17 Density and Relaxation Contrast

17.1 Introduction

In the previous chapter we learned how to create a magnetic resonance image. The basic process is to excite the magnetization vector by applying a magnetic field that is constant in the rotating frame (the RF pulse) and then applying a series of magnetic field gradients that cause the precessing frequency to vary as a function of location. This effect is then used to map out the amplitudes of the spatial frequencies in the object. The signal that is detected in the coil was found to be the Fourier transform of the quantity of interest, the spatial distribution of the transverse magnetization, which, in turn, can be related to spin density. Therefore, once the data was collected, it is Fourier Transformed to create the resulting image. What we see as the final "image" is the magnitude of the Fourier Transform of the signal.

What does this image look like? Well, one of the remarkable things about MRI is that the user has a lot of control over the answer to this question. While the MR signal is a result of the water in a voxel, the MR image need not be a representation of the amount of water in the voxel (though such images can be made). MRI is really reflective of the *state* of water, which depends in a complicated way on the structure and physiology of the tissues. One is often concerned with distinguishing different tissue type, in which case the goal is to produce a significant *difference* in image intensity between different tissues, or tissue *contrast*. The goal of MRI is thus to *create* the contrast of interest. For brain imaging, for example, one might want to create contrast between gray and white matter, or between functionally activated and unactivated regions of the brain, or perhaps between tissues that have different diffusion properties. In standard (i.e., clinical) imaging, the goal is typically to create contrast between different organs or tissue types.

One way in which the tissue structure is reflected in the signal is through the signal decay or recovery, or *relaxation*, of the signal. The most straightforward, and most common, way to create tissue contrast is to exploit the different in tissue relaxation properties (and water content). In this chapter, we show how to do this. We'll find that this can be accomplished in a very straightforward manner using the basic spin echo sequence by varying the timing parameters of the pulse sequence. Since the signal depends so strongly on the relaxation parameters, it is not surprising that changing the timings of the pulses will cause the resulting image to be affected by the relaxation rates. Using this dependence to advantage allows us to create images that depend in a controlled way on the relaxation rates, which, because they are tissue dependent, allow us to create contrast between tissues. It is also possible to directly measure these parameters and display their spatial variations, and we will touch on this briefly below.

17.2 Creating relaxation contrast

While there are many variations on this theme of *relaxation contrast*, the most basic methods for creating relaxation contrast are quite simple to understand, in principle, in the basic spin echo pulse sequence. Creating relaxation contrast in gradient echo images is actually a much more subtle subject, and we'll discuss that in the next chapter.[1]

Relaxation contrast can be understood in term of the Bloch Equations (Eqn **??**), the phenomenological equations for the transverse and longitudinal magnetization in the rotating frame. Recall that these equations, in their most basic form, are derived from the simple vector model of magnetization under the influence of an external static main field and an external field applied at the Larmor frequency (the rotation rate of the rotating frame), with the addition of *phenomenologically* derived exponential relaxation of the longitudinal and transverse components.

17.3 Relaxation during free precession

The time rate of change of the transverse (M_{xy}) and longitudinal (M_z) magnetization in the rotating frame are [2]

$$\frac{dM_{xy}}{dt} = -\frac{M_{xy}}{T_2} \tag{17.1a}$$

$$\frac{dM_z}{dt} = -\frac{M_z - M_z^0}{T_1} \tag{17.1b}$$

where M_z^0 is the longitudinal magnetizaqtion at thermal equilibrium. These can be solved

$$M_{xy}(t) = M_{xy}(0_+)e^{-t/T_2} \tag{17.2a}$$

$$M_z(t) = M_z^0(1 - e^{-t/T_1}) + M_z(0_+)e^{-t/T_1} \tag{17.2b}$$

where $M_{xy}(0_+)$ and $M_z(0_+)$ are, respectively, the transverse and longitudinal magnetizations just *after* the RF pulse.

