capable, how these relate to the management of their patients, and the tools or procedures for controlling exposures on their systems. Artifacts, or the changes needed to manage the patients, may affect diagnostic quality of the examination, especially those examinations relying on high-performance acquisitions. With this in mind, reassessing the appropriateness of the MR examination, and how to modify the examination to achieve the needed goals safely, are skills that will be important for the team to have to maintain appropriate patient access to this advanced imaging service. Understanding the potential issues and tradeoffs in MR imaging can aid teams in providing higher-reliability examination safety and quality through more informed patient selection, protocoling, screening, scanning, and interpretation.

Clinics care points

- An understanding of the physical underpinnings of MRI safety is necessary for informed risk-benefit decision making as well as risk management.
- Applying the physical principles of MRI safety to the management of patients in the MR environment as well as acquisition protocols can aid in reducing patient discomfort, enhancing both patient compliance and image quality as well as safety.

DISCLOSURE

The authors have nothing to disclose.

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