K E Y  F A C T S

- Clinical magnetic resonance imaging (MRI) measures the spatial distribution of protons in the body.
- Gradient coils are used to provide spatial information. The changing gradients are associated with noise produced during imaging.
- Relaxation times T1, T2, and T2* are important tissue characteristics for imaging.
- Low-field magnets have lower signal to noise ratio (SNR); longer scan times, making patient motion a potential problem; decreased resolution; decreased sensitivity to old blood and calcified lesions; lower gadolinium enhancement; and difficulty in spectral fat suppression.
- Gadolinium contrast medium is often used combined with fat-suppressed T1-weighted imaging to increase contrast between enhanced tissue and surrounding tissue.
- Artifacts are numerous in MRI and can lead to erroneous diagnosis if not understood or eliminated. The magic angle phenomenon produces increased signal in portions of tendons oriented at approximately 55 degrees to the main magnetic field. These areas will appear bright on short TE sequences (e.g., T1) and can lead to an erroneous diagnosis of degeneration or tear.
- Patient safety is paramount and can be maximized by thorough prescreening and other safety measures.

GENERAL PRINCIPLES

Imaging Principles

Magnetic resonance imaging (MRI) measures the spatial distribution of specific nuclear spins (usually those of protons) in the body. Electric signals from the spins are measured using precessional motion of the proton spins after they are excited by radio frequency (RF) pulses irradiated in a static magnetic field.

Precession refers to a change in the direction of the axis of a rotating object.

The phenomenon in which the nuclear spins generate or emit electric signals of a specific frequency (Larmor frequency) in a static magnetic field is called nuclear magnetic resonance (NMR).

The electric signal (NMR signal) itself carries no spatial information. The spatial information necessary to generate an image is given by magnetic field gradients that are generated by gradient coils. Because they are driven by pulsed electric currents in a strong magnetic field, the coils receive a repetitive strong force, and a loud sound is produced during the MRI scan.

NMR Signal: Free Induction Decay and Spin Echo

Two kinds of NMR signal are generally used in MRI: free induction decay (FID) and spin echo. FID is elicited by a single RF pulse (e.g., 90 degrees) (Figure 3 1). The FID decays with the time constant T2*. The decay of the NMR signal can be recovered by applying a second RF pulse, called a 180 degree pulse. At a specific time (TE/2) after the second RF pulse, the spin echo signal is observed. The intensity of the spin echo signal decays with the time constant T2.

Relaxation Times

The relaxation times of proton spins are the most important parameters in MRI. Three kinds of relaxation times are generally used: T1, T2, and T2*. T1 or longitudinal relaxation time is the time by which nuclear spins return to thermal equilibrium (initial state) after irradiation by an RF pulse(s). T1 is generally used to visualize the degree of saturation or suppression of NMR signal or image intensity because tissues with longer T1 give suppressed NMR signal in T1 weighted sequences, as described below.

Tissues with long T1 are dark on T1 weighted images.

T2, or transverse relaxation time, describes the lifetime of spin echo signal, as shown in Figure 3 1. T2 is generally used to distinguish pathologic tissues from normal tissues, because proton spins of pathologic tissues usually have longer T2.

T2*, describes the decay rate of FID signal, as shown in Figure 3 1. Although T1 and T2 depend on NMR frequency (magnetic field strength), typical T1 and T2 values of water content in normal tissues are roughly 1000 ms and 50 ms, respectively. T1 and T2 of pathologic tissues usually become longer than those of normal tissues, making MRI very useful in diagnosis of various diseases.

PULSE SEQUENCES AND IMAGE CONTRAST

The contrast of magnetic resonance (MR) images is determined by combinations of relaxation times and pulse sequences. The pulse sequences are divided into two major categories: spin echo and gradient echo sequences.
Spin Echo Sequence

Spin echo (SE) sequences utilize spin echo signal and produce spin echo images, in which image intensity, $I(x,y)$, is expressed as:

$$I(x,y) = k \rho(x,y) \{1 - \exp(-TR/T1(x,y))\} \exp(-TE/T2(x,y)),$$

where $k$, $\rho(x,y)$, $TR$, and $TE$ are a constant, proton density, repetition time of the pulse sequence, and spin echo time, respectively (see Figure 3.1). This equation shows that spin echo images are proton density images modified by the ratios of $TR/T1$ and $TE/T2$. Although $\rho(x,y)$, $T1(x,y)$, and $T2(x,y)$ can be computed by combinations of several spin echo images, three practically useful images are widely used: T1 weighted images (T1WI), T2 weighted images (T2WI), and proton density weighted images (PDWI). Because $T1$ and $T2$ of water content normal tissues are roughly 1000 ms and 50 ms, respectively, the pulse sequences shown in Table 3.1 are used for T1WI, T2WI, and PDWI acquisition. Typical and instructive T1WI, T2WI, and PDWI of a chicken egg are shown in Figure 3.2. The yolk and white of the egg are visualized with various image contrasts because they have different $T1$, $T2$, and proton densities.

