Title
Physical Principles of Magnetic Resonance Imaging

Journal
Clinical Orthopaedics and Related Research®, 244(&NA;)

ISSN
0009-921X

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Publication Date
1989-07-01

DOI
10.1097/00003086-198907000-00003

Peer reviewed

Reprinted from Clinical Orthopaedics, July, 1989
Volume 244

Physical Principles of Magnetic Resonance Imaging

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When a nucleus contains either unpaired protons, neutrons, or both, it has angular momentum. This property provides the basis for magnetic resonance imaging (MRI). Because of its abundance in the human body, hydrogen is used for clinical MRI. Within the MR magnet, the atoms are aligned with the magnetic field. The MR signal is generated by subjecting the atoms to radiofrequency (RF) pulsations, and the pulse sequences selected will determine the appearance of the image. The repitition time (TR) is the time between the RF pulses, and the echo time (TE) is the time between the RF pulse and the recording of the signal. T1-weighted images (short TR/short TE) provide the best anatomic detail, but T2-weighted images (long TR/long TE) are often needed to demonstrate pathology. Many unfamiliar artifacts may be encountered on MR images. These can be due to several factors, including inhomogeneities in the magnetic field (for example, those resulting from metallic orthopedic appliances or prior surgery), extraneous RF interference, and motion. Surface coils are commonly used in musculoskeletal MR imaging to improve the quality of the examination by increasing the signal-to-noise ratio over the area of interest.

Magnetic resonance imaging (MRI) has had a profound impact on the evaluation of several musculoskeletal disorders, including traumatic, neoplastic, degenerative, and inflammatory conditions.[4] With this technique, bone marrow, ligaments, and tendons can be directly imaged with higher sensitivity than has been possible with any other modality. Because MRI does not utilize ionizing radiation, it is an attractive alternative to computed tomography (CT) in cases in which cross-sectional imaging is required.

This article is intended to briefly review those principles that are required to understand the appearance of musculoskeletal MR images. All equations have intentionally been excluded from the discussion. Many excellent references are available to those interested in a more complete understanding of the physics of MRI.[1,3,5,6,8,10,13,14]

THE MAGNETIC RESONANCE PHENOMENON

Radio waves, used in MR image acquisition, and X-rays, employed for conventional roentgenography, are found at opposite ends of the electromagnetic spectrum (Fig. 1). It is only within these extremes of very high energy (X-ray) and very low energy (radio) that medical imaging is possible. The body is not transparent to energy levels in the middle of the spectrum, such as visible, infrared, and ultraviolet light.

Although the chemical properties of an atom depend on its orbital electron structure, its physical properties are mainly dependent on the nucleus. With the exception of hydrogen, the nucleus of all atoms contains both protons and neutrons. In order to maintain electrical neutrality, the number of positively charged protons and negatively charged electrons is usually the same. The number of protons and neutrons, however, may not necessarily be the same.

If a nucleus contains either unpaired protons, neutrons, or both, it has a net spin and angular momentum. The spin of the atom

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Received: December 13, 1988.

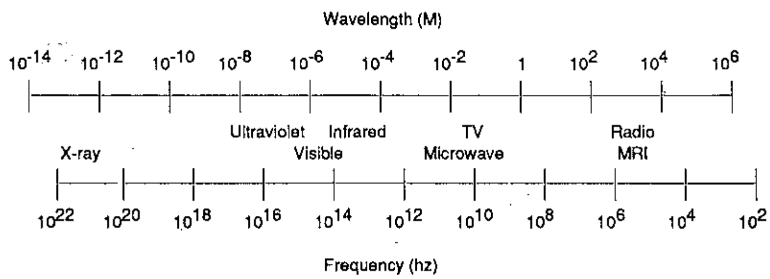


FIG. 1. The electromagnetic spectrum. Wavelengths are shown across the top and frequencies across the bottom. The windows for imaging the human body are at the two extremes of the spectrum, with high-energy X-rays at one extreme and low-energy radiowaves at the other. The energy levels shown are approximate.

represents a current loop, with north and south poles. Angular momentum describes the rotational motion of a body and must be nonzero in order for the MR phenomenon to occur.