17.4 Saturation Recovery

Let's consider a most basic pulse sequence: a series of 90° pulses repeated at time intervals T_r, the *repetition time*, as shown in Figure 17.1. In order to analyze this sequence, we can use the results Eqn 17.2. Because we now have multiple pulses - let's say we have n of them - we'll need to consider not only the magnetization after the n'th pulse, which we'll denote $M_z^n(0_+)$, but also the magnetization $M_z^n(0_-)$ *before* the n'the pulse. Similarly, $M_{xy}^n(0_+)$ and $M_{xy}^n(0_-)$ are the transverse magnetizations before and after the n'th RF pulse. If we assume the system is initially in thermal equilibrium, the longitudinal magnetization before and after the first pulse are

$$M_z^1(0_-) = M_z^0 \tag{17.3a}$$

$$M_z^1(0_+) = 0 \tag{17.3b}$$

[1] Need ps diagram.
[2] Liang3.121

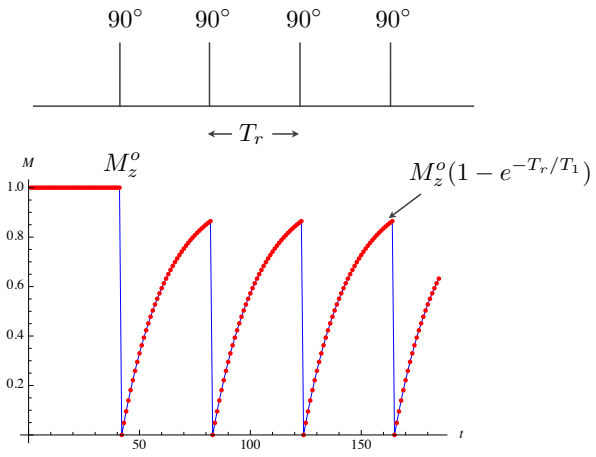

Figure 17.1 The saturation recovery pulse sequence showing the recovery of longitudinal magnetization.

And we know from Eqn 17.2 that the longitudinal magnetization just before the n'th pulse is

$$M_z^n(0_-) = M_z^0\left(1 - e^{-T_r/T_1}\right) + M_z^{n-1}(0_+)e^{-T_r/T_2} \qquad n > 1 \tag{17.4}$$

[3] If, after n pulses, there is no longer any longitudinal magnetization, that is, $M_z^n(0_+) = 0$, then the magnetization is said to be *saturated*, which means that the transverse magnetization created by the 90° pulse has decayed away before the application of the next pulse. This is true if $T_r \gg T_2$ and thus this is known as the *saturation condition*. If this condition is met, then $e^{-T_r/T_2} \approx 0$ and so Eqn 17.4 is approximately

$$M_z^n(0_-) = M_z^0\left(1 - e^{-T_r/T_1}\right) \qquad n > 2 \tag{17.5}$$

so the n'th pulse longitudinal magnetization is just dependent upon constants: the thermal equilibrium value, the pulse sequence repetition time T_r, and the T_1. The sequence is thus said to be in the *steady state*. The sequence in Figure 17.1 is thus called a *saturation recovery*, or *SR*, sequence. The transverse magnetization at the n'th pulse is thus just Eqn 17.5 tipped into the transverse plane:

$$M_{xy}^n(0_+) = M_z^0\left(1 - e^{-T_r/T_1}\right) \qquad n > 2 \tag{17.6}$$

and thus image intensity, which is proportional to the transverse magnetization, is of the form

$$I \propto \rho(\boldsymbol{r})\left(1 - e^{-T_r/T_1(\boldsymbol{r})}\right) \tag{17.7}$$

where we've included the spatial dependence $(\boldsymbol{r})$ of both the relaxation and the spin density. Thus the image intensity in a saturation recovery sequence is proportional to the spin density

[3] The last T_2 is written as T_1 in Liang, which must be an error!

and is weighted by T_1. If the repetition time is made long relative to T_1, $T_r \gg T_1$, then the image intensity is weighted by the proton density:

$$I \approx \rho(\boldsymbol{r}) \tag{17.8}$$

However, if the repetition time is made short relative to T_1, $T_r \ll T_1$, then the image intensity is heavily weighted by T_1 through Eqn 17.7.

