Chapter 1

Magnetic Resonance Imaging: A Preview

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Introduction

The primary purpose of this chapter is to provide a succinct overview of the basic principles involved in the process of using nuclear magnetic resonance for imaging. This overview is in the form of a list of results, without derivation, in order to provide a story line and goals for the reader to follow in the detailed treatment of this material in the coming chapters.

The chapter begins with an explanation of the name, ‘magnetic resonance imaging,’ or MRI, followed by a short and inadequate history of some of the developments that led to the discovery of key imaging concepts. The third, and largest, section is a preview of the subsequent twenty-six chapters. Finally, some relevant reference texts and articles are listed in the suggested reading section.

1.1 Magnetic Resonance Imaging: The Name

Magnetic resonance imaging is a relatively new discipline in the realm of applied sciences. A main thrust has come from the imaging of soft tissues in the human body and metabolic processes therein, such that it occupies a strong position in biomedical science applications.

\(^1\)A great deal of the history of nuclear magnetic resonance and MRI is presented in volume one of the Encyclopedia of Nuclear Magnetic Resonance which is listed in the historical/review references.
Chapter 1. Magnetic Resonance Imaging: A Preview

MRI is a powerful imaging modality because of its flexibility and sensitivity to a broad range of tissue properties. One of the original reasons for the excitement about MRI was, and continues to be, its relative safety, where the ‘noninvasive’ nature of the magnetic fields employed make it possible to diagnose conditions of people of almost any age. Today MRI also offers great promise in understanding much more about the human body, both its form and its function.

MRI stems from the application of nuclear magnetic resonance (NMR) to radiological imaging. The adjective ‘magnetic’ refers to the use of an assortment of magnetic fields and ‘resonance’ refers to the need to match the (radio)frequency of an oscillating magnetic field to the ‘precessional’ frequency of the spin of some nucleus (hence the ‘nuclear’) in a tissue molecule. It might be more accurate to refer to this field as NMRI rather than MRI, but there is widespread concern over any phrase containing the word ‘nuclear.’ Although the nuclear component simply refers to a benign role of the ‘spin’ of the nucleus in the process, the word has been suppressed and the public and the profession have embraced the MRI acronym.

1.2 The Origin of Magnetic Resonance Imaging

To describe the history of any technological advance in a given field is a very difficult and sensitive issue; to offer a brief and incomplete account is fraught with peril. Still, the beginning student may be aided and inspired by even a short historical discussion. It may be said that MRI had its beginnings in 1973 with the seminal papers by Lauterbur and Mansfield. It was already well known that the intrinsic angular momentum (or ‘spin’) of a hydrogen nucleus (the proton) in a magnetic field precesses about that field at the ‘Larmor frequency’ which, in turn, depends linearly on the magnitude of the field itself. Their idea was very simple. If a spatially varying magnetic field is introduced across the object, the Larmor frequencies are also spatially varying. They proposed and showed that the different frequency components of the signal could be separated to give spatial information about the object. This key point of spatially encoding the data opened the door to MR imaging. Others also recognized the importance of this area, with early attention brought to tumor detection by Damadian.

Something may be learned here by the beginning student. Often the basic step toward new developments, which may become quite complicated as a whole, is a simple connecting idea. The concept of using a magnetic-field gradient was one such ‘aha’ that captured the essence of MRI as it is practiced today, much like the coupling of the nuclear spin of the proton to the magnetic field was the key to the early experiments by both Bloch’s group and Purcell’s group in their pioneering work in NMR.

The concept of nuclear magnetic resonance had its underpinnings with the discovery of the spin nature of the proton. Leaning on the work of Stern and Gerlach from the early 1920’s, Rabi and coworkers pursued the spin of the proton and its interaction with a magnetic field in the 1930’s. With this foundation in hand in 1946, Bloch and Purcell extended these early quantum mechanical concepts to a measurement of an effect of the precession of the spins around a magnetic field. Not only did these gentlemen successfully measure a precessional signal from a water sample and a paraffin sample, respectively, but they
explained precociously many of the experimental and theoretical details that we continue to
draw from still today. For this work, they shared the Nobel prize in physics in 1952.

