Review Article

The Biophysical Basis of Tissue Contrast in Extracranial MR Imaging

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So that radiologists can understand better the differences between tissues that determine MR image contrast, the biophysical basis of proton-MR tissue contrast is reviewed. Differences in the molecular sizes of water, fat, and protein macromolecules affect the rate of molecular motion, which in turn influences relaxation times. Differences in the physical state of water within tissues determine the extent to which the motion of water is restricted by binding to hydrophilic macromolecules (e.g., protein), which in turn causes variable amounts of T1 and T2 shortening. Understanding these concepts and knowing the location in the body of free water, cellular tissues, fat, collagen, bone, and tissue iron improve the radiologist's ability to extract useful information from MR images.

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Because of outstanding soft-tissue contrast, MR imaging has become a valuable technique for imaging many areas of the body. Although much has been written about the pulsing sequences and physical principles of MR imaging, little emphasis has been placed on the biological basis of image contrast. The striking soft-tissue contrast available with MR is due primarily to large differences in T1 and T2 relaxation times between various tissues; proton density is thought to have less effect in most soft tissues [1]. These differences in relaxation times can be correlated with differences in tissue composition. Information obtained as a result of extensive basic research in this area can be applied directly to understanding signal intensities in clinical MR images.

Mechanisms of Relaxation

Signal in proton MR depends on the local environment of a proton, which affects the rate of return to equilibrium of the proton's magnetization after it is perturbed by an RF pulse [1]. At equilibrium, macroscopic magnetization is in the longitudinal axis. Immediately after the RF pulse, some of this magnetization (100% if the flip angle is 90°) is transferred into the transverse plane. Both the longitudinal and transverse components of magnetization return to equilibrium by different mechanisms, with different time constants. Longitudinal magnetization is recovered by spin-lattice relaxation, which is described by the T1 relaxation time. Spin-lattice (T1) relaxation involves transfer of energy to the environment (lattice) from spin systems excited by the RF pulse. Transverse magnetization decays because of spin-spin relaxation, described by the T2 relaxation time, which involves the loss of spin coherence among excited nuclei. These relaxation processes occur by specific interactions between nuclei, which are modulated largely by molecular motion and the resultant fluctuation of local magnetic and electrical fields.

The frequency distribution of molecular motion within a sample is described by the correlation time, Tc, which is the average time between collisions for a "typical" molecule within a sample [2]. Two or more correlation times are used if motion about one axis is preferred and if both translational and rotational motions are to

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be described. Because of molecular motion, the local magnetic fields fluctuate and interact with spin systems excited by the RF pulse (macroscopic magnetization). As a result, only motion with a frequency equal to or double the Larmor resonance frequency can contribute to both T1 and T2 relaxation. Slower molecular motion, however, contributes to T2 alone, causing T2 of biological tissues to be significantly shorter than T1 [3, 4].

In nonviscous liquids, molecules move rapidly, described by a short Tc, and have a wide and evenly distributed frequency range. The component with frequency equal to the Larmor frequency is thus relatively weak, and the resulting relaxation is ineffective. Such random and rapid motion averages local fields almost to zero. As a result, T1 and T2 are equal to each other, independent of the Larmor frequency and inversely proportional to the Tc of the liquid [3-6]. T1 and T2 relaxation are more effective in viscous liquids and biological soft tissues, which have longer correlation times.

T1 is minimum when Tc is equal to the inverse of the resonance frequency; in this situation the component of motion with frequency equal to the Larmor frequency is largest. The biological tissue with Tc closest to the inverse of the resonance frequency is adipose tissue, which therefore has a short T1 [1].

Tc is long in macromolecules such as protein, resulting in ineffective spin-lattice relaxation, and thus long T1. The low frequency motion of macromolecules contributes to spin-spin relaxation, however, resulting in shortening of T2 until its plateau, which is characteristic of a completely rigid solid microstructure [3-7]. Because of the extremely long T1 and short T2 of macromolecules, no signal in MR images arises directly from them [1]. Macromolecules such as protein do, however, influence the magnetic relaxation of adjacent water molecules and are thus an important source of tissue contrast, whether dissolved in fluid collections or part of the structure in an intracellular matrix [3, 5-13].

