

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

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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

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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, superparamagnetic, 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 at 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 6×10^{15} .

Basic physics dictate that the energy is proportional to the nuclei's unique resonant frequency in MR; this is called the Larmor frequency. The frequency of the spinning 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. MRI is able to make high-quality images, not because of the energy of the spinning protons, but due to the abundance of hydrogen protons present in the body. The spinning or "resonating" of nuclei occurs because of unpaired electrons in the orbital shell. Each nucleus with this characteristic will resonate at a unique frequency. The spinning protons act like toy tops that wobble as they spin. The rate of wobbling is termed precession. These precess at the resonance or Larmor frequency for hydrogen.

If a radiofrequency (RF) pulse is applied at the resonance frequency, the protons can absorb that energy. The absorption of energy causes the protons to jump into a higher energy state. This causes the net magnetization to spiral away from the main magnetic field, designated B_0 . The net magnetization vector, therefore, moves from its initial longitudinal position a distance proportional to pulse, which is determined by its temporal length and strength. After a certain length of time, the net magnetization vector would rotate 90° and lie in a transverse plane. It is at this position that no net magnetization can be detected. When the RF pulse is turned off, three things start to happen simultaneously:

1. The absorbed energy is retransmitted at the resonance frequency.
2. The spins begin to return to their original longitudinal orientation, termed the T1 relaxation.
3. While the precessions were initially in-phase, they begin to de-phase, termed T2 relaxation.

The return of the excited nuclei from the high energy state to their ground state is termed T1 relaxation (or spin-lattice relaxation). The T1 relaxation time is the time required for the magnetization to return to 63% of its original longitudinal length. The T1 relaxation rate is the reciprocal of the T1 time ($1/T1$). T1 relaxation is dependent on the magnetic field strength that dictates the Larmor frequency. Higher magnetic fields are associated with longer T1 times.

T2 relaxation occurs when spins in high and low energy states exchange energy but do not lose energy

to the surrounding lattice as occurs in T1 relation. It is, therefore, sometimes referred to as spin-spin relaxation. This results in loss of transverse magnetization. In biological materials, T2 time is longer than T1 time. T2 relaxation occurs exponentially like T1 and is described as the time required for 63% of the transverse magnetization to be lost. In general, T2 values are unrelated to field strength. In patients, the magnetic signal decays faster than T2 would predict. There are many factors creating imperfections in the homogeneity of the magnetic field, including the magnet and patient inhomogeneities including surface contours, air-tissue interfaces, and any metal the patients may have within them, including dental work, staples, and orthopedic appliances. The sum effect of all of these inhomogeneities pronounces an effect called T2*. The T2 relaxation comes from random interactions, while T2* comes from a combination of random and fixed causes including magnet and patient inhomogeneity.

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

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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. The time between RF pulses to move the protons out of

Table 1.1.
Relationship of image parameters to signal-to-noise ratio (SNR), resolution, and imaging time

Increase Parameters	SNR	Resolution	Acquisition Time
FOV	+	–	nc
NEX	+	nc	+
Slice thick	+	–	nc
Gap	+	–	nc
TR	+	nc	+
TE	–	nc	nc
Matrix size	–	+	+
Bandwidth	–	nc	nc
Magnet strength	+	nc	nc

nc, no change.

longitudinal alignment is termed TR (repetition time). 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

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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.

SECTION 3

Sequence Selection

Patrick R. Gavin

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 often 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 atten-

uation 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 TI 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 anatomical 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.

SECTION 4

Artifacts

Patrick R. Gavin

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.

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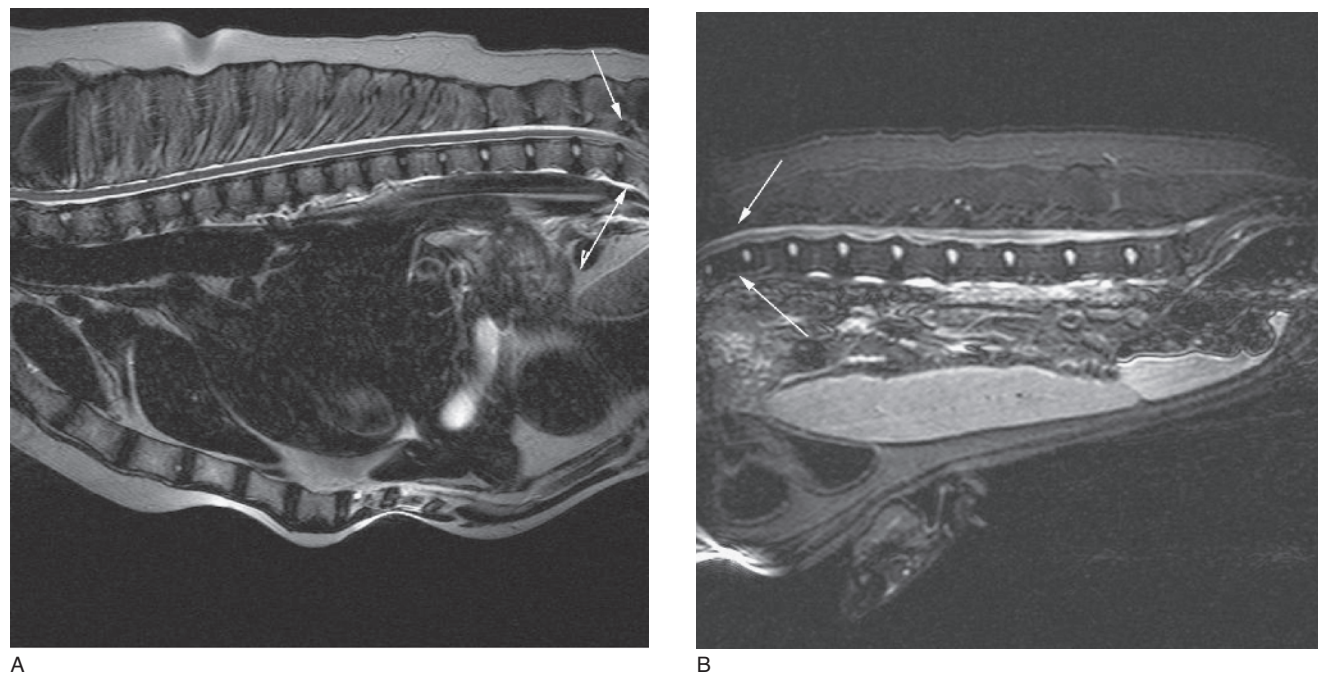


Figure 1.1. (A) T2-weighted sagittal imaging. Arrows show the bending of the field at the edge of the field of view. This is from magnetic inhomogeneity at the periphery of the imaging volume. (B) Sagittal STIR sequence of the lumbar spine of the same patient showing bending at the opposite end of the patient from the same affect.