Because imaging appearances vary depending on the imaging parameters used, attention should be directed to the parameters listed on the image itself. The TR and TE, among other parameters, are indicated adjacent to the MR image (Figure 3.3).

Table 3.2 explains the appearance of common tissues on SE imaging. The signal may change depending on the combination of the TR, TE, and inversion time (TI) used for obtaining the sequence.

In actual clinical settings, SE sequences are usually performed as fast spin echo (FSE) sequences to shorten the imaging time by using multiple spin echoes. Basic image contrasts are similar to those obtained by the traditional or conventional spin echo sequences described above. However, fat tissue is of higher signal on FSE $T2$ weighted imaging than on spin echo $T2$ weighted imaging.

Gradient Echo Sequence

Gradient echo (GRE) sequences utilize FID signal and are characterized by sequence parameters TR, TE, and FA (flip angle). The FA is the angle by which nuclear spins are rotated from the direction of the static magnetic field. However, the image contrasts of GRE images are not determined solely by the sequence parameters but are strongly affected by the pulse sequence design.

Regarding the pulse sequence design, GRE sequences are categorized into three groups: incoherent acquisition sequence (e.g., FLASH, SPGR), partially coherent acquisition sequence (e.g., GRASS, FISP), and coherent acquisition sequence (e.g., TrueFISP, SSFP). MRI manufacturers use different names for their own GRE sequences. However, for simplicity, we use the terms FLASH, GRASS, and TrueFISP to represent the three acquisition methods described above. FLASH is mainly used as a $T1$ weighted sequence. GRASS is used as a $T2^*$ weighted or $T1$ weighted sequence. FLASH and GRASS are faster than the spin echo $T1$ weighted sequence, but the image contrasts are slightly different (Figure 3.4). TrueFISP is a very fast sequence and is mainly used for visualization of fluids such as blood.

VARIOUS TECHNIQUES

Multislice and Three-Dimensional Imaging

As shown in Table 3.1, TR is usually much longer than TE or $T2$ because $T1$ recovery of proton spins takes longer than $T2$ decay. To shorten the scan time for an imaging volume, the pulse sequences are designed to excite multiple planes successively during the repetition time. This technique is called multislice imaging.

Three dimensional (3D) imaging is another solution for shortening the scan time. 3D imaging is usually performed with short TR gradient echo sequences because long TR requires a long acquisition time. 3D imaging has several advantages over multislice imaging, including thin slices, no slice gap, and isotropic voxel.
A voxel is a volume element that forms a small portion of the image.

**Fat Suppression**

Fat is visualized as a high intensity signal on T1WI, PDWI, and even in T2WI. Fat signal therefore frequently conceals slight contrast differences between water content tissues near the fatty tissues. In this situation, fat suppression techniques are used, either by utilizing Larmor frequency difference (~220 Hz at 1.5T) between water and fat signals or by using T1 difference between fat and other tissues. The former technique is the “chemical shift selective” method and is used in high field MRI machines. The latter technique is one of the inversion recovery methods in which image acquisition is performed when the fat signal becomes zero after the fat proton spins are inverted from the direction of the static magnetic field. This technique is called short tau inversion recovery (STIR) and is used mainly in low field MRI machines.

**FIGURE 3.3.** MRI appearances of the knee on (A) coronal T1 weighted image, (B) coronal STIR image, and (C) axial T2 weighted image. MR parameters are indicated adjacent to the MR image. Note that the appearance of fat is bright on the T1 weighted image and dark on the STIR image. Fluid signal is bright on the STIR sequence, making this sequence useful for identifying conditions with increased fluid such as tumor or edema. On the T2 weighted fast spin echo sequence (C) the joint fluid is bright. In fast spin echo images, fat is also bright, which sometimes limits the distinction between these two tissues.
STIR images are clinically useful because the fat signal is suppressed (black) and fluid (e.g., edema) becomes easier to identify (white).

**MRI SYSTEM: FIELD STRENGTH AND MAGNET CONFIGURATION**

**Field Strength**

A variety of magnetic field strengths from 0.2T to 3.0T are used for clinical MRI of arthritis and bone lesions.

- **T** (Tesla) is a measure of magnetic field strength; 1 Tesla is approximately 20,000 times the Earth’s magnetic force.

The major advantage of high field MRI is the increase in SNR, which improves spatial and/or temporal resolution and reduces scan time while preserving image quality. On the other hand, a low field magnet allows a variety of configurations, increases the patient’s comfort by using an open magnet and dedicated MRI system for extremities, and makes possible isocenter imaging even for off center anatomic sites. The open MRI magnets usually have field strength in the range of 0.2T to 0.7T. Disadvantages of low field magnets are lower SNR, increased acquisition time, decreased resolution, decreased sensitivity to old blood and calcified lesions, lower gadolinium enhancement, and difficulty of spectral fat suppression (Table 3.3).

