Chapter 1

Introduction

A journey of a thousand miles must begin with a single step.

Lao Zi

Tomography is an important area in the ever-growing field of imaging science. The term *tomos* ($\tau o \mu o \varsigma$) means "cut" in Greek, but tomography is concerned with creating images of the internal (anatomical or functional) organization of an object without physically cutting it open. To a beginner, it might seem inconceivable, but as your reading of this book progresses, you will appreciate not only the feasibility but also the inherent beauty and simplicity of tomography.

Tomographic imaging principles are rooted in physics, mathematics, computer science, and engineering. However, development of these principles is traditionally tied to solving application problems—particularly biomedical problems. Therefore, their theoretical significance has not been well appreciated by researchers outside this field. Like any other scientific discipline, tomography has a unique history. Radon was perhaps the first to address the tomographic imaging issue, albeit from a purely mathematical standpoint. Unfortunately, his seminal work published in 1917 went unnoticed for half a century. The sixties and the seventies were the formative years of tomography when ground-breaking work was done for both X-ray tomography and magnetic resonance imaging (MRI) . Now, a number of tomographic imaging modalities are available for medical and nonmedical uses. A partial list includes X-ray CT (computer tomography), MRI, PET (positron emission tomography), SPECT (single photon emission computed tomography), MEG (magnetoencephalography), SAR (synthetic aperture radar), and various acoustic imaging systems. Although these systems use different physical principles for signal generation and detection, the underlying signal processing principles for image formation are, to a large extent, the same. Therefore, it is fair to say that understanding how one imaging modality works will provide a significant insight into the working principles of other imaging modalities as well.

This book is about MRI. Its emphasis is on the principles of image formation rather than the hardware to build an MRI system or the applications of such a system. The reader is referred to the book by Chen and Hoult [13] on MRI technology or to the two volumes edited by Stark and Bradley [62] on MRI applications.

1.1 What Is MRI?

Simply put, MRI is a tomographic imaging technique that produces images of internal physical and chemical characteristics of an object from externally measured nuclear magnetic resonance (NMR) signals. Physically, MRI is based on the well-known NMR phenomenon observed in bulk matter independently by Felix Bloch at Stanford and Edward Purcell at Harvard in 1946. Image formation using NMR signals is made possible by the spatial information encoding principles, originally coined *zeugmatography* [182], developed by Paul Lauterbur in 1972. These principles enable one to uniquely encode spatial information into the activated MR signals detected outside an object. To help answer the question "What is MRI?" some notable features of MRI are listed below.

First, like any other tomographic imaging device, an MRI scanner outputs a multidimensional data array (or image) representing the spatial distribution of some measured physical quantity. But unlike many of them, MRI can generate two-dimensional sectional images at any orientation, three-dimensional volumetric images, or even four-dimensional images representing spatial-spectral distributions. In addition, no mechanical adjustments to the imaging machinery are involved in generating these images.

Second, MR signals used for image formation come directly from the object itself. In this sense, MRI is a form of *emission* tomography similar to PET and SPECT. But unlike PET or SPECT, no injection of radioactive isotopes into the object is needed for signal generation in MRI. There are other forms of tomography in use, including *transmission* tomography and *diffraction* tomography. X-ray CT belongs to the first category, while most acoustic tomography is of the diffraction type. In both cases, an external signal source is used to "probe" the object being imaged.

Third, MRI operates in the radio-frequency (RF) range, as shown in Fig. 1.1. Therefore, the imaging process does not involve the use of ionizing radiation and does not have the associated potential harmful effects. However, because of the unique imaging scheme used, the resulting spatial resolution of MRI is *not* limited by the "probing" (or working) frequency range as in other remote-sensing technologies.