17.5 Inversion Recovery

Consider, for a moment, the longitudinal magnetization M_z following excitation by a 90° pulse in the saturation recovery sequence shown in Eqn 17.1. The 90° pulse sets the $M_z = 0$, whereupon it recovers at a rate governed by T_1. So notice then that if we applied pulse larger 90° than so that a component of M_z was negative, the T_1 recovery would result in the longitudinal magnetization recovering through $M_z = 0$, which is called a *null point*. Consider the case that we apply an 180° pulse, which converts an initial M_z into $-M_z$, an thus is called an *inversion pulse*. We still need to follow this with a 90° pulse to measure the signal by converting the longitudinal magnetization to transverse magnetization. Let's call the time between the inversion pulse (the 180°) and the measurement pulse (the 90°) the *inversion time* T_i. We then wait a time T_{rec}, the *recovery time* before repeating the sequence. The repetition time is just the time between successive applications of these sequence, and so is just the time from the centers of the 180° pulses, and so is just $T_r = T_i + T_{rec}$. This *inversion recovery*, or *IR*, sequence is shown in Figure 17.2. You can probably immediately guess the utility of this: Since the null point depends upon the T_1, contrast between two tissues that have different T_1's can be created by nulling the signal from one of the tissues. To do this, the inversion time T_i has to be chosen correctly, which is easy to do once we write down the IR signal equations.

First note from that we really only need to look at the signal between 90° pulses, since after each the longitudinal magnetization is set to 0. So, for an initial longitudinal magnetization M_z^0, the first inversion (90°) pulse sets the magnetization to 0, and then it grows according to Eqn 17.4 and thus at a time $T_r - T_i$ after the 90° pulse when the 180° pulse is applied, the longitudinal magnetization has grown to

$$M_z(t_{180-}) = M_z^0 \left(1 - e^{-(T_r - T_i)/T_1}\right) \tag{17.9}$$

This is inverted by the second 180° pulse, and this then $M_z(t_{180+}) = -M_z(t_{180-})$. In the time interval T_i between the 180 and the second 90, the magnetization recovers according to Eqn 17.4:

$$\begin{aligned} A_{ir} = M_z(t_{90-}) &= M_z^0 \left(1 - e^{-T_i/T_1}\right) + M_z(t_{180+}) e^{-T_i/T_2} \\ &= M_z^0 \left(1 - e^{-T_i/T_1}\right) - M_z^0 \left(1 - e^{-(T_r - T_i)/T_1}\right) e^{-T_i/T_2} \\ &= M_z^0 \left(1 - 2e^{-T_i/T_1} + e^{-T_r/T_1}\right) \end{aligned} \tag{17.10}$$

And this is the signal equation for the inversion recovery sequence. Now we can find the value of T_i that will null a tissue a with longitudinal relaxation time T_1^a - we just set the term in parentheses in Eqn 17.10, with $T_1 = T_1^a$, equal to zero, and solve for T_i, and call it T_i^a. This gives

$$T_i^a = T_1^a \left[\ln 2 - \ln\left(1 + e^{-T_r/T_1^a}\right)\right] \tag{17.11}$$