1.3 A Brief Overview of MRI Concepts

In this section, we attempt to introduce and list for the reader the basic elements that make
MRI possible. The derivations, explanations and related details of these results are covered
principally in Chs. 2-15. Although the later chapters, Chs. 16-27, are focused on advanced
applications and concepts, additional related results are presented there as well.

1.3.1 Fundamental Interaction of a Proton Spin with the Magnetic
Field

We have alluded to the idea that MRI is based on the interaction of a nuclear spin with an
external magnetic field, $\vec{B}_0$. The dominant nucleus in MRI is the proton in hydrogen and
its interaction with the external field results in the precession of the proton spin about the
field direction (see Fig. 1.1). Imaging of humans rests on the ability to manipulate, with a
combination of magnetic fields, and then detect, the bulk precession of the hydrogen spins
in water, fat and other organic molecules.

![Diagram of magnetic field and proton spin](image)

Fig. 1.1: By definition, precession is the circular motion of the axis of rotation of a spinning body
about another fixed axis caused by the application of a torque in the direction of the precession.
The interaction of the proton's spin with the magnetic field produces the torque, causing it to
precess about $\vec{B}_0$ as the fixed axis. When looking down from above the vector $\vec{B}_0$, the precession
of the magnetic moment vector $\vec{\mu}$, which is proportional to the spin vector, is clockwise. For the
customary counterclockwise definition of polar angles, the differential $d\phi$ shown is negative.

The basic motion of the proton spin may be understood by imagining it as a spinning
gyroscope that is also electrically charged. It thus possesses an effective loop of electric
current around the same axis about which it is spinning. This effective current loop is capable of interacting with external magnetic fields as well as producing its own magnetic field. We describe the strength with which the loop interacts with an external field, as well as the strength with which the loop produces its own field, in terms of the same ‘magnetic dipole moment’ vector \( \mathbf{\mu} \). The direction of this vector is nothing other than the spin axis itself and, like a compass needle, the magnetic moment vector will tend to align itself along any external static magnetic field, \( \mathbf{B}_0 \). This is like the initial tendency of a gyroscope to fall in the direction of gravity. However, the tendencies are complicated by the spin in exactly the same way as is the gyroscopic motion. Instead of ‘falling’ along the field direction, the magnetic moment vector, like the spinning gyroscope, will precess around the field direction. In Ch. 2, we find that the precession angular frequency for the proton magnetic moment vector (and for the spin axis as well) is given by

\[
\omega_0 = \gamma B_0
\]  

(1.1)

where \( \gamma \) is a constant called the gyromagnetic ratio. In water, the hydrogen proton has a \( \gamma \) value of roughly \( 2.68 \times 10^8 \text{ rad/s/Tesla} \) (so that \( \gamma = \gamma / (2\pi) \) is \( 42.6 \text{ MHz/Tesla} \)). For a 2 T field, for example, the spins precess at a radiofrequency of 85.2 MHz, just below the FM range for radio broadcasting. (We use SI units throughout the text.) This precession frequency is referred to as the Larmor frequency and (1.1) is referred to as the Larmor equation.

1.3.2 Equilibrium Alignment of Spin

The magnetic moment vector for a typical proton is prevented from relaxing fully to an alignment along the external magnetic field because of thermal energy associated with the absolute temperature \( T \). From the discussions in Chs. 5 and 6, we can compare the magnetic field interaction with the average thermal energy \( kT \), where \( k \) is the Boltzmann’s constant. At human body temperatures, the thermal energy is millions of times larger than the quantum energy difference for parallel alignment (lower energy) versus anti-parallel alignment (higher energy). For a proton with only two quantum spin states, these are the only two possible alignments. Significantly, the frequency in the quantum energy difference, \( \hbar \omega_0 \), is nothing other than the Larmor precession frequency (1.1) where \( \hbar = \hbar / (2\pi) \) in terms of Planck’s quantum constant \( \hbar \).