Several specific mechanisms exist by which a perturbed spin system returns to equilibrium. Three of these are dominant in proton imaging of biological systems [2]: (1) Nuclear dipole-dipole interaction refers to the interaction between the magnetic dipole fields of neighboring nuclei, which fluctuate because of molecular motion. This effect decreases as a power of 6 of the distance between the nuclei. (2) Relaxation via chemical shift anisotropy can occur if electron shielding of the nuclei (chemical shift) depends on the orientation of the molecule in the magnetic field. This effect may be an important cause of the extremely short T2 of collagenous tissues such as tendons [14]. (3) Paramagnetic substances, which have unpaired electrons, induce relaxation by their electron magnetic moments or by transfer of unpaired electrons to the relaxing atom itself [15]. Dipole-dipole interaction from this mechanism is extremely powerful because the electron magnetic moment is approximately 1000 times stronger than the nuclear magnetic moment. Thus, even small amounts of paramagnetic substances can decrease relaxation times significantly.

An important concept in MR imaging is that of magnetic susceptibility, which indicates the degree to which a substance alters the applied magnetic field [16]. Inhomogeneous magnetic susceptibility within a tissue results in nonuniform alteration of the magnetic field, causing water protons to precess at different rates. To some extent the resulting phase spread can be corrected by the 180° refocusing pulse applied in spin-echo imaging. A refocusing pulse cannot, however, correct the phase spread of water molecules that diffuse to regions of different field strength during the pulse sequence. The resulting loss of phase coherence causes an observed decrease in effective T2 relaxation time. This phenomenon increases with the square of the strength of the main magnetic field [16]. When a 180° refocusing pulse is not applied, as in pulse sequences that depend on gradient refocusing, the effects of inhomogeneous magnetic susceptibility remain completely uncorrected [16, 17]. This leads to marked signal loss at boundaries between substances of different susceptibility (such as within and adjacent to hematomas, and at interfaces between soft tissue and air and—to a lesser extent—between soft tissue and bone).

The Biological States of Water

In soft tissues other than fat, water is the principal source of MR signal. Water within most tissues differs from that of simple solutions. Although the precise explanation for this is controversial, most investigators agree that biological water exists in at least two states and that this phenomenon leads to different relaxation times among tissues. Some disagreement exists, however, as to the number of states of biological water, the amount of water in each state, the relative contribution of each state to tissue relaxation rates, and the existence of a component of slow exchange between states. In this paper, we present a basic scheme for understanding how the behavior of biological water affects contrast between tissues in MR images, emphasizing points of agreement among investigators. For more detailed formulations of this phenomenon, the interested reader is referred to several in-depth discussions of this topic [3, 5, 6, 9, 18-20].

The two basic states of water often are referred to as free and bound [1, 3, 5-8, 11-13, 18-20]. Free water has relatively unrestricted motion. Extracellular water exists primarily as free water. Bound water has markedly different properties, which are thought to be due, at least partially, to hydrogen bonding with hydrophilic groups within the complex matrix of intracellular proteins and membranes. This interaction restricts the motion of water molecules and shortens T1 and T2 relaxation times.

Many investigators favor a fast-exchange two-phase model. In this model, the free and bound portions of intracellular water are in rapid exchange with each other, and the relaxation rate of tissue therefore reflects a weighted average of its free and bound components [5, 8, 19]. Three-phase models have been proposed also [3, 6].

Tissues with an increased amount of free water have longer T1 and T2 relaxation times. Increased free water is present in tissues with abundant extracellular fluid. Neoplastic tissues also have an increased amount of free water that is only

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partially accounted for by increased extracellular fluid. Neoplastic cells themselves can have increased free water and longer T1 and T2 relaxation times; this is due to increased total cellular water and/or a decreased proportion of intracellular water that is bound to the surfaces of macromolecules [5, 11, 12, 21]. T1 relaxation time appears to be correlated with the rate of cell division and the phase of mitosis, even when total cell water is constant. T1 of synchronized cell lines is shortest during the S phase, when nuclear microstructure is complete, whereas T1 is longest during the M phase, when chromosomes are most condensed and chromosomal surface area is therefore at a minimum [21].

