

Review Article

Pacing in magnetic resonance imaging environment:

Clinical and technical considerations on compatibility

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Introduction

Magnetic resonance imaging is a widely accepted tool for the diagnosis of a variety of disease states. However, the presence of an implanted pacemaker is considered to be a strict contraindication to magnetic resonance imaging in the vast majority of medical centres, precluding a substantial and growing number of patients from the diagnostic advantages of this imaging modality.

The potential effects of magnetic resonance imaging on cardiac pacemakers are multiple (Table 1)^[1–5]. The static magnetic field may close the reed switch, resulting in asynchronous pacing. The radiofrequency field may induce an undesirable fast pacing rate. The time-varying magnetic gradient fields may induce currents in the pacemaker system of sufficient magnitude to pace the heart. There are other theoretical concerns, including potential thermal injury at the lead tip. However, the most influential factor that supports the current practice is the reported lethal consequences of magnetic resonance imaging in patients with implanted pacemakers^[6–8].

Despite the above-mentioned concerns, the effects of magnetic resonance imaging on cardiac pacemakers remain controversial. Most of the previous studies that prohibit magnetic resonance imaging in pacemaker patients were based on in vitro and animal model data from the 1980's using older pacemaker and lead technology^[1–3]. More recent anecdotal reports describe a small series of pacemaker patients who have safely

undergone magnetic resonance scanning (Table 2)^[9–14]. These studies were performed in patients for whom magnetic resonance imaging was considered to be an absolute diagnostic necessity.

The potential effects of magnetic resonance imaging in patients with implanted pacemakers is outlined in this paper, which provides an extensive review of previous in vitro and animal studies, as well as recently published observations in humans. Based on technical innovations and clinical developments in magnetic resonance imaging and pacemaker systems, safety and compatibility issues are discussed.

Clinical considerations

Basic principles of magnetic resonance imaging

Since the first live human images reported in 1977, the development of clinical magnetic resonance imaging has been rapid^[15,16]. Continuous improvement in magnetic resonance methodology, hardware and software has

Table 1 Potential effects of magnetic resonance imaging on pacemaker systems

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|---|
| 1. Static magnetic field |
| (a) Reed-switch closure |
| (b) Pacemaker displacement |
| (c) Changes in electrocardiograms |
| 2. Radiofrequency field |
| (a) Heating |
| (b) Alterations in pacing rate |
| (c) Pacemaker reprogramming or reset |
| 3. Time-varying magnetic gradient field |
| (a) Induction voltage |
| (b) Heating |
| (c) Reed-switch closure |

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Table 2 Published reports describing the non-lethal consequences of magnetic resonance imaging in pacemaker patients

Author (ref.)	Year	n	Indication for MR scanning	Pacemaker model	Dual/single chamber	Lead polarity	Pacemaker mode	Magnetic field	Outcome
Alagona ^[9]	1989	1	Brain tumour	AFP II (283)	Dual chamber	Unipolar	OOO	1.5 Tesla	Normal PM function
Inbar ^[10]	1993	1	Cerebello-pontine angle syndrome	Paragon II (2016T)	Dual chamber	Bipolar	OOO	1.5 Tesla	Normal PM function
Gimbel ^[12]	1996	1	Heart valve	AFP (261)	Single chamber	Unipolar	OOO	1.5 Tesla	MR image artifact
		1	Brain tumour	Genesis (285)	Dual chamber	Unipolar	DOO	0.5 Tesla	Pause (2 s)
		1	CIA	Paragon II (2016)	Dual chamber	Bipolar	OOO	0.35 Tesla	Normal PM function
		1	Pituitary tumour	Synchrony (2020)	Dual chamber	Bipolar	DOO	1.5 Tesla	Normal PM function
		1	Cervical disk	Synchrony II (2022)	Dual chamber	Bipolar	DDD	1.5 Tesla	Normal PM function
		1	Dizziness	Thera DR (7940)	Dual chamber	Bipolar	VVI	1.5 Tesla	Rapid ventricular pacing
Fontaine ^[49]	1998	1	Cranial nerve palsy	Meta (1254)	Dual chamber	Bipolar	AOO	1 Tesla	Asynchronous pacing
Garcia-Bolao ^[11]	1998	1	CNS (13)	Elite (7077) (3x)	Dual chamber	n.a.	DDD	0.5 Tesla	Asynchronous pacing
Sommer ^[14]	1998	18	Cardiac tumour (2)	Elite (7085) (2x)	Dual chamber	n.a.	DDD	0.5 Tesla	Asynchronous pacing
			Periprosthetic (asc. aorta)	Elite (7076)	Dual chamber	n.a.	DDD	0.5 Tesla	Asynchronous pacing
			Pseudoaneurysm (1)	Thera DR (7940) (2x)	Dual chamber	n.a.	DDD	0.5 Tesla	Asynchronous pacing
			Paravalvular (prosthetic)	Relay (294-03) (2x)	Dual chamber	n.a.	VOO	0.5 Tesla	Normal PM function
			Aortic valve) (1)	Vista (941) (2x)	Dual chamber	n.a.	DDD	0.5 Tesla	Asynchronous pacing
			Constrictive pericarditis (1)	Synergyst (7027, 7071)	Single chamber	n.a.	DDD	0.5 Tesla	Asynchronous pacing
				Dialog (700) (2x)	Single chamber	n.a.	VVI	0.5 Tesla	Asynchronous pacing
				Spectrax (5985)	Single chamber	n.a.	VVI	0.5 Tesla	Asynchronous pacing
				Ceryx 3	Single chamber	n.a.	VVI	0.5 Tesla	Asynchronous pacing