The extreme smallness of the quantum spin energy compared with the thermal energy means that the fraction \( \hbar \omega_0 / (kT) \ll 1 \). In that case, the Boltzmann probability discussion of Ch. 6 demonstrates why the number of spins parallel to the magnetic field exceeding the number anti-parallel to that field, the ‘spin excess,’ is also very small. Specifically, the spin excess is suppressed by a factor involving that fraction:

\[
\text{spin excess} \approx N \frac{\hbar \omega_0}{2kT} \]  

(1.2)

where \( N \) is the total number of spins present in the sample. In the first problem below, it is found that the spin excess is only one in a million spins even for a magnetic field strength as large as 0.3 T.
Problem 1.1

Using $\hbar = 1.05 \times 10^{-34}$ Joule-s, $k = 1.38 \times 10^{-23}$ Joule/K and $T = 300$ K, find the spin excess as a fraction of $N$ at 0.3 Tesla.

Since the spin excess is millions of times smaller than the total number of proton spins, it might be guessed that no significant signal would be detected at room temperature. However, there are Avogadro numbers of protons in a few grams of tissue. Consider the average magnetic dipole density, or 'longitudinal equilibrium magnetization' $M_0$ for the component of the magnetic moment vector along the external field direction. For a sample with $\rho_0$ defined as the number of protons per unit volume (or the 'spin density'), the longitudinal equilibrium magnetization is found in Ch. 6 to be given by the proton magnetic moment component $\gamma \hbar / 2$ multiplied by the relative spin excess (1.2) times the spin density. Noting (1.1), it is thus given by

$$M_0 = \frac{\rho_0 \gamma^2 \hbar^2}{4kT} B_0$$  \hspace{1cm} (1.3)

This equilibrium value, while limited by the spin excess, leads to measurable NMR effects to be described next.

1.3.3 Detecting the Magnetization of the System

Even in a macroscopic body, a bulk nonvanishing spin excess is not enough to guarantee a detectable signal. In a classical picture (which is derived from the quantum underpinnings in Ch. 5), the magnetization vector (the magnetic moment vector density due to the spin population) must be tipped away from the external field direction in order to set it into precession. The magnetic field produced by the aggregate proton spins will precess along with the magnetization yielding a changing flux in any nearby ('receiver') coil (Ch. 7). To accomplish this, as discussed in Chs. 3 and 4, the magnetization can be rotated away from its alignment along the $B_0$ axis (i.e., from its longitudinal direction) by applying a radiofrequency (rf) magnetic field for a short time (an rf 'pulse'). This rf pulse is produced from another nearby 'transmit' coil (which may be the same as the receiver coil, provided its radiofrequency is tuned to the Larmor frequency (see Fig. 1.2). This is the resonance condition in MRI described earlier and it ensures that the precessing spin gets a continuously synchronized push (rotation) away from the longitudinal direction (the $z$-axis, say).

Suppose that $\vec{M}$ has been rotated by an rf pulse to a direction orthogonal to $\vec{B}_0 = B_0 \hat{z}$. (The rf pulse that tips all the original longitudinal magnetization, or $z$-magnetization, through an angle of 90° into the transverse, or $x$-$y$, plane is called a $\pi/2$-pulse). The resulting 'transverse magnetization' has magnitude $M_0$ and begins to precess clockwise in the $x$-$y$ plane. Its rectangular components have sinusoidal time dependence with frequency given by the Larmor frequency. As defined in Ch. 4, the complex magnetization is

$$M_z(t) = M_x(t) + iM_y(t) = M_0 e^{-i\omega_0 t + i\phi_0}$$ \hspace{1cm} (1.4)
in terms of the magnitude and the (polar angle) phase. This shows an important connection between the time-dependence of the complex phase and the rotation of the magnetization. The phase angle gives the direction in the x-y plane of the two-dimensional transverse magnetization vector. The initial phase $\phi_0$ is determined by the choice of rotational axis for the initial rotation into the transverse plane. For the example in Fig. 1.2, $\phi_0 = \pi/2$.