Hydrogen is the ideal atom for medical imaging. Of the stable atoms with unpaired nucleons, it is the simplest (only one nucleon: a proton), and it accounts for two thirds of all the atoms in the human body. The remainder of this discussion will refer only to MRI using hydrogen, although other atoms that are less abundant may also be used (sodium 23, phosphorus 31, carbon 13, and others). The terms "proton" and "hydrogen" will be used interchangeably.

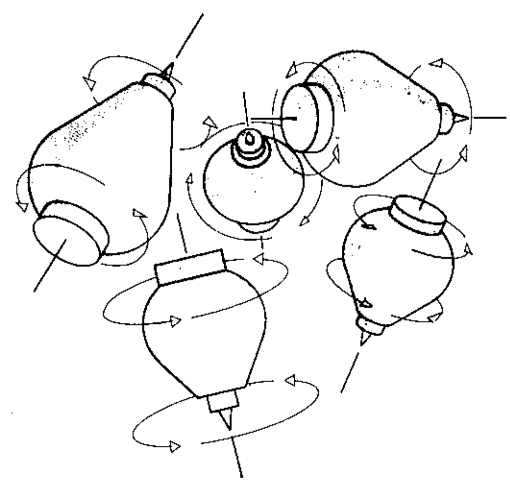


FIG. 2. Protons randomly spinning, represented as spinning tops.

The magnetic, or dipole, moment of an atom is the tendency to produce motion, and is the result of an atom's angular momentum. In the absence of an externally applied magnetic field, the vectors of these magnetic dipole moments are randomly oriented (Fig. 2). Once exposed to a magnetic field such as is present in a magnetic resonance scanner, the dipoles tend to align with the field like tiny bar magnets.

Because of their small size, protons follow the rules of quantum mechanics rather than those of classical Newtonian physics. In Newtonian physical theory, a bar magnet must align precisely with the applied magnetic field. In quantum theory, protons may be in one of two positions that differ from one another by 180°; that is, either parallel or antiparallel to the applied field (Fig. 3). The parallel orientation is the low energy condition, and is called the ground state. The antiparallel orientation is the high energy state and is referred to as the excited state. Because slightly more protons align in the low-energy ground state, a sample of protons placed in a static magnetic field will have a net magnetic field vector parallel to the applied field. In a population of one million atoms, only one more atom will be in the parallel than in the antiparallel position, a difference that, while minute, is sufficient to allow clinical MR imaging.

In addition to spinning, protons precess, or wobble a few degrees off the axis of the applied magnetic field, similar to the spinning of a gyroscope under the influence of the earth's gravitational field (Fig. 4). The frequency of this precession is known as the resonant frequency and is proportional to the strength of the applied magnetic field.

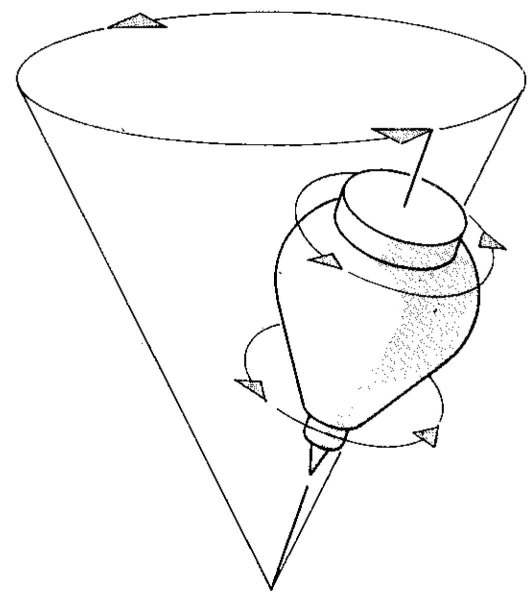


FIG. 4. Precession of a spinning proton about static magnetic field. This is similar to the spinning of a gyroscope under the influence of the earth's gravitational field.