n = number of patients studied; n.a. = data not available; PM = pacemaker

now resulted in whole body imaging systems that are capable of producing high contrast images with a spatial resolution of under 1 mm, and in total imaging times of under 15 min, which permit a physician to complete a diagnostic evaluation^[17]. A magnetic resonance image is the relative response of specific nuclei to absorbed radiofrequency energy. Most magnetic resonance images are designed to observe the hydrogen nucleus because of its relative abundance in the body. A magnetic resonance image is usually a tomographic map of the distribution of protons in the imaged sample. However, image contrast is influenced more by other physical factors, including differences in the ability to re-emit the absorbed radiofrequency signal and flow phenomena. Magnetic resonance imaging is similar to computerized tomography as a cross-sectional imaging modality, but it is non-ionizing and offers additional soft tissue contrast. The imaging sequences can be modified to visualize blood flow and to compensate for the blurring effects of cardiac or respiratory motion. Magnetic resonance also offers the unique ability to acquire images in virtually any orientation, without repositioning the patient. This provides greater convenience for the medical staff and minimizes patient discomfort. In addition, magnetic resonance imaging provides chemical information not measurable with conventional imaging modalities. The above-mentioned unique abilities have accelerated the acceptance of magnetic resonance imaging in current medical practice.

Potential benefits of magnetic resonance imaging in paced patients

General indications

In patients with permanent pacemakers, magnetic resonance imaging may provide important diagnostic benefits, since this technique has advantages over other imaging modalities^[18]. Magnetic resonance imaging has now become the procedure of choice for all congenital, traumatic, hereditary, vascular, infectious, autoimmune, metabolic, and neoplastic disorders of the central nervous system. It is the procedure of choice for further evaluation of the entire range of musculoskeletal disorders, when the physical examination or plain radiography suggests a serious abnormality. Magnetic resonance imaging shows excellent contrast between tumour and other tissues, therefore it has become an indispensable tool in oncology. Cardiovascular magnetic resonance imaging is useful in the evaluation of congenital and acquired diseases of the heart and great vessels and has been a rapidly advancing area of clinical research^[19].

Magnetic resonance imaging in cardiovascular diseases

Magnetic resonance imaging provides substantial information on cardiac anatomy and measurements of cardiac function, such as flow, perfusion, wall motion, and metabolism^[19]. Special magnetic resonance tagging sequences allow non-invasive assessment of myocardial

deformation during contraction/relaxation^[20–25]. Magnetic resonance imaging may also be an invaluable tool with which to study the influence of pacing site and multisite pacing on cardiac function. While clinically used pacing produces considerable deterioration of mechanical performance^[26–32], pacing from various sites have shown that more synchronous activation of the ventricles is associated with improvements in cardiac haemodynamic function^[33–35], which may be demonstrated by magnetic resonance imaging. Cardiac stress studies using magnetic resonance compatible catheters may detect abnormalities in cardiac function and flow that may become apparent only during increased demand^[36–39]. In addition, a stable heart rate imposed by pacing would offer alternative means of preventing artifacts in all magnetic resonance imaging techniques^[40–42].

Effects of magnetic resonance imaging on cardiac pacemakers

In vitro and animal studies

The potential hazardous effects of magnetic resonance imaging in patients with cardiac pacemakers have been studied since 1983. Pavlicek and colleagues were the first to report the effects of magnetic resonance imaging on pacemaker function^[1]. They showed that radiofrequency fields present in an magnetic resonance unit could possibly inhibit demand pacemakers and time-varying magnetic fields could generate pulse amplitudes to mimic cardiac activity. The threshold for initiating the asynchronous mode of a pacemaker was reported to be as low as 17 gauss. The possibility of altering pacemaker parameters was presented as a serious limitation of magnetic resonance imaging. Fetter *et al.* showed that pacemakers reverted from the demand to the asynchronous mode within the magnetic field of the scanner (0.15 Tesla), but microscopic testing showed no evidence of reed switch sticking or magnetizing, or damage to other discrete pacemaker components^[2].

Other investigators studied the feasibility of dual-chamber pacing systems in the magnetic resonance environment. Erlebacher *et al.* tested different DDD pacemakers in a saline phantom, and showed that during scanning at 0.5 Tesla, all units malfunctioned due to radiofrequency interference which caused total inhibition of atrial and ventricular output, or resulted in atrial pacing at very high rates^[3]. The potential for rapid cardiac stimulation during magnetic resonance scanning was also reported in animal studies^[43]. Lauck *et al.* investigated the performance of different stimulation modes (VVI, VVIR, VOO, DDD, DDDR and DOO) during magnetic resonance scanning at 0.5 Tesla^[44]. Reversible activation of the reed switch with consecutive asynchronous stimulation was observed in all pacemakers. Pacemakers in the asynchronous mode were not affected with regard to stimulation rate and capture during scanning. In contrast, pacemakers with

automatic mode switching to demand pacing or programmed inactivation of the reed switch were triggered in the dual chamber mode and were inhibited in the single chamber mode. Thus, the investigators recommended programming into the asynchronous mode prior to scanning, and in those without permanent pacemaker dependency, complete inactivation of the system, if possible.