![Diagram](image)

**Fig. 1.2:** Illustration from Ch. 3 of the effect of an rf pulse on an individual magnetic moment $\vec{\mu}$. (a) In a frame rotating about $\vec{B}_0$ (which is along $\hat{z}$, say) at the Larmor frequency (with coordinates $x'$, $y'$ and $z' = z$), there is no observed precession about $\vec{B}_0$. Upon application of an rf magnetic field pulse applied along $\hat{x}'$, the magnetic moment is rotated about $\hat{x}'$ at a rate corresponding to the frequency $\omega_1 = \gamma B_1$ determined by the amplitude of the rf field, $B_1$. A $\pi/2$ flip relative to its starting position along $\hat{z}'$ is achieved in a time $\tau_{rf}$ provided that $\omega_1 \tau_{rf} = \pi/2$. (b) The behavior of the same magnetic moment rotation is observed to be more complicated in the fixed laboratory frame. This picture has been constructed for the case $\omega_1 = 0.06 \omega_0$. In actual MR applications, the frequency $\omega_1$ would be much smaller in relation to $\omega_0$, but then the spiraling would be too dense to illustrate.

The signal analyzed in Ch. 7 corresponds to the voltage induced in a receive coil from the time-varying magnetic flux that, in turn, is produced by a rotating magnetization. The inductive coupling of the receive coil to the magnetization may be described, according to a reciprocity principle, as equivalent to a constant flux, produced by a unit current flowing around the receive coil, that penetrates the precessing magnetization of the sample. The voltage, or electromotive force (emf), induced in the receive coil is given by

$$ emf = -\oint d\vec{s} \left( \dot{\vec{M}} \cdot \vec{B}_{rf} \right) $$

where $\vec{B}_{rf}$ is the static field produced by the receive coil per unit current. Ignoring any spatial variations and noting that the time derivative of the phase in the transverse magnetization (1.4) dominates the time derivative in (1.5), the emf is proportional to $\omega_0 M_0$. (In other words, the dominant time dependence is due to precession.) From (1.3) and (1.1), it follows that the signal from an MR experiment will depend on the square of the static magnetic field $B_0$

$$ signal \propto \frac{\gamma^2 B_0^2 \rho_0}{T} $$

(1.6)
1.3. A Brief Overview of MRI Concepts

The interest in higher fields stems from the growth of the signal with field strength; we return later in the preview to address the field dependence of the signal-to-noise ratio.

1.3.4 Magnetic Resonance Spectroscopy

Hydrogen protons in different molecules are immersed in slightly different magnetic environments, even in the presence of identical external magnetic fields. That is, different chemical compounds have slightly different local magnetic fields which means that the local Larmor frequency is ‘chemically shifted’ to different values \( \omega_0(j) \) depending upon the molecular species type \( j \). Chemical shift imaging is discussed in Chs. 8, 10 and 17.

Chemical shift imaging may be considered as adding another dimension, corresponding to the frequency range of different species. Each species will have its own contribution to the total signal. For instance, the transverse magnetization (1.4) gives a signal

\[
\text{signal} \propto \sum_j M_0(j) \omega_0(j) e^{-i\omega_0(j)t+i\phi(j)}
\]  

(1.7)

Here, the time derivative of the phase in (1.4) again yielded the factors \( \omega_0(j) \) in the leading terms.

The goal in magnetic field spectroscopy is to find the relative (spectral) amplitudes of the different frequency components, \( M_0(j) \), in (1.7), whose spin densities \( \rho_0(i) \) would enter via (1.3). This may be analyzed utilizing a Fourier transform to map the time domain back into a frequency domain. The goal in chemical shift imaging is the spatial disentangling of the signals of different tissue (spectral) components, such as water and fat.

