PRACTICAL SMALL ANIMAL MRI
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DEDICATION

This work is dedicated to our former and current radiology and neurology residents, the animal patients, and their owners.

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Preface

Dr. Rod Bagley and I discussed writing a textbook on Veterinary MRI for several years. Rod had already published a textbook, and knew the tremendous amount of work involved. We tackled the text with enthusiasm, but also with a degree of trepidation due to the daunting task.

Our goal in writing this textbook was not to have an exhaustive referenced rehash of material that is already present in text. Our goal was to provide a useful, clinical Veterinary MRI text. We were fortunate in starting magnetic resonance imaging of the brain in 1986. Magnetic resonance imaging for spontaneous brain tumors in dogs was a foundation for our research project. MRI continued to be a cornerstone of our research project to monitor the response of the brain tumor treatment and to evaluate normal tissue toxicities.

Washington State University had the oldest teaching hospital in North America in the 1990s. In 1996, we moved into a new facility that included a 1.0 Tesla superconducting magnetic resonance unit. We were the first veterinary medical college in an U.S. university with a state-of-the-art in-house superconducting MR unit. We were enthusiastic to expand beyond the imaging of brain tumors, and our initial efforts were in spinal disease. Our initial studies were not very good, and we had to methodically evaluate numerous pulse sequences and image planes to arrive at a clinically useful study. Magnetic resonance imaging was evolving rapidly in all fields, and some of the anatomical differences with our small patients necessitated a change from previous human protocols. We became comfortable in the spine, and then advanced into imaging for other conditions of the head, orthopedic disease, thorax, and abdomen. We also embarked on the imaging of the limbs of live adult horses. Our equine studies will not be covered in this textbook and will need to be treated in a separate volume.

The superior visualization of soft tissues of the body has allowed for the imaging of virtually all disease processes. The improved conspicuity allows clinicians of multiple disciplines to have clear visualization of the disease process. This text is to provide examples of our experiences that have been gained over the past 21 years. In addition to our experiences, some of the examples come from my active collaboration with other sites. These include the IAMS Pet Imaging Centers in Vienna, Virginia, Raleigh, North Carolina, and Redwood City, California. Examples have also been used from Dr. Michael Broome’s Advanced Veterinary Medical Imaging Center in Tustin, California. These centers, coupled with the Washington State University cases make up the bulk of the material for the figures in the text, but other images come from Dr. Kelley Collins, Veterinary Imaging Center in Ambler, Pennsylvania, Oakridge Veterinary Imaging in Edmond, Oklahoma, Tacoma Veterinary MRI in Tacoma, Washington, and Veterinary Neurological Centers in Phoenix, Arizona.

The images chosen are realistic examples of common abnormalities. We have endeavored to provide good quality images, but not ones that cannot be readily obtained virtually all superconducting magnets. There are no permanent magnet images due to lack of availability of such images in our files. The images are shown with the patients right on the viewers left, unless otherwise indicated. The images are generally shown with the dorsal anatomic area to the top of the image, even when acquired differently. If the dependence of the image is important, that will be given in the figure legend.

There is intentional overlap of some diseases in the various chapters. For example, neoplasia of a peripheral nerve may be covered in Chapter 2 (Section 3) on the peripheral nervous system, Chapter 3 on orthopedic disease, and Chapter 7 on MR imaging for cancer. The studies may have been requested for different reasons, that is loss of function, lameness, or for radiation therapy planning, and this redundancy should help the reader find the material for the clinical problem as presented.

We considered an exhaustive library of normal images, but discarded that notion. All studies have variability due to species, individual variation, technique, and volume averaging. Therefore, it is impossible to show an example that would fit all needs. We have given limited normal information, and have endeavored to illustrate common misunderstandings.

The text has limited information on physics, sequence selection, and artifacts. There are many superb texts that delve into these topics in great detail. We have only provided a skeleton of that material to facilitate the discussion of the case material presented in the
various chapters. Magnetic resonance imaging is just now becoming an accepted modality throughout the veterinary profession. We have been fortunate to be among the early adaptors of this exciting technology. We hope this text will aid you in the continued exploration and discovery of new information.

We would like to thank Dr. Susan Kraft, DVM, PhD DACVR and Dr. Shannon Holmes for their superb contributions. Finally, we would like to thank the many students, interns, residents, and colleagues that helped us learn from our mistakes.
CHAPTER ONE

Physics

SECTION 1

Comparative Imaging

Patrick R. Gavin

Diagnostic imaging has always been a mainstay of the armamentarium for the veterinarian. Veterinarians have limited resources available as regards history and routine screening procedures. Therefore, diagnostic imaging has a major role in the workup of numerous veterinary patients. An overreliance on diagnostic imaging has been observed by numerous clinicians; however, the move toward less invasive diagnostic procedures with a high precision of diagnosis has continued to drive this phenomenon. This chapter deals with the advances in diagnostic imaging through the last 60 years.

