The Physics of Clinical MR Taught Through Images

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Fourth Edition

Thieme
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To my mother and father, kind good physicians that they were, and the reason that my two siblings and I all chose academic medicine as our career.
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The objective of this book, as with the prior editions, is to teach through images a practical approach to magnetic resonance (MR) physics and image quality. Unlike other texts covering this topic, the focus is on clinical images rather than equations. A practical approach to MR physics is developed through images, emphasizing knowledge of fundamentals important to achieve high image quality. Pulse diagrams are also included, which many at first find difficult to understand. Readers are encouraged to glance at these as they go through the text. With time and repetition, as a reader progresses through the book, the value of these and the knowledge thus available will become evident (and the diagrams themselves easier to understand).

The text is organized into concise chapters, each discussing an important point relevant to clinical MR and illustrated largely with images from routine patient exams. The topics covered encompass the breadth of the field, from imaging basics and pulse sequences to advanced topics including contrast-enhanced MR angiography, spectroscopy, perfusion and advanced parallel imaging/data sparsity techniques. Discussion of the latest hardware and software innovations, for example MR-PET, 7 T, interventional MR, 4D flow, CAIPIRINHA, radial acquisition, simultaneous multislice and compressed sensing, is included because these topics are critical to current clinical practice as well as to future advances. Included in the fourth edition are a large number of new topics, keeping the text up to date in this increasingly complex field. The text has also been thoroughly revised to include additional relevant clinical images, to improve the clarity of descriptions, and to increase the depth of content.

The clinical applications and complexity of MR continue to increase. Progress in MR has largely dominated the field of diagnostic radiology for the past three decades. Today MR stands as a major diagnostic subspecialty. The sophistication of this technique and continued advances dictate that MR will continue to play a dominant role in clinical medicine for the foreseeable future.

Val M. Runge, MD
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Portions of the textbooks entitled Imaging of Cerebrovascular Disease (Thieme 2017) and Neuroradiology: The Essentials with MR and CT (Thieme 2015) were incorporated with permission in the current text. Portions of the article entitled Speed in Clinical Magnetic Resonance (Runge VM et al), published in Investigative Radiology (Invest Radiol 2017;52(1):1-17), were also used with permission.
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Section I

Hardware
Components of an MR Scanner

The Magnet

Hydrogen atoms (of particular relevance, those in water molecules) have a nuclear spin, and associated with the nuclear spin is a magnetic moment. An externally applied magnetic field causes these magnetic moments to preferentially align parallel to the field. The unit of magnetic field strength $B_0$ is the tesla (T). A 3 T system (Fig. 1.1) provides a magnetic field 60,000 times stronger than Earth’s, yet with no permanent effects on human physiology and negligible temporary alterations.

The magnetic field is generated by feeding 400 A (Fig. 1.2) into the windings of a superconducting magnet. Superconductivity means that once the current is applied, the power supply can be disconnected, the end and beginning of the coil windings connected, and the current will continue to flow without resistance. Thus, such a magnet will be at field at all times, even during a power outage.

The Transmitting Radiofrequency Coil

Tilting the magnetic moments away from the parallel orientation causes them to precess (spin with a motion in which their axis sweeps out a cone, like a toy spinning top), with a Larmor (resonance) frequency of 42 MHz/T. These rotating magnetic moments induce an electromagnetic signal in adjacent coils: the so-called magnetic resonance (MR) signal. To tilt the magnetic moments, a rotating $B_1$ field is required that rotates with the same frequency as the magnetic moments. Only those “in resonance” will be affected, hence the term magnetic resonance. An antenna transmitting a radiofrequency (RF) field provides the rotating $B_1$ field (Fig. 1.2). A 90-degree RF excitation pulse with a duration of, for example, 2.5 msec provides a $B_1$ field of ~2.3 $\mu$T. The radiofrequencies used are below that of microwaves, yet there will be some warming of the patient. The energy transferred is referred to as specific absorption rate (SAR), which, in normal mode, is up to 1.5 W/kg.
The Gradients

Currents driven through a gradient coil provide a smooth change in magnetic field strength along one direction, causing the magnetic moments to have different precessional frequencies depending on location. An RF pulse with a predefined frequency range can then be applied, with the result being that only those magnetic moments in resonance with these frequencies will be excited. This is the basic principle of slice-selective excitation. Once the magnetic moments rotate, magnetic field gradients can again be established to “encode” spatial information into the signal via the precessional frequencies. Magnetic field gradients are established by sending 400 A (a reasonable value for clinical imaging systems) through the resistive windings of a gradient coil (Fig. 1.2) for a very short interval (e.g., 7 msec), causing a field variation of 40 mT·m⁻¹. The slew rate, reported in T·m⁻¹·sec⁻¹, is a measure of how fast the gradient can be established. All applications benefit from a fast and strong gradient system. Mechanical forces based on electromagnetic interactions between the windings of the gradient coil cause minor distortions in their shape, resulting in the knocking noise heard during an MR exam. This technology has advanced to the point where the patient is the limiting factor. Fast-changing magnetic field gradients induce currents in the patient’s body (which is a poor conductor) that can mimic nerve signals causing unintended muscle contractions. A stimulation monitor assesses the setup and prevents these limits from being reached.

The Receiver Coils

A primary source of image noise in MR is the patient. The smaller the receiving coil (a coil for brain imaging is depicted schematically in Fig. 1.2), the less noise is picked up. Also, the closer the coil is to the source of the MR signal, the greater the induced signal. This has resulted in a progression toward multi-channel coils with small elements that are in close proximity to the patient.

Fig. 1.2 Components of an MR system.
Magnetic resonance imaging (MRI) has become a mainstay of diagnostic imaging. When properly used, it is very safe and effective. Safety concerns do exist, however, with each type of magnetic field associated with an MR system: the static field \( (B_0) \), the radiofrequency field \( (B_1) \), and the gradient magnetic fields used for spatial encoding. In this chapter, we focus on issues relating to the static magnetic field \( (B_0) \).

There are two types of forces exerted on a ferrous object when brought in proximity to an MR magnet: rotational and translational. The rotational force (torque) is that which causes a ferrous object to turn and align with the direction of the main magnetic field \( (B_0) \). Rotational forces are strongest at the isocenter of the magnet. Translational forces are those that cause a ferrous object to be pulled toward the magnet isocenter. Translational forces are actually near zero at the isocenter because translational forces are felt when a ferrous object is in a magnetic field that changes in strength over distance. MRI requires a homogeneous magnetic field over the entire field of view and, thus, there is little translational force within the bore of a magnet. However, as one approaches an MR system, entering from the door of the scanner, the field strength begins to increase. The closer one gets to the magnet, the more rapidly the magnetic field strength increases. As magnetic field strength increases, so does the attractive force on ferrous objects. Most horizontal field (cylindrical) MR systems today are magnetically shielded to bring this fringe field closer to the magnet for siting purposes. As a result, the magnetic field changes very rapidly as one approaches the magnet. Bringing a ferrous object into the room is extremely dangerous and should never be done. Many times, once one feels the pull of the magnetic field it is already too late. Fig. 2.1 shows the unfortunate result of a standard-design wheelchair being brought into an MR scan room. There have been cases of injury and death resulting from ferrous oxygen tanks being brought into the exam room. Floor buffers (used by housekeeping) represent an important additional example of an object that should never be brought into an MR room.

Vertical field (some older low-field or so-called open MR) systems are not safer with regard to translational forces. Even though the magnetic field at the isocenter may be lower than the field strength of horizontal field (cylindrical) magnets, the change in the fringe field is actually very great near the

![Fig. 2.1 Wheelchair accident.](image-url)
magnet poles, going from near zero to the maximum in just a meter or two. The same precautions that one follows for high-field (1.5 or 3 T) cylindrical systems should be followed regardless of the MR system field strength or orientation. It should also be noted that, for the vast majority of magnets, the magnetic field is always "on" (this is true of both superconducting and permanent magnets) and cannot be turned off other than by quenching the system. For all types of MR systems, access by non-MR personnel should be restricted, and warning signs, stating that the magnet is always on, are advocated.

The presence of implants and magnetically and/or electrically activated devices can pose serious hazards for anyone with such an implant or device entering the scan room. For example, some intracranial aneurysm clips (used in the distant past) represent a contraindication to MR due to the force that can be exerted, leading to displacement and potentially death. An MR scan is contraindicated in a patient with an aneurysm clip, such as that illustrated (arrow) on CT in Fig. 2.2, for which the exact type cannot be documented. Screening for an MR exam by necessity must be in-depth and thorough. Top contraindications (among a long list) include metal within the globe (eye), a ferromagnetic aneurysm clip, or a cardiac pacemaker (see Chapter 3). Screening must cover any metal or implants within the body. Anyone entering the scan room (or going beyond the 5-gauss line) must be screened by trained MR personnel. This includes not only the patient but also any family members or support personnel. Most orthopedic implants, fortunately, are made from nonferromagnetic materials and are thus safe for MR.