The Effect of T1 and T2 Relaxation on MR Image Contrast

After a 90° RF excitation pulse, longitudinal magnetization recovers at a rate determined by the average T1 relaxation time of the tissue. Tissues with short T1 recover a greater portion of their longitudinal magnetization within a given time interval and therefore tend to have higher signal intensity than tissues with long T1, unless contrast due to differences in T2 and/or proton density predominate. If the time between excitation pulses (TR) is three times longer than the longest T1 among tissues in an image, then recovery of longitudinal magnetization will be nearly complete for all of these tissues. In such an image, little T1 contrast will be present. If, however, TR is short, then tissues with short T1 will have higher signal than tissues with long T1. Such an image can be considered relatively T1 weighted. The optimum TR depends on the T1 values of the tissues of interest and therefore depends on magnetic field strength. Typically, T1-weighted images have TR less than or equal to 600 msec. In the liver, decreasing the TR to approximately half that amount may increase the T1 contrast, at least when imaging at 0.6 T or less [22].

Another imaging parameter that has an important effect on T1 contrast is the excitation pulse angle [23], a full discussion of which is beyond the scope of this article.

If the time between the centers of the excitation pulse and the echo (TE) is short (e.g., 20 msec or less), most spins will still be in phase, and there will be little contrast based on differences in T2. With long TE (e.g., greater than 60 msec), however, tissues with short T2 will experience much dephasing and will have lower signal intensity than tissues with long T2, unless contrast due to differences in T1 and/or proton density predominate. Images in which the primary source of contrast is differences in T2 can be considered relatively T2 weighted.

For optimum T1 weighting, both TR and TE should be short, although the minimum TE often is set by technical limitations. For optimal T2 weighting, TR and TE should both be long, although TR often is limited by imaging time constraints and TE by reduced signal that results from increasing TE. Images with long TR and short TE have less T1 and T2 contrast and are therefore frequently referred to as "proton-density weighted," although this term does not necessarily reflect the combination of T1, T2, and proton-density differences that govern contrast in these images.

Classification of Tissues and Body Fluids Based on MR Signal Characteristics

Most extracranial soft tissues have similar proton density [1, 24]. T1 and T2 relaxation times vary greatly, however, and tissues therefore can be separated into seven classes on the basis of the preponderant material that determines their appearance on MR images (Table 1, Fig. 1).

Free Water

Extracellular water and neoplasms are rich in free water. Extracellular water can be found in edematous areas; urine (e.g., kidney, bladder); tubules (e.g., testes, prostate); follicles (e.g., ovaries, thyroid); stagnant blood (e.g., venous plexes, spleen, cavernous hemangiomas, penis); and CSF, intervertebral disks, cysts, necrotic areas, and fluid collections (Table 2). These tissues and body fluids, as well as neoplasms, usually will have low signal on T1-weighted images (except on magnitude inversion-recovery images with short inversion time, in which tissues with long T1 have high signal intensity [25]) and high signal on T2-weighted images. The ability to separate fluids or tissues with abundant free or extracellular water from more densely cellular tissues on the basis of signal characteristics is one of the most important advantages MR has when compared with other imaging techniques.

Some neoplasms cannot be clearly distinguished from the organ of origin. Neoplasms that are hypocellular and desmoplastic can have low signal on T2-weighted images because their signal characteristics may be dominated by the fibrous tissue [26]. Neoplasms that originate in organs that have long T1 and T2 relaxation times because of abundant extracellular water, such as kidney [27] or spleen [28], also may have less contrast with the organ of origin. In some instances, such as in the prostate, carcinoma may be manifested as focally decreased signal [29].

Proteinaceous Fluid

Protein in aqueous solutions enhances spin-lattice relaxation and therefore shortens T1. Proteinaceous fluid, such as that found in synovial fluid, abscesses, and proteinaceous cysts, therefore will have shorter T1 relaxation times than "simple" fluids, such as urine or CSF, and will have signal intensity similar to that of normal cellular tissues on T1-weighted images [10–13]. Similarly, neurofibromas, which have a large amount of extracellular water contained in a myxoid matrix rich in mucopolysaccharide molecules, have a shorter T1 than muscle [30].

