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The Underestimated Role of Gradient Coils in MRI Safety

Luca Zilberti,^{1*} Alessandro Arduino,^{1,2} Oriano Bottauscio,¹ and Mario Chiampì²

MRI scanning of patients carrying implants is becoming a reality in many hospitals, because of the wide increase of population with implants, particularly in the orthopedic ones (1). The responsibility for this practice falls on the subscribing physicians or radiologists, who rely on guidance establishing safety and compatibility of implants in the MR environment (eg, (2)), which basically address issues related to magnetically induced mechanical actions and radiofrequency (RF) induced heating of tissues. Even if large-scale MR safety studies on orthopedic implants have been published that show no evidence of specific risks for patients' health (eg, (3)), additional analyses of the interaction between MR fields and metallic implants can contribute to support the scan decision on a firm scientific rationale.

In a recent paper (4), we showed that the tissues surrounding a metallic hip prosthesis exposed to a gradient field (GF) may undergo a nonnegligible heating (up to some degree), as a result of the electromagnetic energy deposited in the implant. Since then, we have realized that in the MR community these results are generally received with some skepticism, probably because of the fact that most of the scientific papers dealing with thermal problems in MRI focus on RF fields (5–16). Similarly, the attention of relevant standardization and regulatory bodies is focused solely on RF-induced heating (2,17). This tendency is absolutely well-grounded when studying native tissues, where GFs are not able to develop significant thermal effects. However, in the presence of metallic foreign bodies the situation is different.

Many papers show that a significant heating can be associated with the exposure of such bodies to RF fields, especially with wire-like structures, which may act as antennas (18–30). In this case, RF typically deposits a relatively low power inside the metallic parts, but their presence may cause the enhancement of the specific absorption rate directly developed in the tissues anyway. What we would like to highlight here is that, in presence of bulky metallic prostheses, GFs can produce a nonnegligible heating too, because of the thermal power generated inside the metal (Joule effect), which involves the tissues indirectly, by diffusion (“indirect effect” (31)). The GF thermal effect, which becomes sizable only for prostheses far from the coils’

isocenter, has been considered up until now by a few articles (4,32–34). To sustain these findings, we exploit an analytical solution (35) that is not affected by possible artifacts of numerical results. The solution provides the current density J (in complex notation) induced within a nonmagnetic metallic sphere, with radius R , conductivity σ and permeability μ_0 , immersed in a nonconductive background. When the sphere is radiated by a homogeneous, time-harmonic, magnetic flux density B (peak value) at frequency f , the current density in an internal point is

$$J = \frac{B}{\mu_0} \frac{3k^2 R}{2\sin(kR)} \left[\frac{\sin(kr)}{(kr)^2} - \frac{\cos(kr)}{kr} \right] \sin\theta \quad [1]$$

with

$$k = \sqrt{-j2\pi f \mu_0 \sigma} = \frac{\sqrt{-2j}}{\delta} \quad [2]$$

where r is the distance from the sphere center, θ is the colatitude measured from the sphere diameter parallel to the magnetic field, δ is the penetration depth, and $j^2 = -1$. The current density is azimuthal everywhere.

This solution is used to simulate the effect of GFs, considering that they are approximately homogeneous in correspondence to the quite small volume taken up by the sphere (which is placed away from the coils' isocenter). The total Joule losses (P) are obtained by integration of the power density $|J|^2/(2\sigma)$ over the sphere and therefore scale as B^2 ; in contrast, for a given B amplitude, it can be proved that P increases monotonically with frequency.

The same solution turns out to be useful also for simulating a homogeneous rotating RF field (like that of birdcage antennas around the system isocenter), decomposable into two time-harmonic fields, with a 90° shift both in phase and geometrical disposition. The total current density is obtained by superposing the two corresponding contributions given by (1) (taking into account their direction and phase). For the rotating field, P is double than that given by a single oscillating field (with the same frequency and amplitude).