**Closed Magnet MRI**

The closed magnet configuration refers to the original tube shape of most MRI scanners (Figure 3.5, A). All high field superconducting MRI scanners are of the closed configuration type. Currently, 90% of MRI machines are traditional closed MRIs. As mentioned previously, high field MRIs produce higher SNR and superior quality imaging. Therefore MRI with a closed magnet would be the first choice in many cases. Problems arise with claustrophobic patients and overweight (over 350 lb) patients, for whom open magnet or extremity MRI would be the next choice.

**TABLE 3-2. Examples of Tissue Appearance on Common Imaging Sequences**

<table>
<thead>
<tr>
<th>T1 weighted</th>
<th>T2 weighted</th>
<th>STIR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone cortex, calcification</td>
<td>Very low signal</td>
<td>Very low signal</td>
</tr>
<tr>
<td>Bone marrow</td>
<td>High signal</td>
<td>High signal</td>
</tr>
<tr>
<td>Cartilage</td>
<td>Iso signal</td>
<td>Slightly low signal</td>
</tr>
<tr>
<td>Joint effusion</td>
<td>Iso signal</td>
<td>High signal</td>
</tr>
<tr>
<td>Acute hemorrhage</td>
<td>Low to iso signal</td>
<td>Low to iso signal</td>
</tr>
<tr>
<td>Subacute hemorrhage</td>
<td>High signal</td>
<td>High signal</td>
</tr>
<tr>
<td>Hemosiderin</td>
<td>Very low signal</td>
<td>Very low signal</td>
</tr>
<tr>
<td>Fat</td>
<td>High signal</td>
<td>High signal if FSE</td>
</tr>
</tbody>
</table>

Comparison is made to the signal of muscle. FSE, Fast spin echo.
Extremity and Open Magnet MRI

Despite the previous considerations, the use of dedicated extremity and open magnet MRI for the evaluation of arthritis and other musculoskeletal pathologic conditions has several advantages over the use of whole body MRI.

Dedicated extremity MRI requires less space than a whole body MRI system, is less expensive, offers greater patient comfort, avoids claustrophobia, and minimizes potential biohazards associated with the presence of metal in or on the patient by placing only the limb of interest in the magnet bore (Figure 3.5). It has been reported that 64% of patients with arthritis of the hand and wrist preferred 0.2T extremity MRI to 1.5T high field MRI because it was more comfortable, less claustrophobic, and quieter.1 Recently, low field dedicated extremity MRI has been used for the evaluation of rheumatoid arthritis (RA) of the hand and wrist. Low field MRI performed well for cross sectional grading of bone erosions, joint space narrowing, and synovitis in RA. Even low field MRI detected approximately twice as many erosions as radiography.2 The volume of synovial membrane determined with extremity MRI was significantly correlated with and not significantly different from that determined with high field MRI with gadolinium injection.1 Therefore low field dedicated extremity MRI may be useful for the evaluation of RA.

Open MRI allows easy access to patients (Figure 3.5, C), making it well suited for patients who are very large, severely anxious, claustrophobic, or in need of constant support during an exam (e.g., children). In addition, open MRI makes it possible to perform interventional MRI, which requires open magnet technology and real time imaging. There are two types of open magnet MRI: vertically and horizontally open magnets. The vertically open MRI system (Signa SP, GE Medical Systems, Milwaukee) allows radiologists and surgeons direct vertical access to the patient through an opening, with near real time imaging. This system is a whole body scanner operating with a 0.5T superconducting magnet with actively shielded gradients. The flexible RF coil with sterile covers is placed on the patient during an intervention. MRI has several advantages over other equipment for interventional guidance. MRI does not expose patients, radiologists, or surgeons to ionizing radiation. Excellent soft tissue contrast aids in selecting a biopsy site with multiplanar imaging capability.

**MRI ARTIFACT**

MRI produces several specific artifacts; familiarity with them is essential for a correct diagnosis.

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### TABLE 3-3. Advantages and Disadvantages of Low Field MRI

<table>
<thead>
<tr>
<th>Advantage</th>
<th>Disadvantage</th>
</tr>
</thead>
<tbody>
<tr>
<td>Length of exam</td>
<td>Longer scan times</td>
</tr>
<tr>
<td>Patient comfort</td>
<td>“Open” magnets may be more comfortable</td>
</tr>
<tr>
<td>Large or claustophbic patients</td>
<td>May be scanned on open or extremity scanners</td>
</tr>
<tr>
<td>Signal to noise ratio</td>
<td>Lower</td>
</tr>
<tr>
<td>Resolution</td>
<td>Decreased</td>
</tr>
<tr>
<td>Fat suppression</td>
<td>Limited</td>
</tr>
<tr>
<td>Gadolinium</td>
<td>May need higher dose</td>
</tr>
<tr>
<td>Cost of unit</td>
<td>Relatively less expensive</td>
</tr>
<tr>
<td>Size of unit</td>
<td>Extremity units are smaller</td>
</tr>
</tbody>
</table>