Protein molecules in solutions are relatively unstructured and randomly distributed rather than fixed in intracellular membranes and organelles. This may explain why the amount of T1 and T2 shortening by protein in solution is similar. This is to be distinguished from normal intracellular water, in which T2 is much shorter than T1. Consequently, proteinaceous fluid has high signal intensity on T2-weighted images. In fact, with most T2-weighted pulse sequences in clinical use (which do not have a long enough repetition time to allow full recovery

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TABLE 1: Classification of Tissues and Body Fluids Based on MR Signal Characteristics

Tissue or Body Fluid	Examples	Signal Intensity	
		T1	T2
Hard tissue	Cortical bone Calculi Tendons Ligaments Fibrocartilage Scar	Low	Low
Fat	Adipose tissue Bone marrow	High	Intermediate to high
Cellular tissue	Liver Pancreas Adrenal gland Muscle Nonneoplastic tumors Hyaline cartilage[a]	Intermediate	Low to intermediate
Free water	See Table 2	Low	High
Proteinaceous fluid	Abscess Synovial fluid Some cysts	Intermediate	High
Hemorrhage	See Table 3	Variable	Variable
Blood flow		Variable	Variable

Note.—T1 = T1-weighted images; T2 = T2-weighted images.
[a] Hyaline tissue is not a cellular tissue, although it may resemble cellular tissues on spin-echo MR images.

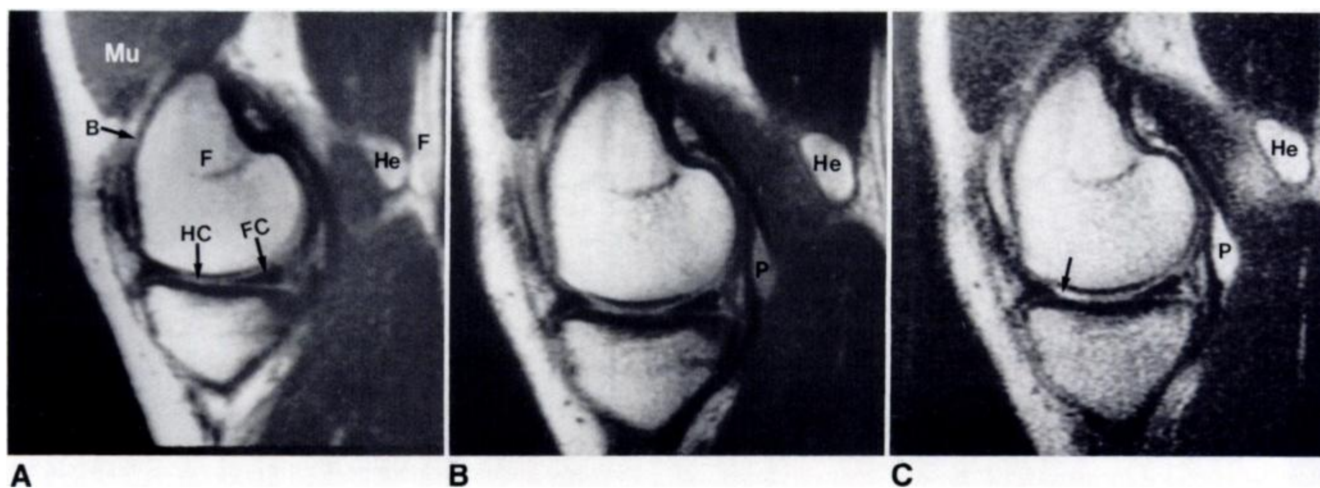

Fig. 1.—Sagittal MR images (1.5-T Signa; General Electric, Milwaukee) of a posttraumatic knee show different signal characteristics of cortical bone, fibrocartilage, muscle, fat, hyaline cartilage, hematoma, and proteinaceous fluid.