The effects of more powerful magnetic resonance scanners (i.e., 1.5 Tesla) on cardiac pacemakers were initially reported by Hayes *et al.*^[45]. In vivo evaluation of different single and dual-chamber pacemakers showed reversion into asynchronous mode and transient reed switch inhibition. Seven of the eight pulse generators paced rapidly when exposed to the radiofrequency signal associated with a marked decrease in blood pressure. Stimulation cycle length was 200 ms (300 beats \cdot min⁻¹) corresponding to the frequency of pulsing. It was proposed that rapid pacing was the result of an 'antenna' effect that couples the radiofrequency energy back into the pacemaker output circuits. More recently, Achenbach *et al.* showed in a phantom study on 11 pacemakers and 25 leads that no pacemaker malfunction was observed in asynchronous pacing mode (VOO/DOO), whereas inhibition and rapid pacing were observed during spin-echo imaging if the pacemakers were set to VVI or DDD mode^[46]. The authors suggested that rapid pacing was caused by induction of currents above the sensing threshold in the atrial lead and consequent triggering of ventricular stimulation. Direct interference with the pacemaker electronics seemed to be an unlikely explanation, because the rapid pacing rate was always equal to the programmed frequency limit. The heating effect of pacemaker leads was also investigated in this study. Continuous registration of the temperature at the lead tip using an optical temperature sensor showed a maximal temperature increase of 63.1 °C during 90 s of scanning. In seven electrodes, the temperature increase exceeded 15 °C. Resultant myocardial necrosis could be demonstrated in histological studies.

Human studies

There are few anecdotal reports of unexpected deaths in patients undergoing magnetic resonance imaging^[6-8]. In one case, the patient had no escape ventricular rhythm and apparently died due to asystole. Another patient developed ventricular fibrillation during the imaging procedure that was not recognized immediately because ECG monitoring was not used^[47]. It is likely that many potential complications are unreported in the literature for various reasons (liability, etc.). On the other hand, there are also reports of pacemaker patients who underwent magnetic resonance imaging safely (Table 2). Therefore, differences exist among clinicians regarding the perceived safety of scanning paced patients^[48].

In patients who underwent magnetic resonance imaging of the head, no pacemaker malfunction was observed with the pacemaker turned off or programmed to an asynchronous pacing mode prior to magnetic

resonance exposure^[9-11]. In another study on five patients with pacemakers, Gimbel *et al.* reported normal pacemaker performance in four patients during magnetic resonance imaging (0.35 and 1.5 Tesla)^[12]. One patient had a pause of approximately 2 s in duration near the completion of magnetic resonance imaging, the cause of which could not be determined. This occurred in a pacemaker dependent patient with a unipolar dual-chamber device programmed to the DOO mode. No rapid cardiac pacing occurred and no patient reported a torque or heating sensation. Fontaine *et al.* reported a case of rapid cardiac pacing during magnetic resonance imaging (1.5 Tesla) in a patient with a dual chamber pacemaker^[49]. The patient developed an irregular ventricular rhythm during radiofrequency pulsing which terminated with the cessation of radiofrequency pulsing.

Magnetic resonance imaging at 0.5 Tesla was shown to have no influence on atrial and ventricular stimulation thresholds, P and R wave amplitudes, electrode impedance, battery voltage, current, and impedance measurements in patients with implanted pacemakers^[13]. In the largest reported series to date, Sommer *et al.* studied 14 pacemaker models in vitro and 18 patients undergoing magnetic resonance scanning at 0.5 Tesla with standard spin, turbo spin, and gradient echo sequences^[14]. All pacemakers switched to the asynchronous mode due to activation of the reed switch in the static magnetic field in vitro. Atrial and ventricular stimulation thresholds remained unchanged. Pacemaker programme changes, damage to pacemaker components, dislocation/torque of the pacemaker and rapid pacing of the pacemaker were observed neither in vitro nor in vivo. The investigators concluded that the presence of an implanted pacemaker should not be considered an absolute contraindication to magnetic resonance imaging at 0.5 Tesla.

Safety issues in patients with retained pacing leads

Many patients have endocardial pacemaker leads left in place after pulse generator removal. The safety of magnetic resonance imaging in patients with retained endocardial pacemaker wires has not been systematically investigated to date. Likewise, the behaviour of pacing pulmonary artery catheters during magnetic resonance imaging is not known. On the other hand, the effect of transoesophageal pacing leads during magnetic resonance imaging were investigated in animal studies which showed heating with consequential necrosis^[50].

Magnetic resonance imaging in patients with retained epicardial wires after cardiac surgery was previously considered to be a relative contraindication^[51]. Temporary pacing wires, usually made of braided stainless steel, are sutured to the epicardial surface of the heart over the right ventricle and right atrium after cardiac surgery, and connected to an external pacemaker if the patient develops bradycardia or atrioventricular block.

Theoretical calculations using a circuit formed by epicardial pacing wires showed induction of currents up to 80 μ A using a magnetic field strength of 1.5 Tesla^[52]. Hartnell *et al.* investigated the safety of 1 or 1.5 Tesla magnetic resonance systems operating with conventional pulse sequences in 51 patients with retained epicardial pacing wires, cut short at the skin, after cardiac surgery^[53]. None of the patients reported symptoms suggesting arrhythmia or other cardiac dysfunction during magnetic resonance imaging, and there were no changes from the baseline ECG rhythms. Therefore, retained epicardial wires do not seem to present a hazard to patients in the magnetic resonance environment. However, this conclusion applies mostly to non-cardiac magnetic resonance examinations^[54]. A survey among neuroradiologists regarding magnetic resonance imaging in patients with retained epicardial pacemaker wires after cardiac surgery yielded varying practice patterns^[55]. Some physicians would perform magnetic resonance examinations even if they know that wires are present, while others would not even screen for epicardial pacemaker wires. Despite the many examinations, no respondents had experienced a problem related to the presence of epicardial wires in the magnetic resonance environment.