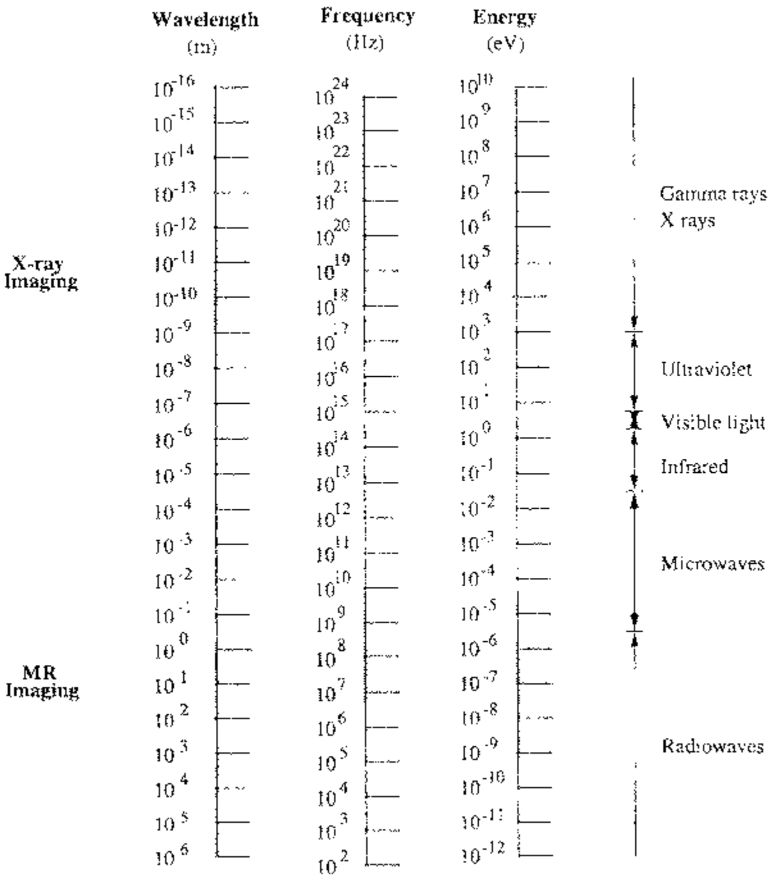

Figure 1.1 Electromagnetic spectrum.

Finally, and perhaps most importantly to an MRI user, MR images are extremely rich in information content. The image pixel value is in general dependent on a host of intrinsic parameters, including the nuclear spin density ρ, the spin-lattice relaxation time T_1, the spin-spin relaxation time T_2, molecular motions (such as diffusion and perfusion), susceptibility effects, and chemical shift differences. The imaging effects of these parameters can be suppressed or enhanced in a specific experiment by another set of operator-selectable parameters, such as repetition time (T_R), echo time (T_E), and flip angle (α). Therefore, an MR image obtained from the same anatomical site can look drastically different with different data acquisition protocols. An example is given in Fig. 1.2, in which the three images shown were obtained from the same cross section of a human head using a so-called spin-echo imaging sequence (discussed in Section 7.4). As can

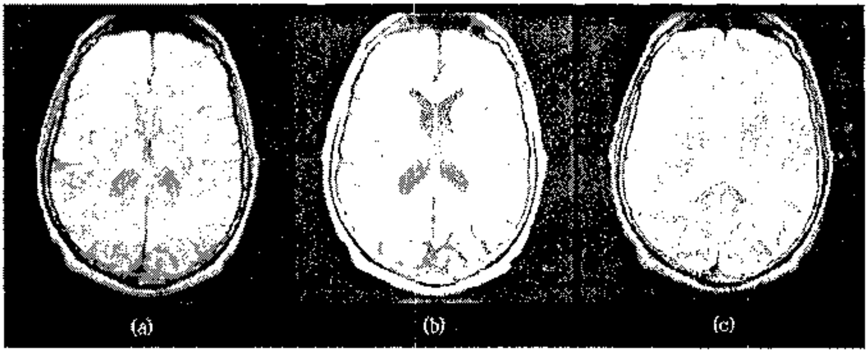

Figure 1.2 Cross-sectional head image obtained using a spin echo excitation sequence with (a) spin density-weighted contrast ($T_E = 17$ ms, $T_R = 2000$ ms); (b) T_1-weighted contrast ($T_E = 18$ ms, $T_R = 400$ ms); and (c) T_2-weighted contrast ($T_E = 80$ ms, $T_R = 2500$ ms).

be seen, the image contrast corresponding to different T_R and T_E values is quite different. Another example is shown in Fig. 1.3, in which signals from stationary spins were suppressed so that only the flowing blood was imaged. In general, an MRI image can be made to be a spatial map of the density of stationary spins or moving spins, or of relaxation times, or of the water diffusion coefficients. These are the subjects of study for subareas known as spectroscopic imaging, diffusion-weighted imaging, angiographic imaging, and functional imaging. Arguably, it is the flexibility in data acquisition and the rich contrast mechanisms of MRI that endow the technique with superior scientific and diagnostic values.

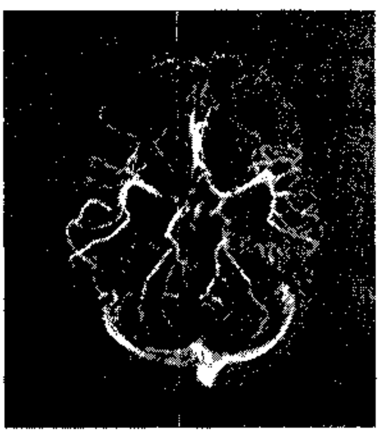

Figure 1.3 An example of angiographic imaging.