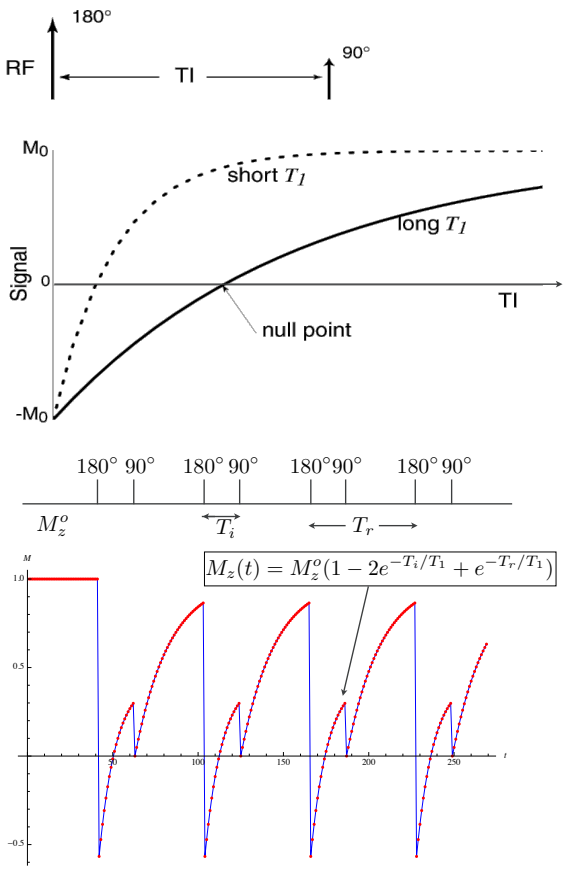

Figure 17.2 The inversion recovery pulse sequence.

An example of a sequence with the inversion time set to null CSF is shown in Figure 17.3.

17.6 Spin echo imaging

The standard spin echo sequence consists of a 90° pulse followed by a 180° pulse at a time $T_e/2$ later as shown in Figure 17.4 (a stripped down version of Figure 16.30) We already know from the last section that the signal at a time T_e after $90° - 180°$ pulse combination is

$$M_z(0_-) = M_z^0 \left(1 - 2e^{-(T_r - T_e/2)/T_1} + e^{-T_r/T_1}\right) \tag{17.12}$$

The amplitude of the spin echo signal is thus

$$A_{se} = M_z(t_{90-})e^{-T_e/T_2} = M_z^0 \left(1 - 2e^{-(T_r - T_e/2)/T_1} + e^{-T_r/T_1}\right) e^{-T_e/T_2} \tag{17.13}$$

If $T_e \ll T_r$, as is typically the case, then

$$A_{se} \approx M_z^0 \left(1 - e^{-T_r/T_1}\right) e^{-T_e/T_2} \tag{17.14}$$

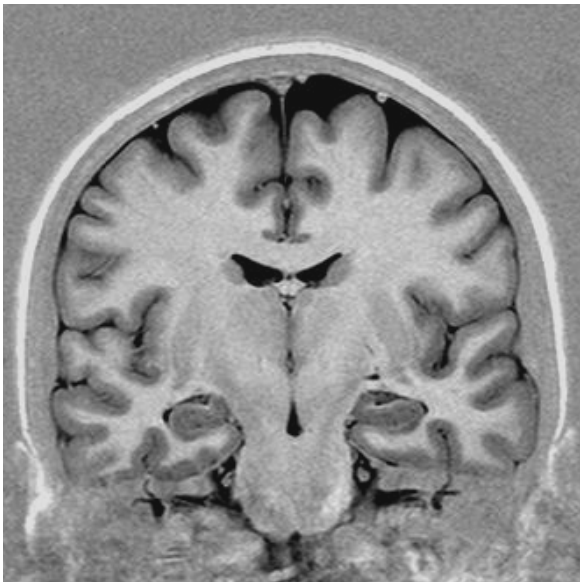

Figure 17.3 An image of the human brain acquired with inversion recovery pulse sequence with the inversion time set to null the CSF. (from www.mr-tip.com)

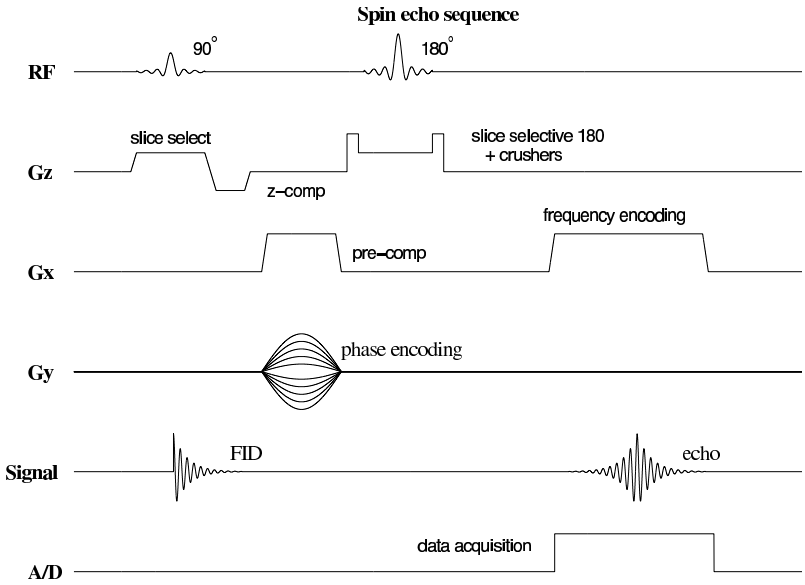

Figure 17.4 The spin echo pulse sequence.