***A*, TR = 600 msec; TE = 20 msec. In this relatively T1-weighted image, "hard tissues," such as cortical bone (B) and fibrocartilage in the medial meniscus (FC), are depicted as signal voids. Soft tissues, such as muscle (Mu), have slightly higher signal intensity. Hyaline cartilage (HC), which is composed of water in a mucopolysaccharide matrix, has a shorter T1 and therefore has intermediate signal intensity. A posterior hematoma (He) has an even shorter T1, presumably because of the paramagnetic effects of methemoglobin, and therefore has higher signal intensity. Fat (F) in the bone marrow and subcutaneous tissues also has a short T1 and has signal intensity similar to that of the hematoma.**

***B*, TR = 2000 msec; TE = 20 msec; in a similar plane to *A*. Less T1 contrast occurs because of longer TR. Proteinaceous cyst (P) has intermediate signal intensity, similar to hyaline cartilage. He = hematoma.**

***C*, TR = 2000 msec; TE = 60 msec; in a plane identical to *B*. In this relatively T2-weighted image, hematoma (He) has higher signal intensity than fat because of its longer T2. Synovial fluid (*arrow*) and proteinaceous fluid (P) also have long T2 relaxation times and therefore have higher intensity than more highly structured articular cartilage. Fat has shorter T2 relaxation than these fluids and therefore has lower signal intensity.**

of longitudinal magnetization for all tissues) proteinaceous fluid may have even higher signal intensity than simple fluid (e.g., urine) [31]. This is because the large difference in T1 relaxation in this case is dominant over the smaller difference in T2 relaxation, even on T2-weighted images.

Soft (Cellular) Tissues

Much of the water in soft tissues (e.g., pancreas, liver, adrenal, muscle) is intracellular and therefore is affected significantly by hydrogen bonding to cellular macromolecules.

TABLE 2: Tissues or Body Fluids with Abundant Free or Extracellular Water

Tissue or Body Fluid	Location
Urine	Kidney
	Bladder
Interstitial water	Edematous areas
Tubules	Testes
	Prostate
	Seminal vesicles
Follicles	Ovaries
	Thyroid
Stagnant blood	Spleen
	Venous plexuses
	Penis
	Cavernous hemangiomas
Neoplasms (nondesmoplastic)	Any location
Other	Cysts
	Fluid collections
	Necrotic areas
	CSF
	Intervertebral disks
	Synovial fluid

Cellular tissues therefore have shorter T1 and especially short T2 relaxation times relative to tissues primarily composed of extracellular water. Their spin-echo intensities are intermediate between fat and bone on both T1- and T2-weighted images.

Cellular tissues that are active in protein synthesis and therefore contain abundant endoplasmic reticulum (e.g., pancreas and liver) have an especially large intracellular surface area and thus have especially short T1 relaxation times [20]. Muscle, which is less active in protein synthesis, has a rigid matrix of actin and myosin that apparently is less effective in shortening T1 relaxation than it is in shortening T2 relaxation. This may be the reason that muscle has lower signal intensity on both T1- and T2-weighted images than most other soft tissues [20].

Certain benign tumors composed of normal cells, such as focal nodular hyperplasia of the liver [32] or nonhyperfunctioning adenoma of the adrenal gland [33, 34], have relaxation characteristics similar to that of normal cellular tissues. Like normal soft tissues, these tumors typically will have signal intermediate between that of bone and fat on both T1- and T2-weighted images.

Hyaline cartilage is a unique tissue that is rich in extracellular water bound to a mucopolysaccharide matrix [35]. Presumably because of extensive hydrogen bonding between water and protein molecules, hyaline cartilage has relaxation characteristics similar to cellular tissues and has intermediate signal on most MR images. Fibrocartilage, with less water content and a higher concentration of oriented collagen fibers, has a long T1 and a very short T2. Fibrocartilage therefore has low signal intensity on both T1- and T2-weighted images [36, 37] and can be considered a "hard" tissue.

The T1 shortening effect of macromolecules in biological tissues appears to be less pronounced at higher magnetic field strength [3, 5, 11, 12, 38], which may theoretically diminish T1 contrast between water-containing tissues. In most cases, T2 relaxation is relatively independent of magnetic field strength [5, 11, 38].

Hard (Mineral-Rich or Collagenous) Tissues

Hard tissues have low signal intensity on virtually all MR images. Although the explanation for the low signal of mineral-rich tissues is different from that of collagenous tissues, these tissues can be considered together because of their similar appearance on MR images.