Technical considerations

Electromagnetic fields in magnetic resonance imaging

There are three types of electromagnetic fields used in an magnetic resonance imaging unit:

1. The main *static magnetic field* is used to align the protons. An intense static magnetic field (B_0) is always present even when the scanner is not imaging. The field strength is 0.5–1.5 Tesla in most of the currently available magnetic resonance imaging units for clinical use. Current state-of-the-art technology is pushing this upper limit to 4 or 5 Tesla in research magnetic resonance imaging systems. This is about 100 000 times the magnetic field strength of the earth. In addition to magnetic field strength, magnetic field homogeneity — the measure of field uniformity within the measurement area of the magnet — is an important parameter in the magnet system. Optimal field homogeneity is crucial to generating images free from distortion and with the maximum possible signal-to-noise ratio.
2. The pulsed *radiofrequency field*, generated by the body-coil or the head-coil, is used to change the energy state of the protons and elicit magnetic resonance signals from tissue. The radiofrequency field is homogeneous in the central region and has two components: (a) The magnetic field is circularly polarized in the axial plane, and (b) the electric field is related to the magnetic field by Maxwell's equations. The radiofrequency field is switched on and off during measurements and has a

frequency of 21–64 MHz depending on the magnetic field strength.

3. The time-varying *magnetic gradient fields*, used for spatial localization, change their strength along different orientations and operate at frequencies in the order of 1 kHz. The vectors of the magnetic field gradients in the x, y, and z directions are produced by three sets of orthogonally positioned coils and are switched on only during the measurements.

Potential problems in magnetic resonance imaging equipments

The potential benefits of magnetic resonance imaging are numerous; however, there are hazards intrinsic to the magnetic resonance environment. These hazards may be attributed to one or to a combination of the three main components that make up the magnetic resonance environment.

1. Static magnetic field

(a) *Reed-switch closure.* When a pacemaker is brought close to an magnetic resonance scanner, the reed switch closes, and asynchronous pacing occurs, which may compete with the underlying cardiac rhythm. For this reason, it is common practice for patients with pacemakers to be prevented from undergoing magnetic resonance examinations; they are also physically restricted to remaining at, or outside of, at most, the 5-gauss line surrounding the magnetic resonance imaging environment^[56]. In patients with unstable conditions, such as myocardial ischaemia, there is substantial risk for ventricular fibrillation during asynchronous pacing. A physician has to decide if asynchronous pacing over a longer time period, e.g., up to 1 h, is justifiable or not. In most modern pacemakers the magnet function is programmable. If the magnet response is switched off, synchronous pacing is possible even in strong magnetic fields. In addition, the reed switch may not close inside the homogeneous magnetic field in some special orientations. Programming the pacemaker to the asynchronous mode will avoid the influence of the reed switch during magnetic resonance imaging.

Damage to the reed switch by the strong static magnetic field is a theoretical concern, but it was not reported in any of the published studies^[2,43]. After exposure to the static magnetic field of magnetic resonance equipment, the reed switch of various pacemakers closed at the same field strength as before. Likewise, extended exposure to static magnetic fields (10 h exposure to 1.5 Tesla) did not result in any detectable change on the reed switch, the telemetric coils, or in the pacemaker software settings^[57].

(b) *Pacemaker displacement.* Some parts of pacemakers, such as batteries and reed switches, contain ferromagnetic materials; thus, mechanical forces occur

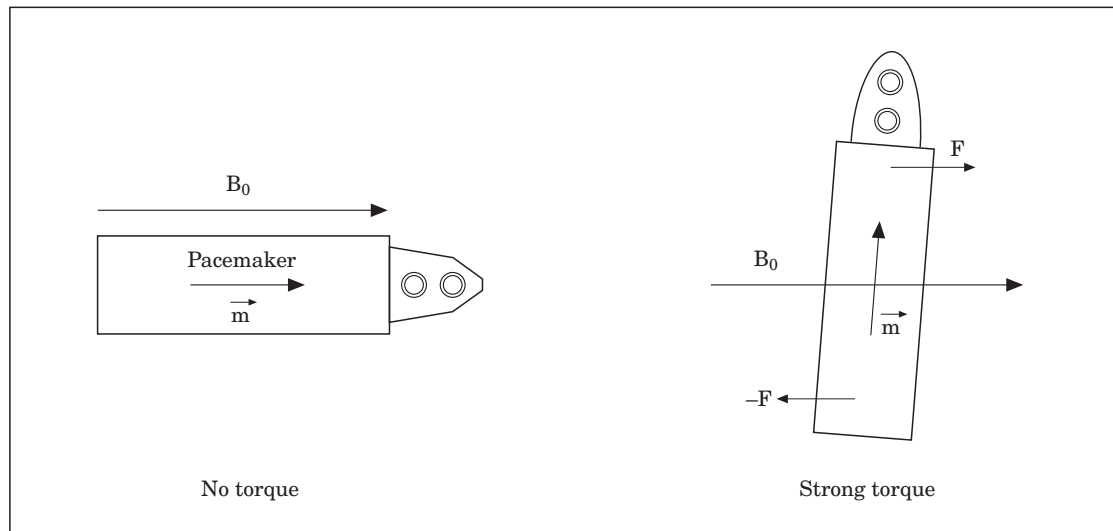


Figure 1 When the orientation of the pacemaker is optimal, the magnetic torque is relatively small (left). Stronger torque occurs in the presence of an orthogonal position of the pacemaker relative to the main magnetic field (B_0) of the magnetic resonance scanner (right).