Diagnostic Radiology

Diagnostic radiology was invented in the late 1800s. The use of diagnostic radiology was rewarding primarily in the study of skeletal structures. However, due to the cost of the equipment, lack of education, and potential risks, the modality did not penetrate veterinary medicine until approximately the 1950s. Initially, these were the colleges of veterinary medicine in North America that possessed the equipment to perform diagnostic radiographic examinations. There were no trained radiologists at that time and in some places the studies were often performed and interpreted by non-veterinarians. Clinicians did not know what to expect as they had no prior knowledge of the diagnostic modality. Clinicians were often asked if they wanted a V/D or lateral and would merely say “yes” at the answer and accept the outcome. Much was to be learned.

Diagnostic radiology advanced rapidly in veterinary medicine, and the first examinations for veterinary radiologists were performed by charter diplomates for the American College of Veterinary Radiology in 1965. Following this beginning veterinary radiology advanced rapidly. Diagnostic radiology was utilized in multiple species throughout colleges of veterinary medicine and in selected practices. By the early to mid-1970s, advanced radiographic procedures including fluoroscopy and angiography were available, though primarily at colleges of veterinary medicine. The use of diagnostic radiology expanded with improved knowledge, especially with better understanding of its diagnosis of various pathologic conditions. The use of diagnostic radiology abated somewhat with the advance of diagnostic ultrasonography; however, it has remained the stalwart of diagnostic imaging in the veterinary profession. At the current time, there is a major push to move from conventional analog film screen technology to computed and/or digital radiography. It is presumed that veterinary radiology will continue to follow the progression realized in human radiology.

Nuclear Medicine or Gamma Scintigraphy

The previously used term, nuclear medicine, fell out of favor with the antinuclear movement of the 1970s. Medical personnel were quick to adopt the softer terminology of gamma scintigraphy that facilitated its continued development as an imaging modality. While gamma scintigraphy has the advantage of visualizing physiologic and temporal pathologic changes, for the most part its greatest use in veterinary medicine has been static studies for the diagnosis of skeletal disease. The use of the modality for the diagnosis of skeletal disease is well documented. The challenges of using nuclear isotopes, radiation safety concerns, and time delays are well documented. Some studies have become
rather routine in veterinary medicine. These include studies of the thyroid gland that have been published and have led to a better understanding of thyroid disease.

While this modality has been present since the turn of the century, it became rather commonplace in veterinary medicine in the 1980s. Its involvement as a diagnostic modality has undergone little evolution in the last two decades.

**COMPUTED TOMOGRAPHY**

Computed tomography (CT) was first utilized in the mid-1970s in veterinary medicine, primarily for the diagnosis of intracranial disease. The modality was modified for the study of large animal species shortly thereafter. CT has had a large expansion in the veterinary medical field. Virtually all colleges of veterinary medicine provide this diagnostic modality. In the last 10 years, extension into private veterinary practices has significantly expanded its availability. There are now numerous large specialties, and even general practices, with CT on site. Many units were purchased as used equipment, but many include state-of-the-art helical units.

CT uses the same basic physical principles as diagnostic x-ray, except it depicts the shades of gray in cross-section. It is also possible to better visualize different tissues and the pathologic change within them, if present. Therefore while the modality is similar to diagnostic x-ray, CT is superior in diagnosis because the axial images are far superior to the two-dimensional radiographic projections. CT has led a renaissance in the understanding of three-dimensional anatomy and physiologic principles.

**ULTRASONOGRAPHY**

Ultrasonography became a clinical imaging modality in veterinary medicine in the late 1970s. It languished in veterinary colleges through much of the 1980s as the technology advanced. The initial technology of static B-mode machines was replaced by real-time machines that allowed an approximate 80% reduction in scanning time. The resolution and utility of the studies improved at the same time. However, diagnostic ultrasonography did not hit its stride and become mainstream in the United States until approximately the 1990s. Now, most large veterinary practices (and certainly referral practices) have diagnostic ultrasonography. This modality is also available in many smaller private practices. There have been numerous technologic advancements that have improved the quality of this modality. Increased availability of traveling diagnostic radiologists and/or interpretation via teleradiology have improved diagnostic outcomes.

Other specialists utilizing diagnostic ultrasonography, including cardiologists and internists, have further fueled the expansion of this modality in veterinary medical practice. Currently, most ultrasonographic examinations are performed by licensed veterinarians. It is this author’s opinion that in the future, many of these procedures will be performed by trained ultrasonographers and interpreted by radiologists, just as occurs in the human field. In the human field, there is a greater medical liability issue, and if physician radiologists can make it work, certainly veterinary radiologists can work in this format to further advance this modality’s utility in the diagnosis of our veterinary patients.

**MAGNETIC RESONANCE IMAGING**

Magnetic resonance imaging (MRI) came into clinical utility in the mid-1980s. It was utilized in veterinary medicine primarily as a research tool in the 1980s and early 1990s. In the mid-1990s, some areas began to use MR as a routine clinical modality. The procedure was applied to large animal imaging a few years later. However, the attitude of “not invented here” plagued the inclusion of MRI for the diagnosis of veterinary patients at many sites in the early years. Many veterinary sites had antiquated equipment or equipment with poor reliability, which gave it the aura of an unreliable diagnostic modality. However, as more sites gained modern diagnostic equipment, the utility of the modality became apparent.