MAGNETIC RESONANCE IMAGE FORMATION

Because spin-echo (SE) imaging is the most commonly used method of MR image formation, this technique will be described. Before the MR signal of a sample of tissue within a magnet can be generated and detected, the protons must undergo three additional manipulations: (1) the net magnetic vector of the protons must be flipped 90° from the parallel (longitudinal) position into the transverse plane; (2) the spins of the protons must be coherent, or spinning in phase together; and (3) the spins must be moved into a higher energy level such that there are an equal number in the high and low energy states. All of these conditions are accomplished by the application of a second magnetic field offset 90° from the main magnetic field, which is oscillated with a frequency synchronized to the resonant frequency of the precessing spins. This oscillating magnetic field is a radiofrequency (RF) pulse. In clinical practice, the second pulse is followed by a 180° rephasing pulse in order to compensate for inherent inhomogeneities in the static magnetic field of the scanning system. This, however, is not necessary to generate or detect an MR signal.

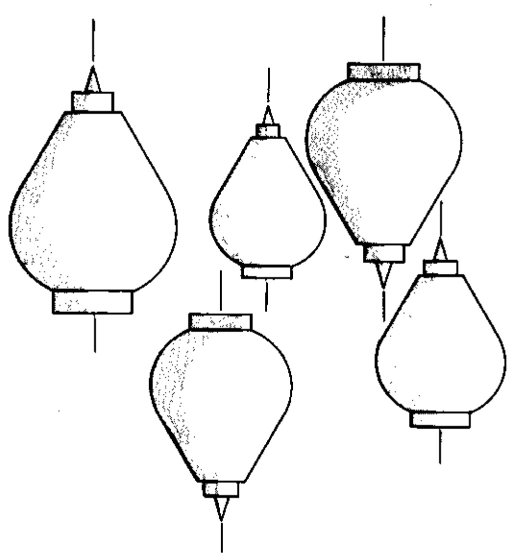


FIG. 3. Once subjected to an externally applied magnetic field, the protons align either parallel or antiparallel to the field. Slightly more align in the parallel, or low energy state.

When the second magnetic field (the RF pulse) is turned off, the protons move back to the lower energy state and thus emit RF energy of their own. This release of RF energy is the MR signal. Among other things, the amplitude of the MR signal is proportional to the number of spins in the sample, or its proton density. When more protons are present, the intensity of the magnetization is greater, and the signal detected by the RF receiver coils is greater.

In order to obtain an anatomic image, the spins are subjected to three magnetic field gradients within the magnet. The slice-select gradient locates the position and specifies the thickness of a particular image within a scan sequence. The remaining two gradients are used to construct the two-dimensional image. The frequency, or read, gradient locates a given proton by the frequency of its spin. The phase-encoding gradient specifies the location in space of a proton within the slice according to the phase of its spin. Depending on the selection of imaging plane and type of imaging sequence chosen, any of these three gradients may be along the X, Y, or Z axis. An appreciation for these three different gradients assists in understanding MR artifacts, which are discussed below.

PULSE SEQUENCING

For any sample, the recovered signal intensity is less than its original amplitude as determined by the proton density. This is due to irreversible signal losses within the sample, known as T1 and T2 relaxation. The rate of these relaxations depends on the local environment of the sample and thus reflects its chemical structure.

T1 relaxation, also called longitudinal or spin-lattice relaxation, characterizes the interaction of a nucleus with its environment. After the RF stimulation, the spins are moved into the higher-energy transverse orientation. As they fall back into the lower-energy longitudinal orientation, energy is dissipated into the surrounding environment. T1 in milliseconds is the time required for 63% of the longitudinal magnetization to recover following the RF pulse. T1 relaxation times vary with the main magnetic field strength of the imaging system and increase slightly with stronger magnets. In clinical practice, however, the effect on image contrast is of little consequence when compared with what is possible with pulse sequence manipulation.

T2 relaxation, also known as transverse or spin–spin relaxation, is due to randomly varying inhomogeneities in the magnetic field created by adjacent nuclei within the sample. It characterizes the interaction of a nucleus with surrounding nuclei of the same kind. T2 in milliseconds is the time necessary to reduce the transverse magnetization to 37% of its original value following the RF pulse. Because it reflects the chemical environment of a proton, T2 relaxation is independent of magnetic field strength.