and the pacemaker may be displaced. Pacemaker displacement may occur in response to magnetic force or magnetic torque. There exists a magnetic force only if the magnetic field changes from place to place. It will be strongest in an area with the greatest field change over a short distance. In the centre of a magnetic resonance scanner, where the magnetic field is maximal, no magnetic force can be measured. Magnetic force increases with increased distance from the magnet isocenter. In contrast, magnetic torque tries to turn ferromagnetic or paramagnetic material parallel to the magnetic field. It has a linear association with the magnetic field strength and will be strongest in the middle of the magnetic resonance scanner.

The magnetic forces exerted on the pacemakers are relatively small. In a preliminary study, we evaluated the magnetic force and torque on 32 pacemaker models (15 dual-chamber and 17 single-chamber units) in a 1.5 Tesla magnetic resonance unit^[58]. The measured magnetic force was in the range of 0.05–3.60 Newton. The newer generation of pacemakers had significantly lower magnetic force values, even lower than the gravity, as compared to the older devices. Likewise, the torque levels were significantly reduced in modern pacemakers. Pacemaker displacement by the magnetic force of the magnetic resonance equipment was not reported in any of the published studies^[9–14].

Pacemakers are paramagnetic, leading to a preferred orientation parallel to a strong magnetic field. If a patient lies parallel to the main magnetic field, as in most magnetic resonance scanners (some 'open' magnetic resonance scanners are exceptions), the orientation of the pacemaker is nearly optimal, so the magnetic torque that rotates the pacemaker may be relatively small (Fig. 1). However, exceptional situations may arise when pacemakers are near the entrance of the magnetic

resonance scanner. The position of the thorax (where the pacemaker and the corresponding leads are located) with respect to the magnet bore is also important.

The metallic parts of the leads are usually composed of MP35N. This alloy of nickel, cobalt, chromium, and molybdenum is non-ferromagnetic; therefore, there is no concern that such leads will move or dislodge^[59].

(c) *Changes in electrocardiograms.* An electric potential is produced by a moving conductor, such as flowing blood, beating heart, or respiring lungs in a static magnetic field. The induced electrical potentials from this motion are signals superimposed on the surface electrocardiogram^[60,61]. Polarization of the flowing blood in the magnetic field may often result in increased T-wave amplitude. Gaffey *et al.* demonstrated reproducible T wave alterations with a threshold near 0.3 Tesla, the extent of which increased linearly with field strength^[62]. Electrocardiogram alterations have also been observed at field strengths as low as 0.1 Tesla^[63]. These observations are in agreement with theoretical calculations that the electrocardiographic changes are from the induced electromagnetic force associated with high velocity blood flow perpendicular to the static magnetic field and presumed to be caused when the negatively charged red blood cells course around the aortic arch^[64]. The induced electrical potentials from the blood motion can be formulated as:

Potential (mV) =

$$\frac{1}{10} \text{ Magnetic field (T)} \cdot \text{Blood velocity (cm/s)} \cdot \text{Vessel diameter (cm)}$$

It was concluded theoretically that the impact of magnetohaemodynamic effects is not important physiologically, because the current densities associated with

these potentials are much lower than that needed for electrical excitation of the heart^[65]. No arrhythmias or changes in heart rate or blood pressure were induced in vivo, and the normal electrocardiogram appeared immediately on cessation of magnetic field exposure^[66]. However, the existence of T wave alterations may be confused with signs of disease if the magnetic influence is not recognized. Occasionally, there may be sufficient elevation of T waves to produce technical difficulties while attempting to gate or trigger the radiofrequency and the gradient pulse timing of the magnetic resonance imaging procedure relative to the timing of the QRS complex of the cardiac cycle.

The strong static magnetic field of an magnetic resonance scanner may also slightly change the intracardiac electrograms. However, pacemakers use electrogram deflections which correspond to P and R waves on surface electrocardiogram for sensing. The shape of these deflections are not greatly altered, because in this time period blood flow and heart motion are relatively small.

2. Radiofrequency field

(a) *Heating.* At the frequencies of interest in magnetic resonance imaging, some of the radiofrequency energy is absorbed and converted to heat. In fact, the main biological effects associated with exposure to radiofrequency radiation are related to the thermogenic qualities of the radiofrequency field^[67–74]. The power deposited by radiofrequency pulses during magnetic resonance imaging is dependent upon multiple factors, including the power and duration of the radiofrequency pulse, the transmitted electromagnetic field frequency, the number of radiofrequency pulses applied per unit time, the type and configuration of the radiofrequency transmitter coil used, the volume of tissue imaged, the electrical resistivity of the tissue, the configuration of the anatomical region imaged, etc^[56]. The causes of heating in the magnetic resonance environment are twofold: (a) radiofrequency field coupling to the lead can occur, inducing significant local heating, and (b) currents induced during the radiofrequency transmission can cause local Ohm's heating next to the tip of the lead.