Mineral-rich tissues, such as bone or calculi, contain few mobile protons and consequently result in little signal regardless of the pulse sequence used. Collagenous tissues (e.g., tendons, ligaments, fibrocartilage, or fibrosis) also contain less water because of their high proportion of protein molecules. Decreased density of water cannot by itself, however, account for the extremely short T2 and long T1 relaxation times observed for collagenous tissues [8, 14] or the extremely low signal observed clinically on both T1- and T2-weighted images [36, 37, 39].

Some investigators think that the extremely short T2 relaxation time of collagen is a result of the orientation of collagen fibers, which causes motion of adsorbed water molecules relative to the magnetic field to be anisotropic. This phenomenon is thought by some investigators to account for the extremely short T2 times observed for collagenous tissues [8, 14].

Fat (Adipose Tissue)

Although in vivo measurements have disclosed a slight but significantly higher proton density of adipose tissue relative to some other soft tissues [24], differences in T1 and T2 relaxation times are more important in most images. Most protons in fat are present within hydrophobic CH_2 groups contained in intermediate-sized molecules, which tumble at a rate close to the Larmor frequency. Interaction with the environment is efficient, and T1 relaxation is short [1]. Signal therefore will be high in most T1-weighted images.

Because of the hydrophobic nature of fat, hydrogen bonding to macromolecules does not affect its physical or magnetic state [12, 13]. T1 relaxation times for most soft tissues increase with higher field strength because the T1 shortening effects of intracellular macromolecules are less pronounced. This is not true for fat, however [13, 38]. At higher magnetic field strength, T1 differences between fat and most soft tissues therefore may be increased, resulting in higher contrast between fat and water on T1-weighted images.

Protons within fat have a shorter T2 than free water because of the larger size and longer correlation time of lipid molecules. Because of its shorter T2, the signal from fat will be decreased with respect to water on images that are sufficiently T2 weighted. On many "T2-weighted" sequences, fat may appear isointense or hyperintense relative to water (e.g., urine or cyst fluid) [40]. This is presumably because image contrast in these sequences is dominated by the large differences in T1 between fat and water, which oppose the differences in T2. Decreased signal of water in T2-weighted

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images also may be due to off-resonance effects from selective RF pulses, which result in decreased signal intensity, especially in tissues with long T2 (e.g., free water) [41].

Fat is also a unique tissue in that most of its signal arises from protons within CH_2 groups. These protons precess approximately 3 ppm slower than protons in water. This phenomenon also can be used in various techniques to separate fat from water [42, 43].

Hemorrhage

Iron atoms within hemorrhage can exert powerful paramagnetic and/or susceptibility effects, which depend on the physical and oxidative state of hemoglobin [16] (Table 3). Hemoglobin in oxygenated blood has only one unpaired electron and is therefore only minimally paramagnetic. Thus, oxygenated blood has relaxation characteristics similar to free water, accounting for the appearance of sluggish flowing blood (e.g., spleen [28], cavernous hemangiomas [44], venous plexuses [29, 31]) on T1- and T2-weighted images and causing potential difficulties in differentiating fresh hemorrhage (less than 1 day old) from other fluid [45].

Deoxyhemoglobin in unoxygenated blood is paramagnetic because of the four unpaired electrons in the ferrous (Fe^{2+}) ion [16]. These unpaired electrons, however, are not physically accessible to water molecules. Therefore, deoxyhemoglobin has little if any effect on spin lattice (T1) relaxation [16, 45–47]. Intracellular deoxyhemoglobin in acute hemorrhage can decrease signal in T2-weighted images however. This occurs because the slowly moving deoxyhemoglobin molecules within intact RBCs create zones of different magnetic susceptibility. The resulting field gradients are partially compensated for by a 180° refocusing pulse used in the spin-echo technique. A refocusing pulse cannot, however, correct losses in phase coherence resulting from diffusion of water molecules along these field gradients. This results in greater spin dephasing, which shortens the effective T2. The dephasing effect of deoxyhemoglobin increases if field strength is high [16, 46] or if a 180° refocusing pulse is not applied [17]. In these situations, acute hemorrhage will have low signal intensity on both T1- and T2-weighted images [46].