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

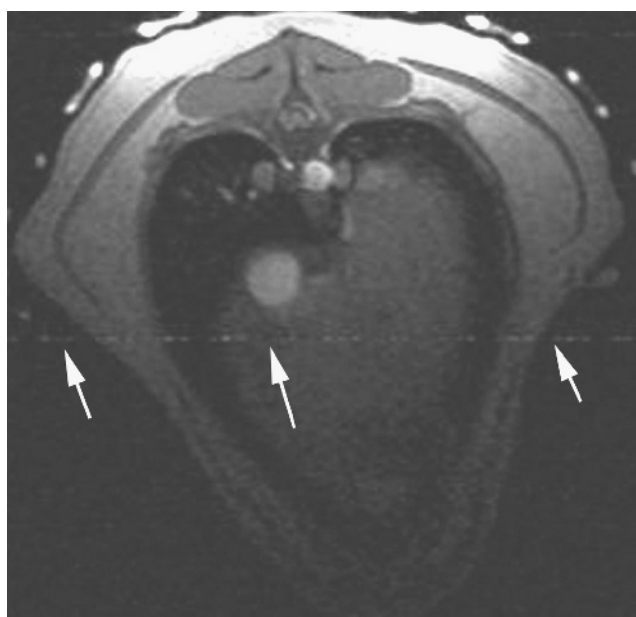


Figure 1.2. Zipper artifact: T2-weighted transverse image with a horizontal zipper artifact (arrows).

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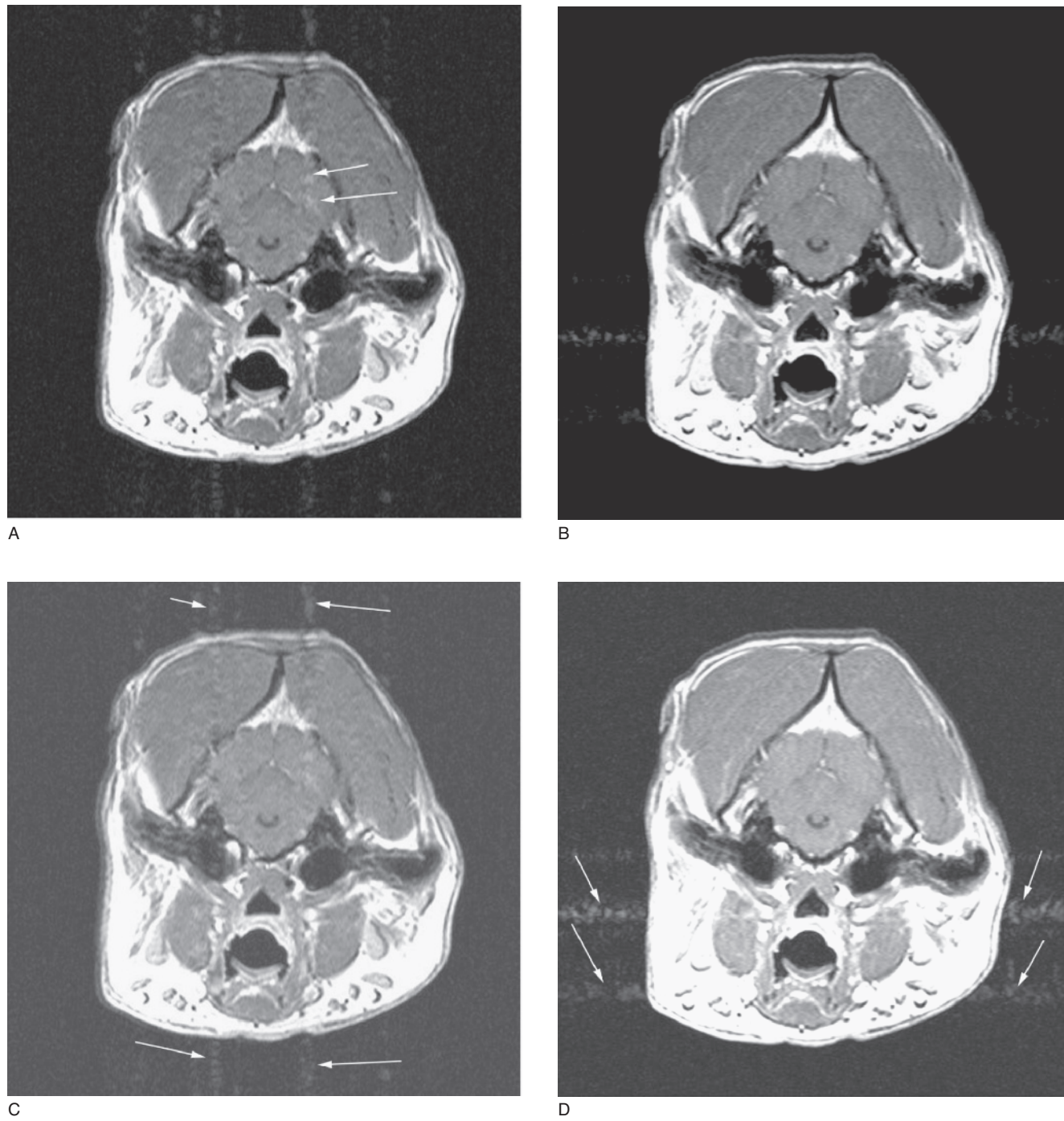


Figure 1.3. Motion artifact. (A) Apparent small lesion in the left occipital cortex on T1 postcontrast. (B) Lesion gone. (C) Brightened background and phase is ventral to dorsal and flow artifact is seen in alignment with the "lesion." (D) Brightened background with phase left to right.