Guidelines have been established on the allowable whole body exposure, expressed as specific absorption rate (SAR) given as watts per kilogram. The maximum temperature elevation (ignoring cooling effects) can be determined by the following formula:

$$\Delta T (^{\circ}\text{C}) = \frac{\text{SAR (W/kg)} \cdot \text{time (s)}}{\text{specific heat (J/g} \cdot ^{\circ}\text{C)}}$$

For a specific absorbed power of $4 \text{ W} \cdot \text{kg}^{-1}$ for 10 min, the maximum temperature rise equals 0.7°C using the soft tissue specific heat of $0.83 \text{ kcal} \cdot \text{kg}^{-1} \cdot ^{\circ}\text{C}^{-1}$. Because of specific absorption rate limitations, magnetic resonance for diagnostic imaging has not been associated with any detrimental effects, and it appears that this level of whole-body specific absorption rate is acceptable. However, the local electric field can be amplified near conducting

instruments making the peak specific absorption rate difficult to predict. The radiofrequency field in an magnetic resonance scanner has sufficient energy to cause local heating of long conductive wires, such as pacemaker leads, which could destroy parts of the adjacent myocardial tissue. The effects of electrode heating are not detectable by monitoring during magnetic resonance imaging. However, an increase in pacing threshold, myocardial perforation and lead penetration, or even arrhythmias caused by scar tissue are among the potential concerns long after scanning, but such long-term heating effects of magnetic resonance imaging have not been studied yet.

Theoretical calculations, in agreement with in vitro experimental findings, showed that no heating problem occurs in patients with large compact metallic implants^[75]. Therefore, pulse generators without leads are not expected to cause heating problems. In a preliminary study, we investigated the heating effects of magnetic resonance imaging at 1.5 Tesla on 10 various leads (screw-in/passive, uni/bi-polar, coaxial/coradial) in vitro^[76]. Temperature increase exceeded 10°C in eight leads and 20°C in five leads. Under special conditions with the lead tip at the surface of the saline tank, maximal temperature increase of 69°C was observed. None of the tested leads appeared to be particularly safe against heating. The findings were similar when the leads were isolated or connected to pulse generators. Therefore, retained pacemaker leads (yet without a pacemaker) may present a hazard to patients in the magnetic resonance environment.

Estimation of the heating problem in leads. A thin linear wire has a maximal absorption rate with half the length of λ , where λ represents the wavelength of the electromagnetic waves. In a vacuum, the wavelength of the radiofrequency field in a 1.5 Tesla magnetic resonance scanner is approximately 4.7 m, hence maximal absorption will occur in a wire of 2.35 m. In case of saline water, the critical length will be reduced to 30 cm. For a shorter wire ($l < \lambda/2$), the absorption of energy is reduced by the factor $[l/(\lambda/2)]^3$. However, there are other factors that may influence the degree of heating, such as the geometrical structure of the wires, the placement in the body, the insulation of the wires. The strongest heating occurs in areas with high changes of the electric field. Tissue next to sharp edges and points will be exposed to the highest dose of thermal energy. On the other hand, the constant flow of blood around the leads may have a cooling effect. The impact of magnetic resonance scanning at 0.5 Tesla with respect to heating should be less than that of 1.5 Tesla, since the power absorption of a linear antenna is proportional to the square of the frequency of the radiofrequency field^[77]. Hence, the use of a 0.5 Tesla system reduces the power absorption considerably^[78]. Likewise, magnetic resonance sequences with a low specific absorption rate (e.g. gradient echo preferred over spin echo) may decrease heating. An accurate calculation of the heating problem could not be performed to date.

(b) *Alterations in pacing rate.* The radiofrequency field may induce an undesirable fast pacing rate^[3,44,45]. Two mechanisms have been proposed to explain rapid pacing: direct interference with pacemaker electronics may be possible, because pacing of up to 300 beats \cdot min⁻¹ synchronized to the radiofrequency pulses has been observed in several studies^[3,44]. In contrast, rapid pacing at the upper tracking limit in DDD pacemakers may be caused by induction of currents above the sensing threshold in the atrial lead and consequent triggering of ventricular stimulation^[45].

(c) *Pacemaker reprogramming or reset.* Changing the programmed parameters (phantom programming) or resetting a pacemaker during magnetic resonance investigation may also be a concern. The newer generation of pacemakers have 'security checks' at each sequence of pacemaker programming that may avoid changing parameters. In addition, the telemetry frequencies of most manufacturers (32–175 kHz) are out of the range of the frequencies of the radiofrequency and gradient fields.

3. Time-varying magnetic gradient fields

(a) *Induction voltage.* The contribution of the time-varying gradient fields to the total strength of the magnetic field is negligible; however, an effect may occur because these fields are rapidly applied and removed. The important factors are the rate with which the magnetic field changes and the length of time this changing field is applied. Budinger has calculated that an electrical current density of 1 μ A \cdot cm⁻² will be induced for a time-varying field of 1 Tesla \cdot s⁻¹^[66]. Even using today's gradient systems with a time-varying field up to 50 Tesla \cdot s⁻¹, the induced currents are likely to stay below the biological thresholds for cardiac fibrillation (in the range of 100 to 1000 μ A \cdot cm⁻²).

Using the induction law:

$$V = dB/dt \cdot \pi \cdot r^2$$

where V=induced voltage (volts), dB/dt=time rate of magnetic field change (Tesla \cdot s⁻¹), and πr^2 =area of wire loop (m²), the amount of induced voltage is proportional to the area of the wire loop. The gradient field can induce a critical voltage in unipolar leads. In bipolar leads this danger should be smaller.

