An illustrative review to understand and manage metal-induced artifacts in musculoskeletal MRI: a primer and updates

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Abstract This article reviews and explains the basic physical principles of metal-induced MRI artifacts, describes simple ways to reduce them, and presents specific reduction solutions. Artifacts include signal loss, pile-up artifacts, geometric distortion, and failure of fat suppression. Their nature and origins are reviewed and explained through schematic representations that ease the understanding. Then, optimization of simple acquisition parameters is detailed. Lastly, dedicated sequences and options specifically developed to reduce metal artifacts (VAT, SEMAC, and MAVRIC) are explained.

Keywords MRI · Artifacts · Prosthesis

Introduction

Magnetic resonance imaging (MRI) studies the magnetic properties of tissues (proton density ρ , and relaxation times T_1 and T_2) by applying different calibrated magnetic fields: a static field B_0 , an alternative field B_1 , and three-directional gradient fields.

These fields, used to generate and receive signals from the tissues, must be as homogenous as possible, and every field default will inexorably induce, causing spatial encoding mismatches and artifacts. Metal interferes massively with a magnetic field and induces local inhomogeneity that is responsible for characteristic artifacts, particularly troublesome in musculoskeletal imaging [1–3]. Metallic implants such as prosthesis or rods, due to their high magnetic susceptibility, impede correct spatial coding, leading to distortion, formation of areas of signal loss, and of areas of abnormal high signal (“pile-up” artifacts) [4].

Although these artifacts are easily recognizable (Fig. 1), their origins are more complex to handle, and are generally poorly understood by common users. To reduce these artifacts, basic strategies are available on all routine MR units and can be easily set up. Besides those tips, manufacturers continue to develop and propose specific reduction options.

In this review, we aimed to help radiologists to understand the physical mechanism of metal-induced artifacts, to adapt MR protocols in order to reduce artifacts in daily practice, and to present advanced reduction strategies developed by manufacturers.

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Physics of artifacts

Magnetic susceptibility

Magnetic susceptibility is a dimensionless proportionality constant that indicates the degree of magnetization of a material (M) in response to an applied magnetic field (H). Magnetic susceptibility can be expressed in parts per million ($1 \text{ ppm} = 10^{-6}$). It is defined by the following relationship [5, 6]:

Fig. 1 Metal-induced MR artifacts in spinal instrumentation and knee prosthesis. Spine instrumentation in a 50-year-old man: lateral radiograph shows intersomatic cage and rods with screws (*arrows*) (**a**) while sagittal T2-weighted fast spin-echo MR images (**b** and **c**) show artifacts. Total hip

replacement in a 62-year-old woman: hip radiography reveals head and shaft landmarks (*stars*) (**d**) while coronal T1-weighted fast spin-echo MR images (**e** and **f**) demonstrate distortion and signal loss

$$X = \frac{M}{H} \quad (1)$$

X is the magnetic susceptibility of the material (M)

M is the magnetization of the material measured in amperes per meter (A/m)

H is the magnetic field strength measured in amperes per meter (A/m)

Depending on the value of X , three magnetic effects can be distinguished: dia-, para-, and ferro-magnetism (Table 1) [5].

- *Diamagnetism* ($X < 0$ ppm): is a property of all materials and always makes a weak contribution to the material's response to a magnetic field. For materials that show

some other form of magnetism (paramagnetism and ferromagnetism), the diamagnetic contribution may be regarded as negligible. Diamagnetic materials induce magnetic fields in the direction opposite to that of the applied magnetic field.

- *Paramagnetism* ($0 < X < 300$ ppm): some materials are attracted by an externally applied magnetic field and form internally induced magnetic fields in the direction of the applied magnetic field. Paramagnetic materials have a small, positive susceptibility to magnetic fields and are thus slightly attracted by a magnetic field. Paramagnetic properties are due to the presence of unpaired electrons and to the realignment of the electron paths caused by the external magnetic field. Unlike ferromagnets,

Table 1 Magnetic susceptibility (in ppm) and type of magnetism of human tissues and materials

Material	Magnetic susceptibility (ppm)	Magnetism
Water (soft tissue)	−9.05	Diamagnetism
Cortical bone	−8.86	
Liver	−8.76	
Blood (deoxygenated)	−7.9	
Liver (hemochromatosis)	0	Paramagnetism
Air	0.36	
Aluminium	20.7	
Titanium	182	
Gadolinium contrast agent (0.1 mol/l)	249	Ferromagnetism
Chromium-cobalt	900	

paramagnets do not retain any magnetization in the absence of an externally applied magnetic field because thermal motion randomizes spin orientations.