*In comparison to high-field units.*

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**Motion Artifact**

Motion artifact is presumably the most common artifact in MRI. It causes ghosts and blurring on MR images, as the phase gradient cannot anticipate and encode signals from moving structures. Its sources are voluntary motions, involuntary motions, and physiologic motions.2 Voluntary motions by the patient can be minimized by explaining the importance of keeping still. Children may have lower compliance, and sedation might be necessary. Involuntary motions are more difficult to handle, as they cannot be suppressed through the patient’s own will. There is a broad range of causes, from mental illness to neurodegenerative processes such as Parkinson’s disease or Huntington’s chorea. Physiologic motions in the patient’s body are multifactorial. For example, great difficulties in thoracic imaging have been caused by respiration and cardiac action. Shorter sequences and electrocardiogram (ECG) controlled picture acquisition help counteract these problems.3 Other physiologic motions such as pulsation in arteries or bowel peristalsis are more difficult to handle.4 At least the latter can be controlled to a certain extent by antispasmodics. Using short sequences such as single shot fast spin echo (SSFSE) helps reduce the likelihood of motion artifacts.4

**Flow Artifact**

Flow artifact is one type of motion artifact caused by motion of liquids within the human body, usually blood or cerebrospinal fluid (CSF). Arterial flow artifact has not only a flowing component but also a pulsating one. The reasons for flow artifacts are multiple, and their appearance varies. Blood flowing through a slice can undergo excitation from an incoming RF pulse but might already have left the slice before readout. As a result, the vessel would appear empty or at least less bright than expected. It is more difficult to record an adequate signal from within vessels with a laminar pulsatile flow. Possible reasons for low signal intensity are: (1) fast flow, (2) intravoxel phase dispersion from different velocities in the voxel, (3) odd ordered echo dephasing, (4) displacement effects related to in plane flow during acquisition, and (5) saturation from prior RF pulse.5

The artifacts caused by flow might even appear bright. If blood flow is slow, a certain amount of unsaturated blood might follow the saturated blood, which has experienced a prior RF pulse. When the unsaturated volume flows into the slice just in time to experience the 90 degree pulse, it creates a stronger signal than expected. Hence possible reasons for high signal intensity are: (1) slice related
inflow enhancement, (2) even echo rephrasing, (3) diastolic pseudo
gating, and (4) pseudoflow related to methemoglobin.3

Possible techniques to reduce artifacts include flow compensa
tion, saturation pulses, and cardiac triggering.4,8 Flow compensation
uses a series of gradient pulse sequences to eliminate the interfering
effects of fluids in motion. With saturation pulses, signals are
added parallel to slices to suppress the blood signal.8 Cardiac trig
gering works by synchronizing the imaging sequences with cardiac
action.4 Sometimes this artifact overlies normal or pathologic struc
tures, making diagnosis difficult. The switching of phase and fre
quency direction may help in such cases9 (Figure 3.6).

**Wrap-Around Artifact**

Wrap around is a preventable artifact caused by improper choice of
parameters in an MRI scan. If the field of view (FOV) is made too
small, the tissue surrounding the FOV might become excited and pro
duce interference signals during readout. As phase encoding gradients
are gauged for the FOV alone, they cannot integrate this “external”
Thus the signal is not correctly registered as to location, but instead gets wrapped around to the opposite side of the FOV (Figure 3 7). This phenomenon is also called “aliasing.” In a clinical setting, the frequency direction is usually chosen along the long axis of the object to be scanned to avoid wrap around artifacts. There are other approaches to avoid this common artifact. The simplest way is to add presaturation pulses to tissue you do not want to image before applying the pulses for excitation.10 Another solution is the use of low pass and high pass RF filters, which filter out initial signals that exceed the bandwidth. Increasing the FOV is a possible solution, with the caveat that it decreases spatial resolution of images.3 Finally, a "no phase wrap" option is provided by some manufacturers.9 This doubles the FOV, thereby doubling the phase encoding steps (phase over sampling) to keep resolution at the same level while halving the number of excitations to keep scan time constant.

Chemical Shift Artifact

At the boundary between tissues high in fat and those high in water, protons of fat can be incorrectly imaged, an effect called chemical shift artifact. It can occur in MRI because of slight differences in the precession frequency (also known as Larmor frequency, \( v = \gamma * B_0 \)) of these protons. These different frequencies are caused by slight inhomogeneities of the main magnetic field (\( \Delta v = \gamma * \Delta B_0 \)) and get worse with increasing field strengths. In a 1T magnet, the difference in frequency is 147 Hz, whereas in a 1.5 T magnetic field the difference is 224 Hz.11

Because the computer assumes all protons precess at the same frequency, the signal from fat is mapped to a different location corresponding to the frequency at which it is precessing. Narrow receive bandwidths accentuate this by assigning a smaller number of frequencies across the image (Figure 3 8). This artifact may appear as high intensity areas when signals of water and fat overlap and low intensity areas when their signals spread apart.12 As a result, the affected structures may be incorrectly imaged and thereby misinterpreted. The chemical shift artifacts in musculoskeletal imaging are seen in vertebral end plates, fluid filled cysts, fat containing tumors, and at the cartilage bone marrow interface.13,14