Table 1 indicates the power developed in spheres of different radii, considering two conductivity values close to those of typical alloys adopted for prostheses (eg, 0.58 MS/m for Ti-6Al-4V). The results refer to a GF of 6 mT (eg, the dominant longitudinal component produced by a 30 mT/m gradient, at 0.2 m from the coils' isocenter) at 100 Hz or 1 kHz (the most significant harmonic components of many real sequences fall within this range) and a RF rotating field of 10 μ T at 64 or 128 MHz.

As can be seen, in some cases GFs give rise to power values higher than, or at least comparable to, those resulting

¹Istituto Nazionale di Ricerca Metrologica, Torino, Italy.

²Politecnico di Torino, Dipartimento Energia, Torino, Italy.

*Correspondance to: Luca Zilberti, PhD, Istituto Nazionale di Ricerca Metrologica, Strada delle Cacce 91, I-10135, Torino, Italy. E-mail: l.zilberti@inrim.it
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Table 1
Total Joule Losses P (in watt) Developed within the Sphere

R (mm)	Gradient field (6 mT)				RF field (10 μ T)			
	$f = 100$ Hz		$f = 1$ kHz		$f = 64$ MHz		$f = 128$ MHz	
	$\sigma = 10^6$ S/m ($\delta = 50.3$ mm)	$\sigma = 10^5$ S/m ($\delta = 159$ mm)	$\sigma = 10^6$ S/m ($\delta = 15.9$ mm)	$\sigma = 10^5$ S/m ($\delta = 50.3$ mm)	$\sigma = 10^6$ S/m ($\delta = 62.9$ μ m)	$\sigma = 10^5$ S/m ($\delta = 199$ μ m)	$\sigma = 10^6$ S/m ($\delta = 44.5$ μ m)	$\sigma = 10^5$ S/m ($\delta = 141$ μ m)
40	$3.00 \cdot 10^{-1}$	$3.05 \cdot 10^{-2}$	12.8	3.00	$3.03 \cdot 10^{-2}$	$9.50 \cdot 10^{-2}$	$4.30 \cdot 10^{-2}$	$1.35 \cdot 10^{-1}$
20	$9.52 \cdot 10^{-3}$	$9.53 \cdot 10^{-4}$	$8.70 \cdot 10^{-1}$	$9.52 \cdot 10^{-2}$	$7.55 \cdot 10^{-3}$	$2.40 \cdot 10^{-2}$	$1.07 \cdot 10^{-2}$	$3.36 \cdot 10^{-2}$
10	$2.98 \cdot 10^{-4}$	$2.98 \cdot 10^{-5}$	$2.96 \cdot 10^{-2}$	$2.98 \cdot 10^{-3}$	$1.90 \cdot 10^{-3}$	$5.90 \cdot 10^{-3}$	$2.67 \cdot 10^{-3}$	$8.35 \cdot 10^{-3}$
5	$9.30 \cdot 10^{-6}$	$9.30 \cdot 10^{-7}$	$9.30 \cdot 10^{-4}$	$9.30 \cdot 10^{-5}$	$4.70 \cdot 10^{-4}$	$1.40 \cdot 10^{-3}$	$6.63 \cdot 10^{-4}$	$2.06 \cdot 10^{-3}$

from RF (in particular for bulky prostheses). This happens because B is much lower for RF fields than for GFs. Note that, at RF, P is higher when the implant is less conductive, and vice versa for GFs. The case $R = 20$ mm is representative for the femoral head of a hip prosthesis, whose P at 1 kHz is similar to that found in previous realistic simulations (4,36), in which nonnegligible heating was predicted (4). Because the duty cycle of GFs is typically higher than that of RF fields, their contribution would appear even more significant in terms of energy.

In conclusion, these findings may contribute to the theoretical understanding of induced heating in metallic implants, demonstrating that GFs can provide a measurable effect, particularly when the implant is at some distance from the coil's isocenter, also in light of the tendency of designing higher and higher gradient-strength systems.

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