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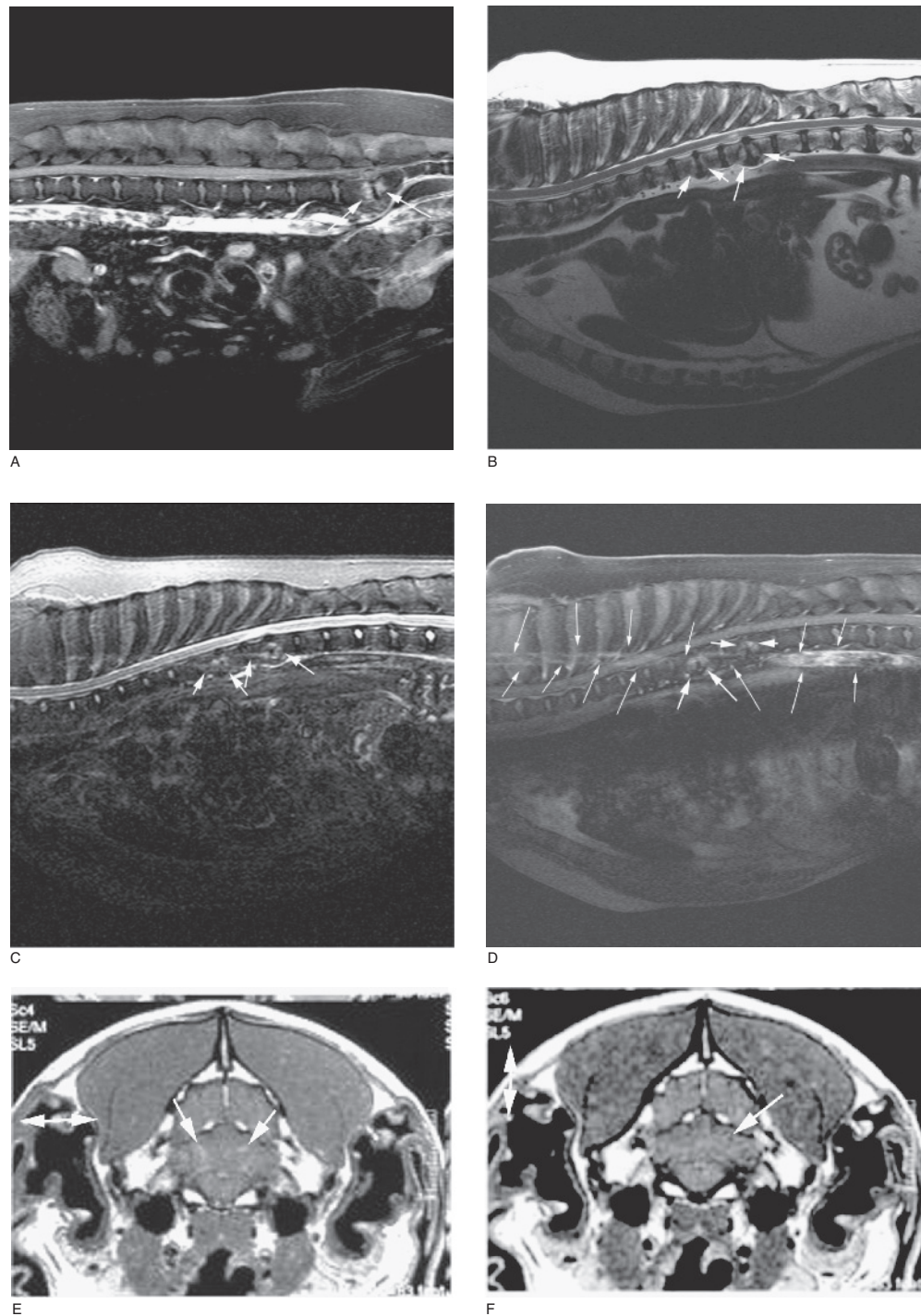


Figure 1.4. (A) STIR sagittal sequence of the lumbosacral spine showing the presence of hyperintensity of the endplates at L7-S1. (B) T2-weighted sagittal image of the thoracic spine showing hyperintensity of the endplates at two sites. (C) STIR sequence of the same location as (B). Hyperintensity is 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.

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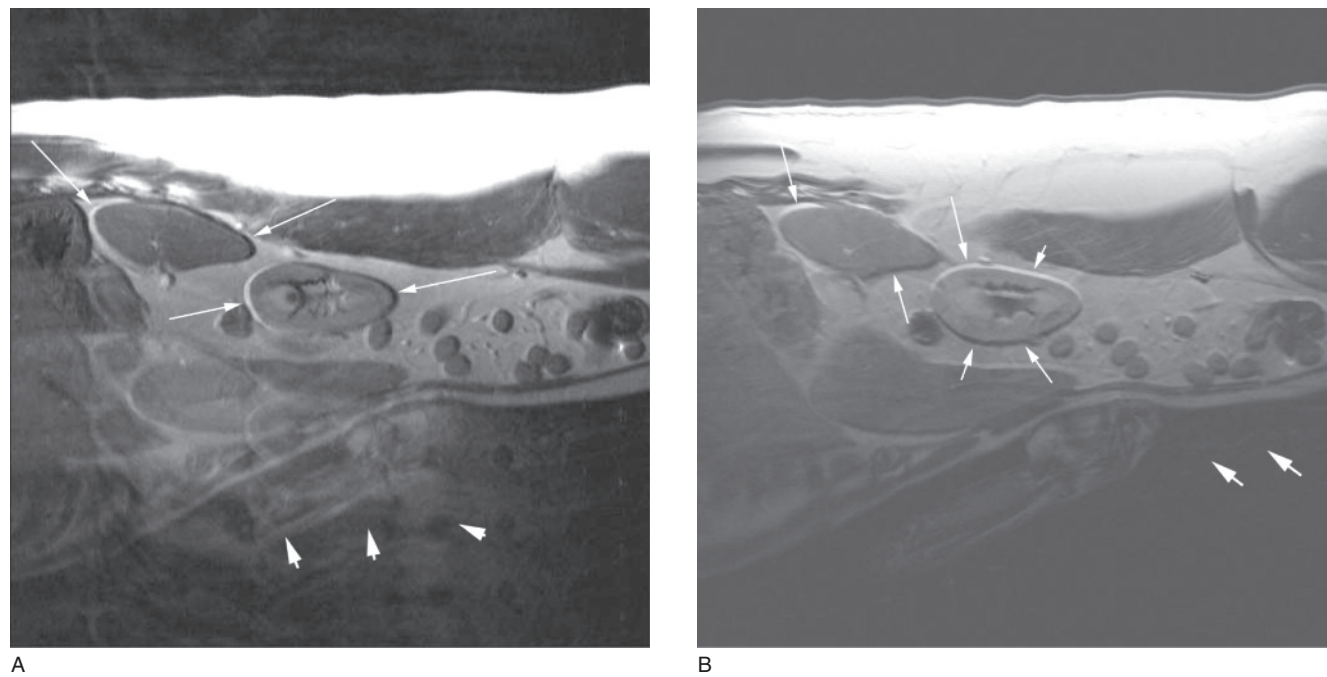


Figure 1.5. (A) Phase direction is ventrodorsal, which means the frequency direction is foot to head. This chemical shift artifact can readily be seen as the black line at the posterior aspect of the spleen and the white line at the cranial aspect. This chemical shift artifact occurs due to the water 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.