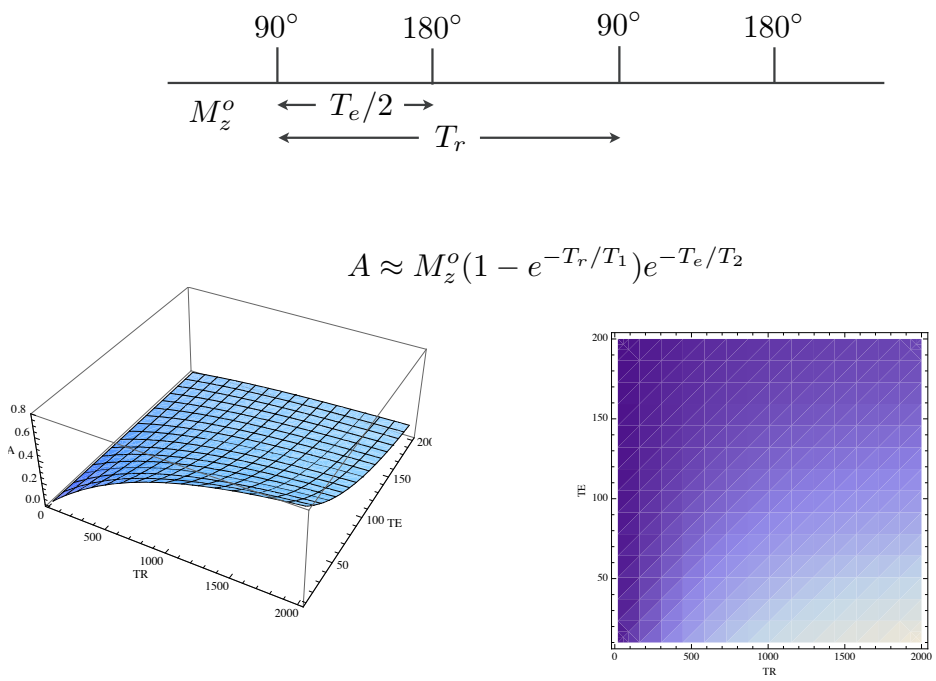

Figure 17.5 The contrast regimes in a spin echo pulse sequence.

Eqn 17.14 shows that the spin echo signal is weighted simulataneously by the spin density (e.g. M_z^0), T_1, and , T_2. These contrast regimes for the spin echo sequence are shown in Figure 17.5 Images from the four standard contrast regimes for the spin echo sequence are shown in images of the knee in Figure 17.6 Variations in image contrast can be created by changing the pulse sequence timing parameters to exploit these contrast mechanisms and selectively enhancing the contrast of the different relaxation mechanisms, as shown in Figure 17.7. The variations in image contrast produced by the manipulation of the imaging parameters to give different weightings is shown in images of the brain in Figure 17.8 .

17.7 Imaging relaxation rates directly

All MR images are intrinsically weighted by the proton density, and the relaxation parameters T_1 and T_2. The standard clinical protocols can emphasize contrast based upon these variations, but it also is possible to directly measure the actual relaxation parameters themselves and construct an image of their spatial distribution. This has important clinical implications because certain tissues, such as cartilage, exhibit relaxation rate changes in tissue degeneration before they exhibit gross anatomical variations (**?**). The way to image the relaxation rates is relatively simple and intuitively clear from Figure **??**[4]. The T_r recovery curve varies appreciably as a function of T_1, and the signal decay with TE depends strongly on T_2. The simplest way to measure T_1 is to

[4] This refers to the bottom left and bottom right figures, which should be made into their own subfigures and referenced accordingly.