IMAGE CONTRAST

MR signal intensity is a reflection of T1 and T2 relaxation values as well as several other parameters that are beyond the scope of this discussion. The relative contributions of some of these factors may be manipulated by controlling the timing of the RF pulses. The timing values themselves are the TR (repetition time) and TE (echo time). TR refers to the time in milliseconds between the 90° RF pulses. TE is the time in milliseconds between application of the 90° RF pulse and recording the signal (echo) produced by the sample. A short TR and TE will result in an image that emphasizes the T1 characteristics and minimizes the T2 characteristics of a tissue and is said to be T1-weighted, for example, TR = 200 and TE = 15. A longer TR and TE will result in a more T2-weighted image, for example, TR = 2000 and TE = 85. Proton-density images are obtained by using a pulse sequence with a short TE and a long TR. This type of pulse sequencing may be the first echo of a double-echo sequence (a long TR and two different TEs, the first short and the second long), or it may be used in situations in which true T1- or T2-weighting is not needed.

Tissues behave differently according to the pulse sequence chosen for imaging (Table 1). This property assists in characterization of normal and abnormal tissue when both T1- and T2-weighted imaging are utilized.

T1-weighted imaging is useful for identifying abnormal tissue within structures of high fat content, such as marrow and subcutaneous fat. T1-weighted imaging also provides

TABLE 1. Tissue Signal Intensity on T1- and T2-Weighted MR Images

Tissue	*T1-Weighted*	*T2-Weighted*
Cortical bone	Void	Void
Ligaments, tendons	Void	Void
Fibrocartilage	Void	Void
Normal fluid	Low	High
Tumor	Low to intermediate	High
Abnormal fluid (*e.g*, pus)	Intermediate	High
Hyaline cartilage	Intermediate	Intermediate
Muscle	Intermediate	Intermediate
Fat (marrow)	High	Intermediate to high

the highest signal-to-noise ratio and is therefore optimal for evaluating anatomic subtleties. Substances with a short T1 value (high-signal intensity) include fat and lipid-containing materials and proteinaceous fluid. Tissues with a long T1 value (low-signal intensity) include normal body fluids (cerebrospinal fluid, urine), calcium (cortical bone), and most ligaments and tendons. Most other soft tissues, including muscle, tumors, and infected tissue, have an intermediate-signal intensity in T1-weighted images.

T2-weighted images assist in distinguishing normal from abnormal soft tissues. Tissues with a short T2 value (low-signal intensity) include calcium (cortical bone) and most ligaments and tendons. Tissues with a long T2 value (high-signal intensity) include neoplasms, inflammation, and most fluids. The signal intensity of normal muscle on T2 weighting is low to intermediate. These differences in T2 values allow for the identification of boundaries of tumor and infection within soft tissues.

It should be emphasized that there are degrees of T1 and T2 weighting, which allows for a wide spectrum of tissue appearances. The specific type of pathology under investigation will determine the ideal choice of pulse sequencing for imaging any given patient.

One of the prohibitive aspects of MR imaging is the time required for image acquisition, especially for T2-weighted imaging. In order to overcome this, a variety of new scanning methods have recently been developed. The most widely used fast scan is gradient-echo imaging. For this technique, a flip angle smaller than the 90° pulse of conventional spin-echo imaging is used, and the 180° rephasing pulse is replaced with an echo generated by gradient reversal. Gradient-echo imaging cannot always substitute for true T2-weighted imaging, and the appropriate use of this technique will depend on the type of pathology under clinical investigation. Gradient-echo imaging has been shown to be useful in the spine, but applications to other parts of the musculoskeletal system have not yet gained wide acceptance.

ARTIFACTS

Because magnetic resonance uses an approach to imaging totally different from that of ionizing radiation, many unfamiliar artifacts may be encountered. Several excellent articles have been written on the subject of MR artifacts.[2,9,11,12] The discussion below will concentrate on those that are either common in musculoskeletal imaging or can be a source of critical misdiagnoses.

The majority of MR artifacts may be categorized according to the aspect of image acquisition that is primarily responsible for it. The four general types are (1) static magnetic field artifacts, (2) RF magnetic field artifacts, (3) gradient-field artifacts, and (4) motion artifacts.