1.2 A System Perspective

An MR imager is shown in Fig. 1.4, which resembles an X-ray CT scanner. However, beyond the system appearance, an MR scanner and a CT scanner have little in common in terms of hardware components. An MR scanner consists of three main hardware components: a main magnet, a magnetic field gradient system, and an RF system. This section briefly describes their functional characteristics. For a more in-depth discussion, the reader is referred to the literature [8, 64].

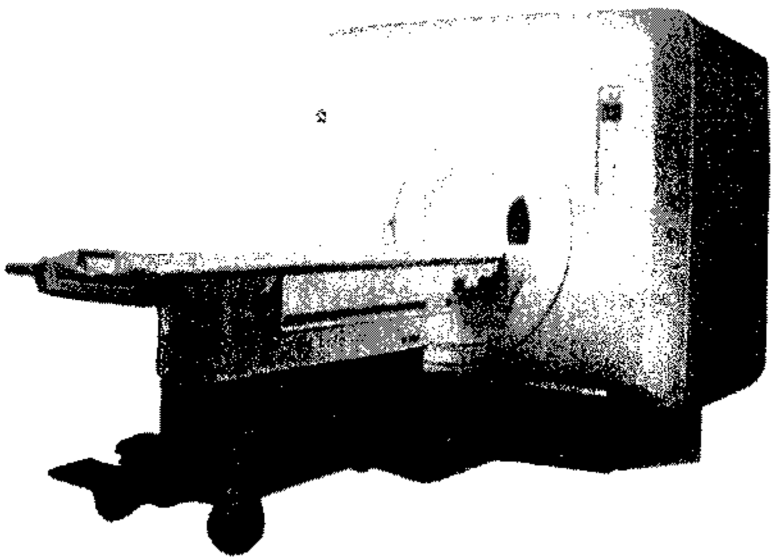

Figure 1.4 A typical clinical MRI scanner. (Images courtesy of GE Medical Systems.)

1.2.1 The Main Magnet

The main magnet is either a resistive, a permanent, or a superconducting magnet. Its primary function is to generate a strong uniform static field, referred to as the B_0 field,[1] for polarization of nuclear spins in an object. Resistive magnets are generally used at low field (< 0.15 T); permanent magnets can operate at field strengths up to 0.3 T; superconducting magnets are normally used for generating

[1] A magnetic field is a vector quantity, but we will use the conventional scalar notation for a magnetic field when only its magnitude is concerned and its direction need not be made explicit. Following convention, the B_0 field is assumed to point along the z-direction.

higher field strengths.[2] The optimal field strength for imaging is application-dependent. Advantages of high fields are better signal-to-noise ratio and spectral resolution. Disadvantages include RF penetration problems and higher costs. The field strength commonly encountered in a clinical whole-body system ranges from 0.5 to 2 T.

The spatial homogeneity of the main magnetic field is defined as the maximum deviation of the field over a given volume within the region of interest:[3]

$$\text{Homogeneity} = \frac{B_{0,\max} - B_{0,\min}}{B_{0,\text{mean}}} \tag{1.1}$$

An imaging magnet requires moderate homogeneity over a large volume to provide good image quality. A typical requirement for a human system would be 10 to 50 parts per million (ppm) over a 30 to 50 cm diameter spherical volume. For spectroscopic imaging, the requirement for field homogeneity is much more stringent. In practice, the main magnet alone is not capable of generating such a highly homogeneous field. The common approach to overcoming this problem is to use a secondary compensating magnetic field generated by a set of so-called shim coils to bring the overall field to the level of desired homogeneity.

1.2.2 The Gradient System

The magnetic field gradient system normally consists of three orthogonal gradient coils, an example of which is shown in Fig. 1.5. Gradient coils are designed to produce time-varying magnetic fields of controlled spatial nonuniformity, whose formal definition is given in Section 4.4.1. The gradient system is a crucial component of an MRI scanner because gradient fields are essential for signal localization, as will become evident in Chapter 5.

Important specifications for a gradient system include maximum gradient strength and the rate at which this maximum gradient strength can be obtained. Gradient strength is normally measured in units of millitesla per meter (mT/m), and the higher it is the better. Most clinical imaging systems can provide a maximum gradient strength of approximately 10 mT/m. The lower limit of the gradient strength required is determined by the criterion that the gradient field must be stronger than the main field inhomogeneity.