Following the change of the millennium, MR became the modality of choice for the veterinary neurologist for the examination of disease processes involving the brain and spinal cord. Efforts to expand the use of the modality included corporate sponsorship of diagnostic facilities. At the time of this writing, this author is aware of more than 40 sites dedicated to MR imaging of animals using what would be considered modern state-of-the-art equipment. One limitation has been the non-availability of appropriately trained veterinary radiologists with expertise in this modality capable of providing accurate diagnoses of clinical conditions. Currently, the American College of Veterinary Radiology does not require training time minimums in MRI for their core
curriculum, as not all training sites have this modality available. Therefore, many veterinary radiologists, and others, must essentially undergo “on the job training” in the use of this modality.

There is a broad spectrum of equipment options. These options span from the currently available best, including machines capable of functional MRI, commonly utilized super conducting magnets, cost-effective mid-field units, to even less expensive but less capable low field permanent magnets. It is this author’s opinion that equipment generally costs what it is worth. Therefore, equipment that is more expensive is of more diagnostic worth, and conversely, equipment that costs less has less diagnostic capability. The equipment purchase balance will be finding equipment that provides the utility required for the financial reality of the practice. There has been a rapid development of equipment in the last few years.
SECTION 2

Basic Physics

Patrick R. Gavin

It is beyond the scope of this text to do an extensive treatise of the physics of MRI. There are several excellent texts, as well as numerous study guides, and even impressive volumes of free information on the Internet that can be consulted for more in-depth information on patient MR physics. This chapter outlines the salient features of the physics of MRI to allow a better understanding of image, and artifact, production, and visualization.

Current clinical applications for MRI rely on visualization of the hydrogen atom’s nucleus. This physical property was previously known as nuclear magnetic resonance, that is, the hydrogen atom nuclei resonate. The word nuclear does not refer to radioactivity, but merely refers to the nucleus of the atom. For more politically correct names it has become known as MRI. The basic physical principle is that a moving electrical charge produces a magnetic field. The size of the magnetic field is dependent on the speed of movement (magnetic movement) and the size of charge. While the hydrogen nucleus has a small electric charge it spins very fast. These physical attributes in concert with the abundance of the hydrogen nucleus within the body produce a detectable magnetic field.

Magnetic field strengths are measured in units of gauss (G) and tesla (T). One tesla is equal to 10,000 gauss. The earth’s magnetic field is approximately 0.5 G. The strength of MRI is similar in strength to the electromagnets used to pick up large heavy scrap metal. Materials can be ferromagnetic, paramagnetic, supraparamagnetic, or diamagnetic. Ferromagnetic materials generally contain iron, nickel, or cobalt. These materials can become magnetized when subjected to an external magnetic field. In MR images, these materials cause large artifacts characterized by the properties of signal and distortion of the image. These artifacts can be seen in MR images even when the ferromagnetic substances are too small to be seen on conventional radiography. Commonly seen sources of these artifacts are microchips, ameroid constrictors, certain bone plates, gold-plated beads, and colonic contents.

Paramagnetic materials include ions of various metals such as iron (Fe), manganese (Mg), and gadolinium (Gd). These substances can also have magnetic susceptibility, but only about 1/1,000 that of ferromagnetic materials. These substances increase the T1 and T2 relaxation rates. Because of this property, chelates of these elements make ideal components of MR contrast agents. Gadolinium chelates are the most common agents and generally cause an increase in T1-weighted signal. This is seen as increased hyperintensity (brightness) in T1-weighted images. At very high gadolinium concentrations, as seen in the urinary bladder, loss of signal can be seen as a result of T2 relaxation effects dominating.

Supraparamagnetic elements are materials that have ferromagnetic properties. The most commonly used is super paramagnetic iron oxide (SPIO), which is an iron (Fe) based contrast agent for liver imaging. These have been used minimally in veterinary MR. Diamagnetic materials have no intrinsic magnetic moment, but can weakly repel the field. These materials include water, copper, nitrogen, and barium sulfate. They will cause a loss of signal and have been seen as a loss of MR signal in images made after the administration of barium sulfate suspensions.

Since hydrogen is the common element used to make an MR clinical image, we will discuss the process of image formation. When hydrogen is placed within a large external magnetic field, the randomly spinning protons (hydrogen nucleus) will come into alignment with the external field. Some of the protons align with the field and some align against the field, largely canceling each other out. A few more align with the field than against it. The net number aligning with the magnetic field is very small. Approximately, three protons align with the field for every one million protons as 1.0 T. This number is proportional to the external magnetic field strength. While this number appears very small,
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the abundance of hydrogen allows for high-quality images. For example, in a typical volume imaging element termed a voxel, the number of protons aligned with the field would be roughly $6 \times 10^{15}$.

Basic physics dictate that the energy is proportional to the nucleus’s unique resonant frequency in MR; this is called the Larmor frequency. The frequency of the precession of the hydrogen nuclei is relatively low. The resonance frequency is proportional to the external magnetic field, which for hydrogen is equal to $42.56$ MHz/T. 3. The Larmor frequency is proportional to the external magnetic field, which for hydrogen is equal to $42.56$ MHz/T. 