- *Ferromagnetism* ($X > 300$ ppm) is the basic mechanism by which certain materials form permanent magnets, or are attracted to magnets. Paramagnetism and diamagnetism forces are usually so weak that they can only be detected by sensitive instruments in a laboratory, whereas ferromagnetism typically creates strong forces that are responsible of the common attraction phenomena of magnetism encountered in everyday life. Table 1 reports various tissues and materials according to magnetic susceptibility and type of magnetism.

Artifact types and origins

MRI is based on the recording of signals that are frequency-encoded by gradient fields applied in three orthogonal directions [4–10]. The application of a gradient in a homogeneous B_0 field generates a linear variation in order to make spatial encoding achievable. Therefore, in a homogeneous magnetic field, the spatio-frequency relationship (the correspondence of spatial location and frequency encoding) is linear: to each spatial coordinate of the studied volume corresponds a single frequency coordinate. Metallic objects with high magnetic susceptibility placed within the magnetic field will generate spatial encoding errors that explain artifacts. Indeed, the spatio-frequency relationship then loses its linearity and a spatial encoding error offset takes place along the gradient's slope (Fig. 2) for protons exposed to higher frequencies in the center of the object. A signal accumulation appears in the offset front head (pile-up artifacts) (Fig. 2) associated with a loss of signal at the other extreme (Fig. 2) [4–7]. At the edges of the metallic object, the same encoding error offsets are observed, but in the opposite direction of the gradient slope,

due to lower magnetic field: as presented in Fig. 3, the magnetic field is locally more intense at the north and south poles and lower at the extreme borders of the object. The co-existence of areas of signal accumulation (dark stars) and signal loss (white stars) explains the typical “boomerang” appearance. As slice selection similarly required the application of a selection gradient, the above detailed mechanisms explain both in plane and through slice artifacts (Fig. 4) [10].

Basic optimization of sequence parameters

Read-out bandwidth increase

One simple way to reduce in-plane artifacts is to increase the gradient read-out bandwidth: in our own experience, significant improvements are seen with a read-out bandwidth higher than 500 Hz/pixel: (Fig. 5a). Nevertheless, if this method seems obvious on a simple object such as a cylinder, the result is less predictable with metallic hardware such as prosthesis (Fig. 5b) [3, 4, 9–11]. As it can be seen in the relationship below (Eq. 2), the signal-to-noise ratio (SNR) is inversely proportional to the square root of the read-out bandwidth [12]. In this formula, the K factor includes hardware-dependent parameters such as coil, pre-amp and noise power spectrum, field strength-dependent factors, pulse sequence-type parameters, and tissue-dependent parameters (spin density, T1, T2). For 2D SNR where K is a constant, FOV_x and FOV_y are the field-of-view in x and y directions, N_x and N_y are the number of frequency and phase encoding steps, Δ_z is the slice thickness, NEX is the number of signal averages, and BW is the read-out bandwidth.

$$SNR = K \cdot \left(\frac{FOV_x}{N_x} \cdot \frac{FOV_y}{N_y} \cdot \Delta_z \right) \sqrt{\frac{N_x \cdot N_y \cdot NEX}{BW}} \quad (2)$$

Slice thickness

Through-slice artifacts can be reduced by increasing the slice selection bandwidth [5, 11, 13–16]. Consequently, this will reduce the number of excited protons and therefore the slice thickness, with decrease of the signal-to-noise ratio. To maintain a sufficient SNR, the use of thin slices will therefore require increasing the number of averages (NEX) with inherent longer examination time. A way to increase the slice selection bandwidth without impacting slice thickness would consist of enlarging the RF bandwidth, however, modification of the RF bandwidth is in general not user-accessible.