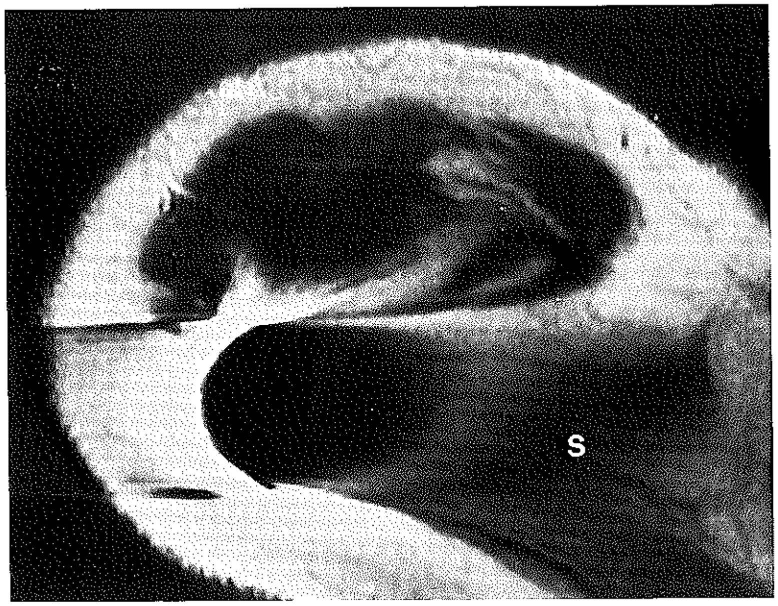


FIG. 5. An axial T1-weighted MR image of the shoulder at the level of the supraspinatus muscle(s). Anterior is at the top of the image. The patient had a small pin in her bra strap. An artifact of a combination of a signal void and adjacent bright signal is most severe in the transverse plane (right to left), reflecting the direction of the frequency-encoding axis used for acquiring this image.

STATIC MAGNETIC FIELD ARTIFACTS

Static magnetic-field artifacts are due either to intrinsic inhomogeneities within the static magnetic field itself or to external environmental influences on this field. Intrinsic field inhomogeneities most frequently result from improper coil design or thermal