Estimation of the induced voltage in unipolar leads. The area to be considered in the induction law has been reported to be between 200 to 500 cm²^[79]. A theoretical upper limit for the induced voltage is 20 V. Such a voltage during more than 0.1 ms will be enough to pace the heart. If the regular gradients are applied and the z-projection of the area is taken into account, the induced voltage will be reduced.

Estimation of the induced voltage in bipolar leads. If bipolar leads are used, the area between the two wires is much smaller. Stimulation of the heart should not

longer be possible. The induced voltage, however, may influence sensing.

(b) *Heating* The induced current may lead to local heating. However, assuming that the magnetic field of the gradients and the radiofrequency are orthogonal, the calculated heating effect of the gradient fields is much less compared to that caused by the radiofrequency field, and therefore, may be neglected.

(c) *Reed-switch closure.* It may be possible to open and reclose the reed switch in the main magnetic field by the gradient field. Theoretically, few reed switch orientations inside the homogeneous main magnetic field exist, in which case it does not close. In such a case, it is possible that the gradient fields could close and reopen the reed switch.

Magnetic resonance imaging of different body structures

A critical issue concerning safety of patients with pacemakers undergoing magnetic resonance imaging is the structure of the body to be scanned (e.g. thorax vs head or extremities). Most of the data collected in this regard has been of the brain. In such cases, the heart and pacemaker system are outside the isocentre of the magnetic resonance scanner, leading to a reduction in the radiofrequency field strength at the device. With the use of a transmit/receive head coil, the radiofrequency field at the area of the lead and the pacemaker will be further reduced. On the other hand, the pacemaker will be close to the portal of the magnetic resonance scanner where the highest magnetic forces on the device occur, as discussed previously^[58].

A similar reduction of the radiofrequency field interactions will occur with magnetic resonance imaging of the lower extremities. Recently, a specially designed, low field strength (0.2 Tesla magnetic resonance system, Artoscan, Lunar Corp., Madison, WI/Esate, Genoa, Italy) magnetic resonance system has become available for imaging of extremities^[80]. This magnetic resonance system has a small magnet bore where only the extremity will be placed (while the rest of the body remains outside), leading to very low radiofrequency, gradient and static magnetic fields at the thorax. Because of the unique design features of the extremity magnetic resonance system, it has been suggested that it may be possible to perform extremity magnetic resonance imaging in patients with cardiac pacemakers and implantable cardioverter defibrillators^[81].

The clinical use of magnetic resonance for cardiovascular imaging is much less common today than 'non-cardiac' magnetic resonance, which may be needed in a patient with an implanted pacemaker. The safety of using magnetic resonance imaging for cardiac evaluation in pacemaker recipients has not been tested on any significant scale to date.

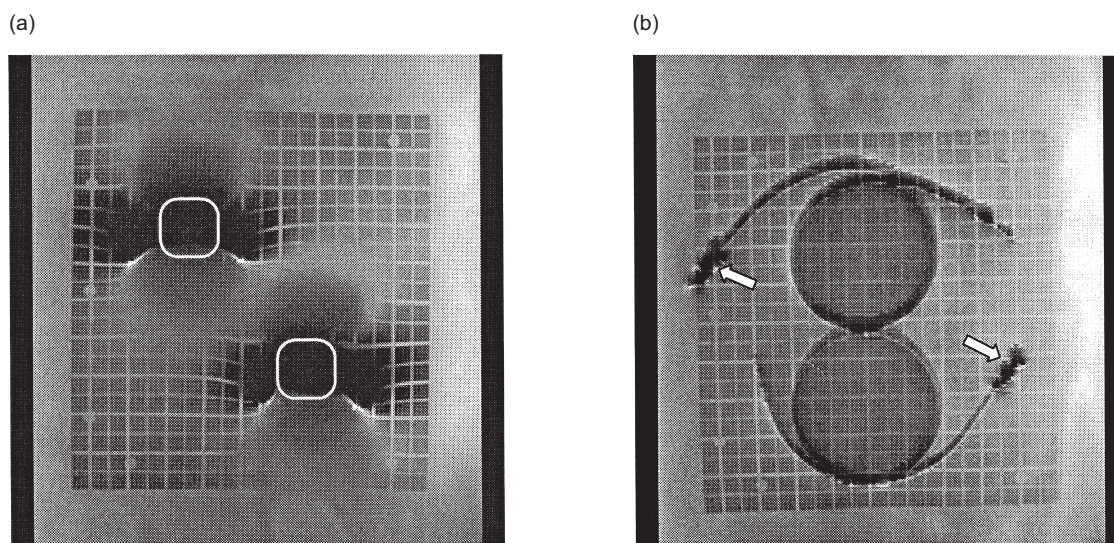


Figure 2 Magnetic resonance image artifacts caused by pacemakers and endocardial pacing leads: (a) Marked image disturbances around the pacemakers may be observed (pacemaker positions are shown by the rectangles). Bending and obliteration of contours within the imaging phantom (saline container) are noted. (b) The non-ferromagnetic leads show only mild artifacts appearing as small signal-void areas surrounded by or adjacent to high signal zones (arrows).

Image artifacts due to pacemaker system components

Image artifacts can be caused by the presence of pacemaker system components that are in or near the imaging field of view in the magnetic resonance environment. This is due to the static magnetic field of the implanted components that can perturb the relationship between the position and frequency essential for accurate image reconstruction. The pulse generators have a magnetic susceptibility that is significantly different from that of tissue, therefore marked distortion may result (Fig. 2). The amount of image artifact produced by the leads may only be of concern if imaging anatomy is in the immediate vicinity.