To attempt to negate the fixed causes, a 180° refocusing pulse is used. Consider the following analogy, three cars in a race going at different speeds. At the start, all the cars are obviously together, and can be thought of as being in-phase. At some time after the start of the race, there is a noticeable difference between them due to different speeds; they are in essence out-of-phase. At that time, everybody will turn around and go back toward the starting line. If it is assumed that everyone is still going at the same rate as before, then they will all arrive at the starting line together and in-phase. The time required for the atoms to come back in-phase is equal to the time it took for them to lose phase. This total time is called the “TE” or echo time. The 180° pulse is used to reverse the T2* de-phasing process. As soon as the spins come back into phase, they will immediately start to go out-of-phase again. The two variables of interest in spin echo (SE) sequences are (1) the repetition time (TR) and (2) the echo time (TE). All SE sequences include a slice-selective 90° pulse followed by one or more 180° refocusing pulses. This refocusing pulse can be applied multiple times. The use of multiple refocusing pulses is the basis for fast or turbo spin-echo imaging, FSE, or TSE respectively.

Images of T1 and T2 relaxation are produced by sampling the signal at various times. Both effects are always present, however, we will often accentuate one effect over the other such that the sequences are often properly termed T1-weighted or T2-weighted images. To produce the cross-sectional images, gradient coils are needed, which produce deliberate variation in the main magnetic field. There is one gradient coil in each Cartesian plane direction (X, Y, and Z planes). These
slight variations in the magnetic field will allow for slice selection and phase and frequency encoding. The slice selection gradient will be the Z, X, and Y gradients for a patient in supine position for the transverse, sagittal, and dorsal plane sequences, respectively.

A term commonly used in discussing image formation is the signal-to-noise ratio (SNR). SNR determines the appearance of the image. This ratio is measured by calculating the difference in signal intensity between the area of interest (the patient) and the background. The difference between the signal and background noise is divided by the standard deviation of the signal from the background, which provides an indication of the variability of the background noise. SNR is proportional to the volume imaged, called a voxel, and the square root of the number of signal averages and the number of phase encoding steps. Since signal averages and phase steps are temporal parameters, SNR is closely related to image acquisition time. Decreasing the voxel size (by decreasing the field of view), increasing the phase encoding, and decreasing the slice thickness will all decrease the SNR. Increasing the voxel size (by increasing the field of view, increasing the slice thickness, or decreasing the matrix size), or decreasing the phase encoding steps will all improve the SNR. The slice selection gradient will set the slice thickness. The two dimensions of the image are then mapped depending on emitted frequency in the phase and frequency encoded directions.

All of the frequencies in the frequency encoded direction can be encoded at one time; whereas, the number of phase encodings increases the time of acquisition in a directly proportional manner. Therefore, it is common to map the signal with fewer phase encodings compared to frequency encodings (e.g., 192 × 256) to reducing scan time. Dividing the field of view by the matrix size gives the voxel area, which represents the displayed element called a pixel. The depth of the voxel is determined by slice thickness. Slice thickness is almost always the largest dimension of this imaging voxel. Therefore, the resolution perpendicular to the image plane is generally the poorest. The signal obtained for the image can be improved by increasing the number of signal averages. This is done by increasing the number of RF pulses to knock the protons out of alignment. The scan time is directly proportional to the number of signal averages, sometimes termed number of excitations (NEX). While doubling the signal averages will double the acquisition time, the increase in the signal obtained will be the square root of 2, or only a 40% increase.

The T1 and T2 relaxation rates affect the SNR. Changing TR will affect the T1-weighting. Since T1 is relatively short, a T1-weighted image has a short TR and a short TE. To improve T2-weighting, the TR is long to allow for a longer TE.

As can be seen from above, the relationship between signal-to-noise, resolution, and acquisition is complex. Changing one element affects the others. A table is given to illustrate these direct features (Table 1.1).

To develop protocols, a very thorough understanding is needed of these interrelationships. Protocol development is beyond the scope of this text. However, familiarity with these basic principles is needed in order to maximize the protocol for the individual patient. This is more challenging in veterinary patients, which can vary tremendously in size. The ultimate goal is to maximize these relationships to provide the best possible image in a clinically viable acquisition time. While it seems counterintuitive, the smaller patient may require thicker slices to maintain sufficient SNR. Another counterintuitive imaging principle is the need to reduce the matrix to improve visualization because of its effects on SNR.

Image sequences occur as two main types. The first is the SE sequence. This is the most commonly used sequence for T1-weighted, proton density, or T2-weighted images. The variables of interest include TR and TE. SE sequences use a 90° RF pulse, followed by one or more 180° refocusing pulses.