1.3.5 Magnetic Resonance Imaging

The goal of imaging is to correlate a series of signal measurements with the spatial locations of the various sources. When all protons are represented by just one chemical species such as water, then the above spectroscopic analysis simply gives the total signal from all spins regardless of their spatial location in the static magnetic field, as long as that field is uniform.

We now utilize the fact that the addition of a spatially changing magnetic field across the sample produces a signal with spatially varying frequency components according to

\[
\omega(x) = \gamma B(x)
\]

(1.8)

where \( x \) denotes the spatial coordinate along the direction of the gradient of the field. This means that the spectral components now represent spatial information and, in turn, leads to the possibility that the signal could be ‘inverted’ and the physical object could be reconstructed (Chs. 9 and 14) or ‘imaged.’

The inversion of the signal is greatly facilitated through a connection to Fourier transforms (Chs. 9, 10 and 11). By constructing an additional coil (a linear gradient coil) that changes the original field \( \vec{B}_0 \) linearly in some direction, the phase in (1.4) becomes linear in the coordinates of that direction, so that the mapping back and forth between signal space and the image position space may be carried out with a Fourier transform. With more gradient coils, data reconstruction by inverse Fourier transformation can be carried in more
spatial dimensions. Two- and three-dimensional ‘imaging’ in MRI is elegantly realized with this powerful mathematical tool (Ch. 10).

In particular, the application of a finite bandwidth rf excitation centered at the Larmor frequency of the combined static field plus a gradient field leads to the excitation of a layer, or slice, of spins orthogonal to that gradient with slice thickness $TH$, say (see Fig. 1.3). By employing different configurations of gradient coils, the choice of gradient direction is completely flexible, a powerful procedure allowing slices to be acquired in any orientation. No physical rotation of the sample is required.

![Diagram](image)

Fig. 1.3: The precession frequency ($f = \frac{\omega}{(2\pi)}$) in the laboratory frame is a function of position along the slice select axis. The original static field $B_0$ has been supplemented with a field with constant gradient $G_z$ in the $z$-direction. The central frequency and spectral bandwidth of the rf pulse ($\Delta f \equiv BW_{rf}$, the shaded horizontal strip) are such that the slice of thickness $\Delta z \equiv TH$ (the shaded vertical strip) is uniformly ‘excited’ (i.e., all spins in the slice have the resonance condition satisfied). The fact that the slice is offset from the origin in the $z$ direction by $z_0$ implies that the center frequency of the rf pulse must be offset from the static Larmor frequency $f_0 = \gamma B_0$ by $\gamma G_z z_0$ as has been shown along the frequency axis.

### 1.3.6 Relaxation Times

An important factor in the strength of the signal has been omitted in the above discussions and must be considered. It is the ‘spin-lattice’ decay or relaxation of the signal due to the interactions of the spins with their surroundings. After the magnetization has been rotated into the transverse plane, it will tend to grow back along the direction of the static field $B_0$, chosen here to be $\hat{z}$. This rate of regrowth can be characterized by a time constant $T_1$ called the longitudinal relaxation time and arises from the interaction between the spins and the atomic neighborhood. The magnetization time evolution is described by the solutions given in Ch. 4 for the famous Bloch equations which incorporate both relaxation and precession effects. For an initial situation where $M_z(0) = 0$ (for example, the condition achieved following the application of a $\pi/2$-pulse), the subsequent regrowth of $M_z$ is given by

$$M_z(t) = M_0(1 - e^{-t/T_1})$$  \hspace{1cm} (1.9)
If the data are sampled following the application of another rf pulse at a time $\tau$ short compared to $T_1$, the longitudinal magnetization $M_z(\tau)$ is suppressed according to (1.9). Therefore, any transverse magnetization obtained by an rf rotation of $M_z(\tau)$ into the transverse plane will also be suppressed.