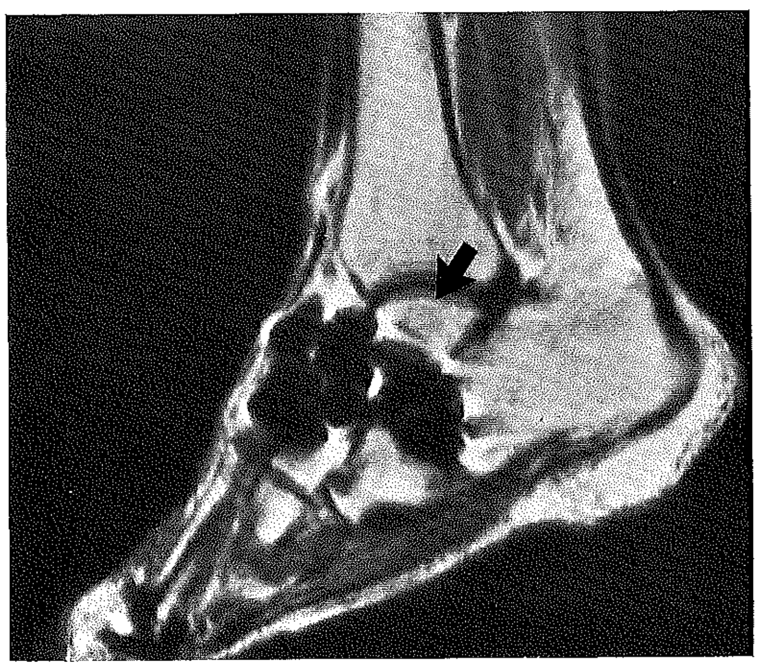
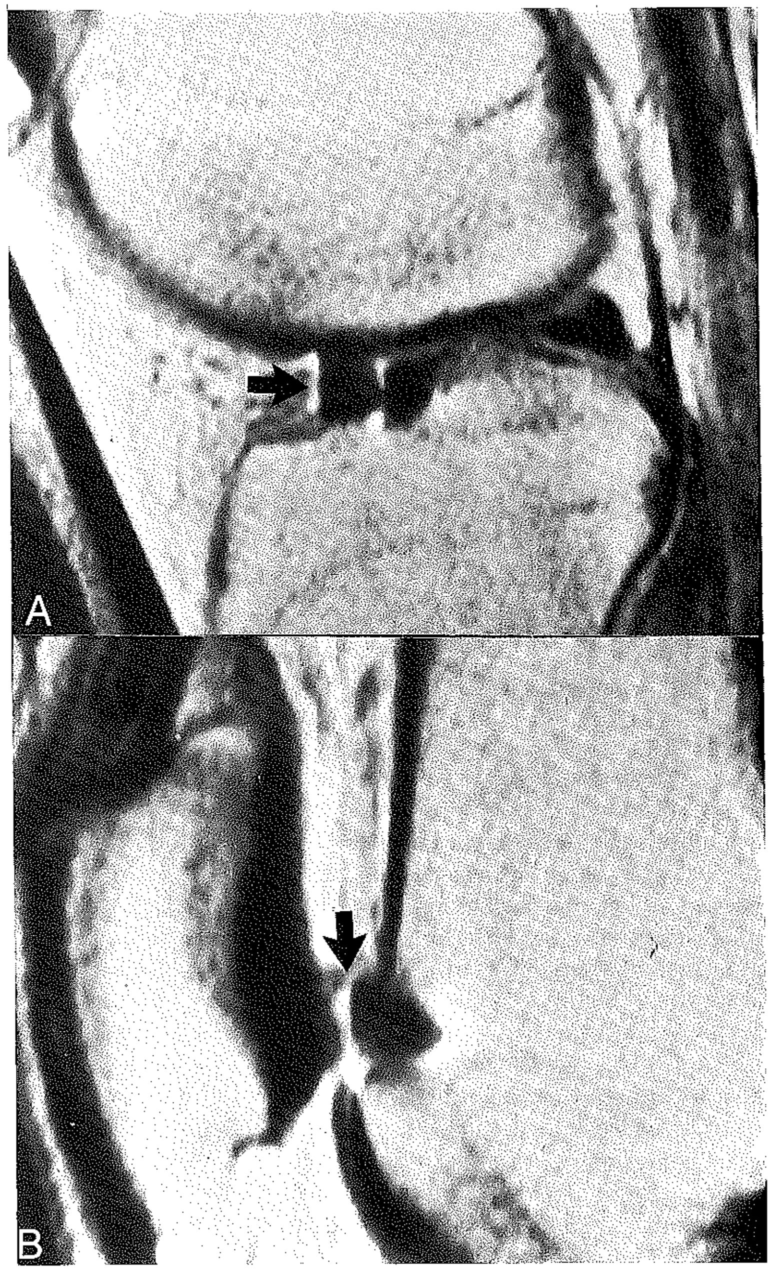


FIG. 6. A sagittal T1-weighted MR image of the ankle of a woman who had had triple arthrodesis. MRI was requested to evaluate the talar dome for osteonecrosis. By aligning the frequency-encoding axis in the anterior-to-posterior direction, artifact from the screws did not interfere with acquisition of diagnostic images. The talar dome bone marrow (arrow) is normal.



FIGS. 7A and 7B. Sagittal T1-weighted MR images of the knee. The patient had had previous arthroscopic surgery for meniscal repair. (A) Metal artifact obscures the anterior horn of the meniscus (arrow). (B) Artifact (arrow) is also seen within the joint, posterior to the patella.

changes in the case of resistive magnets. Environmental interference may be secondary to large metallic objects far removed from the scanner (*e.g.*, passing automobiles) or small metallic objects introduced into the scanner with the patient (Fig. 5). Even when metallic objects are not attracted to the magnet and therefore present no potential danger to the patient during scanning, severe image degradation may occur secondary to minute amounts of impurities. A metal artifact characteristically appears as a region of signal