A subset of SE sequences includes the inversion recovery (IR) sequences. IR sequences can be used to null any substance, but are most commonly used to null out cerebral spinal fluid, termed the fluid attenuation inversion recovery or FLAIR sequence, or fat, using the short tau inversion recovery or STIR sequence. An IR sequence is a 180° prepulse; time is allowed such that
the tissue to be nulled has its vector in the horizontal magnetization plane. Then, the 90° RF pulse will only affect those tissues that were not at the zero or horizontal magnetization plane. Another way to null fat is through fat saturation. These sequences consist of multiple 90° RF pulses that have relatively short TR.

The other basic type of sequence acquisition is the gradient echo (GE or GRE) sequence. The basic sequences are varied by adding de-phasing and re-phasing gradients at the end of the sequence. The variables include TR and TE, but there is also the variable of flip angle. Generally, flip angles of less than 90° are used. GE sequences can be used to acquire images rapidly and are often used for breath holding techniques and visualization of moving structures, including the cardiovascular system. GE sequences generally have less contrast than SE sequences. Lower field MR units often rely on GE sequences due to short TR and TE, permitting short imaging duration. The lack of standard T1 and T2 contrast can limit the utility of these sequences in multiple anatomical regions.
It is not the intent of this text to go through all the various imaging sequences that could be utilized with MR. These sequences are often explained similar to a recipe in a cookbook. Just as there can be an exhaustive number of recipes to cook with any given list of ingredients, the same is true for the number of imaging sequences.

Imaging sequences are generally either SE sequences or GE sequences. The majority of imaging protocols for conventional clinical MR imaging use SE sequences. GE sequences do have some specific uses. Low field magnets are often heavily dependent on GE sequences to provide shorter examinations with relatively thin sections. However, many GE sequences suffer from lack of contrast or increased magnetic susceptibility artifacts. Because of these limitations, this author favors traditional SE sequences over GE sequences. GE sequences and their specific use(s) will be highlighted throughout the book, but the coverage in this book is not exhaustive.

Standard clinical imaging sequences most commonly utilize T2-weighted sequences, STIR sequences, and T1-weighted sequences. T1 sequences are fundamental in contrast studies with the administration of a paramagnetic gadolinium-based contrast agent.

Other sequences that are commonly utilized are the FLAIR sequence, the GE sequence for the detection of hemorrhage, and heavily T2-weighted images for the visualization of fluid structures, including the subarachnoid CSF columns, the biliary system, or the fluid containing inner ear structures of the cochlea and semicircular canals.

T2 Sequence

T2-weighted sequences are the bulwark of imaging protocols. When performed with fast SE techniques, reasonable imaging time is achieved and it produces images in which both fat and fluid are seen as relatively high signal intensity. Some systems use T2-weighted fat suppression to further increase conspicuity of fluid, and have the advantage of negating the need to additionally acquire STIR images. Bright fluid in images is desirable as most pathologic abnormalities have an increased fluid signal. The fluid can be from either intracellular fluid, in the case of cellular abnormalities including neoplasia or granulomatous conditions, or intercellular fluid from diseases such as abscessation or edema.

T1 Sequence

T1-weighted sequences are generally utilized with contrast agents. The T1-weighted precontrast study is “always” necessary. One cannot definitively assess contrast enhancement without the pre-enhanced study, and a shortcut eliminating this sequence can lead to serious misinterpretation. In T1-weighted images, fat is hyperintense and fluid is hypointense. Following the administration of contrast, abnormal tissue often has an increased vascular supply leading to increased signal intensity. In some cases there are breaks in tissue structure, such as the blood–brain barrier, that allow the contrast agent to leak into the tissue and change the relaxation of the tissue leading to increased signal intensity. It must be remembered that the gadolinium contrast agent is not visualized. The only element that can be visualized at this time is the element hydrogen. Therefore, the gadolinium-based agents affect the relaxation of the protons in the molecules. This fact needs to be remembered, as the amount of contrast required for the paramagnetic effect on the proton relaxation is not as concentration dependent as iodine-based contrast agents for CT.

If a T1-weighted image (prior to the administration of gadolinium) has hyperintensity in tissues that are not related to fat, then paramagnetic substances must be present. The only paramagnetic substances within the body are iron and manganese. Since the amount
of manganese present is in a very small degree, the only reasonable element that could be present would be iron. For iron to be bright on T1-weighting, it requires a degradation of iron through normal processes until it reaches extracellular methemoglobin. The various stages of the iron degradation process that can be seen in MR images will be given with examples utilizing the brain. Again, since T1-weighted images result in high signal intensity with fat, it is often preferable to perform T1-weighted images with fat suppression. However, following the administration of contrast, it is possible that the lesion can have a relaxation time similar to that of fat and its signal can be nulled. Therefore, it is advisable to always have some postcontrast studies used without fat suppression to make certain that lesions are not lost.

**STIR Sequence**

The STIR sequence is a workhorse sequence as it allows for a T2-weighted type of image with uniform loss of the fat signal. The IR sequence is an easily performed study utilizing a 180° prepulse, prior to the 90° excitation pulse. The relaxation time of fat is known for all magnet strengths. Therefore, it is easy to set the time of inversion (TI) for a specific magnetic field strength, which will ensure uniform and generalized suppression of the fat signal. STIR sequences should always be performed prior to the administration of contrast. It is possible that contrast enhancement could change the relaxation time of the tissues similar to fat, and again the tissue’s signal will be nulled on a STIR sequence if performed after contrast administration. STIR sequences are utilized as they display normal vascular or other fluid-filled structures as bright on a generalized dark background. Typically, pathologic changes in tissue are easily detected as “stars” in a dark sky.