Increasing bandwidth and using low field magnets are options to reduce chemical shift artifact. Other feasible solutions to reduce...
Susceptibility artifacts are caused by microscopic gradients or by substances with different magnetic susceptibilities at the boundary between contiguous tissues. The difference in magnetic susceptibility can lead to minor inhomogeneities in the magnetic field strength, which in turn cause distortion in terms of spatial frequency or signal intensity. A ferromagnetic object residing in a diamagnetic structure like the human body is sensitive to magnetic susceptibility. This object induces eddy current due to the incident RF magnetic field, altering the RF field near itself and thereby causing distortion. This in turn creates gradients that produce dephasing of spins and frequency shifts in surrounding tissue.\textsuperscript{18} Susceptibility artifacts on MR images appear as areas with profuse signal intensity or are totally devoid of signal.\textsuperscript{18}

Susceptibility artifacts obscure surrounding normal structures and may also mask areas of abnormality (Figure 3.9). Large susceptibility artifacts can be seen around prosthetic joints with GRE sequences.\textsuperscript{19} Long echo times also exacerbate these artifacts.\textsuperscript{18,19}

\textbf{FIGURE 3.8.} Chemical shift artifact. A, MR image with narrow bandwidth (72.4 Hz) shows more prominent chemical shift artifact than (B) that with wide bandwidth (446.4 kHz). The chemical shift artifact makes it difficult to evaluate cartilage (arrows). Note the apparent change in the thickness of the anterior femoral cortex on the two images. (Courtesy of Philips Medical Systems.)

\textbf{FIGURE 3.9.} Magnetic susceptibility artifact over the cartilage surface of the knee (arrows). A, GRE image is more sensitive to difference in the magnetic susceptibility than (B) FSE image. These artifacts (arrows) typically result from prior surgery and may not be visible on radiographs.
SE or FSE may help minimize these artifacts, as do high bandwidth and short echo times.

**Magic Angle Effect**

This effect is responsible for producing increased signal (and therefore possible erroneous diagnosis) in certain tissues such as tendons. The magic angle effect is a phenomenon related to collagen anisotropy in MRI.\(^{34}\)

Anisotropy is the property of being directionally dependent.

If the angle between the main magnetic field (B\(_0\)) direction and the collagen fiber increases from 0 degrees, the signal intensity on short TE sequence changes as a result of increasing T2 relaxation time.\(^{27}\) T2 relaxation time is at its maximum at an angle of almost 55 degrees relative to B\(_0\).\(^{20,22-24}\) It occurs in any tissue that contains anisotropically arranged collagen fibres such as tendons, menisci, and hyaline cartilage.\(^{22,25,26}\)

The water content of cartilage varies from 70% in deep layers to 84% in superficial layers.\(^{27,28}\) The short T2 relaxation time in cartilage depends on the dipolar orientation of water molecules, which are linked to collagen macromolecules. Histologically, hyaline cartilage has multiple layers (superficial, transitional, deep radial, calcified cartilage) that are distinct from the layers seen on high resolution MRE.\(^{29}\) Microscopy studies reveal that collagen fibrils in the deep radial layer of cartilage are arranged perpendicular to the subchondral bone, but more superficially, fiber orientation parallels the articular surface.\(^{30}\) This arrangement induces the magic angle effect. If cartilage is placed in the magnet, the area of anisotropic arrangement of collagen fibers increases signal intensity at magic angle.\(^{20,22,23}\) It can occur in any depth of cartilage.\(^{29}\) The increased signal and inhomogeneity of signal in the articular cartilage created by this artifact should not be confused with early degenerative changes in the cartilage substance.

The magic angle effect is also seen in various tendons. The rotator cuff, in particular the supraspinatus tendon, is frequently examined in MRI (Figure 3 10).\(^{30,32}\) Magic angle effects in healthy tissue can look similar to signal abnormalities caused by degenerative processes or a partial tear and can lead to difficulties in diagnosis. To avoid the magic angle effect, long TE sequences with and without fat suppression may help or, if necessary, repositioning of the patient may be tried.\(^{21,35}\)

**Truncation Artifact**

Truncation artifacts are also known as *Gibbs ringing artifacts* (in honor of Josiah W. Gibbs). They appear in MR images as alternating dark and bright lines that run parallel to a sharp change in signal intensity.\(^{34}\) For example, this change can be produced at the boundary between layers of fat and muscle tissue. Truncation artifact was also frequently observed in the cartilage of both the patellofemoral compartment and the posterior region of the femoral condyles on fat suppressed 3D SPGR images. This laminar appearance does not indicate degenerative ative change of the articular cartilage, nor does it reflect the anatomic layers of the cartilage; it is merely an artifact (Figure 3 11).\(^{35-37}\)