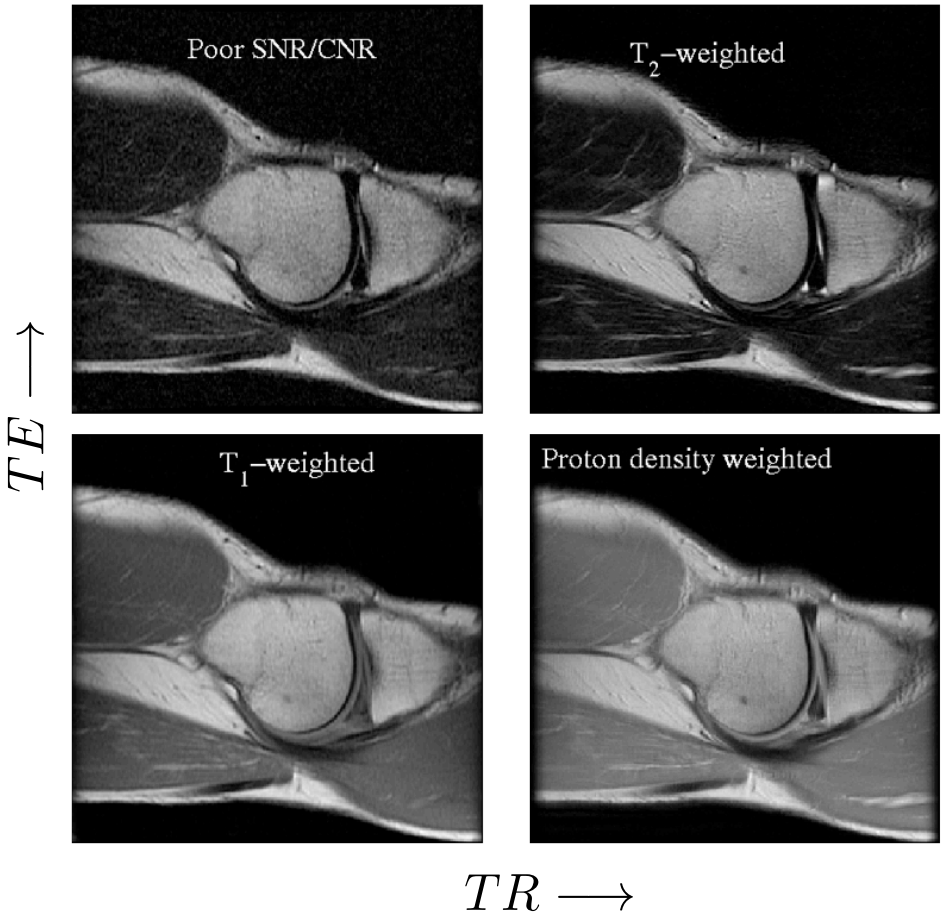

Figure 17.6 Images from the four contrast regimes in a spin echo pulse sequence in the human knee (left) and from the three most useful of these regimes in the human brain (right).

collect images for a range of T_r values and then fit the recovery curve (Eqn **??**) to estimate T_1. The simplest way to measure T_2 is to collect images for a range of TE values and then fit the decay curve (Eqn **??**) to estimate T_2. The result is a spatial map of T_1 and T_2, such as is shown in Figure 17.9 (**?**).

17.8 Dipolar contrast

Dipolar effects are particularly influential in tissues, such as cartilage and tendon, that are composed of large, macromolecular structures that are highly oriented. The water that is tightly bound to such structures, such as the proteoglycan-rich matrix fraction of water in cartilage, is more constrained in its motion than that of free water, and therefore has a longer rotational correlation time than free water. The correlation frequency is approximately proportional to molecular weight (**?**), so that macromolecules will have long correlation times. The long T_1 results