**FLAIR Sequence**

The FLAIR sequence is similar to the STIR sequence except it uses an inversion time to get null fluid signal. In general, the FLAIR is utilized in the brain and gets rid of the usually hyperintense fluid signal from the cerebral spinal fluid. Therefore, lesions that are periventricular are easier to detect as increased signal intensity, adjacent to a black or darkened cerebral spinal fluid. The attenuation appearance of cerebral spinal fluid is somewhat dependent on time of inversion as well as other factors specific to MR unit. With some protocols, one is capable of detecting abnormal cerebral spinal fluid from its appearance on the FLAIR sequence. The abnormal signal appearance could be due to increased protein content and/or cellularity or associated CSF flow.

This sequence can be useful when applied to other fluids. A FLAIR sequence can be used to get a T2-weighted image of a urinary bladder tumor. Since urine is basically acellular and with no proteins, the T1 for CSF can be used. The nulling out of fluid from conditions such as hydrothorax allows darkening of the effusion while still allowing visualization of T2-weighted image characteristics of the thoracic wall and organs. IR times for such studies vary due to many parameters, but essentially all fluid, including urine, synovial fluid, thoracic and abdominal effusions, or cerebral spinal fluid can be nulled with the FLAIR technique.

**Gradient Echo Sequence**

GE techniques are the most commonly used sequence for rapid studies, and as such are often utilized for localization sequences. In the brain, its most common clinical application is to verify the presence of hemorrhage. GE sequences are very sensitive to magnetic field inhomogeneities. Therefore, the iron concentration within hemorrhagic tissue is detected as a magnetic field inhomogeneity. Unfortunately, this same degree of inhomogeneity can cause massive image artifacts from small metallic implants including BBs, steel bird shot, the wire around microchip placement, or simple anatomic tissue differences including the frontal sinuses (air) next to the brain. Other than the benefit of detecting hemorrhage, the overall tissue contrast is poor with GE sequences, even though T1- and T2-weighted GE sequences are available. Therefore, this author tends to limit them to the detection of hemorrhage in studies of the CNS. GE sequences are commonly used in the thoracic and abdominal studies to minimize motion artifact.

In the subsequent chapters, we will utilize a few additional sequences for the visualization of specific structures. These are often heavily T2-weighted images that allow the visualization of structures including vascular structures, the equivalent of an MR myelogram, or the fluid in the semicircular canals and cochlea.
The goal of all imaging modalities is to aid in visualization of normal anatomy and disease states. Unfortunately, all imaging modalities have some artifacts that can mimic pathologic change and lead to misdiagnosis. MR is no different. Knowledge of the common MR artifacts and the ability to distinguish artifactual change, which may be mimicking pathologic change from a true pathologic abnormality, are critical to accurate interpretation of MR images.

**Hardware Artifacts**

Some artifacts come from the magnetic field inhomogeneity. The artifacts can be intensity, spatial, or both. An artifactual bending of the spine may be seen when at the edge of the main magnetic field (Figure 1.1). The patient should be repositioned in the gantry if this is creating a diagnostic dilemma.

Artifacts can also occur from defects in the RF shielding. The shield could be faulty but often these artifacts are from transient breaks in the shielding and most often are seen if someone enters the magnet room during a sequence. This type of “zipper” artifact can be avoided by waiting until the end of a sequence to enter the room (Figure 1.2).

The advantage of MRI over CT is its ability to visualize the body in any plane. As mentioned, the three common planes are (1) transverse (axial), (2) dorsal (coronal), and (3) the sagittal plane. The images are made from different slices within these planes, which are formed from the three magnetic gradients used. When understanding the orientation gradients, it is useful to assume that the patient went into the bore of the magnet head first and supine for the imaging study. One of the gradients is selected for the slice selection to provide the desired plane. In this scenario, the gradient in the Z direction is used for transverse or axial slices, the X direction is used for sagittal slices, and the Y direction is used for coronal or dorsal slices. All gradients can be modified depending on the patient positioning. For instance, if the patient is in the right decubitus position, then the Y gradient would be used for the sagittal slices of the body.

The other two gradients map the signal in the two dimensions of the slice plane. The signal is mapped according to its phase and frequency. The frequencies of the signals are similar to the range frequencies of different radio stations. Think of the phase as a time zone. There could be an FM 101.1 station in Denver, Colorado, and a station with the same frequency in Los Angeles, California. Then, if one were to realize that the time zones are a continuum from east to west, this allows for many more time zones than the current artificially drawn time zones for a 24-h clock. The number of frequencies and the number of phases are often 256, but can go much higher.