The time interval for a gradient to ramp up to its full strength is called the *rise time*; the shorter the rise time, the better the gradient system. For conventional imaging methods, rise times of approximately 1.0 ms from 0 to 10 mT/m are considered to be good. For some fast imaging methods (especially echo-planar imaging methods to be discussed in Chapter 9), a shorter rise time is needed.

[2] The strength of a magnetic field is measured in the units of gauss (G) or tesla (T) with 1 T = 10^4 G. The earth's magnetic field is approximately equal to 0.5 G.

[3] Outside the region of interest, the B_0 field is highly inhomogeneous. Therefore, various expressions in the book are valid only within the region of interest and should be used as such.

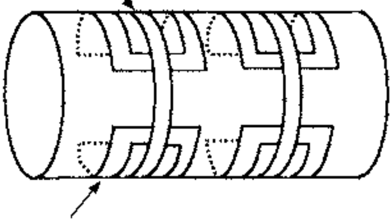

Figure 1.5 Schematic representation of the Maxwell coil pair (z-coil segment indicated) and the y-saddle coil set used to establish the z- and y-gradient, respectively.

1.2.3 The RF System

The RF system consists of a transmitter coil that is capable of generating a rotating magnetic field, often referred to as the B_1 field, for excitation of a spin system, and a receiver coil that converts a precessing magnetization into an electrical signal. Sometimes, a single coil can be used as both a transmitter and receiver coil, thus the name *transceiver* coil. Both the transmitter and receiver coils are usually called *RF coils* because they resonate at a radio frequency, as required by spin excitation and signal detection.

A desirable feature of the RF component is to provide a uniform B_1 field and high detection sensitivity. To do so, an MR system is often equipped with RF coils of different shapes and sizes for different applications. Some common examples are solenoidal coils, saddle coils, birdcage coils, and surface coils, as shown in Fig. 1.6. A long solenoidal coil consists of many closely spaced turns on a cylindrical form with a diameter much less than its length. It can produce a uniform B_1 field in its interior. A saddle coil has a pair of coils wound on a cylindrical surface and is able to generate a relatively homogeneous field near its center. A birdcage coil consists of a series of identical loops connected together and located on the surface of a cylinder giving the appearance of a birdcage. It provides the best RF field homogeneity of all the RF coils currently in use. Surface coils come in different forms and sizes. The simplest is a loop of wire that is useful for imaging of a limited region.

1.3 A Signal Processing Perspective

While an MRI system is rather complex, the imaging principles that such a system implements are much less intimidating. Especially when viewed from a signal processing standpoint, the imaging process essentially involves a pair of trans-

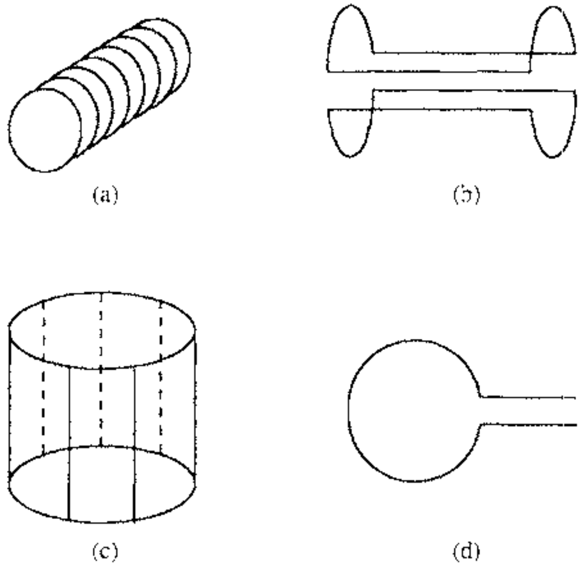

Figure 1.6 Some examples of RF coils: (a) solenoidal coil, (b) saddle coil, (c) birdcage coil, and (d) surface coil.

formations as shown in Fig. 1.7. The first transformation, often referred to as the *imaging equation*, governs how the experimental data are collected, and the second, often referred to as the *image reconstruction equation*, determines how the measured data are processed for image formation. In engineering, the first step is known as the *forward* problem and the second step is the so-called *inverse* problem. This book will adopt this transform-based approach to describe the principle of MR image formation. Specifically, we will focus on how MR signals are generated, detected, manipulated, and processed into an image. In doing so, we start with the microscopic magnetic moments $\vec{\mu}$ in an object and then trace step-by-step how they are subsequently converted to a bulk magnetization $\vec{M}$, a transverse magnetization $\vec{M}_{xy}$, an electrical signal $S(t)$, a k-space signal $S(\vec{k})$, and finally the desired image $I(\vec{x})$. Therefore, the main thread that links the book's materials together is

$$\vec{\mu} \longrightarrow \vec{M} \longrightarrow \vec{M}_{xy} \longrightarrow S(t) \longrightarrow S(\vec{k}) \longrightarrow I(\vec{x})$$