Truncation artifact occurs when the echo at the edges of the acquisition window does not return to 0. This happens especially when a small acquisition matrix is used. One way to reduce the severity of this effect is to increase the resolution of images, but this reduces the SNR or extends the imaging time. Another possibility is to use filters on images, although this can be associated with decreased image resolution. Changing the frequency and phase encoding directions may help to reduce truncation artifact.\(^{38-40}\)

**CLINICAL APPLICATIONS**

**Osteoarthritis**

MRI is increasingly being utilized to evaluate lesions of the articular cartilage, and numerous imaging sequences have been advocated for this purpose. Early studies suggested that T1 weighted and T2 weighted images were indispensable for detailed evaluation of articular cartilage degeneration.\(^{21}\) Subsequently, several new imaging sequences have been developed. Magnetization transfer contrast (MTC) imaging can separate articular cartilage from adjacent joint fluid by suppressing the signal produced from cartilage.\(^{42-44}\) FSE imaging with fat suppression for proton density weighted images and T2 weighted images can depict articular cartilage abnormalities in osteoarthritis with higher accuracy than arthroscopic grading.\(^{45,46}\) (Figure 3 12). Fat suppressed 3D spoiled gradient recalled acquisition in the steady state (SPGR) has been reported as a more sensitive imaging sequence for the detection of articular cartilage defects in the knee.\(^{47-50}\) In recent studies, driven equilibrium Fourier transform (DEFT) imaging has been shown to provide contrast between cartilage and joint fluid by enhancing the signal from joint fluid, rather than by suppressing the signal from cartilage as other sequences do.\(^{50}\) Delayed gadolinium DTPA\(^2\) enhanced MR imaging is also a promising method that has potential for monitoring the glycosaminoglycan content of cartilage in vivo.\(^{51,52}\)

The relative signal intensity of the normal articular cartilage is dependent on the pulse sequences used. T2 weighted SE imaging, proton density weighted and T2 weighted FSE imaging, MTC imaging, and DEFT imaging can show synovial fluid of high signal intensity (bright) and cartilage of intermediate to low signal intensity (dark), whereas fat suppressed 3D SPGR sequences produce bright cartilage and dark synovial fluid. However, the signal intensity of the normal articular cartilage may not be uniform due to artifacts and other phenomena such as magic angle effect, truncation artifact, chemical shift artifact, magnetic susceptibility effect, and regional anatomic variations.\(^{53-55}\) The laminar appearance within the articular cartilage on fat suppressed 3D SPGR images is predominantly attributable to truncation artifact rather than to histologic zonal anatomy, as mentioned above.\(^{35,50}\) Thus the MR appearance of the articular cartilage is highly variable, and understanding normal variations is clinically important in order to improve diagnostic accuracy and avoid misdiagnoses.\(^{34}\)

**Rheumatoid Arthritis**

MRI is much more sensitive than radiography or ultrasonography for the diagnosis of rheumatoid arthritis (RA) especially in its early stages. Previous studies have reported MRI is seven to nine fold more sensitive than radiography for detecting erosions in early disease and is able to detect lesions 6 to 12 months before they appear on radiographs.\(^{55-59}\) MRI identified more than twice as many erosions than did ultrasonography and radiography and was more sensitive than ultrasonography for detecting synovial disease.\(^{60}\)

Because MRI can provide excellent soft tissue contrast, it can detect synovitis, erosions, and bone marrow edema due to RA very well. Previous studies comparing various imaging sequences found that dynamic imaging or fat suppressed T1 weighted imaging with gadolinium contrast medium was useful in diagnosing synovial inflammation of early stage RA.\(^{58,61-65}\) (Figure 3 13). However, enhancement of synovium in patients with RA is time dependent, and it is necessary that MR images be acquired within 5 minutes following contrast medium administration to differentiate active synovitis from fibrosis or joint effusion.\(^{62,65-67}\)
Extremity MRI may play an important role in the diagnosis of RA. Conventional whole body high field MRI is expensive and inconvenient for patients and has some contraindications, such as implanted metal objects (pacemakers, aneurysm clips, and cochlear implants) and claustrophobia. Low field dedicated extremity MR machines are now commercially available and have been applied to the evaluation of RA. In some reports, the diagnostic accuracy of low field dedicated extremity MRI for synovitis, bone marrow edema, joint effusion, and bone erosion accompanying RA is comparable to that of the high field MRI.\(^1,2\) Even at low field, the sensitivity to bone damage of a portable MRI system was superior to that of radiographs of the wrists and metacarpophalangeal joints.\(^6,8\)

MRI identified bony erosion in 95% of patients with inflammatory arthritis, whereas radiographs identified only 59%. The introduction of effective therapies for RA has increased the importance of imaging in rheumatology, and low field extremity MRI offers adequate performance but at lower cost and with greater comfort and convenience for the patient.\(^69\) However, a recent review of in office MRI scanning concluded that additional study is warranted.\(^70\)

**Osteoporosis**

Osteoporosis is a metabolic bone disease characterized by bone loss and structural deterioration of bone tissue, leading to bone fragility.
and increased susceptibility to fractures, especially of the hip, spine, or wrist. According to National Osteoporosis Foundation estimates, osteoporosis is a major public health threat for an estimated 44 million Americans, or 55% of people 50 years of age and older. In the United States today, 10 million individuals are estimated to already have the disease. Radiographs or MRI may be used for diagnosis of fractures secondary to osteoporosis. Critical to the evaluation of vertebral fractures on imaging studies is the fact that not all vertebral fractures are due to osteoporosis. In particular, antecedent trauma, infection, and tumor must be excluded. In many cases, MRI is useful for differentiating osteoporotic fractures from pathologic fractures by showing abnormal contrast enhancement of bone marrow and adjacent soft tissues in pathologic fractures.