Spin echo versus gradient echo sequences

It is commonly thought that gradient-echo sequences (GE) increase image distortion due to the presence of metal. In fact,

Fig. 2 Origins of metal-induced MR artifacts. **a** The spatio-frequential relationship (the correspondence of spatial location and frequency encoding) is linear in a homogeneous magnetic field. The relationship lost its linearity when a metallic object (limits on *dotted yellow line*) is present within the magnetic field: protons (in *red color*) exposed to non-expected frequencies in the center of the object present encoding error offset (gray arrow) along the gradient's slope direction. As a consequence, signal accumulates in the offset front head (= pile-up artifacts) (*dark stars*) associated with a loss of signal at the other extreme (*white stars*). **b** At the edges of the metallic object, the same encoding error offsets are created, but in the opposite direction of the gradient slope. Areas of signal accumulation (*dark stars*) and signal loss (*white stars*) explain the characteristic "boomerang" appearance

Fig. 3 The "boomerang" aspect. **a** A cylinder (white arrow) fulfilled with gadolinium contrast agent at 0.1 mol/l (249 ppm) is embedded in gelatinous water (black arrow) and placed in a MR field (B_0). At this concentration, gadolinium provides no signal and simulates metal-induced artifacts. **b** Coronal (xz plane) spin-echo image of excited slice (* in **a**) shows in-plane artifacts, while dotted circle represents the real limits of the cylinder, a typical "boomerang" appearance is seen with

pile-up hyperintensities localized in the sharpest bends (black stars) and lack of signal seen near the less pronounced curves of the boomerang (white stars). **c** Schematic representation of the magnetic dipole field outside the cylinder (grey arrowheads): the resulting forces (red arrows) are along B_0 at the center, and opposite to B_0 at the edge of the object, explain the boomerang shape

Fig. 4 Experimental demonstration of in-plane and through-slice artifacts. **a** Experimental phantom containing a hip prosthesis (CoCr29Mo) immersed in gelatinous water (8 g agar-agar/l_{water}). **b** Coronal T2w SE MR image through the hip prosthesis (1.5-T MRI system (Magnetom Aera, Siemens Medical Solutions, Erlangen, Germany; slice thickness 3 mm) shows in-plane artifacts with pile-up (dark stars) and signal loss (white stars). **c** Sagittal reconstructions following dotted line show through-plane artifacts

the type of sequence does not impact directly image distortion, which is a consequence of the imaging mechanism. The differences between GE and SE sequences are faster spin–spin dephasing induced by the presence of metal: GE (under T2 * decay curve) are more sensitive to local field heterogeneity than spin-echoes (SE) (under a T2 decay curve). Thus, GE

sequences are very sensitive to field heterogeneity and generate a massive loss of signal, well beyond the limits of metal-induced distortions observed with SE sequences (Fig. 6). Prosthesis induces two kinds of artifacts: distortions (boomerang shape with signal pile-up and signal loss) and signal decrease due to premature spin–spin dephasing.

Fig. 5 Bandwidth influence. **a** Schematic illustration of readout bandwidth increase: increasing bandwidth reduces signal misregistration (white arrow). **b** Corresponding bandwidth changes on

CrCo prosthesis phantom: the “boomerang” appearance decreases as bandwidth increases (*BW* bandwidth)

Fig. 6 Comparison of coronal T1w GRE image (a) and coronal T1w SE image (b) of a 50-year-old male patient with left total hip replacement (Cr-Co). Spatial distortions are present on the two images but with major signal decrease (T2*) on gradient-echo sequence

Therefore, SE sequences have to be preferred to GE when exploring metallic hardware. Moreover, fast SE techniques that employ long echo-train-length avoid premature signal decrease and are even more recommended [6, 13, 15, 16].

Fat suppression

There are, roughly speaking, two main groups of sequence to remove fat signal on MR. The first is based on the chemical shift phenomenon that occurs between protons (H) of water and fat (208 Hz at 1.5 Tesla): pulse-selective saturation techniques (Fat Sat, Spir) and phase shifts techniques (water exCitation, ProSet, Dixon, etc.). Metal affects preferentially this group of sequences exploiting chemical shift (Fig. 7). Nevertheless, among them, multipoint sequences (Dixon, Ideal) provide better results than conventional fat-sat techniques [6, 15, 17, 18] (Fig. 8). Heterogeneity may be partially limiting, but not totally by applying a manual shimming.

The second group of techniques is based on inversion recovery, the most commonly used begin short tau inversion recovery (STIR). STIR is less sensitive to magnetic field inhomogeneity and provides a more homogeneous fat suppression than techniques based on chemical shift (Figs. 7 and 8). Unfortunately, due to short inversion time (150–200 ms at 1.5 T), STIR only allows T2 (long TE) or Rho (short TE)-weighted contrasts [18, 19].

