Imaging the Knee in the Setting of Metal Hardware

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KEYWORDS

• Metal artifact reduction sequences • Metal imaging • Metal hardware • View angle tilting • Slice encoding for metal artifact correction • Multiacquisition variable-resonance image combination • Ultrashort echo time-MAVRIC • Knee

KEY POINTS

- Magnetic resonance (MR) imaging will be increasingly used to evaluate the knee in the setting of metal hardware as techniques continue to improve.
- Protocols that are optimized for the evaluation of the painful knee without hardware are suboptimal in the presence of metal.
- Knowledge of the basic principles behind MR imaging metal artifact reduction sequences will allow the radiologist to tailor examinations to match the degree of metal artifact.
- MR imaging metal artifact reduction sequences can show diseases that are occult on other imaging modalities.

INTRODUCTION

Magnetic resonance (MR) imaging is an established modality for the evaluation of musculoskeletal structures, including the knee. This situation is because of the superior soft tissue contrast of MR imaging compared with radiography or computed tomography (CT). In addition, MR imaging can frequently detect osseous abnormalities that may not be visible with other imaging modalities.[1,2] Over the past several years, clinical guidelines for the use of knee MR imaging have broadened.[3,4] Furthermore, metallic implants are increasingly used, particularly for total knee replacement (TKR).[5–7] In part because of prosthetic device wear over time, the number of revision surgeries has also increased.[8] Coupled with continuing improvements in techniques designed to minimize artifacts, it is not uncommon for the radiologist to encounter a patient presenting for MR imaging of the knee in the setting of metal hardware. This article reviews the general principles behind the effects of metal on MR imaging, various MR techniques that are available or have been recently described in the literature, and diseases that can be encountered on MR imaging of patients related to knee arthroplasty, hardware after internal fixation, or hardware used for soft tissue fixation.

Conflicts of interest: none.

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Magn Reson Imaging Clin N Am 22 (2014) 765–786
http://dx.doi.org/10.1016/j.mric.2014.07.009
1064-9689/14/$ – see front matter Published by Elsevier Inc.

GENERAL PRINCIPLES

Magnetic Susceptibility

Magnetic susceptibility, expressed in parts per million (ppm), is a measure of the tendency of a material to interact with and distort the main magnetic field (B_0).[9] Negative susceptibility values indicate magnetism that opposes B_0, or diamagnetism, and positive susceptibility values indicate the tendency to increase B_0, including paramagnetism or ferromagnetism. In clinical MR imaging, the material of interest is typically water, and therefore, the scanner hardware is tuned to the resonance frequency of protons attached to water molecules, which is 64 MHz at 1.5 T (termed the on-resonance frequency).[10] Water has a magnetic susceptibly value of –9 ppm[9] and materials of different magnetic properties than water create perturbations in the local field, causing precession frequency to increase or decrease (off-resonance frequencies). Susceptibility differences can be small and in the order of a few ppm, such as the differences between biological tissues (water, bone, and fat).[9,11] In contrast, implanted metals show marked susceptibility differences to human tissue in the order of hundreds to thousands of ppm. Implants containing ferromagnetic materials such as nonmagnetic (MR-safe) stainless steel alloys can show a positive shift of 6700 ppm, and cobalt-chromium alloys can show a shift of 1370 ppm.[12,13] Paramagnetic materials show lower susceptibility, such as titanium (182 ppm) and zirconium (109 ppm).[9] **Table 1** provides a list of approximate magnetic susceptibilities of materials commonly encountered during imaging of orthopedic implants about the knee. The susceptibility is heavily dependent on the composition of the implant, and this information is often not easily ascertained. Radiographically similar appearing prostheses can have vastly different composition and susceptibility, such as cobalt-chromium alloys and oxidized zirconium femoral components.[14] In addition, the terms cobalt-chromium alloy, cobalt-chromium-molybdenum alloy, and titanium alloy do not specify the precise composition of a prosthesis. For instance, cobalt-based and cobalt-chromium have both been used to describe implants containing various amounts of cobalt, chromium, manganese, nickel, molybdenum, iron, carbon, and silicon.[13,15] Minute differences in implant composition beyond what is typically controllable through the manufacturing process can cause differences in measurable magnetic susceptibility.[9]

Table 1
Comparison of magnetic susceptibilities

Material	Approximate Magnetic Susceptibility (ppm)	Reference
Zirconium oxide	−8.3	9
Water	−9.1	9,126
Bone (cortical)	−8.9 to −12.8	9,126,127
Fat	−7.8 to −12	9,128
Polyethylene	~0.0[a]	129
Air	0.4	9
Titanium alloys	14.6	130
Zirconium	109	9
Tantalum	178	9
Titanium	182	9
Cobalt-chromium alloys	900 to 1370	12,13
Stainless steel alloys (nonmagnetic, MR-safe)	3520 to 6700	9

Values represent a guide because susceptibilities differ depending on the precise composition of metal alloys.
[a] Value converted from originally reported value of mass susceptibility.

Imaging Artifacts in the Setting of Metal Hardware

When a clinical magnet is appropriately shimmed, the B_0 field is typically homogeneous over a 45-cm diameter to at least 1.5 ppm.[12,16] In the setting of orthopedic implants, the static magnetic field becomes inhomogeneous, typically because of magnetic susceptibility but also influenced by implant size, shape, and orientation. These inhomogeneities cause 3 main types of artifacts on conventional, clinical MR images: T2* dephasing, displacement artifacts, and failure of fat suppression (**Fig. 1**A).[17] These artifacts are described in further detail later, and conventional techniques used to decrease these artifacts are summarized in **Table 2**.

T2 dephasing*

T2* dephasing occurs because of varying rates of precession inside a voxel (see **Fig. 1**B). Dephasing effects can be minimized through the use of refocusing pulses, such as those used with spin-echo or fast spin-echo (FSE) imaging. T2* dephasing can also be minimized with decreasing effective echo time (TE),[18] such as with ultrashort TE (UTE) techniques (**Fig. 2**).

Displacement artifacts

Displacement artifacts arise because of frequency variations in both the slice selection (through-plane, z-axis) and readout (in-plane, x-y–axis) directions. In clinical MR imaging, spatial variation