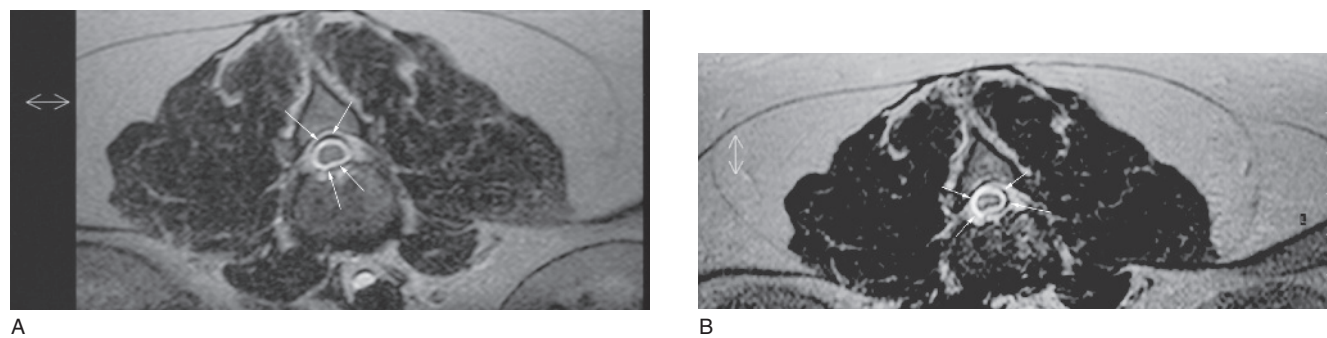


Figure 1.6. (A) Chemical shift artifact in the spinal column. The phase direction is in the direction of the arrows on the left-hand side of the image, in a left-to-right direction. Therefore, the chemical shift is of a ventrodorsal nature. There is a decreased signal intensity at the dorsal aspect of 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.

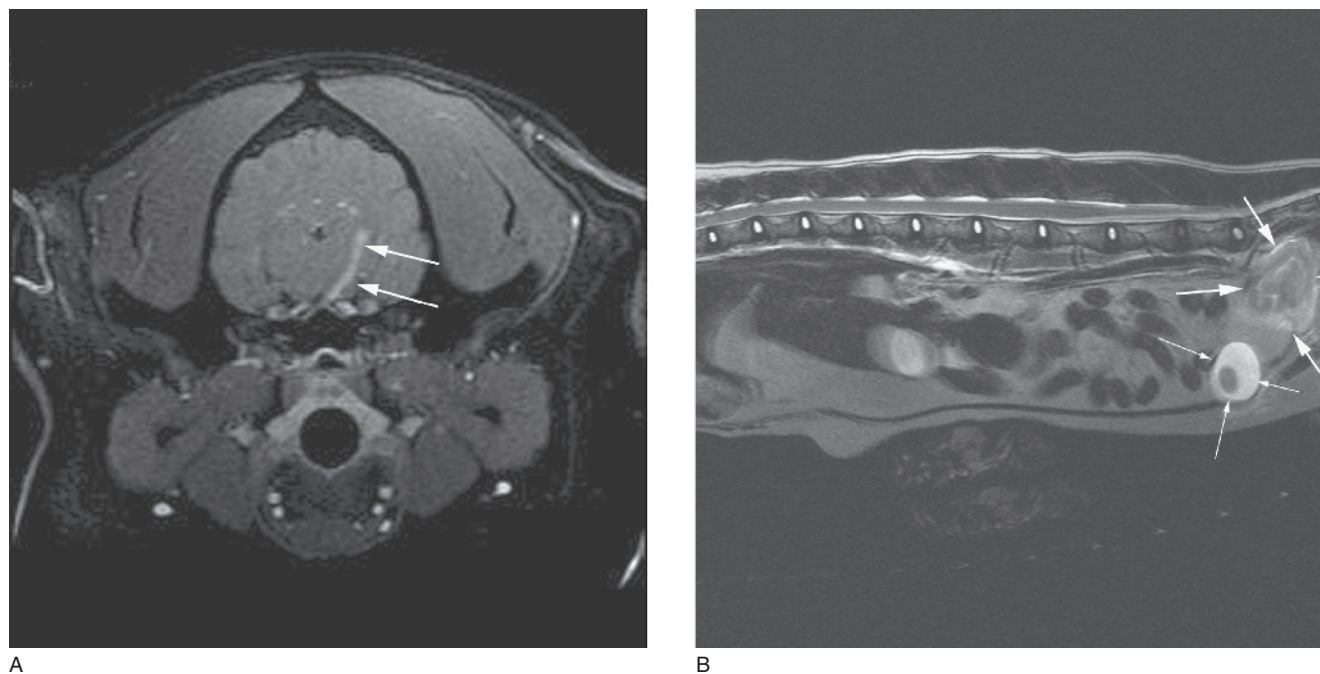


Figure 1.7. (A) Fold-over or aliasing artifact of the pinna of the ear superimposed over the brain. These artifacts can be negated by a slight increase in the field of view to include the pinnae. One should also try to keep the pinnae close to the surface of the head when imaging to help alleviate this problem while maintaining a relatively small field of view. (B) Aliasing artifact where the head is being superimposed over the caudal portion of the abdomen. The small arrows show the eye and the larger arrows the brain, superimposed over the region of the urinary bladder. While the eye makes it relatively easy to see this fold-over, if one was at a different sagittal plane, the artifact could be easily misinterpreted for a pathologic lesion. One of the best ways to prevent these confusions is to always obtain sequences in multiple planes, and be able to confirm lesions on more than one sequence and more than one plane.