Influence of magnetic field and metal materials

Any non-metallic object placed inside the magnet induces a low variation of B_0 due to diamagnetic (tissues) or

paramagnetic properties (Table 1). This moderate increase in heterogeneity, on the order of 100 Hz, which can be compensated with electromagnetic shim gradient but metal causes heterogeneity of several tens of kHz that cannot be fully compensated with shimming. Moreover, the stronger the magnetic field, the higher the magnetization of a metal object [4, 5, 20–22] and the higher the magnetic field distortion will be. Increasing the magnetic field is therefore detrimental for prosthesis imaging: it is recommended to preferentially scan patients with metallic implants on a 1.5-T rather than a 3-T magnet.

The composition of the materials will directly impact the level of images distortion: metal-induced artifacts increase with the magnetic susceptibility of materials [4, 5, 20–22] (Table 1). For instance, titanium implants (magnetic susceptibility of 182 ppm) generate less artifacts than chromium-cobalt (900 ppm) prostheses. In practice, users should know that it is easier to reduce metal artifacts from titanium materials than chromium-cobalt [23, 24].

Dedicated MR sequences new developments

Specific sequences are being developed to reduce metal artifacts; some of them are currently available at MR clinical units. The three main solutions offered by manufacturers are view angle tilting (VAT), multi-acquisition variable-resonance image combination (MAVRIC), and slice encoding metal artifact correction (SEMAC). We only aimed to describe below the basic principles of these methods; complete technical reviews are available in the literature and are beyond the scope of this article.

Fig. 7 Experimental comparison of conventional (frequency selective) fat saturation and STIR techniques. Coronal fat-saturation (a) and STIR (b) MR images of a phantom composed of a femoral metallic shaft (*), a specimen of pure gadolinium chelates (black arrow) and a sample of fat (white arrow). Frequency selective fat saturation leads to imperfect fat suppression, but maintains signal-to-noise ratio (SNR), whereas STIR provides uniform fat suppression but decreases SNR.

Fig. 8 In vivo comparison of fat suppression techniques coronal STIR (a), selective frequency fat-saturated T2w (b), and Dixon T2 (c) images of a 47-year-old female patient with left total hip replacement confirm the

more homogenous fat suppression of STIR sequence. Dixon T2 sequence is an interesting alternative, thanks to higher SNR

View angle tilting

During the signal readout, a constant spatial encoding error exists for a given object, due to signal heterogeneity, and it can be estimated by a preliminary field mapping. The VAT technique applies, simultaneously, to the read-out gradient, a VAT gradient on the slice selection axis that tilts the signal sampling. The slice is effectively viewed at an angle $[\tan^{-1}(G_z/G_x)]$. The frequency shift due to field heterogeneity during readout is therefore compensated [23]. This option limits in-plane distortions but remains ineffective in reducing through-plane artifacts. However, if VAT limits in-plane artifacts, it generates blurring on interfaces along the readout axis (Fig. 8) that can be reduce in implementing a large read-out gradient bandwidth (Fig. 9) [3][25–28].

Multi-acquisition variable-resonance image combination

When a metallic object is present within a magnetic field, protons of the same plane spin at different frequencies around the Larmor frequency (Fig. 10). This spectral dispersion of spins near metal implants is well beyond (150 ppm) levels

typically encountered in magnetic resonance studies (3 ppm) and this inhomogeneity will distort the selected slab from its intended geometry. The MAVRIC technique uses a spectral approach: with judicious choices of RF spectral bands and increments of subimage frequency offsets, a complete composite image can be obtained via either maximum intensity projection or a sum of squares computation in the image domain (Fig. 10) [29]. The MAVRIC method successfully images in extreme off-resonance conditions by breaking abnormally broad spectral distributions into discrete and independently imaged frequency bins [29, 30]. However, acquisition-time increases with the number of subimages used for the combination step. Currently, this method takes more than 10 min, and can be shortened by optimizing the number of phase encoding steps and the RF pulse bandwidth.

Slice encoding for metal artifact correction (SEMAC)

The SEMAC technique aims to correct through-plane artifacts by sampling slice thickness with a phase encoding gradient. This allows protons location along the slice-selecting axis using phase coordinates (Fig. 11) [21, 31].

Fig. 9 VAT sequence. **a** Conventional spin-echo sequence: coronal spin-echo T1w MR image of a 57-year-old female patient with a total hip replacement (Cr-Co) shows in-plane artifacts. **b** VAT method applies a gradient on the slice-selecting axis during readout, with amplitude equal

to that of the slice-selecting gradient. When viewed at this angle, shifts in the slice-selection plane compensate for shifts during readout, so that in-plane shifts are registered again in the images

a.

b.