If the patient has some type of implant or device, then it should be positively identified so that the physician can determine if it is MR safe or if there is a significant risk such that the patient should be excluded from entering the MR scan room. It is the ultimate responsibility of the MR radiologist/physician to determine if a patient can safely undergo an MR procedure. With the advent of higher field MR systems (3 and 7 T), it is important to remember that implants and/or devices that have been found safe at 1.5 T are not necessarily safe for imaging at higher fields. However, most such implants and devices are indeed safe at 3 T. It should be noted, regardless, that rotational forces on a ferromagnetic object inside a 3 T magnet increase by a factor of 4 compared with those of a 1.5 T magnet and that translational forces approximately triple. Up-to-date information about implant safety and testing can be found at the website.
Gradient magnetic fields are used primarily for spatial encoding (localization) of the MR signal. Additional coils of wire located within the magnet bore (but underneath the plastic housing, and thus not visible to the patient or technologist) produce these fields. During scan acquisition, the current in the gradient coils is switched on and off rapidly, causing in turn quick changes in amplitude and polarity of the gradient magnetic fields. There are four primary performance-related characteristics that define a gradient system. These are slew rate (how fast one can drive a gradient to a specific amplitude, dB/dt), maximum amplitude (how high a gradient field one can actually achieve, regardless of how long it takes), spatial linearity (how far away from isocenter the gradient field reaches before its strength begins to fall off), and duty cycle (how often one can drive the gradient without failure). All four are very relevant to clinical performance. High slew rates permit shorter echo times (TEs) and echo spacing, improving the image quality of fast spin echo scans and contrast-enhanced MR angiography. High-amplitude gradients markedly improve diffusion-weighted scans.

Above a certain threshold in switching of the magnetic field gradients, peripheral nerve stimulation (with muscle twitching) and pain may occur. On modern MR systems, dB/dt values are calculated prior to scan execution and thus monitored to prevent initiation of a scan if safety standards are exceeded. One very noticeable result from the rapid switching of the gradient coils is acoustic noise. Pulsing the gradients creates a force (due to the change in the magnetic field created around the wires) that generates pressure waves and thus the audible sound (knocking). Generally speaking, higher magnetic fields, higher gradient amplitudes, and faster gradient ramping all increase the acoustic noise. Many strategies exist, and continue to be developed, to reduce the sound to acceptable limits, with the end result that some 3 T scanners today are quieter than older generation 1.5 T systems. For high-field MR systems, hearing protection is offered and encouraged for all patients and anyone in the scan room.

The radiofrequency (RF) field used in MRI is also referred to as the $B_1$ field. Its purpose is to excite the spins, creating an MR signal that can be detected by a receiver coil. The frequency at which the MR phenomenon can be induced is termed the Larmor frequency, which for protons (hydrogen nuclei) is 42.6 MHz/T, and thus 63 MHz at 1.5 T. The amount of RF power necessary for imaging is dependent on several factors. These include the size and type of RF coil used for transmission, the distance of the coil from the patient, the field strength ($B_0$), and the number and type of RF pulses in the imaging sequence. For example, a 180-degree RF pulse requires four times the RF power of a 90-degree pulse, if the waveforms used are identical.

The rate at which energy (RF) is deposited into the body is defined as the specific absorption rate (SAR), measured in watts per kilogram body weight. U.S. Food and Drug Administration (FDA) guidelines limit RF power deposition to 4 W/kg averaged over 15 minutes for the whole body, 3 W/kg over 10 minutes for the head, 8 W/kg over 5 minutes per gram of tissue for the head or torso, and 12 W/ kg over 5 minutes per gram of tissue for the extremities. Body core temperature is not to increase beyond 1°C. Standards developed by the International Electrotechnical Commission (IEC) differ slightly.
This organization comprises members from all over the world (including the United States) and is equivalent to the FDA in the rest of the world (for electrical, electronic, and related technologies). Safety features on current clinical MR systems prevent these limits, for both dB/dt and SAR, from being exceeded.

A major safety issue due to applied RF, mandating careful patient screening, is that metal, outside or inside the patient, may experience rapid and extreme heating under certain circumstances. Reported incidents include (up to) third-degree burns, coma, and death. Extreme care should be taken with any electrically conductive material that must remain within the bore of the MR scanner during imaging. Dangers exist due to contact of any conductive material with the patient, and leads or wires close to the wall of the magnet bore. Skin-to-skin contact points forming a closed loop can also allow current flow within the body with the risk of burns at contact points (e.g., when a patient forms a closed loop by touching his index fingers together). Detailed specific recommendations exist and should be followed closely. The reader is referred to *The Reference Manual for Magnetic Resonance Safety, Implants, and Devices* (Biomedical Research Publishing Group, 2018). First- and second-degree burns have also been reported with decorative tattoos.

Cardiac pacemakers (*Fig. 3.1*) and deep brain stimulators (*Fig. 3.2*) represent a relative contraindication to MRI. Although reports exist in the literature in which patients have been scanned safely with these and similar implanted electrical devices, this has been done under very closely controlled circumstances (and sometimes with complications, regardless). As such, it is important to monitor all patients during an MR exam: visually, verbally, and through the use of instruments such as an MRI compatible pulse oximeter. Some of the newer cardiac pacemakers, implanted defibrillators and neurostimulators have been developed to allow MR imaging. In each of these categories there exist today devices that have received a ‘conditional 5’ recommendation, permitting an MR exam (in most cases only at 1.5 T) to be performed - however only under specific well-defined guidelines. Care must be taken to assure model type and specific magnetic field safety issues before scanning any patient with any type of implantable device.
The closer the receiving antenna (or coil) is to the source of the MR signal, in general, the better the signal-to-noise ratio (SNR). As a natural consequence, an impressive variety of radiofrequency (RF) coils are available today: all designed to match the geometry and to allow optimal coverage of the anatomic region being examined. There are, however, important differences in internal design, the topic of this case, that are often not readily apparent on visual inspection. Also, although only rigid coils are illustrated, many modern coils (even the most advanced in design) are flexible, made so they can be wrapped closer to the body part of interest.

◆ **Linearly Polarized Coils**

The simplest coil, although rare nowadays, is a single conductive loop (Fig. 4.1). The sum of the magnetic moments (the bulk magnetization) as it rotates causes a magnetic field fluctuation inside the loop area. This, according to Maxwell’s law, induces a voltage. The latter is digitized and analyzed to provide the information necessary to create an image.

◆ **Circularly Polarized Coils**

A circularly polarized (CP) coil, illustrated with a basic 1.5 T head coil in Fig. 4.2, is in essence a combination of two linearly polarized coils arranged in such a way to detect the rotation of the bulk magnetization rather than just a linear field fluctuation. As the sum of the magnetic moments rotates, the signal is detected initially by the first coil and subsequently by the second, with the two coils arranged so that one detects the component of the net vector in the x-axis and the other in the y-axis. Because each coil acquires an independent signal, SNR is improved by a factor of $\sqrt{2}$ compared with that of a linearly polarized coil.
Transmit and Receive Coils

Usually the body coil is employed for RF transmission, and anatomic-specific coils for RF reception. To provide a 90-degree excitation pulse, the transmit coil will need to create an oscillating $B_1$ field—for example, $2.3 \mu T$ for 2.5 msec. For some applications, as illustrated with a 1.5 T transmit (Tx) and receive (Rx) knee coil (Fig. 4.3), it is useful to excite a smaller volume of interest. This requires less RF power compared with that for whole body excitation. Also, little signal will be generated from adjacent anatomic structures (e.g., the other knee) and no further actions will be necessary to avoid wrap-around artifacts from those regions. Transmit coils are also usually circularly polarized in design because such coils require less power to provide a rotating $B_1$ field compared with that required for a linearly polarized coil.

Phased Array (Matrix) Coils

A small coil has increased signal pickup near to the coil and can be placed very close to the area of interest, with the advantage of improved SNR but the disadvantage of limited anatomic coverage. The solution is simply to use many small coils together. The term phased array refers to this design (illustrated with a 64-channel 3 T head/neck coil in Fig. 4.4), in which there are multiple coils with multiple independent receiver channels. Phased array coils can also be combined—for example, for the examination of different body parts that require different coverage in length as well as depth. Taken to the extreme, this permits high-resolution whole-body MR imaging. In addition to the larger coverage and improved SNR due to sampling the same signal source independently with each coil, the geometric arrangement of the coils contains spatial information that can be used to reduce the number of encoding steps. Because the coils acquire the signal in parallel, this approach is termed parallel imaging.
The demand in MRI for increased anatomic coverage, higher spatial resolution, and decreased acquisition time has required the development and implementation of hardware that improves the signal-to-noise ratio (SNR) and maximizes the efficiency of data handling. The SNR of an MR image is influenced by a variety of factors including the selected pulse sequence and magnet strength. However, the size of a signal-receiving coil element, its proximity to the tissue being examined, and the number of RF receive channels also greatly affect the quality of an image and the time necessary to acquire it.