Additional problems with implantable cardioverter-defibrillators

Problems with implantable cardioverter-defibrillators in the magnetic resonance environment may be expected to be similar to those observed with pacemakers. However, implantable cardioverter-defibrillators use different and larger batteries that may cause higher magnetic forces^[58]. Despite a dramatic reduction in size and weight, new generation implantable cardioverter-defibrillators may still pose problems due to strong magnetic torque^[58]. The sensitivity of implantable cardioverter-defibrillators is normally much higher than that of pacemakers. Therefore, implantable cardioverter-defibrillator devices may falsely detect a ventricular tachyarrhythmia and subsequently deliver antitachycardia pacing, cardioversion or defibrillation

therapies, which can lead to an actual ventricular tachyarrhythmia. In addition, magnetic fields may prevent detection of ventricular tachycardia or fibrillation. The heating problem of implantable cardioverter-defibrillator leads can be expected to be the same as in pacemakers.

Future directions for magnetic resonance imaging compatible pacing

Any component of a pacing system, which is introduced inside the patient, must fulfil a number of stringent requirements. Instruments to be designed must satisfy general safety considerations unique to the magnetic resonance environment. In addition, they should distort the image as little as possible. It is possible that the magnetic resonance environment exerts less magnetic force and torque on newer pacemakers with smaller battery and size, as compared to the older pacemakers. Future studies are needed to investigate the safety of magnetic resonance imaging in patients with modern pacemakers.

Currently available pacing leads contain metal parts, which may cause hazardous side effects and distort image quality, therefore they are not suitable for use in the magnetic resonance environment. Non-ferromagnetic materials such as copper, with susceptibilities close to body tissue are desirable for use in pacing catheters. Copper is an excellent conductor and represents an almost ideal material for pacemaker leads. However, it is not biocompatible, a problem which can be solved by appropriate insulation. Several investigators have developed magnetic resonance imaging

compatible leads and reported the initial in vitro and in vivo results. Jerzewski *et al.* designed a pacing lead by modifying a commercially available 6F pacing catheter^[82]. The stainless steel shaft braiding was omitted and 90% platinum/10% iridium electrodes were used at the tip. The electric wiring was made of nearly pure copper. In phantom experiments, the lead was observed to cause only minor image artifacts, and in vivo, external pacing with this lead proved to be effective. Recently, Halperin *et al.* have also shown that diagnostic electrophysiological studies and catheter ablation under magnetic resonance guidance may be feasible using leads made from non-magnetic materials^[83]. However, the potential for tissue heating around the leads was not considered in these studies. Therefore, absolute compatibility of these leads remains of concern and requires further investigation.

Conclusions

Previous studies showed that magnetic resonance imaging in patients with implanted pacemakers may be hazardous. However, there are newer anecdotal reports of safe magnetic resonance imaging when this diagnostic technique was an absolute necessity. While it is possible to image a person safely, one must be aware of all variables to ascertain safety. The effects of magnetic resonance systems on the function of pacemakers and implantable cardioverter-defibrillators depend on various factors including the strength of the static magnetic field, the pulse sequence used, the anatomic region being imaged, and many other factors. In addition, each manufacturer's pacemakers and each pacing modality may behave differently. Therefore, knowledge of the behaviour of the specific pacemaker model in vitro may be helpful. More extensive in vivo testing using different models of pacemakers from various manufacturers is needed to identify device-specific susceptibility to the electromagnetic fields induced by magnetic resonance imaging.

At present, magnetic resonance imaging should be avoided in patients with pacemakers and implantable cardioverter-defibrillators until more information (e.g. results of large studies/registries) is known. If non-magnetic resonance imaging modalities are not adequate to make a diagnosis, after a careful risk-benefit assessment, magnetic resonance imaging at low field strengths (0.5 Tesla or less) with continuous monitoring (e.g. ECG/pulse oximetry) during the procedure may be considered but only in experienced centres. Body scans and magnetic resonance sequences with a high specific absorption rate should be avoided whenever possible. Changes in pacemaker programming (ideally OOO mode or programming off in non-pacemaker dependent patients or asynchronous mode in pacemaker dependent patients) may be performed prior to scanning. Introducing the patient slowly into the magnet with monitoring, and starting magnetic resonance imaging with graded

scanning sequences (single slice, low resolution) and then eventually progressing to conventional sequences should be considered. In any case, clinicians should keep in mind the potential for severe adverse effects with pacemaker use in magnetic resonance systems. Therefore, experienced cardiology assistance and full resuscitation facilities should be available during scanning. While a few patients (and physicians) may have been fortunate in the past, it does not mean that magnetic resonance scanning of pacemaker patients is safe, since a future patient may present with a pacemaker configuration that corresponds to one of the extreme scenarios. Even with the same pacemaker configuration and magnetic resonance settings, a safe procedure at one time does not guarantee a safe repeat examination.

With continuing progress in magnetic resonance imaging technology, higher gradient changes over time are being routinely applied to acquire thinner slices and more rapid images. Therefore, the feasibility and safety of magnetic resonance imaging in some studies may not be extrapolated to techniques using faster field strengths or other imaging protocols that are in use or yet to be developed. However, innovations in pacemaker and lead technology may enable magnetic resonance-compatible pacing systems in the foreseeable future.

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