With aging of hemorrhage, RBCs lyse, and deoxyhemoglobin is oxidized to its ferric (Fe^{3+}) form, methemoglobin. Methemoglobin is also paramagnetic, and its five unpaired electrons are more accessible to water molecules than are the unpaired electrons in deoxyhemoglobin [16]. Thus, T1 relaxation of the nearby water molecules is shortened. Subacute

TABLE 3: MR Signal Characteristics of Hemorrhage

Age	Dominant Component	Signal Intensity	
		T1	T2
Fresh (<1 day)	Free water	Low	High
Acute (1–6 days)	Deoxyhemoglobin	Low	Low
Subacute/chronic	Methemoglobin		
(>7 days)	Intracellular	High	Low
	Extracellular	High	High
Scar	Hemosiderin	Low	Low

Note.—Signal intensity at 1.5 T. T1 = T1-weighted images; T2 = T2-weighted images. Age in days is approximate and may vary with location.

(greater than 7 days) hemorrhage therefore typically will have high signal on both T1- and T2-weighted images [16, 46, 47].

If oxidation to methemoglobin occurs before RBC lysis, a transitional signal pattern will be noted. Intracellular methemoglobin has restricted motion and therefore causes spin dephasing in a manner similar to that of intracellular deoxyhemoglobin. If this occurs, both T1 and T2 relaxation will be significantly enhanced, and the hematoma will have high signal on T1-weighted images and low signal on T2-weighted images [16, 31, 46].

If hemosiderin is present because of chronic hemorrhage (or because of iron storage disease such as hemosiderosis [48]), pronounced spin dephasing and T2 shortening will occur owing to inhomogeneous magnetic susceptibility. Signal will be low, even on relatively T1-weighted spin echo images, because of the T2 contrast inherent in the spin-echo technique [16, 46].

Although the signal characteristics of hematomas can vary, in most cases hemorrhage can be distinguished from other tissues if more than one pulse sequence is used. This constitutes another advantage of MR over other imaging techniques. In isolated cases in which high-signal hematomas cannot be differentiated from fat, chemical shift–sensitive techniques may be of benefit.

Blood Flow

T1 and T2 relaxation of blood are determined by its extracellular water content, as evidenced by the low signal on T1-weighted images and high signal on T2-weighted images of the spleen, venous plexuses, and cavernous hemangiomas. Flow can alter the signal intensity obtained, however, depending on multiple factors such as TR; TE; number of echoes; position within a "stack" of images; and speed, pulsatility, and/or turbulence of flow [49]. Further discussion of flow effects is beyond the scope of this review.

Conclusion

In T1-weighted images, fat has such high signal relative to other tissues that these images can be thought of crudely as "fat images"; much anatomic detail is evident because of contrast with the surrounding fat. Similar to CT images, pathologic lesions often are seen as low-signal (MR) or low-attenuation (CT) defects within fat or soft tissue. Subacute or chronic hemorrhage, however, has high signal and therefore can be characterized on T1-weighted images.

In a similar respect, T2-weighted images can be thought of crudely as "water images"; pathologic lesions, by virtue of their high content of free water, frequently can be seen as high-signal "hot spots." Identification of free water is a major advantage of MR, an advantage that can be exploited best by full knowledge of the location of free water in normal and pathologic tissues.

Designation of images as T1-weighted or T2-weighted is a simplification that can lead to image misinterpretation. Virtually all images have contributions from both T1 and T2 contrast, as well as from proton density. Occasionally, T2

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contrast can predominate in T1-weighted images, such as in hemochromatosis in which the extremely low signal on T1-weighted images is due to the pronounced shortening of T2 relaxation by the effects of inhomogeneous susceptibility. Similarly, T1 contrast can predominate in T2-weighted images, causing proteinaceous or hemorrhagic fluid to have higher signal than urine because of shorter T1.

Signal intensity on MR images reflects the physical characteristics of protons in the tissues being imaged. Fat, cellular tissues, hard tissues, extracellular and neoplastic water, hemorrhage, and flowing blood all have distinct signal characteristics and can be separated if the proper pulse sequences are used. An approach to MR image interpretation that is based on knowledge of this classification of tissues can optimize the tissue-characterizing abilities of MR.

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