As will become evident later in this book, $\vec{\mu} \longrightarrow \vec{M}$ is accomplished by exposing the object to the B_0 field; $\vec{M} \longrightarrow \vec{M}_{xy}$ is done with RF excitations; $\vec{M}_{xy} \longrightarrow S(t)$, known as signal detection, is based on Faraday's law of induction; $S(t) \longrightarrow S(\vec{k})$ is the core of MRI, which involves the use of magnetic field gradients to encode spatial information into the transient responses of a spin system upon RF excitations; and finally, $S(\vec{k}) \longrightarrow I(\vec{x})$ is the well-known image reconstruction problem common to many tomographic imaging modalities. This book will focus on elucidating the underlying principles of each of these steps.

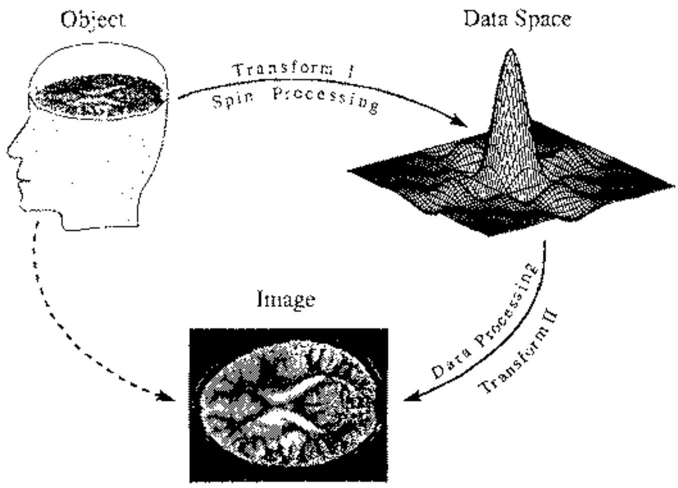

Figure 1.7 The MR imaging process viewed as two mathematical transformations.

1.4 Organization of the Book

This book is intended to be useful to upper-level undergraduate and graduate students in engineering or physical sciences as a textbook for a course on this subject, as well as to practicing MRI scientists as a reference. It emphasizes the fundamental principles and concepts underlying signal generation and detection, spatial information encoding, and image reconstruction. The remainder of the book is organized as follows.

Chapter 2 reviews some of the fundamental mathematical concepts that are key to understanding MRI principles. A primary objective is to make this book self-contained so that the reader will not require many reference books to review these materials.

Chapter 3 covers the basic physical principles for signal generation and detection.

Chapter 4 discusses the excitation requirements and general characteristics of transient signals of various types generated from a spin system after pulse excitations.

Chapter 5 discusses physical and mathematical principles for spatial localization of activated MR signals.

Chapter 6 discusses signal processing principles and techniques for image reconstruction from spatially encoded signals.

Chapter 7 describes the rich mechanisms unique to MRI for manipulating image contrast.

Chapter 8 deals with practical imaging issues such as limited resolution, signal-to-noise ratios, and image artifacts.

Chapter 9 discusses the principles and techniques of fast-scan imaging.

Chapter 10 introduces the concept of constrained image reconstruction.

Appendix A provides a summary of some commonly used mathematical formulas.

Appendix B contains a glossary of technical terms that frequently appear in the MR literature.

Appendix C presents a partial list of mathematical symbols used in the book.

Appendix D presents a partial list of abbreviations used in the book.

Appendix E contains the definitions of several physical constants.

The bibliography provides a list of relevant references. Although MRI has a relatively brief history, the number of publications on this subject is overwhelming. The bibliography included is by no means comprehensive. However, it should provide the reader with some useful guidance in searching for further details on the main ideas discussed in this book.

Exercises

1.1 What is the radio-frequency range?

1.2 What is roughly the strength of the earth's magnetic field? What is the strength of the main magnet of a clinical MR imager?

1.3 What are the main differences between emission, transmission, and diffraction tomography?

1.4 In what sense can MRI be viewed as a form of emission tomography?

1.5 MR imaging requires the use of three types of magnet fields: a strong uniform static field, an oscillating field, and three gradient fields.

(a) Identify the hardware components in an MRI system that produce these magnetic fields.

(b) Describe briefly the primary roles of these magnetic fields.