**Phase and Frequency Artifacts**

Some artifacts are propagated in the frequency direction while others are propagated in the phase direction. Therefore, a prior knowledge of the direction of these encodings is needed to determine the image artifact. Some institutions print this information in the images. It is always part of the DICOM header information that can be assessed if one can access this file. It is often simpler to find some ubiquitous motion artifact from flowing blood, for example, that will be propagated in the phase direction. By changing the background of the image, one can readily depict this in the background (Figure 1.3). Motion artifact can be from gastrointestinal motion, respiratory motion, blood flow, or patient motion. Attempts to limit some of this motion can be made by gating the acquisition to the respiratory or cardiac cycles. This form of image acquisition will prolong the study to a degree and may limit its clinical utility. Therefore, some motion is generally an accepted consequence of MRI.
The signal from flowing blood is often accentuated following the administration of contrast due to the increased signal intensity of the blood with the contrast agent. Pseudolesions can be seen that would mimic pathologic change. Swapping the phase and frequency encoding directions allows one to ascertain, with certainty, if this is artifactual or real (Figure 1.4).

**Chemical Shift Artifact**

Another artifact that can be confused with pathologic change is a chemical shift artifact. This artifact is a mis-mapping that occurs at water–fat interfaces and in the frequency direction. The artifact is often easily recognized in abdominal studies, but in other areas can mimic pathologic change (Figure 1.5). The same ability to swap the phase and frequency encoding directions and to change the direction of the chemical shift artifact helps clarify the existence of the artifact versus pathologic change (Figure 1.6).

**Fold-Over or Aliasing Artifact**

Fold-over or aliasing is another artifact that occurs in the phase direction. This artifact occurs when a portion of anatomy is outside the selected image field of view. This anatomy can be wrapped around to the opposite side of the image as a mirror image into the area of interest. This can confuse the interpretation (Figure 1.7). Field of view should be large enough to encompass...
Figure 1.3. Motion artifact. (A) Apparent small lesion in the left occipital cortex on T1 postcontrast. (B) Lesion gone. (C) Brightened background and phase is ventral to dorsal and flow artifact is seen in alignment with the “lesion.” (D) Brightened background with phase left to right.
Figure 1.4. (A) STIR sagittal sequence of the lumbosacral spine showing the presence of hyperintensity of the endplates at L7-S1. (B) T2-weighted sagittal image of the thoracic spine showing hyperintensity of the endplates at two sites. (C) STIR sequence of the same location as (B). Hyperintensity is somewhat obscured due to flow artifact from the aorta. Phase direction is in the foot-to-head direction such that the aortic signal is bleeding into the spine at this site. (D) Contrast enhancement T1-weighted sequence with fat suppression. The aortic signal can readily be seen bleeding through the spine, spinal cord, and the dorsal spinous processes of the first few thoracic vertebrae. This type of flow artifact could be prevented by changing the phase direction, but then the entire aortic signal would have motion artifact into the spinal cord. Therefore, foot-to-head direction as in this case is greatly preferred, but one must be cognizant of the artifact. (E), (F) T1 postcontrast images of a brain. Brain: phase is going in the direction of the arrows on the left-hand side of the image. The phase is left to right in this image and the arrows point to hyperintensities within the cerebellum. These are flow artifacts, probably from the internal carotid artery, and as in part (F) these hyperintensities are not present when the phase direction is changed to a ventrodorsal direction. Any time contrast enhancements cannot be substantiated on multiple planes, they should be suspect. If one needs to prove the presence of artifacts, the change in phase direction with a repeating of the sequence, as in this case, can be helpful.
Figure 1.5. (A) Phase direction is ventrodorsal, which means the frequency direction is foot to head. This chemical shift artifact can readily be seen as the black line at the posterior aspect of the spleen and the white line at the cranial aspect. This chemical shift artifact occurs due to the water signal from the spleen interacting with the abdominal fat signal. When the phase direction is changed, as in (B), to a ventrodorsal orientation, the change in direction of the chemical artifact is readily seen.

Figure 1.6. (A) Chemical shift artifact in the spinal column. The phase direction is in the direction of the arrows on the left-hand side of the image, in a left-to-right direction. Therefore, the chemical shift is of a ventrodorsal nature. There is a decreased signal intensity at the dorsal aspect of the subarachnoid space, which is artifactual and slightly brighter ventrally, which is also artifactual. (B) The phase direction is ventrodorsal and the frequency direction is left to right. Now, there is a black line at the left-hand side of the subarachnoid space and a bright line at the right side. These are both artifactual due to the chemical shift between the water of the subarachnoid space and the epidural fat. This type of chemical shift artifact is often seen in large dogs due to the amount of epidural fat. This should not be mistaken as a dural lesion.
the anatomy visualized or techniques for fold-over suppression need to be employed to avoid this artifact. Unfortunately, many of these suppression techniques result in increased scan time. Therefore, where possible, fold-over suppression should not be used to either reduce acquisition time or improve signal-to-noise levels. Since fold-over is in the phase direction, the phase must often be set in a certain direction to prevent blood flow related artifacts within the area of interest. For instance, in sagittal images of the lumbar spine, it is preferred to have phase oriented in a foot-to-head direction and, thus, the frequency going anterior to posterior. If one were to have the phase encoding direction going ventral to dorsal, the blood flow artifact from the aorta would superimpose on the spinal cord leading to erroneous interpretation. In this instance, the phase must be oriented foot to head and fold-over suppression is needed to prevent wrap-around artifact.