Early MR systems collected signal through linear polarized, single-element coils and transferred data to the computer that performed the Fourier transformations through one low-bandwidth RF receive channel. Obtaining adequate SNR required that data be collected with lower imaging matrices and multiple signal averages leading to extended acquisition times. Additionally, because increasing the size of a single coil element decreases SNR (and is thus undesirable), the volume of anatomic coverage was limited.

The introduction of circularly polarized coil architecture led to a 40% increase in SNR through the use of two independent coil elements and allowed for greater anatomic coverage. However, emerging applications (e.g., functional imaging) once again demanded further improvements in spatial resolution, SNR, and data transfer rates. Thus, there remained the need for greater efficiency in coil technology and RF channel hardware, leading to the development of the multichannel architecture existent today.

One application of multichannel technology is illustrated in Fig. 5.1. In this implementation, a coil containing eight elements is configured as a phased array with overlapping, circumferential coverage of the entire imaging volume, transferring signal through eight designated RF receive channels. Each of the individual coil elements acquires MR signals from the entire brain with the highest signal obtained from that portion of the body in closest proximity to the element. The small size of each element leads to a higher signal received from the adjacent tissue and a higher overall signal upon reconstruction. Illustrated in Fig. 5.1 is the image reconstructed from a single

![Fig. 5.1 Architecture for an 8-element coil.](image)
coil (element no. 8), together with the final displayed) image, using specifically an eight-element, eight-channel design. Multielement coils are of many different types today, with a 1.5 T 12-channel head coil illustrated schematically in Fig. 5.2. The coil is composed of two adjacent rings of six elements each. Combined head and neck coils composed of up to 64 elements are available commercially today.

When coils with large numbers of elements are employed, the signals from several coils may be combined prior to image reconstruction. Fig. 5.3 illustrates the four images initially reconstructed using a 12-channel coil when employed in a mode where the signal from sets of three adjacent coils is combined. The final single image is depicted in the middle, with the other images usually transparent to both the technologist and the radiologist. Illustrated is an axial T2-weighted fast spin echo scan from a patient with long-standing multiple sclerosis (MS). Multiple periventricular white matter plaques, with abnormal high signal intensity, are depicted, characteristic for MS. Transferring the signal data through a reduced number of RF receive channels (as opposed to 12 channels in this instance) requires that the information be combined through a method called multiplexing. However, this combining method causes an undesirable loss in signal and speed of transfer. For this reason, the signal from each element should, if possible, be transferred through a designated high-bandwidth RF channel allowing for faster transfer of large volumes of information, thereby maintaining speed and also, importantly, signal.

In such a scenario, during image reconstruction, the signal from each element is corrected to minimize element-to-element signal variations and then combined to form the final single image. Advanced reconstruction and storage hardware are also necessary to process the rapid inflow of information, reducing the limitations on data volume. Fig. 5.4a illustrates an image reconstructed from one channel (of a 12-channel coil), and Fig. 5.4b illustrates the final combined image (that viewed by the radiologist), of the same patient as in Fig. 5.3, depicting multiple chronic MS plaques on T1-weighted images with low signal intensity, and the single ring enhancing active plaque (arrow, Fig. 5.4b) noted at this anatomic level.

Modern MR systems make possible acquisition of images with high SNR and spatial resolution within an acceptable scan time. The higher SNR achieved with multichannel technology allows greater flexibility in sequence parameter selection including increased spatial resolution or reduction in acquisition time—the latter used in part to minimize motion-induced artifacts. Fig. 5.5 presents axial T1-weighted images of the brain at the level of the Sylvian fissure acquired with a standard circular polarized coil (Fig. 5.5a) and an eight-element, phased array coil (Fig. 5.5b). The MR system was equipped with eight high-bandwidth RF receive channels. All pulse sequence parameters were held constant to demonstrate the improvement in SNR. This demonstration shows clearly the benefit of multichannel technology. There is a marked increase in SNR achieved with the eight-element coil (visualized as a decrease in “graininess” of the
image), leading to improved gray–white matter differentiation and, on this basis, the potential for improved lesion detection. Superior anatomic definition can be appreciated as well, as illustrated by the improved visualization of cortical gyri (Fig. 5.5b).

Multichannel technology offers many benefits. Higher SNR provides the user with the ability to increase spatial resolution while maintaining acceptable levels of SNR, or combinations of increased spatial resolution and reduced scan time can be achieved. Additionally, applications collecting large amounts of information in a short time frame, such as functional imaging, benefit from the higher SNR as well as the faster data transfer and storage (required to make routine imaging with multichannel systems feasible). Parallel imaging, discussed in depth later, requires the use of multielement coils, with the signal from different coil elements substituting for phase encoding steps. Advances in multielement/multichannel technology (to 128 elements and beyond) will continue to play a role in the development of MR imaging techniques with higher spatial resolution, faster scan times, and increased diagnostic quality.

Fig. 5.3 Multiple sclerosis on T2-weighted imaging, multielement image acquisition.
Fig. 5.4 Multiple sclerosis on T1-weighted imaging, single element and final image.

Fig. 5.5 Higher SNR achieved with a multielement coil.
A major motivation for the introduction of surface coils in the mid-1980s was the improvement in signal-to-noise ratio (SNR), resulting from a combination of increased coil sensitivity and, to a lesser extent, reduced noise. A limited volume, unfortunately, also implied limited anatomic coverage. To compensate for this, combined surface coils or phased array coils were introduced. Further extension of this concept led to the development of multichannel coil technology.

Illustrated in Fig. 6.1 is an axial scan of the liver acquired using a 12-element coil. The coil itself is composed of two rings of six elements, similar in concept to the 12-channel head coil illustrated in Chapter 5’s Fig. 5.2, but designed for abdominal imaging and flexible in nature. Each of the six surface coils (from one ring of elements) acquires signal from the anatomic region adjacent to the coil, with low sensitivity for noise outside the coil profile. The combined image takes advantage of the high signal provided by each individual coil, the latter due to the close proximity to adjacent tissue. Arrayed around the periphery in Fig. 6.1 are the images acquired for a single axial slice by each of the six coil elements, with the final composite image in the center. The scan sequence employed was true fast imaging with steady-state free precession (TrueFISP), applied with spectral fat saturation, with two large liver hemangiomas well depicted.

Fig. 6.1 Imaging of two hemangiomas in the liver with a 12-element coil.
should be noted that on current multichannel systems only the final composite image is routinely provided for viewing, with the steps in between transparent to the user.

Apparent from Fig. 6.1 is the spatial information that is contained in the coil sensitivity profile. The coarse structure of an object, which is commonly acquired with a low phase encoding step, can be derived from the coil sensitivity profile. The only prerequisite is that the sensitivity profiles of multiple coils have to extend in the direction of phase encoding. Spatial information provided by the coil sensitivity profile can then be used to replace that provided by phase encoding. This is the basis for parallel imaging, with one application being a reduction in number of Fourier lines that must be measured, while maintaining matrix size and thus spatial resolution. It should be kept in mind that each measured Fourier line is considered an additional acquisition, because each Fourier line contains information from the whole object. Omitting the measurement of Fourier lines will thus decrease the SNR. The drop in SNR is something that is typical for parallel acquisition techniques. Use of parallel imaging, specifically for replacement of Fourier lines, is thus only feasible in applications that provide sufficient SNR. Its use today is ubiquitous, principally to reduce measurement time. One example - in cardiac MR - is that images of the heart are typically acquired in a breath-hold, with the implementation of parallel imaging critical to reduce the number of heartbeats during which imaging must be performed.

Breath-hold axial images through the liver and gallbladder, acquired with fast spin echo technique and spectral fat saturation, are illustrated in Fig. 6.2. With the introduction of fast spin echo imaging, breath-hold T2-weighted abdominal scans became feasible. Parallel imaging can be used to further reduce scan time, as in cardiac imaging. Alternatively, with fast spin echo technique, parallel imaging can be used to reduce the echo train length while keeping scan time the same. One advantage of using a shorter echo train is that more slices can be acquired within the same scan time. Whereas 29 echoes were used to acquire Fig. 6.2a, only 19 echoes were used for Fig. 6.2b (with scan time constant), which allowed a substantial increase in number of slices. The missing Fourier lines for Fig. 6.2b were reconstructed using parallel imaging, with the penalty being a loss in SNR.

Fig. 6.2  Multielement coil use, enabling implementation of parallel imaging.
In the early development of clinical MR, certain low-field systems were advocated as more patient friendly, due to the design of the magnet being that of two horizontal plates, as opposed to a cylinder. Such systems were often marketed as "open.” Low-field MR systems typically suffer from poorer gradient performance and inferior receiver coil design (fewer elements and channels) when compared with current high-field systems (≥ 1.5 T). This, together with lower SNR, limited clinical throughput and quality. These systems are still marketed today, having been further refined in design, and represent a niche area, due to lower cost (Fig. 7.1). The first true high field (1.5 T) open system was introduced in 2004, an ultrashort cylindrical magnet complemented by an increase in bore diameter (Fig. 7.2).