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

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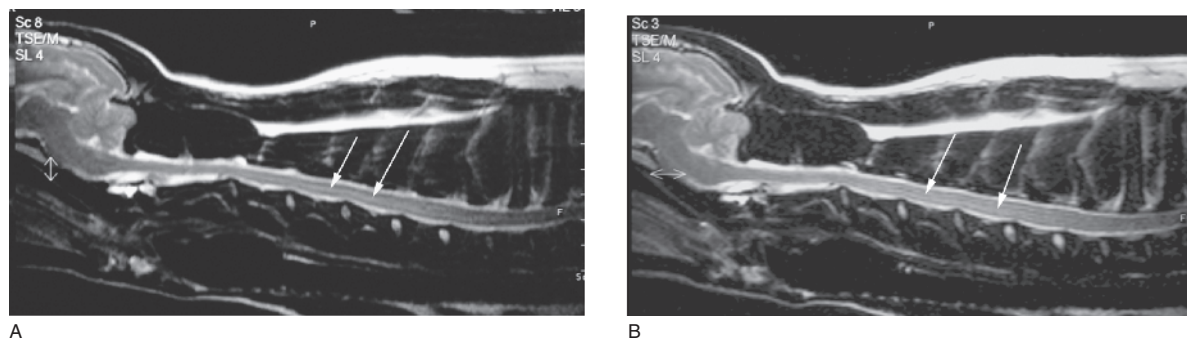


Figure 1.8. (A) Truncation artifact. Truncation artifacts create a mis-mapping of the signal. In this case, the subarachnoid signal is being mis-mapped into the center of the spinal cord making it appear as an enlarged central canal or syrinx. This happens when the phase and coding steps are reduced too much in an effort to save time. By increasing the phase and coding steps, as in (B), the hyperintense signal in the center is negated.

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 when 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 orthogo-

nal 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.

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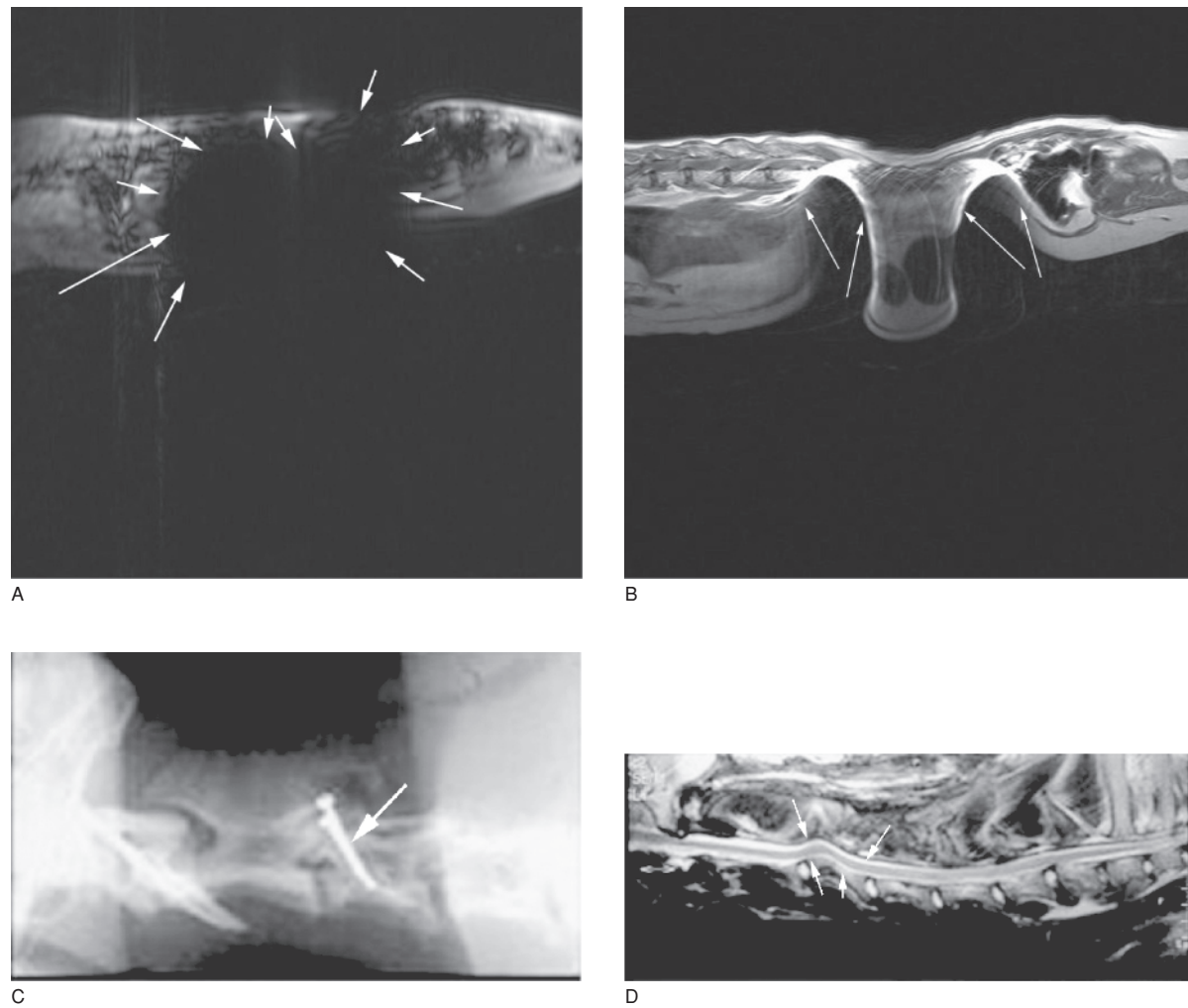


Figure 1.9. (A), (B) Artifact from a small BB in the region of the subcutaneous tissues of this cat. A small BB still causes a very large magnetic susceptibility artifact negating visualization of the lumbar spine in this patient. Part (A) is the gradient echo localizer sequence. Gradient echoes are 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.)