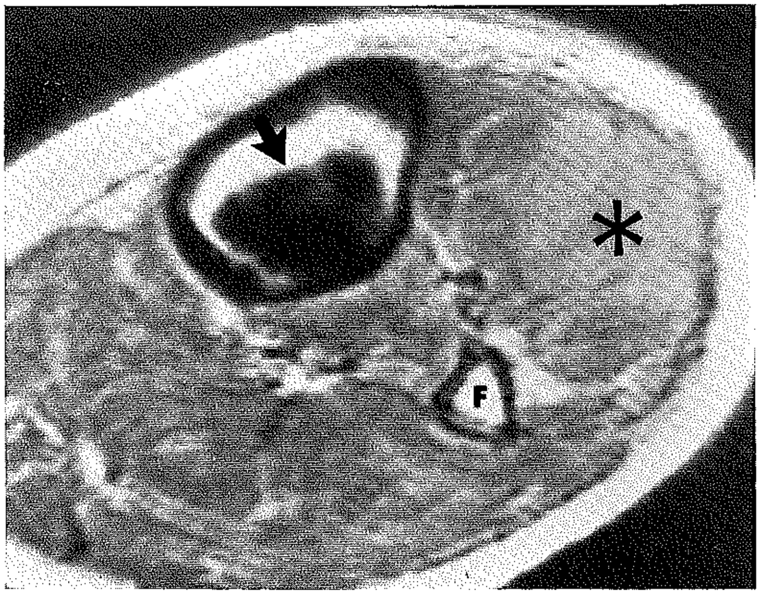


FIG. 8. Methylmethacrylate. An axial T1-weighted MR image through the proximal tibia of a patient who had had limb salvage procedure for osteogenic sarcoma. Methylmethacrylate fixating the tibial component is shown as a signal void (arrow). Recurrent tumor (asterisk) is present within the soft tissues. F = fibula.

void with adjacent very high-signal intensity. The geometric image distortion from the magnetic field nonuniformity caused by ferromagnetic materials is worse along the frequency-encoding axis. In some cases, artifacts produced by internally implanted metallic materials, such as prostheses and fixation devices, can be minimized or altered so as to allow diagnostic examination of the area of interest (Fig. 6). Therefore, it is often essential for the radiologist to have an adequate patient history and access to plain roentgenograms at the time of the examination.

It is not unusual to encounter significant artifacts in images of patients who have had prior orthopedic procedures during which no metallic foreign bodies were implanted and whose roentgenograms display no evidence of retained metal (Fig. 7). These artifacts may be caused by minute metallic particles left from surgical instruments.[7] Similarly, artifacts are also encountered in images of patients who have had their internal fixation devices removed. In these cases, the metallic fragments may have been left at the time of the original operation or may have been left within pin, nail, rod, or screw holes after removal of the appliance.

Methylmethacrylate does not cause an MR artifact[7] and is imaged as a region of signal void (Fig. 8).

Radiofrequency Field Artifacts

The second type of artifact, RF artifact, is produced because MR imaging uses RF pulses that are within overlapping frequencies from many common sources, including commercial radios, computers, and a variety of motors. While great efforts are made to shield scanning systems from these extraneous sources, interference is not uncommon. The appearance of RF interference will depend on the specific frequency and the bandwidth of the outside source (Fig. 9). In cases of wide bandwidth interference, the entire image will be distorted. A narrow bandwidth source will be evident in the image as a line perpendicular to the frequency-encoding axis. The exact position of this line will be determined by the frequency of the noise, the resonant frequency of the scanning system, and the strength of the frequency-encoding gradient.

Gradient Field Artifacts

Gradient field artifacts arise within the three gradients used to construct the two-dimensional MR image. In general, the phase-encoding axis is the most resistant to artifacts from inherent inhomogeneities within the static magnetic field.

Two common gradient-induced artifacts include chemical-shift misregistration artifact and aliasing. Chemical-shift misregistration artifact is found at the interface between tissues with a high fat content and those with a high water content. Because the protons in fat are more tightly bound than those in water, their resonance frequencies differ. The effect is a localized misregistration in the formation of the image at the fat–water junction. Dark and light bands are found at this interface, which are offset along the frequency-encoding axis. In musculoskeletal imaging, chemical-shift misregistration artifact is often seen at the interface between facial planes containing loose areolar tissue and adjacent muscle. This effect is more pronounced at high field strengths.