This was the first increase in bore diameter for cylindrical magnets since their introduction in 1982 and represented a substantial technological innovation. It is interesting to compare the internal dimensions of this system, the Magnetom Espree (Siemens Healthineers), with a 125 cm bore length and a 70 cm bore diameter, to that of a low field (0.35 T) design (during the same time period), which featured a 137-cm magnet pole diameter (length of the magnet plate placed above and below the patient) and 38.5 cm table top to upper pole cover distance. Although the measurements are not completely comparable, this innovative design (the Espree) thus represented a substantially more open magnet.

Fig. 7.1 A current generation 0.35 T open system.

Fig. 7.2 The first-generation open high field 1.5 T system.
design both on the basis of length (shorter) and diameter (larger). Combining open design with the image quality of high-field MR, it should come as no surprise that this design rapidly became the best-selling MR system of its day, with a marked preference for this unit expressed by both patients and technologists.

The first open 3 T system was launched in late 2007, featuring a 70-cm bore diameter (the same as for the 2004 1.5 T design). This was subsequently followed by the current generation design, illustrated in Fig. 7.3. On such units, there is no compromise in image quality for clinical MR when compared to a standard design 3 T. A sample brain image, acquired in a multiple sclerosis patient using a current generation open 3 T MR, the Magnetom Skyra (Siemens Healthineers), is illustrated in Fig. 7.4. The markedly improved SNR of 3 T, in comparison to a low-field open scanner (and as well when compared to a 1.5 T system), permits scan time to be reduced and image quality to be increased—the latter due to a reduction in voxel size (higher spatial resolution). In many areas of the body, this improvement in spatial resolution is critical for diagnosis.

The increase in bore diameter for cylindrical magnet designs, combined with the decrease in bore length, make these units much more flexible in regard to many different clinical applications. For critically ill patients, monitoring and access is markedly improved. The larger diameter bore (open) cylindrical magnets used for these systems have also permitted new procedures such as MR-guided therapy, launching interventional MRI, and kinematic examinations, where space is an absolute prerequisite.
Magnetic Field Effects At 3 T and Beyond

There are many technical challenges associated with creating MRI systems with field strengths greater than 1.5 T. First, the magnetic field ($B_0$) in an MRI system can vary slightly across large imaging volumes. The measure of the extent of this variation is termed magnetic field homogeneity. Highly homogenous fields are important in achieving images with consistent signal and chemical fat suppression. At higher field strengths, maintaining highly homogenous fields within and around the imaging volume requires special considerations in magnet design. New approaches in the design of the magnet hardware of an MR system can affect a wide variety of applications including body, breast, orthopedic, and neurological applications by improving the consistency of signal intensities in the image, maintaining a consistent level of chemical fat suppression, and minimizing distortions at the edges of the images. Fig. 8.1b, when compared with Fig. 8.1a (acquired on a unit with an older magnet design), demonstrates the improvement in image quality near the edges of the imaging field (arrows) achieved with a new magnet design.

At a magnetic field strength equal to or greater than 3 T, image inhomogeneities have been documented that are in general attributed to RF field ($B_1$) inhomogeneities, which result in spatially dependent altered excitation and refocusing angles. To some degree the effect can be attributed to the RF wavelength at 3 T and above being less than the cross-sectional diameter of the body part being imaged. At 3 T, the effect is prominently seen in liver imaging in patients with ascites. Presented in Fig. 8.2 is an axial HASTE image through the liver and spleen, with two different window and center settings. There is a prominent loss of signal intensity centrally, markedly degrading

Fig. 8.1 Improved magnet homogeneity leading to better image quality.
diagnostic quality. With a different window/center, the liver and spleen can be visualized, but the low SNR therein is readily evident. The interaction of biological matter with the electromagnetic field depends on magnetic permeability, electric permittivity and electric conductivity, which are a function of the RF-frequency and thus field strength. The interacting E-field component of the RF-field will also be a function of location with respect to the position of the transmitting coil. The term dielectric effect refers to the interaction of matter with the E-field component of the RF field and has been used to explain some of the observed shading effects, thus the term ‘dielectric effect’ or ‘artifact’, although the appropriateness of that terminology has been questioned.

The ‘dielectric effect’ is accentuated further as the field strength is increased from 3 to 7 T, due to the shorter RF wavelength. The artifact can be quite prominent at 7 T even in brain imaging, if not somehow compensated. It has been suggested and successfully implemented that appropriate control of amplitude and phase of the feeding ports of a transmitter coil can reduce these B1-field inhomogeneities, even patient specific, thus leading to a more homogeneous B1-distribution, reducing in the final image signal variation and the chance of poor chemical fat suppression. The utility of this approach is demonstrated by the example provided in Fig. 8.3. Images of the lower leg in axial cross-section are depicted, with shading noted in the upper set of images (arrow), induced by ‘dielectric resonances’, which is eliminated by the use of a modified RF distribution scheme (lower set of images). Another sophisticated and therefore more expensive approach is the utilization of multiple transmitter channels (termed ‘parallel transmit’) with separately controlled amplitude and phase of the feeding ports for the purpose of generating a patient specific homogeneous B1-field distribution.

Fig. 8.2 The ‘dielectric effect’ in a patient with ascites at 3 T.

Fig. 8.3 Correction of ‘dielectric effects.’
The history of MRI is marked by a progressive increase in utilized magnetic field strength over time. Research systems at pioneering institutions have usually been the first to go to high magnetic field strength, to explore the potential advantage in current and future applications. The main motivation to go to higher magnetic field strength is the increase in signal intensity: allowing for improvement in image quality (increasing the competitiveness of an imaging center), or selecting a higher spatial resolution, thus reducing partial volume effects and increasing diagnostic confidence, or, taking the signal gain to reduce measurement time and increase patient throughput. Last but not least, imaging at high magnetic fields has only been possible due to the progress in engineering that has provided affordable and siteable magnets.

The MR signal is induced into an antenna system by a rotating nuclear magnetization. That magnetization in turn is roughly proportional to the difference in the population of parallel aligned spins versus antiparallel aligned spins. That difference is approximately proportional to the energy difference between the two possible states which, in turn, depends on the field strength used. In addition, the signal induced depends on the rotational frequency of the nuclear magnetization. Thus, signal goes with the power of two of the field strength used. As for the electromagnetic noise level during a MRI, the patient is considered the main source of that noise. Unfortunately, the noise emitted by the patient is also a function of field strength. As a result, the gain in SNR is “only” approximately linear with the magnetic field strength. The gain in SNR may be used to shorten measurement time or to improve image quality.

The influence of magnetic field strength on T1 relaxation times is a curse and a blessing. T1-relaxation depends on $B_1$-fluctuations close to the Larmor frequency. Those fluctuations are primarily dependent on the tumbling of the water molecule and the intramolecular dipole–dipole interaction of the magnetic moments of the spins of the adjacent hydrogen nuclei. The Larmor frequency is proportional to the primary magnetic field strength. Thus, moving to a higher magnetic field results in a prolongation of the otherwise tissue-specific T1-relaxation time. Depending on the tissue imaged an increase in T1-relaxation time may be beneficial or detrimental; for example, the contrast (CNR) in MR angiography is improved and the detectability of T1-shortening contrast material such as gadolinium chelates is increased, but contrast (CNR) between gray and white matter is decreased. Fig. 9.1 shows MIP time-of-flight MRA images (see Chapter 77) of the circle of Willis, obtained at 1.5 T, 3 T, and 7 T. The longer T1 tissue times at higher magnetic fields lead to improved background suppression, which in combination with the intrinsically higher SNR lead to improved spatial resolution with better detail and small vessel depiction.

The influence of magnetic field strength on T2$^*$ relaxation times is a curse and a blessing. The signal decay characterized by T2$^*$ in gradient-echo sequences depends on the dephasing of the transverse nuclear magnetization as a consequence of local magnetic field inhomogeneities. These are in general based on differences in magnetic susceptibility between tissues (e.g., deoxyhemoglobin [paramagnetic] in comparison to the surrounding brain tissue [diamagnetic]). The local magnetic field will be a product
of the magnetic susceptibility in question multiplied with the magnetic field strength used. Local field inhomogeneities and the correlated signal loss due to the T2* effect will scale with the magnetic field strength utilized, thus improving the T2* sensitivity with higher fields. In susceptibility weighted imaging the frequency shift based on local field inhomogeneities will reflect itself in the phase image used as a mask in SWI in order to enhance the visibility of small susceptibility changes. **Fig. 9.2** presents susceptibility weighted images (see Chapter 64) acquired at 1.5 T, 3 T, and 7 T. With increasing magnetic field strength, there is improved depiction and detection of small veins, together with greater sensitivity to iron, the latter effect best seen within the basal ganglia.