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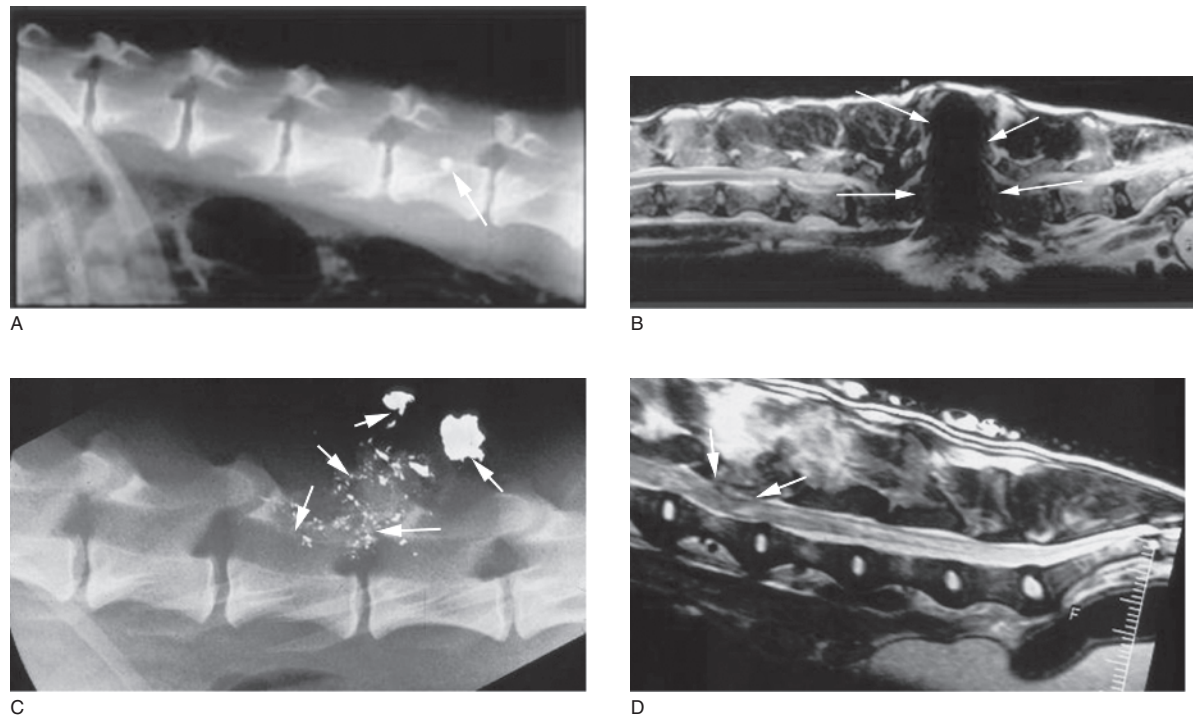


Figure 1.10. This image shows the difference between types of metal and the artifacts that are created. (A) Radiograph showing a small steel shot next to the vertebra. (B) This small piece of steel creates a huge magnetic susceptibility artifact negating visualization of the spine in the L4 through L6 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.)

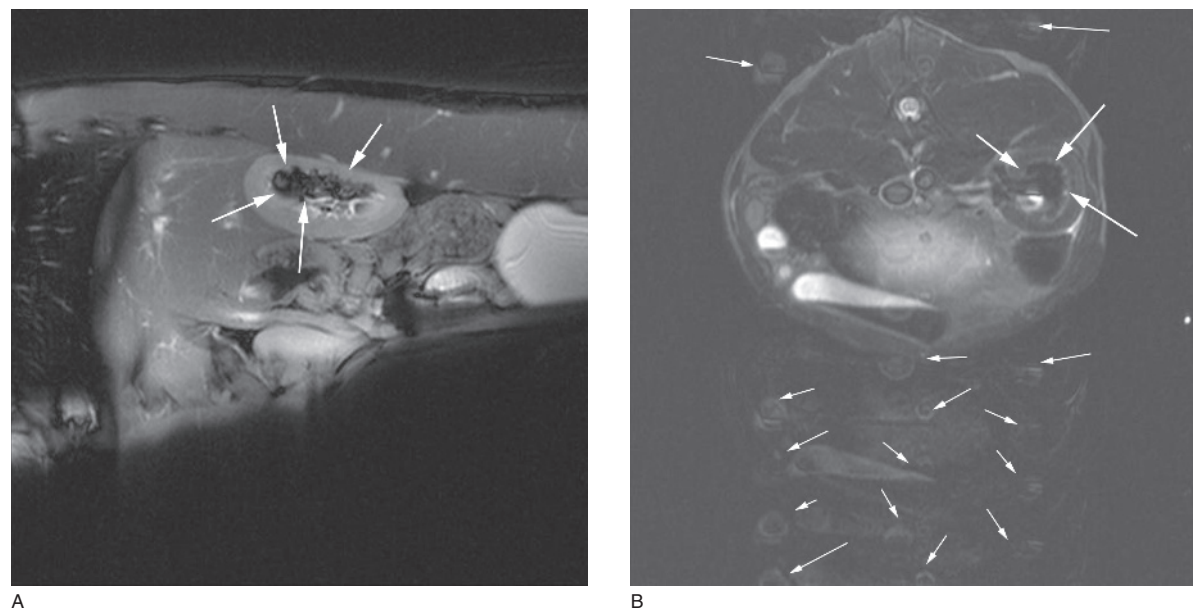


Figure 1.11. (A) T1 fat-saturated postcontrast sagittal image. The very low signal intensity within the renal pelvis is due to the high concentration of gadolinium contrast agent that is being excreted by the urinary system. The concentration is so high that instead of being “enhanced,” it actually gets 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.