Osteoporosis screening with MRI is a challenging area. Dual x ray absorptiometry (DXA) scanning is used for screening but does not allow determination of the microstructure of bone. The methods available for quantitatively assessing microstructure of trabecular bone noninvasively include high resolution or micro computed tomography (CT) and high resolution or micro MRI. MRI can be used to assess the properties of trabecular bone in two different ways. The first is an indirect measure, often termed relaxometry or quantitative magnetic resonance (QMR). This method takes advantage of the fact that trabecular bone alters the adjoining marrow relaxation properties in proportion to bone density and structure. The second is the direct visualization of the dark trabecular bone, which, because of its low water content and short MR relaxation times, appears in stark contrast to the bright marrow fat and water in high resolution MRI. Currently two primary sequences used for micro MRI of trabecular bone are variants of the basic GRE and SE based fast large angle spin echo (FLASE) sequences.

MR SAFETY

MRI is noninvasive and does not involve radiation. However, special safety issues have to be considered. The main risk associated with MRI is the effect of the strong magnetic field on ferromagnetic objects on or inside a patient’s body, such as pacemakers, aneurysm clips, cochlear implants, neurostimulators, metal implants, surgical staples, some artificial heart valves, and foreign metal objects in the eye. Most orthopedic implants such as total joint prostheses do not present a hazard for MRI, although they do distort the magnetic field, potentially limiting the delineation of tissues near the implant.

The safety of pregnant patients should be considered. In 1997, the American College of Radiology issued a statement on the safety of MRI in pregnant patients. The statement is that in light of the

FIGURE 3 11. Truncation artifact. Linear low signal intensity in the cartilage of the femoral trochlea and patellar facet is seen due to truncation artifact on fat suppressed SPGR image.

FIGURE 3 12. Normal MRI of the knee cartilage: A, fat suppressed FSE PDW image; B, fat suppressed FSE T2 weighted image; and (Continued)
lack of data demonstrating deleterious effects of MR on the developing human fetus, MRI should be recommenced for evaluating pregnant patients when any alternative imaging procedure involves ionizing radiation. The question also arises about how to advise pregnant health care practitioners appropriately regarding exposures related to the MRI environment. One survey of reproductive health among female MR workers suggested that the data do not demonstrate a correlation between working in the MR environment and offspring gender or changes in the prevalence of premature delivery, infertility, low birth weight, or spontaneous abortion. However, sufficient safety has not been fully proven at this time.

In addition, certain metallic objects are not allowed into the examination room. Items such as jewelry, watches, credit cards, and hearing aids can be damaged (Box 3.1). Pins, hairpins, metal zippers, and similar metallic items can distort the images. Patients with a history of potential exposure to small metal fragments will be screened for metal shards within the eyes by orbit radiographs or by a radiologist’s review and assessment of contiguous cut CT. For patients with tattoos, it is recommended that cold compresses or ice packs be placed onto the tattooed areas in order to decrease the potential for RF heating of the tattooed tissue. Several websites are available for reference. These include http://www.mrisafety.com, which includes a listing of implants, materials, and medical devices that can be referred to for screening patients prior to MRI, http://www.radiology.upmc.edu/MRsafety, and http://www.ismrm.org.

Time varying gradient magnetic fields may have biologic effects with the introduction of rapid echo planar imaging and the use of high performance gradient systems, as it is known that rapidly switching magnetic fields can stimulate muscle and nerve tissue. At present, however, there is no known mechanism that would suggest an irreversible biologic effect caused by rapidly switching magnetic fields.

RF burns are related to contact between electrically conductive materials such as wires, leads, and implants and the patient’s bare skin during an MRI procedure. Care should be taken to place thermal insulation between the patient and the electrically conductive material during imaging. The rapidly changing magnetic field will induce an electromotive force or voltage in the conductor that causes a flow of current. The flowing current in a conductor with electrical resistance will result in heating the conductor, thus causing a burn if it contacts the skin. Heating also occurs at the point of skin to skin contact. The patient’s bare skin should not be allowed to form a large conductive loop, as occurs when crossing arms and legs while in the magnet.

The specific absorption rate (SAR) is a measure of the absorption of electromagnetic energy in the body (in watts per kilogram hour).

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The specific absorption rate (SAR) is a measure of the absorption of electromagnetic energy in the body (in watts per kilogram hour).