The influence of magnetic field strength on chemical shift is a curse and a blessing. The Larmor precession of a nuclear magnetization in a specific biochemical compound scales with the magnetic field strength. This will lead to a better delineation of different frequency lines in MR spectroscopy. The Larmor frequency of hydrogen atoms contained in lipid compounds is ~3.5 ppm below the Larmor frequency of the spins of the hydrogen nuclei within a water molecule. The frequency in MRI is used for spatial encoding, fat and water image will be shifted depending on the field strength and bandwidth of the imaging protocol. To reduce the field strength–dependent increase in chemical shift, a higher imaging bandwidth is selected in general, slightly diminishing the SNR advantage.

The fact that the SAR value scales with the power of two with the resonance frequency represents a stimulating challenge to scientists and engineers to work on appropriate solutions, as the patient is not allowed to receive more than 4 W/kg (which is roughly the metabolic rate of a marathon runner). The SAR also scales with the power...
of two with the $B_1$-amplitude of the excitation and refocusing pulse. Reducing the $B_1$-amplitude would be a helpful measure. Low SAR-RF pulses have been introduced using a longer duration and a lower $B_1$-amplitude in order to achieve the desired flip angle, but the prolonged duration will lead to a longer minimum echo time and a longer echo train length, thus diminishing image quality. An interesting concept has been introduced in 1988 to change the magnetic field gradient during excitation in order to reduce the effective $B_1$ field still maintaining a short RF duration. Integrating over a range of frequencies will lead to a SINC-function, the typical shape of an RF pulse, with a peak $B_1$ value in the center. With VERSE - variable-rate selective excitation, the RF pulse is selected to be short at the beginning and at the end of the SINC-shape, where it switches to longer duration in the middle of the excitation scheme allowing a drop in $B_1$-amplitude. With the introduction of high field systems, VERSE pulses are almost mandatory.

Another measure to reduce the SAR is the selection of an incomplete refocusing angle. Selecting a refocusing angle smaller than 180° will linearly change the required $B_1$ amplitude and reducing the correlated SAR. Unfortunately, and obvious, the incomplete refocusing will also compromise image quality. As the center $k$-space lines are more dominant in dictating image contrast and quality, 3D multi echo sequences have been designed with a variation of the refocusing angle throughout a 3D acquisition in order to achieve optimal image quality. Such approaches have been named SPACE, CUBE or VISTA. Another measure to reduce the SAR (not only in high field imaging, but at high magnetic fields the motivation is greater as compared to lower magnetic fields) is the utilization of parallel imaging. Using the spatial information of distributed coils allows the omission of encoding lines. As SAR scales with the number of RF pulses (per unit time), parallel imaging is helpful, if the measurement time is kept constant. It has also to be pointed out that parallel imaging per se is correlated with a reduction in SNR (as less is measured). It is mainly due to the utilization of higher magnetic fields that parallel imaging as taken off.

In 1986 David Hoult presented “arguments concerning an optimal field strength”, with the implication that 1.0 T would be optimal. Arguments against a higher field strength pointed to the increase in motion- and flow-related artifact. Again, this has been a stimulating challenge to scientists and engineers to come up with motion artifact reduction techniques.

The RF field penetration into the conductive human body is proportional to the tissue permeability and conductivity, and inversely proportional to the angular frequency

Fig. 9.2 SWI at 1.5, 3, and 7 T.
applied and main magnetic field strength. The decrease in RF penetration with tissue (skin) depth is seen as image shading, and becomes noticeable at high and ultra-high fields. The RF component compromised is the $B_1$ amplitude and the challenge is called $B_1$ shimming—to produce a homogenous image appearance. This can be addressed with specialized coil design, phase and amplitude modifications at feeding points of the transmit coil, or multiple transmit coils also referred to as parallel transmit (pTX).

At ultra-high fields, the wavelength of the RF field becomes comparable with the size of the patient’s body or organs. Reflected electromagnetic waves at boundaries may cause destructive and constructive interference patterns resulting in inhomogeneous $B_1$ distribution. Some authors refer to this phenomenon as “dielectric resonance” whereas others point to the fact that fields within a patient are attenuated as they propagate, so that reflected waves are of little importance and true dielectric resonances are severely damped. Nevertheless, interference patterns based on wavelength effects, standing waves, or standing wave envelopes have to be admitted and addressed at ultra-high fields in the same fashion as cited previously when discussing skin depth.

The cost for a MR system also increases approximately linearly with the strength of the magnetic field used. Adding to the cost and logistics is ~360 tonnes of iron shielding, which is required to confine the fringe field of an ultra-high-field system. This 360 tonnes is in addition to the system weight itself of ~38.6 tonnes. An actively shielded magnet will be even more expensive. Whether any potential new applications or improvements in diagnostic confidence will result in clinical benefit to the patient and merit the increased system cost will remain a controversial discussion for quite some time.

At high magnetic fields, small electric fields can be generated in the human body if there is motion in the magnetic field. This typically occurs in the patient during positioning and can also affect medical staff walking around the magnet or taking a closer look inside the bore. The generation of these small electric fields depends on the speed of the moving person within the magnetic field and may result in various sensory perceptions such as vertigo and nausea. The speed of the patient table at ultra-high-field systems has to be adjusted accordingly and staff has to be educated about these temporary sensory effects. Large static magnetic fields (3 T and higher) can also cause mild vertigo even without movement of the head but solely based on Lorentz forces applicable to ionic currents occurring naturally in the endolymph of the vestibular system.

Last but not least, it must be pointed out again that magnetic pulling and torque forces are a product of the magnetic susceptibility of any implanted object multiplied by the volume of the object, the magnetic field strength at the current location, and the fringe field. For non-ferromagnetic but conductive implants, interactions with the electrical component of the electromagnetic wave can cause voltage buildup and induce currents within the object. These induced currents can result in heating of the conductive implant and surrounding tissue.

Field strength–specific applications include all approaches that require a high temporal and/or spatial resolution that would be prohibited at lower field strength due to an intrinsic limitation in SNR. This includes high spatial resolution musculoskeletal imaging as well as increased detail in diffusion tensor imaging. There are already multiple publications pointing to increased lesion detection in patients with multiple sclerosis. Changes in iron homeostasis as demonstrated in patients with Parkinson’s disease are also more easily documented on ultra-high-field systems due to the increased $T2^*$ sensitivity.
Advanced Receiver Coil Design

New innovative receiver coil designs provide markedly improved SNR together with the ability to perform both parallel imaging and simultaneous multislice imaging (SMS) in any direction. In addition, such coils can be effectively used with higher parallel imaging factors. These advances make possible a marked reduction in scan time and/or improvement in image quality, as well as enabling scan techniques that would otherwise not be feasible due to acquisition time.

**Fig. 10.1** illustrates sagittal diffusion-weighted imaging (DWI) obtained with a 32-channel head coil at 3 T, using (a) no parallel imaging as compared to a factor of (b) 4. Note the marked reduction in bulk susceptibility artifact (black arrow) and anatomic distortion (pons and cervical cord, white arrow) with the use of parallel imaging. Application of the coil in this fashion markedly improves image quality with DWI at 3 T, with magnetic susceptibility effects (the origin of the artifacts noted) double at 3 T as compared with 1.5 T.

In **Fig. 10.2**, axial DWI, obtained without (a) and with (b) SMS (necessitating use of an advanced coil with multiple elements in the slice select direction, achieved in this instance using a 64-channel head/neck coil) depicts a small subacute cerebellar infarct at 3 T with high image quality due to the use of readout segmented technique and a slice thickness of 2.5 mm. SMS enabled a decrease in scan time from 5:46 to 3:26 min:sec with full brain slice coverage.

**Fig. 10.3** illustrates, in a patient with a large enhancing brain metastasis, use of a 32-channel coil alone to achieve a decrease in scan time. Post-contrast T1-weighted (a) 12- and (b) 32-channel coil scans, with equivalent image quality, are illus-
trated. Two averages were required for (a) and only one for (b), with scan time decreasing from 1:52 to 0:56 min:sec.

**Fig. 10.4** illustrates sagittal and coronal reformatted images from a 32-channel coil scan in a patient with a posterior fossa arteriovenous malformation. A 3D T2-weighted SPACE acquisition (with iPAT of 2 in both phase encoding directions) was employed, providing $0.9 \times 0.9 \times 0.9$ mm resolution in a scan time of 2:12 min:sec (an example of making a very high spatial resolution scan possible in a reasonable scan time).