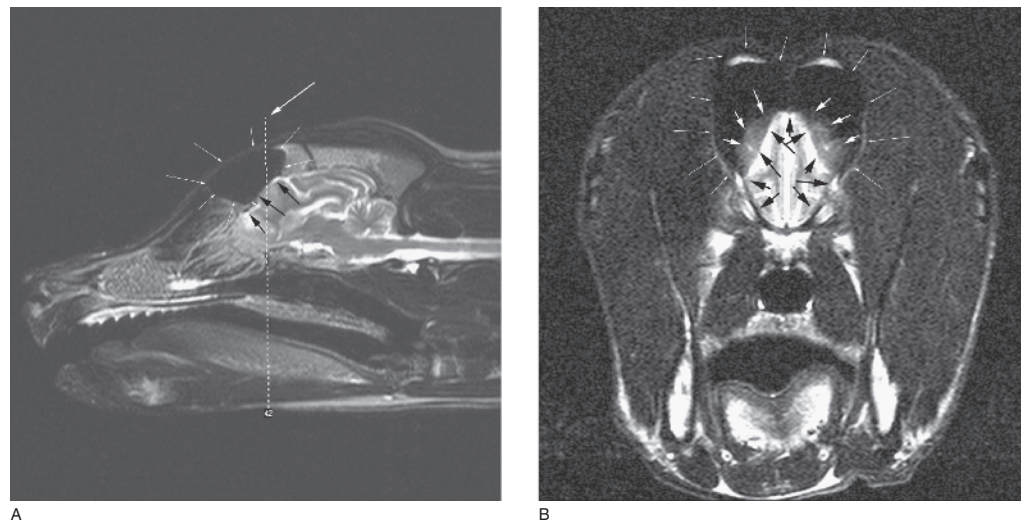


Figure 1.12. Volume averaging artifact. (A) Sagittal T2 image. Dotted line (large white arrow) is the location of (B). The smallest white arrows depict the air-filled frontal sinus. The black arrows point to the periphery of the olfactory bulb of the cerebrum. (B) T2 transverse image. This image is the average of the signals from a 4 mm slice thickness, 2 mm on either side of the dotted line in (A). The smallest white arrows depict the air-filled frontal sinus. The black arrows point to the periphery of the olfactory bulb of the cerebrum. The larger white arrows indicate the volume averaging of the brain and frontal sinus in this 4-mm-thick section.

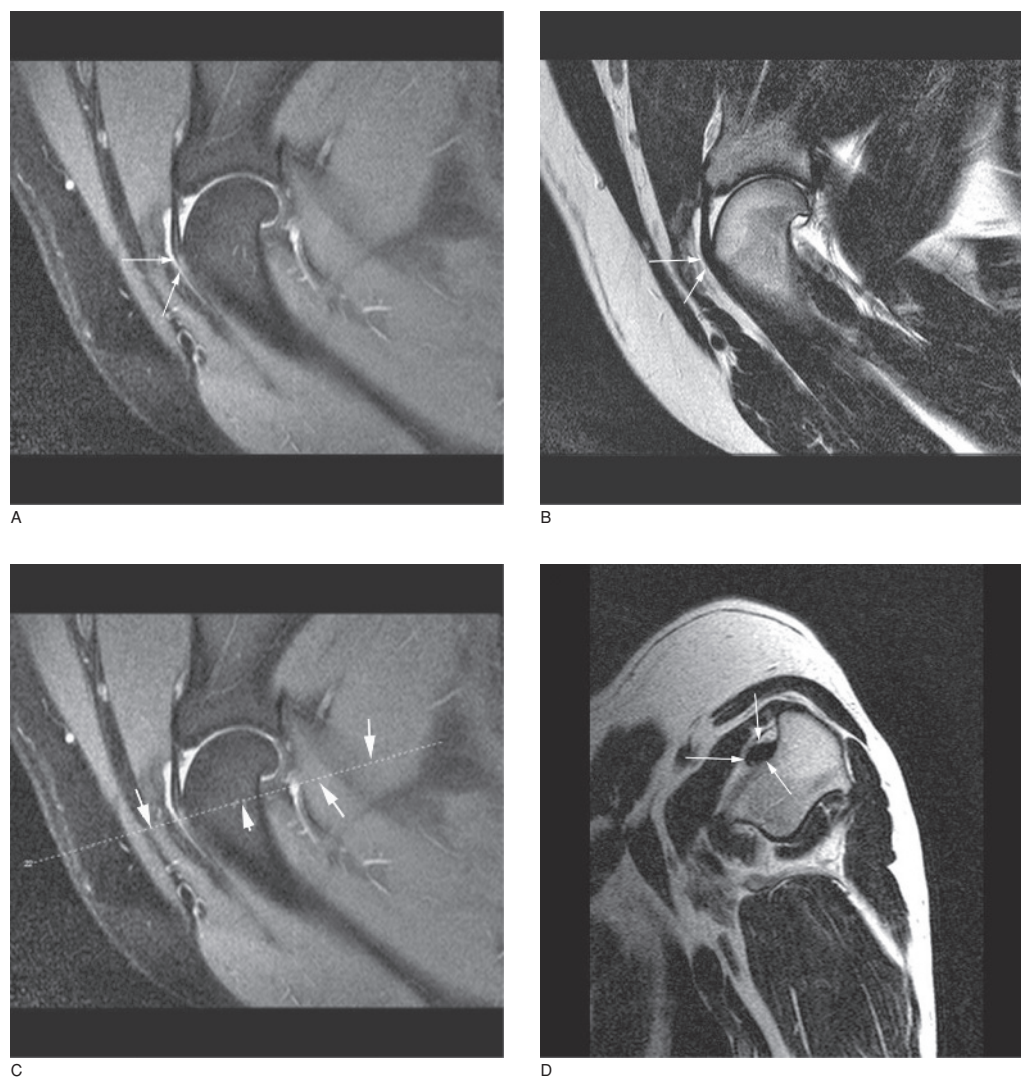


Figure 1.13. Magic angle artifact. (A) Proton density fat-saturated image with hyperintensity in the biceps tendon (arrows). (B) T2 image of same slice as in (A) with normal intensity of the tendon (arrows). (C) Same image as (A) showing location of (D). (D) Transverse T2 image of the biceps tendon with uniform signal.