With the recognition of another relaxation effect, a more realistic assessment of the MRI signal may be achieved. The ‘dephasing’ of clusters of spins represents a ‘spin-spin’ decay of the transverse magnetization before data sampling can occur. Consider an experiment (Ch. 8) where a $\pi/2$ rf pulse is applied at interval time $T_R$ where any previous transverse magnetization has decayed away due to the spin-spin effect and only the longitudinal magnetization corresponding to (1.9) remains to be rotated into the transverse plane. If the signal data are instantaneous sampled at a time $T_E$ (‘echo time,’ see below for an explanation of this nomenclature) following the rf pulse, the signal is proportional to the magnitude of the transverse magnetization given by

$$M_1(T_E) = M_0 (1 - e^{-T_R/T_1}) e^{-T_E/T_2}$$

(1.10)

In (1.10), $e^{-T_E/T_2}$ is the spin-spin decay factor characterized by the time constant $T_2$; it is caused by the decorrelation between (dephasing of) the different spins. Their phases disperse due to variations in the local precessional frequencies.

In general, signals would suffer additional suppression due to dephasing from external field inhomogeneities ($T_2$ would be replaced by a smaller relaxation time $T_2^* < T_2$). But a ‘rephasing’ or ‘echoing’ of this source of dispersion has been assumed in (1.10) such that the additional suppression has been avoided. This can be achieved by an additional rf pulse application, where the basic idea is to flip all the spins $180^\circ$ in the transverse plane. The dephasing is reversed and the refocusing of any external field dispersion occurs at the echo time $T_E$.

The three tissue parameters (the spin density and the two relaxation times, $T_1$ and $T_2$) play principal roles throughout the book. Their specific measurements using MR techniques are the subject of Ch. 22.

### 1.3.7 Resolution and Contrast

An interesting aspect of MRI is the fact that resolution (the size of the spatial features that can be distinguished) does not depend upon the wavelength of the input rf field. Radiofrequencies generally have wavelengths on the order of meters, yet resolution in an MR image is on the order of millimeters. In fact, the inherent resolution in MR is a function of the way the signal and noise are sampled and filtered (see Chs. 12, 13 and 15) and it is ultimately limited only by the diffusion (Ch. 21) of the protons through the tissue and the local magnetic field nonuniformities around the proton.

The success of MRI goes beyond resolution and is understood by recognizing its large number of useful variables. MRI can be used to differentiate between materials because of its sensitivity to proton densities, relaxation times, temperature, proton motion, the chemical shift in the Larmor frequencies, and tissue heterogeneity, as examples. This large set of variables permits images to be generated with different levels of contrast based upon the desired usage. Therefore, MRI is more versatile than those imaging techniques restricted to only one type of contrast.
Contrast-to-noise and contrast mechanisms are first described in Ch. 15 but important aspects of image contrast already can be understood from (1.10). Examining the behavior of the exponentials, we see that for long $T_R$ (relative to $T_1$) and short $T_E$ (relative to $T_2$), the image will be sensitive only to the tissue spin density (Fig. 1.4a). For $T_E \approx T_2$ and long $T_R$, the image is weighted by both spin density and $T_2$ (Fig. 1.4b) and the contrast between tissues with different $T_2$ is enhanced. Often the spin density-weighted images and $T_2$-weighted images exhibit similar contrast features, the latter enhancing the former. Lastly, for $T_R \leq T_1$, and short $T_E$, the image is weighted by both spin density and $T_1$ (Fig. 1.4c).

1.3.8 Magnetic Field Strength

The interest in higher fields stems from the fact that the signal-to-noise ratio (SNR is the subject of Ch. 15) increases with field strength. While machines providing lower fields (less than 0.5 T) are less expensive, they produce lower SNR, as compared to mid fields (0.5 T to 1.0 T) and high fields (higher than 1.0 T). The signal (1.6) exhibits quadratic growth with $B_0$ but this is partially offset by the fact that the noise has linear $B_0$ dependence at high fields. In the range from 0.5 T to 4.0 T, the implied linear growth of SNR with field strength has been experimentally validated in human experiments.