BOX 3-1 Contraindications for MRI*

**ABSOLUTE CONTRAINDICATION:**
- Pacemaker
- Otic implant
- Metal in eye or orbit
- Implanted cardiac defibrillator

**LIKELY CONTRAINDICATION:**
- Heart valve or aneurysm clip installed before 1996

**POSSIBLE CONTRAINDICATION:**
- Heart valve or aneurysm clip installed after 1996

**USUALLY ALLOWABLE 6-8 WEEKS AFTER IMPLANTATION:**
- Passive implants, weakly ferromagnetic (e.g., coils, filters, and stents; metal sutures or staples)
- Rigidly fixed passive implants, weakly ferromagnetic (e.g., bone/joint pins, screws, rods)

**CAUTION:**
- Tattoos

**USUALLY ALLOWABLE IMMEDIATELY AFTER IMPLANTATION:**
- Passive implants, nonferromagnetic (e.g., bone/joint pins, screws, or rods; coils, filters, and stents; metal sutures or staples)

**Passive implants, nonferromagnetic (e.g., bone/joint pins, screws, or rods; coils, filters, and stents; metal sutures or staples)**

**Passive implants, weakly ferromagnetic (e.g., coils, filters, and stents; metal sutures or staples)**

**CAUTION:**
- Tattoos

**Tattoos**

<table>
<thead>
<tr>
<th>Term</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field strength</td>
<td>Static magnetic field within the scanner, measured in Tesla (T).</td>
</tr>
<tr>
<td>Field of view (FOV)</td>
<td>The distance of anatomic coverage in a given imaging direction.</td>
</tr>
<tr>
<td>Fringe field</td>
<td>“Stray” magnetic field extending outside the imaging bore of the magnet. The distance this field extends outside the bore is a major safety consideration in designing the size and shielding requirements of MRI rooms.</td>
</tr>
<tr>
<td>Gradient</td>
<td>Variation in magnetic field strength with change in distance, used to determine voxel location when making an image. Measured in milli Tesla per meter (mT/m).</td>
</tr>
<tr>
<td>Image plane</td>
<td>May be selected based on anatomic considerations. The most common imaging planes are axial, coronal, and sagittal.</td>
</tr>
<tr>
<td>Matrix</td>
<td>The number of “in plane” pixels along each given image direction. In combination with FOV, determines the in plane image resolution.</td>
</tr>
</tbody>
</table>

**Pulse sequences**
- Timing of MRI parameters (RF pulse strength and spacing, magnetic field gradients, and signal collection) used to create MR images with varying degrees of tissue contrast.

**Radiofrequency (RF)**
- Energy deposited in the patient in order to produce MRI signals (usually in the megahertz frequency range at typical magnetic field strengths used). A side effect is unwanted heating of tissues, which limits the amount of allowable energy deposition.

**Selective fat saturation**
- Also known as chemical shift fat saturation, a method of removing fat signal based on the different signal frequencies of fat and water. More subject to non uniform fat suppression than STIR imaging.

**Signal to noise ratio (SNR)**
- Quantitative value to describe the image quality of a detected signal relating the true signal and superimposed background noise signal.

**Slice thickness**
- Definition of the smallest structures that can be differentiated on an image, generally related to pixel or voxel dimensions, although voxels can be interpolated to artificially increase display resolution from the true image resolution. True in plane resolution equals field of view divided by matrix.

**Spatial resolution**
- The through plane voxel dimension.

**STIR**
- “Short Tau Inversion Recovery” pulse sequence; a popular and robust method used for suppression of MRI signal from fat.

**Slice thickness**
- Tesla (T)
- Unit of magnetic field strength. 1 Tesla equals 10,000 Gauss (the earth’s magnetic field strength is approximately 0.5 Gauss).

**TR**
- Repetition time; the time between successive pulse sequences applied to the same slice.

**TE**
- Echo time; the time between the initial pulse and the peak of the echo signal.

**T1 weighted**
- Represents image contrast due to differences in T1 relaxation time. T1 weighted image is created by using short TR and TE (see Table 3 1).

**T2 weighted**
- Represents image contrast due to differences in T2 relaxation time. T2 weighted image is created by using long TR and TE (see Table 3 1).

**T1 relaxation time**
- Time constant that the longitudinal magnetization returns toward equilibrium after RF excitation. Each tissue has a characteristic T1 time.

**T2 relaxation time**
- Time constant that the transverse magnetization decays toward zero after RF excitation. Each tissue has a characteristic T2 time.

**Voxel**
- “Volume element,” the 3D size of each point in an image, generally determined by two in plane pixel dimensions (in turn determined by FOV and matrix) and the slice thickness.

Portions of this table are courtesy of Aaron D. Sodickson, MD, PhD, Brigham and Women’s Hospital, Boston, MA and were borrowed with permission from American College of Rheumatology Extremity Magnetic Resonance Imaging Task Force: extremity magnetic resonance imaging in rheumatoid arthritis. Arthritis Rheum 54:1034–1047, 2006.
REFERENCES