As noted in prior sections, multiple elements within a given coil improve the signal close to each element and thus increase SNR. Due to spatial variations in the location of the signal, this information can also be spatially encoded. This realization gave birth in the early 2000s to parallel acquisition techniques—PAT, also known as SENSE, mSENSE, GRAPPA, or ARC—primarily aiming for and leading to a significant reduction in measurement time. Also, an important point (in terms of ease of operation of the MR system and improved patient throughput) is that system software automatically selects and deselects coil elements as required for the specific clinical application, enabling changes during the exam: for example of the anatomy to be studied or patient position in the magnet, without the need to reposition the patient and/or mechanically change coils.

**Fig. 10.3** Brain metastasis: more coil elements (12 vs. 32) allow faster imaging.

**Fig. 10.4** High resolution 3D T2 SPACE (cerebellar AVM).
The idea for a multidimensional, spatially selective RF excitation was first published in 1989, but it was largely abandoned due to impracticable long RF pulse durations and/or ‘excitation ghosting’ similar to pulsatility ghosting in imaging. This challenge can be addressed using multiple separate RF-channels allowing for the excitation, refocusing, or saturation of arbitrary shapes. As a useful side benefit, multiple RF channels are also a valuable aid to ease the effort in compensating for $B_1$ inhomogeneities commonly observed in high field imaging.

$B_1$ inhomogeneities cause signal loss at the center of the imaging volume that is commonly seen in abdominal MRI at 3 T. These signal losses are especially apparent in patients with a diminished amount of body fat. This is a common problem when imaging athletes, where fat is replaced by water-containing muscle, and in patients with ascites. Current tissue RF excitation techniques for MR imaging are in general slice-selective or volume-selective (see Chapter 12 and Chapter 51). Current RF excitation schemes may be referred to as utilizing one-dimensional (1D) RF pulses.

Transmit coils capable of multidimensional RF excitations were conceived to enable the application of bi-dimensional (2D) and tri-dimensional (3D) selective RF pulses to the imaged tissue. Similar to how raw data acquisition in $k$-space may be transformed using the Fourier transformation to provide an image, deposition of RF energy may be tailored into an "excitation $k$-space" that will provide any desired excitation pattern within the object being imaged. Unfortunately, in order for multidimensional RF transmission to achieve its full potential, magnetic field gradients have to be altered simultaneously at very high velocities. The performance of current gradient systems is sometimes curbed in order to avoid peripheral nerve stimulation, which is also demanded by the U.S. Food and Drug Administration (FDA), thus limiting gradient system capabilities and hampering multidimensional RF transmission. Moreover, 3D spatially selective RF pulses are of significant duration (20 to 30 msec as compared with current durations of 2.5 msec for 1D RF pulses) and are likely to be impractical for most current clinical applications.

The more recent idea to utilize the advantages of multidimensional RF transmission, published in 2003, was to apply the same strategy as with parallel imaging acquisition to the concept of “parallel excitation.” Similar to how parallel imaging is based on the reconstruction of undersampled data using the spatial information from multiple receiver coils, truncated (undersampled) RF waveforms utilizing spatially distributed transmit coils could allow for the excitation, refocusing, or saturation of arbitrary shaped volumes. This permits the application of shortened RF pulses, using a reduced excitation $k$-space, without sacrificing spatial definition.

Using higher magnetic fields for imaging has several important advantages, like better SNR and enhanced $T_2$* sensitivity. However, as the RF wavelength approaches the size of the patient’s head or body, complex RF interaction patterns result in so-called patient-related $B_1$ inhomogeneities. Although these inhomogeneities can be addressed using adiabatic RF pulses, these tend to involve large amounts of RF power and are thus limited to certain clinical applications. These $B_1$ inhomogeneities can be better addressed using a spatially distributed transmit array.
Due to tissue conductivity, there is a loss of RF penetration toward the center of an object being imaged in MRI. This attenuation of the RF amplitude leads to an effectively higher $B_1$ amplitude at the periphery of an object when compared with the center of the same object. The attenuation of RF amplitude increases with magnetic field strength. It has been suggested that with the use of spatially distributed transmit arrays, the loss of RF penetration can be overcome or at least minimized, especially in MR systems with higher magnetic field strengths (e.g., 7 T).

Fig. 11.1 shows an axial T1 image through the liver without $B_1$ shimming. Areas of signal loss are present within the left lobe of the liver and left kidney secondary to $B_1$ inhomogeneity. Fig. 11.2 shows the same image using $B_1$ shimming with homogeneous signal throughout the liver and renal parenchyma, illustrating what could potentially be achieved with multidimensional RF transmission.

Fig. 11.1 Signal loss due to $B_1$ inhomogeneity.

Fig. 11.2 $B_1$ shimming for improved homogeneity.
Section II

Basic Imaging Physics
The signal for MRI comes from the nuclei of hydrogen. Because the hydrogen nucleus consists of a single proton, it is common practice to refer to the signal as coming from protons. On exposing the patient to a magnetic field, more nuclear magnetic moments (of protons) will be aligned with (parallel to) the main magnetic field rather than against it. This results in a net longitudinal magnetization. Tilting of that magnetization can be done with an RF pulse. Once the magnetization is tilted away from the parallel alignment, it will start to precess around the direction of the main magnetic field with the Larmor frequency. That frequency is a function of the local magnetic field strength. To tilt the magnetization, the frequency of the RF pulse must match the rotational frequency of that magnetization. Establishing a magnetic field gradient along the direction of slice selection (e.g., z) and using an RF pulse with a limited frequency range will result in a slice-selective excitation (Fig. 12.1). In this and in the figures that follow, GS is the slice-selective gradient, GP is the phase encoding gradient, and GR is the readout gradient. Magnetizations of different resonance frequencies remain untouched. Following a 90-degree RF pulse, the longitudinal magnetization is converted into transverse magnetization. The latter is responsible for the observed MR signal and will be encoded with spatial information. Fig. 12.1 is an illustration of a slice-selective excitation utilizing the dependency of resonance frequencies on local magnetic field strength; (1) and (2) mark the lower and upper frequency range covered with the slice-selective RF pulse.

Fig. 12.1 The use of a slice-selective gradient to define slice thickness.
Essential to spatial encoding in MRI is the fact that the resonance frequency of the magnetization is a function of the local magnetic field strength. Establishing a magnetic field gradient across an object (e.g., $B_0 + GR \cdot x$) will result in different frequencies along that direction (e.g., $x$). Fig. 12.2 is an illustration of frequency encoding by means of a magnetic field gradient. The alternative term readout gradient is commonly employed, because the gradient is on during readout of the data. The rotating transverse magnetizations induce an MR signal in an adjacent coil. The frequency-encoding gradient spreads the Larmor frequency over a sufficiently wide range to distinguish the individual voxels specified in that direction. An adjacent RF receiver coil will see one transverse magnetization being the sum of all transverse magnetizations of each individual voxel. The range of different frequencies causes a rapid dephasing of the transverse magnetization, resulting in the induced signal rapidly decaying.

Analyzing the received signal for frequency contributions is called a Fourier transformation. The amplitude of the contribution is assigned to a pixel intensity at the location of the measured frequency (Fig. 12.3). The signal course is called an echo, and because gradient switching (in this instance) has been used it is called a gradient echo (GRE). The algorithm (fast Fourier transformation [FFT]) analyzes the collected data for frequency contributions. Because the spatial distribution of frequencies is known (dictated by the magnetic field gradient applied), the FFT can assign the amplitude of the different frequency contributions to specific locations, in this example the location of a one-dimensional (1D) object (Fig. 12.3).

The data acquired during one readout period is called a Fourier line. Because each data point along the line has an index referred to as a "$k$" value in mathematics, the line is also called a $k$-space line. For 2D-encoding the single Fourier line is expanded to a Fourier space or $k$-space. The data structure in the second dimension is similar to the data structure along a Fourier line.

The number of data points taken in either direction (GR and GP) has to be equal to or larger than the matrix resolution of the image to unambiguously assign the signal to

![Fig. 12.2 Frequency encoding for one-dimensional spatial localization.](image)
Amplitude of Frequency contribution = "Image" of the 1D object

Frequency-Analysis = (Fast) Fourier Transformation

Fig. 12.3 Use of a Fourier transformation to determine frequency contributions.

a location. For example, a $256 \times 512$ image resolution requires that at least 256 phase encoding steps are acquired and 512 data points are sampled during the readout period. If fewer Fourier lines are measured, as in half Fourier or parallel imaging techniques, additional algorithms have to supplement the missing information. The important message is that the center of $k$-space contains nothing other than the information of how much signal the whole excited slice is sending. Adjacent to the center of $k$-space in either direction is the information about the coarse structures of the objects within the slice. The information about the requested highest resolution is found in the outer borders of $k$-space. The spatial resolution is given by the selected field of view (FOV) divided by the matrix size in either direction. Fig. 12.4 illustrates how

Fig. 12.4 Dependence of spatial resolution on number of k-space lines.
spatial resolution gradually improves with increasing numbers of Fourier lines around the center of $k$-space.