There also has been concern about rf heating and nonuniform rf fields, where wavelengths finally play a role, as higher fields are considered. (Table 1.1 shows a comparison of free-space wavelengths for the different frequency ranges of familiar electromagnetic wave categories.) However, we have noted that the field strength of 1.0 T corresponds to 42.6 MHz, and this implies a rather long seven-meter free-space wavelength for imaging protons. Hence the rf field wavelengths for higher magnetic fields fall below 1 m only above 7.0 T. On the other hand, in humans, the relative electrical permittivity $\epsilon_r$ is about 50 near 1 T to 2 T, and the interior wavelengths are therefore reduced by a factor of $1/\sqrt{\epsilon_r}$ inside the body. This reduces the effective wavelength to about 1 m at 1 T for in vivo imaging and rf field nonuniformity must be considered at higher $B_0$ values.

More detailed rf pulse considerations are the subject of Ch. 16. The design issues for the coils producing the static field, the gradient fields and the rf fields are discussed in Ch. 27 where SNR is also revisited in terms of the high field dependence.

Problem 1.2

Find the frequency and free-space wavelength associated with the rf field required for proton magnetic resonance at each of the different $B_0$ values of a) 0.04 T, b) 0.2 T, c) 1.5 T and d) 8 T.

1.3.9 Key Developments in Magnetic Resonance

An important mechanism in MRI is the 'echoing' capability discussed earlier for the recovery of some of the signal lost to transverse relaxation. (We should also mention 'gradient
Fig. 1.4: Images of the human head with different forms of contrast: (a) a spin density-weighted image, (b) a $T_2$-weighted image and (c) a $T_1$-weighted image. These different acquisitions can be seen to create different contrasts between white matter, gray matter and cerebrospinal fluid. They all reveal excellent anatomic detail.
Table 1.1: Range of radio and microwave frequencies. The letters F, L, M, H, V, U and S refer to frequency, low, medium, high, very, ultra and super, respectively. Associated free-space wavelengths and NMR field strengths for protons are given here.

<table>
<thead>
<tr>
<th>Category</th>
<th>Subcategory</th>
<th>Frequency (MHz)</th>
<th>Field strength (T)</th>
<th>Wavelength (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>radio waves</td>
<td>LF (long wave)</td>
<td>0.03-0.3</td>
<td>$7 \times 10^{-4}$-$7 \times 10^{-3}$</td>
<td>$10^4$-$10^3$</td>
</tr>
<tr>
<td></td>
<td>MF (medium wave)</td>
<td>0.3-3</td>
<td>$7 \times 10^{-3}$-$0.07$</td>
<td>$10^3$-$10^2$</td>
</tr>
<tr>
<td></td>
<td>AM radio (MF)</td>
<td>0.54-1.6</td>
<td>0.013-0.038</td>
<td>555-188</td>
</tr>
<tr>
<td></td>
<td>HF (short wave)</td>
<td>3-30</td>
<td>$\sim$ 0.07-0.7</td>
<td>$10^2$-10</td>
</tr>
<tr>
<td></td>
<td>VHF (short wave)</td>
<td>30-300</td>
<td>0.7-7 $\varnothing$</td>
<td>10-1</td>
</tr>
<tr>
<td></td>
<td>FM radio (VHF)</td>
<td>54-216</td>
<td>1.27-5.07</td>
<td>5.55-1.39</td>
</tr>
<tr>
<td></td>
<td>UHF</td>
<td>300-3×10^3</td>
<td>7-70</td>
<td>1-0.1</td>
</tr>
<tr>
<td></td>
<td>SHF</td>
<td>3×10^3-3×10^4</td>
<td>70-700</td>
<td>0.1-0.01</td>
</tr>
<tr>
<td>microwaves</td>
<td></td>
<td>$10^4$-3×10^5</td>
<td>$233-7 \times 10^3$</td>
<td>$0.3-10^{-3}$</td>
</tr>
</tbody>
</table>