FIG. 9. An RF artifact from an unidentified source. An axial T1-weighted MR image of the supraclavicular region. A band of noise is present at the bottom of the image. The artifact is outside the region of interest. The location of the artifact in the image is dictated by the frequency of the outside source, and the width is determined by the bandwidth of the offending interference.

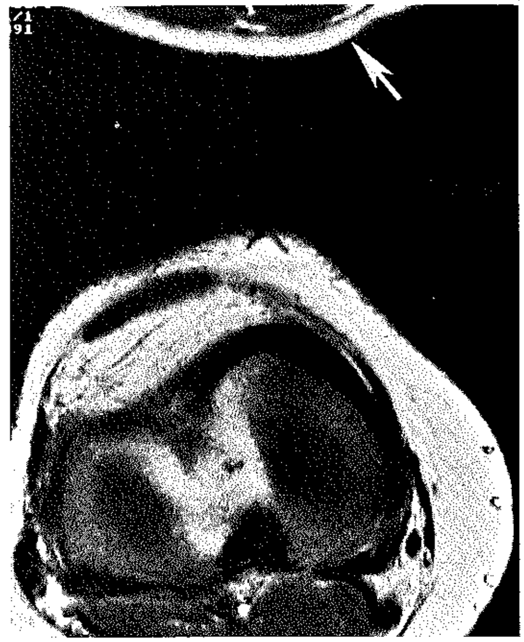


FIG. 10. A wraparound artifact in an axial T1-weighted MR image of the knee. Anterior is at the top of the image. The posterior aspect of the knee was outside of the imaging field and can be seen at the very top of the image (arrow). If a larger body part had been imaged, structures would have been superimposed.

Aliasing, or wraparound artifact, is found in situations in which the body part being imaged is larger than the field of view chosen for image acquisition (Fig. 10). Those parts of the subject that are outside of the imaging field will be folded back in the image and superimposed on the area of interest. Because filters can be used to eliminate this artifact from the frequency-encoding axis, aliasing is usually found only along the phase-encoding axis.

MOTION ARTIFACTS

Motion artifact is overall the most common cause of MR image degradation. Unlike CT scanning, in which one image is acquired at a time, MRI data are incorporated into all images simultaneously. Therefore, any type of motion will degrade all images in a given imaging sequence. Because of the differing sensitivities of the MR image axes to motion,

motion artifacts occur along the phase-encoding axis regardless of which direction the movement takes.

Patient motion, even though seemingly minimal, often serves to render the MR examination useless. There are also several unavoidable types of physiologic movement, such as cardiac, respiratory, and vascular motion. Cardiac and respiratory motion may degrade image quality when scanning the spine but are generally of no consequence in imaging extremities. Vascular flow may at times be evident in the imaging of extremities. Its pulsatile nature may cause linear bands of noise along the phase-encoding axis. Flow may be evident as foci of very high-signal intensity within a vessel, an effect that is seen when unsaturated blood enters the imaging region. Because this volume of blood has not previously been subjected to RF pulses, it has a higher signal than the blood in the central portion of the imaging volume. As the center of the scan is approached, the high-signal intensity representing flow will no longer be evident.

SURFACE COILS

Quite often, diagnostic MRI requires high-resolution scanning of relatively small body parts. Simply magnifying images obtained with the imaging system's built-in body coil will only give bigger pictures without an increase in the signal-to-noise ratio. In order to improve image quality, a wide variety of surface coils is now in clinical use. These coils are small RF receivers that are wrapped around or placed on the body part of interest. By excluding signals generated from tissues outside the region of the coil, thin-section, small field-of-view, high-resolution images can be acquired.

DISCUSSION

As experience with MRI increases, the applications of this modality to the musculoskeletal system will continue to soar. However, because of the possibility of obscuring diagnostic information by the wrong choice of pulsing sequences or imaging planes, MRI provides an additional challenge. There are few situations in which scanning is routine, and imaging must be tailored to each patient's problem. It is essential for the radiologist to have knowledge of the clinical history and suspected pathology for each individual patient, and preliminary plain roentgenograms are often necessary for appropriate image acquisition and interpretation.

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