To summarize a measurement (Fig. 12.5):

1. A slice-selective gradient is established in the direction of slice selection (GS; e.g., $z$).
2. Once that gradient is established, an RF excitation pulse will excite the slice—that is, it will tilt the longitudinal magnetization to become transverse magnetization.
3. A phase encoding gradient, GP, will establish partial spatial information in the direction of the gradient.
4. A readout gradient, GR, will be activated to prepare the starting point for the first data point. During data acquisition, GR is switched on, creating a gradient echo (GRE) with a signal maximum close to the center of $k$-space. A 2D Fourier transformation (2D FFT) then assigns pixel intensities within the image based on a 2D frequency analysis.

For the advanced reader, some of the stated prerequisites are not mandatory. For example, it is common in GRE techniques to shift the echo center to an earlier time in $k$-space. The signal will be stronger due to the shorter TE, with the artifacts caused by the $k$-space asymmetry outweighed by that benefit.

**Fig. 12.5** Summarizing the MR measurement, from excitation to image formation.
A pixel represents the smallest sampled 2D element in an image. It has dimensions given along two axes in mm, dictating in-plane spatial resolution. Pixel sizes range in clinical MRI from mm (e.g., squares with sides of 1 millimeter) to sub-mm. A voxel is the volume element, defined in three dimensions. Its dimensions are given by the pixel, together with the thickness of the slice (the measurement along the third axis). Slice thicknesses in clinical MRI vary from a maximum near 5 mm, achieved using 2D multislice imaging, to sub-mm, achieved with 3D scan techniques.

MRI spatial resolution, which determines the radiologist’s ability to distinguish structures as separate and distinct from each other (together with image contrast), is inherently related to the acquired voxel volume. In the simplest case, the field of view (FOV), acquisition matrix, and the slice thickness determine voxel volume. The pixel size (FOV/matrix) determines the in-plane resolution. Reducing the FOV, increasing the matrix number, or reducing the slice thickness results in an image with reduced voxel volume. SNR is directly proportional to voxel size (assuming that the number of phase encoding steps is held constant). Small voxels produce MR images with high spatial resolution but a lower signal-to-noise ratio (SNR), and thus may appear “grainy” compared with images acquired with a larger voxel volume.

The images shown in Fig. 13.1 demonstrate the effect of altering pixel size. The scans were acquired using T1-weighted technique at 3 T and illustrate a small enhancing lesion (arrow). By appearance alone, the lesion would be consistent with a metastasis, but in this patient it represents a small tumor focus in a multcentric glioblastoma. The pixel dimensions were (a) 0.9 × 0.9 mm versus (b) 0.5 × 0.5 mm. The slice thickness was held constant. Note the improved anatomic detail in (b)—for example, the “sharpness” with which small enhancing vessels are depicted—due to the improved in plane spatial

Fig. 13.1 Smaller pixels improve anatomic detail, but decrease SNR.
resolution (smaller pixel size). To achieve a high-quality diagnostic image, however, scan time was tripled (three averages as opposed to one). From basic MR physics, the reduction in pixel dimension resulted in a factor of three loss in SNR, which was only in part compensated for by the increase in scan averages (SNR being proportional to the square root of the number of scan averages). To summarize, reducing the pixel size increases spatial resolution but also markedly decreases SNR (assuming all other factors are held constant).

The images in Fig. 13.2 demonstrate the effect of altering slice thickness (and thus voxel size). Illustrated on 2D T2-weighted fast spin echo scans is a small, early subacute, left-sided, anterior cerebral artery territory infarct (black arrow). Slice thickness was (a) 5 mm vs. (b) 2.5 mm, with all other scan parameters held constant. The 5-mm section appears “smooth,” reflecting the high SNR, as compared with the 2.5 mm section. However, the border between cortical gray and white matter (white arrow) is better depicted on the thinner section, despite the lower SNR, due to less partial volume imaging.

A comment is warranted here concerning 3D scan techniques, which are commonly used to achieve thin sections (e.g. ≤ 1 mm in the brain). A 3D acquisition excites an entire slab or volume of tissue rather than a slice. The slices are produced by the application of an additional phase encoding gradient in the slice (z) direction. The number of slices (sometimes referred to as “partitions”) desired determines the number of phase encoding steps to be applied in the slice direction and thus directly affects scan time (as well as SNR). 3D acquisitions are useful for acquiring thin contiguous slices. In addition, reformatted images from a 3D data set (e.g., in the sagittal and coronal planes, from an axial acquisition) will be of high quality if the voxel dimensions are near isotropic (equal in all three dimensions).

SNR in MR can be a very complicated subject. But it is simply the signal divided by the noise in an image (see Chapter 14). Looking only at image resolution and matrix size, the signal is directly proportional to the acquisition voxel volume. The noise is proportional to the inverse of the square root of the number of 2D phase encoding steps times the number of phase encoding steps in the third dimension (if the scan is a 3D scan). Thus, more simply, SNR is directly proportional to voxel volume, if the number of phase encoding steps is held constant.

Fig. 13.2 Thinner sections similarly improve anatomic detail, and decrease SNR.
14 Imaging Basics: Signal-to-Noise Ratio

The images in this case illustrate the critical concept of signal-to-noise ratio (SNR) in MR. SNR, as the term implies, is the ratio of MR signal to noise, specifically for the spatially encoded voxel. MR images acquired with a low SNR appear somewhat “grainy” to the eye (Fig. 14.1a; Fig. 14.2a), especially when compared with images acquired with higher SNR (Fig. 14.1b; Fig. 14.2b). However, what is most relevant clinically is the contrast-to-noise ratio (CNR), not the SNR. The higher the contrast between two structures, the less SNR required for differentiation. To distinguish tissues that have similar contrast, high SNR is required. For example, the difference between gray and white matter (small white arrows, Fig. 14.2b) is much easier to see in Fig. 14.2b (with higher SNR) than in Fig. 14.2a.

Fig. 14.1 Higher SNR, due to signal averaging, illustrated on T1-weighted scans.

Fig. 14.2 Improved lesion depiction on T2-weighted scans due to higher SNR.
What determines the signal in a spatially encoded voxel? MR signal is directly proportional to the size of the voxel. The larger the voxel, the greater the number of hydrogen protons and thus the greater the MR signal. Larger voxels, however, result in reduced spatial resolution (see Chapter 13). The parameters that affect the size of the voxel in a 2D acquisition are field of view (FOV), the number of phase encoding steps (acquired phase encoding matrix), the number of frequency-encoding steps (acquired frequency or read matrix), and the slice thickness. As stated, SNR is directly proportional to the size of the voxel. So, for example, doubling the slice thickness will double the voxel volume and double SNR. However, the FOV affects the voxel volume in two dimensions. As such, reducing the FOV by a factor of 2 would reduce the voxel volume and SNR by a factor of 4.

Turning to a consideration of noise, the noise in a voxel is proportional to the square root of the sampling bandwidth (an operator-defined variable) for the voxel divided by the square root of the total number of times the voxel is sampled. The parameters that affect the number of times the voxel is sampled, and thus the noise in the voxel, are the number of phase encoding steps, the number of slices encoded (for a 3D acquisition only), and the number of signals averaged (NSA). NSA is the number of times each line of k-space is sampled and is also known as the number of excitations (NEX) or acquisitions, depending on vendor. NSA is the parameter most often increased in the clinical setting to increase SNR. The problem with increasing NSA to increase SNR (in this case by reducing the noise) is that the total scan time is directly related to the NSA whereas the SNR is related to the square root of the change in NSA. For example, if NSA (and thus scan time) is increased by a factor of 4 (the difference between (a) and (b) in Fig. 14.1 and Fig. 14.2), SNR will only increase by a factor of the square root of 4 (and thus 2).

As noted, SNR is also inversely proportional to the square root of the sampling (receiver) bandwidth of the encoded voxel. For example, if one were to reduce the receiver bandwidth by a factor of 2, SNR would increase by a factor of the square root of 2. Although receiver bandwidth does not affect scan time, reducing the receiver bandwidth does increase chemical shift artifact (see Chapter 112), which can be more problematic at higher field strengths.

Fig. 14.1 presents thin section sagittal T1-weighted images, with the only difference between (a) and (b) being NSA. One signal average was used for (a), and four for (b), with the scan times being 1:12 and 4:48 min:sec, respectively. Referencing the prior discussion, increasing NSA by a factor of 4 results in a twofold increase in SNR, reflected by the reduction in “graininess” in the image, but at the cost of a fourfold increase in scan time. The patient has long-standing compensated hydrocephalus due to a small tectal mass (arrow) causing obstruction of the cerebral aqueduct.