Fig. 10 MAVRIC sequence. **a** With judicious choices of RF spectral bands and increments of subimage frequency offsets, a complete composite image can be obtained via either maximum intensity projection or a sum-of-squares computation in the image domain.

c.

Coronal proton density-weighted 3-T MR images of a 37-year-old male with a right hip without (**b**) and with (**c**) MAVRIC option where artifacts are notably reduced

Protons with the same phase are combined to obtain a non-distorted slice. To further limit in-plane distortions, the SEMAC technique is in general combined with a VAT option [32–35] (Fig. 11). The use of this complementary phase encoding gradient increases acquisition times up to 10 min.

Perspectives

The main trends are currently to optimize the above-presented technique [33, 35] notably in the hope of achieving a better compromise between acquisition time and artifact reduction. Recently, a hybrid acquisition sequence has been developed by combining SEMAC and MAVRIC techniques (“MAVRIC SL”) [36]. This sequence aims to reduce metal artifact even

with a field of higher strength than 3 T [37]. Compared with standard MRI, MAVRIC SL significantly improved the detection of clinically relevant findings, such as periprosthetic osteolysis [19, 37].

In order to avoid a premature loss of signal due to magnetic susceptibility, ultra-short echo-time sequences (UTE) could be interesting. These sequences were initially developed to study short-T2 tissues such as tendon, cortical bone, meniscus, and cartilage, but studies have highlighted their ability to reduce magnetic susceptibility artifacts [37]. Obtaining an instantaneous TE of about 0.1 ms is possible using a k-space radial-ray filling. In this case, the free induction decay signal (FID) is directly placed in the center of the k-space. Combining this approach with MAVRIC allows the use of gradient echo sequences [38].

Fig. 11 SEMAC sequence. **a** Diagram of conventional MR sequence demonstrates through-plane distortions (dotted circle) observed in a patient with cobalt chromium total knee prosthesis. **b** SEMAC sequence adds a z-phase encoding that resolves the distorted excitation

profiles of encoded slices. Compared to standard FSE sequence (**c**), the VAT-SEMAC combination reduces both in-plane and trough-slice artifacts (**d**). (MR images with kind permission of Siemens Healthcare)

Earlier studies have demonstrated that low-field imaging (from 0.1 to 0.5 Tesla) can reduce metal-induced artifacts, especially during interventional procedures requiring the use of needles and other surgical materials [20, 39]. Unfortunately, this type of equipment is very rarely available, but it would have been interesting to test low-field magnets with current sequences, options, and coil technologies (multichannel coils, optical-fiber transmission, etc.)

Conclusions

Understanding the occurrence of these artifacts remains essential for MRI users, both during sequence programming and interpretation steps. Metal-induced artifacts can be limited with simple solutions (Table 2), accessible to users (e.g., increasing read-out bandwidth, preferring STIR over fat-sat techniques, etc.), thanks to the recent availability of specific

Table 2 Common current tips and options to reduce metal-induced artifacts

Tips and options	Consequences	Limits
1 Increase the readout bandwidth	In-plane distortion (pile-up and signal loss) reduction	SNR decrease => acquisition time increase (to reach SNR)
2 Reduce the slice thickness	Trough-slice distortion reduction	
3 Prefer SE, FSE rather than GE	Reduce signal loss from dephasing	No access to dynamic contrast enhancement techniques (rare)
4 Prefer STIR or Dixon (less effective) techniques for fat suppression	Efficient and homogeneous fat suppression	SNR decrease
5 Use 1.5-T rather than 3-T magnet	Reduce all metal-induced artifacts	Radiologic department organization
6 Use VAT option	In-plane distortion (pile-up and signal loss) reduction	Not available on all MR scanners
7 Use VAT-SEMAC combination	In-plane distortion (VAT) and Trough-slice distortion reduction (SEMAC)	Not available on all MR scanners, acquisition time increase
8 Use MAVRIC (-SL)	In-plane distortion and Trough-slice distortion reduction	

options and sequences [16]. Nowadays spine imaging with titanium implants is achievable without troublesome artifacts [23, 24, 28, 40–42], while MR imaging of chromium-cobalt prosthesis could remain challenging. Nevertheless, optimization of standard sequences as detailed above and availability of manufacturer-licensed dedicated techniques might help radiologists to overcome metal-induced artifacts in clinical practice [13, 14, 17, 19, 38].

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Compliance with ethical standards

Conflict of interest The author(s) declare that they have no competing interests.

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