Echoes' where dephasing brought about by the external gradient field itself is countered by reversing the gradient direction during its application.) When Hahn happened on the 'spin echo' concept, it was through an application of multiple rf pulses. This concept made it possible to collect data in what would otherwise be considered poor experimental conditions (inhomogeneous magnetic fields) where little signal would remain. What he had found initially to be a spurious signal later became a workhorse in NMR and in clinical MRI where disease states are often clearly diagnosed in $T_2$-weighted images (see Fig. 1.4b).

In a somewhat related manner, what presents itself as an unexpected and even bothersome image inaccuracy, or an ‘artifact,’ may end up opening new doors and even a new direction of research. Two fairly recent examples of this include MR angiography (the study of blood vessels using MRI) and MR brain functional imaging. The first example came about because of a thrust to eliminate flow and motional blurring. In so doing, early researchers came up with the concept of flow compensation. Eliminating the artifacts in this way led to the very pleasant surprise of enhanced images of blood vessels. Thus began the subfield of MR angiography. The phenomena of blood flow and heart and lung motion and the methods for obtaining their images are detailed in Chs. 23 and 24.

The second example occurred when the loss of signal due to the susceptibility of a contrast agent was so large it led to a dramatic loss of signal on its first pass through the brain. It was soon realized that veins also had a different susceptibility than the rest of the brain and, hence, veins could be recognized by their phase alone or the signal loss they caused at long echo times. When blood flow changes, so does the blood's deoxyhemoglobin concentration and, therefore, the blood's susceptibility and its effect on the signal also changes. Evidently, blood flow changes occur when the brain activates, which, in turn, leads to signal changes.
1.4. Suggested Reading

Today, we can measure these changes in a matter of seconds, i.e., in a sense, we can detect the brain function. This is the basis of MR brain functional imaging known as fMRI and has led to a major focus in the fields of neuroscience and neuroradiology. It is considered in Ch. 25.

Besides artifacts, there were other reasons to improve the MR methodology. Spin echo scans took a long time to acquire and could not offer dynamic imaging to study, for example, the beating heart. To speed up data acquisition, researchers pushed toward fast ‘gradient echo’ imaging (where no extra refocusing rf pulses are applied) which reduced scan times from minutes to seconds (see Ch. 18) and to ‘echo planar’ methods which could acquire data in a fraction of a second (see Ch. 19). These methods had direct implications on MR angiography and functional brain imaging as well as cardiac MRI. However, they still suffer from inhomogeneous fields which cause signal loss and image distortion (see Ch. 20). These are less of a problem today because of improvements in main magnet design (see Ch. 27), but some remnant inhomogeneity effects remain caused by the local fields in the body itself (see Ch. 25). Even these remnants are being addressed today with better sequence designs (see Ch. 26) and improvements in our understanding of the principles of magnetic resonance imaging.

We turn now to the details and the exposition of these and other key concepts in MRI in the coming chapters. It is hoped that the following discussions will help elucidate those remarks that were inadequately explained in this brief overview.

1.4 Suggested Reading

Certain references that offer a good background for this opening chapter are given next. More general references are listed later.

Chapter 1. Magnetic Resonance Imaging: A Preview

There are a number of texts and articles on magnetic resonance. We have grouped these according to (1) primary technical references, which contain a great deal of relevant information to this text, (2) secondary references, both technical and clinical which also contain much useful information, (3) tertiary references, which will broaden the reader's perspective on MR in general and (4) some interesting review articles and texts to add overview and historical perspectives.

Primary Technical References


Secondary Technical and Clinical References


1.4. Suggested Reading


Tertiary References


Historical/Review References