**Truncation Artifact**

Truncation artifact occurs when the number of phase encoding steps is decreased in relation to the frequency encoding steps to save time. With excessive reduction in phase encoding steps there may be a mis-mapping of the image in the phase direction. Truncation artifacts can make conditions such as a dilated central canal within the cervical spinal column or the appearance of a syringohydromyelia appear in an image when, in fact, none exists. Often, this artifact is readily seen and ignored when the change is only seen on one plane and cannot be confirmed on an orthogonal view (Figure 1.8).

**Magnetic Susceptibility Artifact**

One of the more sizable artifacts is from magnetic susceptibility. This artifact occurs when magnetic material is present within the patient. Ferrous metal is especially problematic, including BBs and steel bird shot. Other sources of this artifact can come from the spring on the identification microchips, orthopedic devices, or small bits of metal left behind from a surgery. It is
important to appreciate that small metal fragments that are too small to be seen on a conventional radiograph can create very large artifacts in the MR image. Other problem substances come from ingested rocks containing minerals and even from barium sulfate suspension. High concentrations of gadolinium, especially in the urinary tract from excretion of the contrast agent, will result in a dark artifact on T2- and T1-weighted sequences, which is a magnetic susceptibility artifact (Figure 1.9–1.11).

**Volume Averaging Artifact**

The signal intensity of the voxel is the average of all the different signal intensities from different tissues within the slice thickness. When a slice “cuts” through areas of disparate intensity or contour, the intensity mapped into the voxel is a misrepresentation of the different structures. This is referred to as volume average or slice thickness artifact. It can greatly affect spatial resolution. This is often seen where the change or difference is of small volume (some disc herniations) or where there is a marked change in contour. The curvature of the calvarium at the frontal sinus and brain interface can make lesions, like a meningioma, appear to cross the bone and occupy the sinus (Figure 1.12). The orthogonal views must be assessed to evaluate partial volume average artifact.

**Magic Angle Artifact**

This artifact affects structures of low signal intensity that are oriented at a 55° angle off the main magnetic field. Their true signal is misrepresented as hyperintense on short TE sequences. There are two 55° cones, one positive and one negative within the bore of the magnet. This artifact is common in orthopedic studies, recognized most commonly in tendinous or ligamentous structures. These, in a normal state, have low signal intensity and in diseased states are hyperintense. Thus, this artifact produces lesion in certain orientations, and is avoided by the inclusion of a T2-weighted (long TE) SE sequence in all studies (Figure 1.13).

**Cross-Talk Artifact**

This artifact occurs when setting up multiple stacks of images and their fields of view intersect, which results in loss of signal (Figure 1.14). This can be avoided by adding an additional sequence to allow for proper placement of all slices. This is especially a problem in the lumbosacral region.
Figure 1.9. (A), (B) Artifact from a small BB in the region of the subcutaneous tissues of this cat. A small BB still causes a very large magnetic susceptibility artifact negating visualization of the lumbar spine in this patient. Part (A) is the gradient echo localizer sequence. Gradient echoes are more prone to magnetic susceptibility artifact and a very large black hole can be seen. Part (B) is a T2-weighted sequence showing some visualization of the spine, but marked warping of the image is due to the magnetic susceptibility artifact. (C) Two stainless steel orthopedic screws placed across the facets. While stainless steel creates some magnetic susceptibility artifact, it is nowhere near that seen with the steel of a BB. However, part (D) shows the warping of the image from these stainless steel screws in the facets. The curvature of the spinal cord that appears in C3 is artifactual due to the magnetic susceptibility artifact. (See Color Plates 1.9C,D.)
Figure 1.10. This image shows the difference between types of metal and the artifacts that are created. (A) Radiograph showing a small steel shot next to the vertebra. (B) This small piece of steel creates a huge magnetic susceptibility artifact negating visualization of the spine in the L4 through L6 region. Part (C) is a radiograph of an animal that has suffered a gunshot wound. In this case, the metal is lead. While the radiograph shows fragmentation of the lead, the radiograph cannot depict the spinal cord. Numerous small fragments of lead are identified with the arrows. (D) The MR shows that the lead does not create a magnetic susceptibility artifact and the spinal cord can be seen. The arrows point to a small osseous fragment that has been created from the gunshot wound. The hyperintensity on this STIR sequence is hemorrhage and edema from the gunshot wound. (See Color Plates 1.10A–D.)

Figure 1.11. (A) T1 fat-saturated postcontrast sagittal image. The very low signal intensity within the renal pelvis is due to the high concentration of gadolinium contrast agent that is being excreted by the urinary system. The concentration is so high that instead of being “enhanced,” it actually gets a low signal intensity with this high concentration. (B) T2-weighted sagittal image following the administration of contrast showing the same low signal intensity of the renal pelvis due to the high concentration of gadolinium. This is also commonly seen in the urinary bladder and should not be mistaken for a lesion.