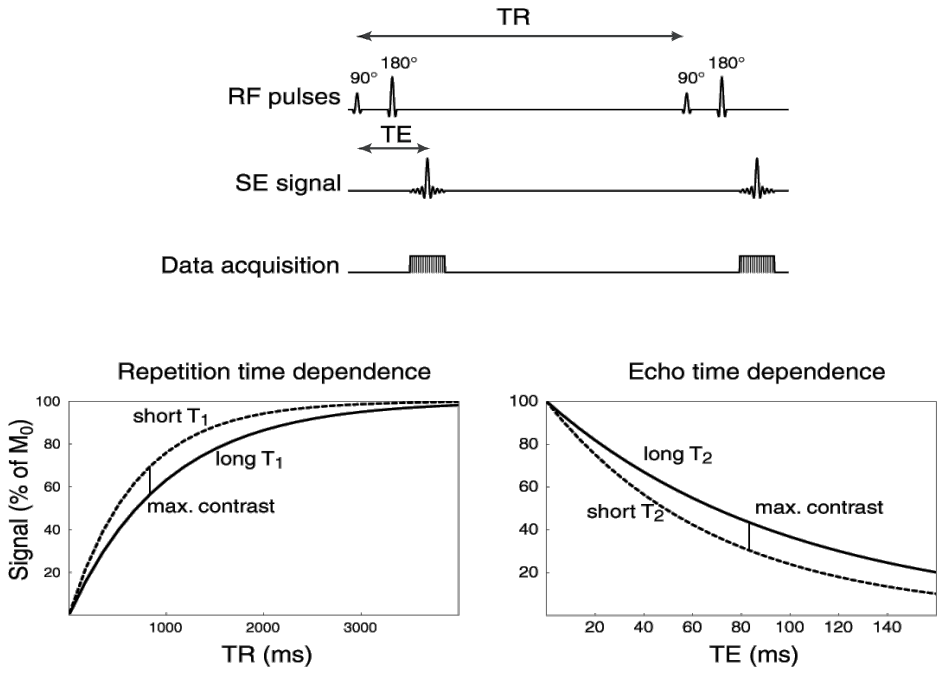

Figure 17.7 The relationship of the pulse sequence timing parameters and the relaxation contrast.

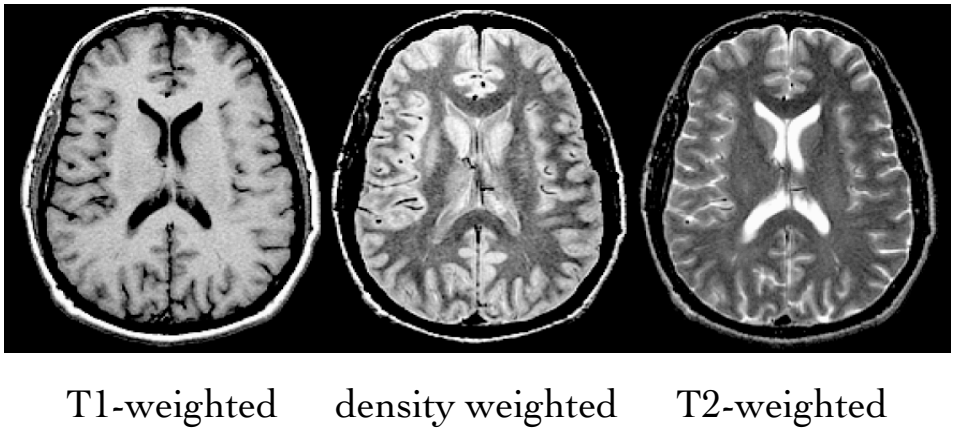

Figure 17.8 Images from the four contrast regimes in a spin echo pulse sequence in the human knee (left) and from the three most useful of these regimes in the human brain (right).

from the fact that few protons are rotating near the resonant frequency at which relaxation is promoted, and the short T_2 is a consequence of the local magnetic field inhomogeneities whose influence on the "static" field term remains important because of the lack of motional averaging due to its tightly bound state.