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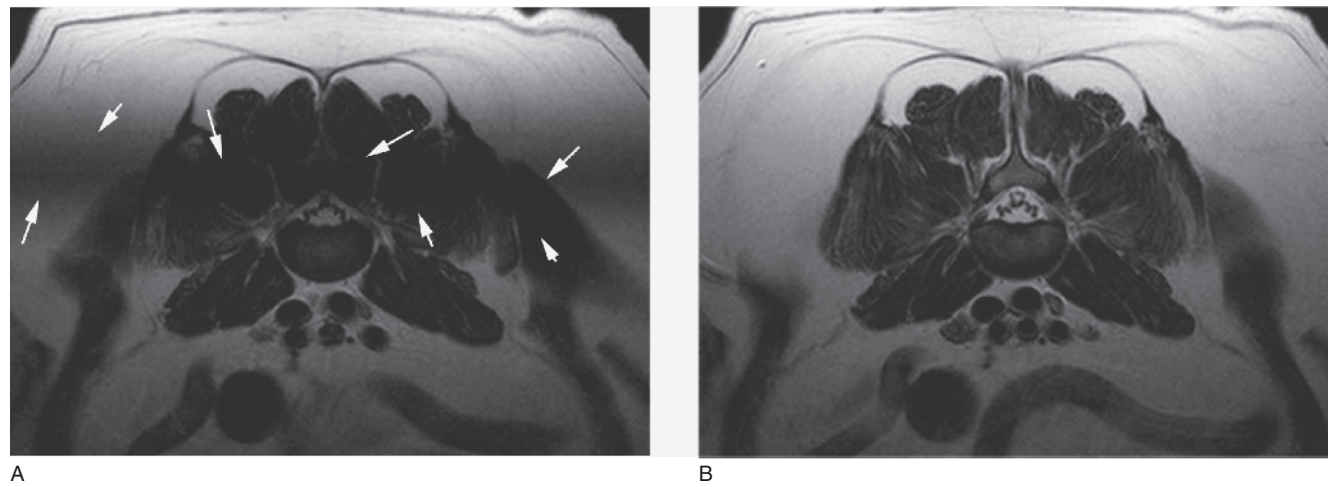


Figure 1.14. Part (A) shows the lack of signal homogeneity due to cross-talk when multiple stacks intersect each other. This commonly occurs at the lumbosacral area due to the change in the angulation. The transverse images should be perpendicular to the spinal canal. This can result in intersection between the images through the caudal lumbar area and those through the lumbosacral junction. The homogeneity of the signal can be seen in (B) where this problem was eliminated by a separate series of slices through the lumbosacral area with no intersection of neighboring slices.

SECTION 5

Equipment Consideration and Selection

Patrick R. Gavin

The main component of the magnetic resonance unit is obviously the magnet. The main magnetic field is called the B_0 field. Within the magnet are gradient coils needed to produce a GE in the X, Y, and Z directions or a gradient in B_0 . Within the gradient coils are RF coils that provide the RF energy needed to rotate the nuclear spins by 90° or any other value selected by the sequence. The strength of the gradient coils determines the ability of an MR unit to change the magnetic field per unit distance. The strength of the gradient coils will have the largest contribution to end-plane resolution. The RF coil, besides an emitting RF coil, is also a receiving coil capable of detecting the signal from the spins within the body.

Most modern magnets are self-shielded with an opposing magnetic field such that they do not need to be magnetically shielded. All magnets do, however, need to have some form of RF shielding. This shield not only prevents RF pulses from radiating out from the magnet, but also prevents RF signals from television, radio, cell phones, etc. from being detected by the imager.

MRI requires powerful computers since every component is under the control of the computer. The computer controls the shape and amplitude of the gradient fields, and the strength and duration of the RF pulses. The computer also provides the necessary method to convert the received RF energy into an image.

The imaging magnet is the most expensive component. Permanent magnets are present as smaller field units. These magnets must be pure and uniform. Permanent magnets have the advantage of not requiring cryogenics. They are heavy but their main drawback is their field strength. Most higher field MR units are superconducting magnets. A superconducting magnet keeps the temperature close to zero kelvin by immersing it in liquid helium. Once electrical current is initiated in the coil, it will continue so long as it is maintained at this temperature. Liquid helium is at -269°C or 4° above absolute zero. Large volumes of liquid helium are required, which is also costly with regard to

maintenance. In initial designs, the liquid helium was surrounded by liquid nitrogen to decrease helium consumption. Currently, cryo-coolers are used to maintain the liquid helium temperature and have eliminated the need for liquid nitrogen. Early magnets required approximately four refills per year of liquid helium. Current machines can be filled less than yearly and some every ten years.

The gradient coils within the main magnet are fundamental to image creation. While the “body coil” can be used for a receiver coil, often specialized RF detector coils designed specifically for certain body areas are utilized to receive the image. There are numerous types of coils. Some of these are volume or quadrature coils with the antenna coils running at right angles to each other to better capture the signal. Other coils are linear coils. Linear coils are often flexible to allow for contouring of the coil to the patient. Circular coils are commonly used for extremity work. The depth of penetration of a circular coil is equal to its radius. One of the more useful coils for veterinary imaging is a multi-element spine coil. Multiple element spine coils allow for the entire patient’s spine to be imaged without physically repositioning the patient. This can greatly speed the imaging process while providing various fields of view for proper examination of the spinal column.

SAFETY

Safety is of critical importance with MRI. The safety issues are related to dangers associated with ferromagnetic objects near the magnet. Ferromagnetic objects that are often forgotten include pagers, cell phones, hoof knives, scissors, and other sharp or heavy objects. Obviously, items such as ferrous oxygen tanks, standard ECG machines, etc. can become flying, and potentially lethal, projectiles inside an MR suite. All personnel entering an MRI suite should be given instructions on MR safety. Of special concern are those people that

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would rarely need to enter the suite, including maintenance personnel.

Some safety issues of extreme importance that are much less commonly experienced by veterinary radiologists deal with patients having pacemakers or aneurysmal clips being exposed to the magnetic field. This is less common in veterinary medicine; however, pacemakers are a definite contraindication. Typical orthopedic appliances may cause an artifact, but do not create a hazard for the patient. Similarly, small objects that are ferrous, including BBs, "gold beads," and steel shot used for water fowl hunting, all create large imaging artifacts but actually create no problem for the patient. They will not become dislodged or move significantly within tissue, but will create a large artifact and may prevent imaging in an area of interest. Similarly,

the small wire used in identification crystals can also create an artifact. The veterinary profession will need to find a better site for implantation of these crystals, other than the neck of small dogs and cats, as the popularity of MR continues. Cervical spinal studies can be compromised by these identification chips in small and toy breeds. These chips must occasionally be removed to allow for proper evaluation of the study.

The amount of energy absorbed during an examination is of concern. However, for the time utilized in veterinary imaging due to the anesthetic concerns, the amount of energy for MR studies has not been a problem to date. If medically indicated and anesthesia concerns can be answered, there is no reason why pregnant patients and neonates cannot be imaged with conventional MR units.