There is a curious result as indicated in Eqn 15.1. If the tissue is oriented such that $3\cos^2\theta = 1$,

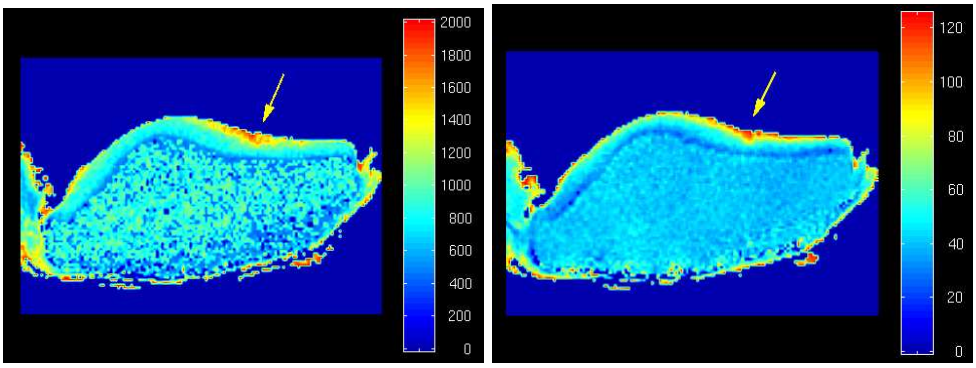

Figure 17.9 Measured relaxation rates (Left) T_1 and (Right) T_2 calculated from a set of high resolution ($FOV = 9.4\,\text{cm}$, thickness $= 5\,\text{mm}$, $256 \times 256, 1NEX$) long T_r double echo images ($T_r = 4000\,\text{ms}$, $TE_1 = 20\,\text{ms}$, $TE_2 = 80\,\text{ms}$), and a short T_r image ($T_r = 600\,\text{ms}$, $TE = 20\,\text{ms}$) in degenerate cartilage specimens. (Reprinted with permission from Frank et al. (**?**))

then $g(\theta) = 0$ and there is no mutual interaction and hence relaxation is not promoted at all! The angle for which this occurs is $\theta = \cos^{-1}\sqrt{(1/3)} \approx 55°$ and is called the *magic angle*. Conversely, the angles $\theta = 0°$ and $\theta = 90°$ both produce a strong influence, with $\theta = 0°$ being the maximum. Therefore a highly structured fibrous tissue such as a tendon can have a very short T_2 (in tendon $T_2(\theta = 0) \approx 100\,\mu\text{s}$) (**?**) but this T_2 value varies according to its orientation in the main field(**?**). At the magic angle, the tissue can appear bright as relaxation becomes longer. This effect is a particular nuisance in fibrous tissues that curve, such as articular cartilage, because the orientation of the tissue relative to the main field changes throughout the tissue(**?**). In many MR imaging applications, such as in brain imaging, the orientation of the subject relative to the main field is rarely given consideration, but in musculoskeletal applications this orientation can be important when looking at fibrous tissues. An example is shown in Figure 17.10.

Suggested reading

An excellent discussion of relaxation contrast is given in (**?**).

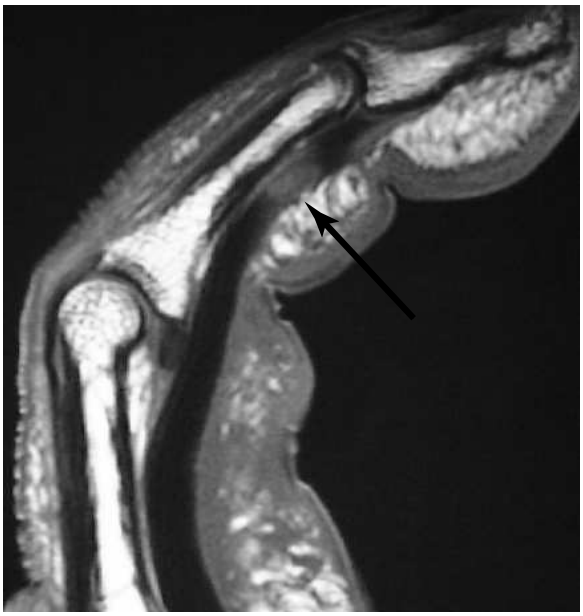

Figure 17.10 Magic angle phenomenon. Focus of abnormal signal intensity (arrow) in the mid portion of the flexor digitorum tendon is due to the magic angle phenomenon. Tendon demonstrates normal morphology. (Image courtesy of Nittaya Lektrakul and Olivier Hauger.)