Fig. 14.2 presents axial T2-weighted images with the difference between (a) and (b) again being a fourfold increase in scan time, translating to a twofold improvement in SNR. An expansile mass fills the left cavernous sinus (black arrow, b), representing residual tumor (meningioma) in this postoperative patient. The lesion is encasing the carotid artery (the central black flow void). Note the improved depiction of the carotid artery and its interface with surrounding tumor in (b), due to the improved SNR. The “graininess” of the image in (a) can be appreciated, when compared with (b), due to the lower SNR. However, the difference is less evident than in Fig. 14.1. If SNR in a scan is already fairly high, then increasing SNR further will be of little benefit (and be difficult to appreciate), in part explaining this observation.
The contrast-to-noise ratio (CNR) is defined as the difference in signal contribution (signal intensity [SI]) between different tissues, divided by the background noise (N). For example, the contrast between cerebrospinal fluid (CSF) and white matter (WM) is given by \((SI_{\text{CSF}} - SI_{\text{WM}})/N\). Note that this measurement is specific to the tissues being compared.

**Fig. 15.1** presents axial T1-weighted, T2-weighted, and FLAIR scans acquired at 3 T in a 38-year-old woman with a history of relapsing/remitting multiple sclerosis (MS) dating back more than 12 years. There are multiple punctate, partially confluent, periventricular white matter lesions consistent with this diagnosis. These lesions have low signal intensity on the T1-weighted scan, and high signal intensity on both the T2-weighted and the FLAIR scans, relative to normal adjacent white matter. These three scans rank in the following order, \(T2 > T1 > \text{FLAIR}\), in terms of CNR for CSF versus normal WM. Translating this to words, the difference in signal intensity between CSF and normal WM, taking into account the noise in the image, is greatest for the T2-weighted scan and least for the FLAIR scan. Note that the scans rank differently, however, when looking at lesion CNR. The three scans rank in the following order, \(T2 > \text{FLAIR} > T1\), in terms of CNR for MS plaques versus normal WM. To make things more complicated, if one calculates CNR for MS plaques versus CSF, the ranking is \(\text{FLAIR} > T1 = T2\). So, the tissues of interest are critical to stating the CNR for an image, and an image may have high CNR for one type of comparison and low CNR for another.

Lesion conspicuity is related to CNR. Of the three scan types illustrated, FLAIR demonstrates the highest CNR for MS plaques when compared with CSF, and thus the widespread use of this technique for screening in MS of the brain.
Fig. 16.1 depicts the signal-to-noise ratio (SNR) for white (WM) and gray matter (GM), and their contrast-to-noise ratio (CNR) for identical measurement times but different TR values. Scan time was held constant, despite TR being twice as long, by using two acquisitions (averages) for the first scan. Increasing the TR from 430 msec (Fig. 16.1b) to 860 msec (Fig. 16.1d) on a 1.5 T system will lead to a ~7% increase in SNR for WM, but a ~30% drop in CNR (WM/GM)! Selecting a TR of 860 msec will increase the overall signal from both GM and WM, but the CNR will be significantly lower. As a consequence—illustrated on cropped sagittal T1-weighted images—visualization of a brain metastasis (white arrow) and accompanying vasogenic edema (black arrow), as well as normal gray-white matter differentiation, is improved with the shorter TR (higher CNR).
The signal intensity of each voxel displayed in MRI depends on multiple tissue characteristics, two of these being T1 and T2, as well as the parameters specified for the acquired pulse sequence; for example, TR, TE, TI, and flip angle. However, there are many additional factors that also influence signal intensity. As mentioned in other sections, the signal intensity of each voxel is directly proportional to the number of mobile protons within the volume element. Increasing the voxel size thus increases the signal-to-noise ratio (SNR); however, this decreases spatial resolution: an undesirable effect in clinical imaging. This means that the higher the resolution, the lower the SNR of the underlying displayed volume element. A further factor influencing SNR is the field strength utilized for MR imaging, with the desire for higher SNR principally driving the move to ever-higher clinical field strengths.

◆ Field Strength

With increasing magnetic field strength, the energy difference between the two levels that protons (spins) can occupy is increased. As a consequence, a greater number of spins occupy the lower energy level prior to any excitation. Thus, with increasing field strength, more spins are available to be excited to the higher energy level, together with more energy needed to excite a spin from the lower to the higher level, resulting in a stronger signal being emitted when the spins return after excitation to the lower energy level. This gain in SNR, according to theory, is directly proportional to the increase in field strength.

The improvement in SNR with high-field imaging (e.g., when moving from 1.5 to 3 T) can be used in general to either increase spatial resolution or to reduce the scan acquisition time, or a combination thereof. In theory, an increase in field strength by a factor of 2 increases SNR by a factor of 2 as previously stated, assuming that all other imaging factors are held constant. But in practice, this does not hold true for several reasons.

◆ Chemical Shift

In MR imaging, phase encoding in one direction and frequency encoding in the other provide the required in-plane spatial information. In tissues with only water protons, this leads to a properly encoded image. However, fat protons have a lower resonant frequency as compared with water protons. Due to this difference, the signal from fatty tissue may be shifted by several pixels, but only along one axis, that of frequency encoding. Depending on the number of pixels the signal is shifted, these so called chemical shift (misregistration) artifacts may be seen as hypo- or hyperintense bands or even as an entire ghost artifact of the fatty tissue adjacent to the object being imaged. These pixel shift artifacts mainly depend on two factors: the imaging bandwidth and the field strength. Chemical shift artifacts increase with field strength, all other factors being held constant, and can be reduced by using a higher imaging bandwidth.
Returning to the topic of improved SNR at 3 T, in most applications chemical shift must also be accounted for, reducing in part this benefit. MR imaging at 3 T is thus typically performed with a higher imaging bandwidth, which leads to lower SNR, diminishing the SNR gained by going to the higher field strength.

**Through-Plane Resolution**

Slice thickness in MR has decreased steadily since its clinical introduction. For example, in the early 1980s, when only low-field systems were available, 10 mm was the standard slice thickness for brain imaging. However, 5-mm thick images are now routinely acquired on 1.5 T systems. With the improved SNR at 3 T, it is possible to further decrease slice thickness while preserving image quality. Depending on site and user preference, the standard slice thickness for brain imaging at 3 T is currently 3 to 4 mm. If some of the available SNR is used as well to reduce scan time, then an additional benefit of 3 T as compared with 1.5 T is decreased artifacts due to patient movement, leading overall to improved image quality. In particular, in brain imaging, these advantages allow for better workup of critically ill and/or uncooperative patients, such as in the diagnosis and management of acute stroke patients.

**Fig. 17.1** illustrates the improvement in SNR at 3 T and its impact on slice thickness. Axial T2-weighted images of the brain are illustrated at 1.5 T with (a) a slice thickness of 5 mm and compared with images in the same patient at 3 T with a slice thickness of (b) 5 mm and (c) 2.5 mm, respectively. A small early subacute infarct along the midline in the left anterior cerebral artery territory is noted (arrow). As expected, there is a substantial improvement in SNR at 3 T when all imaging parameters, including slice thickness, are held constant. This is reflected by the graininess of (a) when compared with (b), with both scans being 5 mm in slice thickness but the first at 1.5 T and the second at 3 T. The SNR in (c), with a 2.5-mm slice thickness at 3 T, is much closer to that of (a), with a 5-mm slice thickness at 1.5 T. This would be expected, given that a reduction in slice thickness by a factor of 2 should halve SNR, which is made up for by the doubling of field strength. A further major reduction in slice thickness at 3 T (e.g., to 1 mm sections, using 2D imaging technique) would lead to an unacceptable, noisy image, and thus is not employed on a routine basis for screening exams.

![Image of brain slices](image-url)

**Fig. 17.1** SNR for the brain at 1.5 vs 3 T with slice thickness the same, then halved.
Slice Orientation

A major strength, clinically, of MR is the ability to acquire thin slices at any angle or orientation within the body. Gradient and RF hardware make such scans possible. This difference, relative to CT, has however been in part negated due to the development of multidetector scanners.

In MR, nuclei (protons) subjected to a homogeneous, static magnetic field resonate with a frequency related to that field. At 1.5 T, the frequency for hydrogen ($^1$H) is ~63 MHz. An RF pulse generated at this specific frequency causes all the resonating hydrogen nuclei within the homogeneous field to absorb and release energy, making spatial localization of a specific tissue impossible. However, when a magnetic field gradient is applied, the nuclei experience different magnetic field strengths and begin to resonate at different frequencies based on their position. An RF pulse can thus be tuned to a specific frequency to excite only nuclei in a desired location based on their resonant frequency. This is the concept that makes slices in any orientation possible.

MRI systems are equipped with three spatial encoding gradients ($x$, $y$, and $z$) made of loops of wire that either add to or subtract from the main magnetic field when a current is passed through them. A gradient change applied along only one axis provides the ability to create a slice orthogonal to that gradient. Turning on the gradient in the $x$ direction results in sagittal slices, $y$ in coronal slices, and $z$ (along the bore of the magnet) in transverse (axial) slices. The thickness of a slice can be defined by 1) adjusting the bandwidth of the RF pulse to increase or decrease the range of frequencies included and therefore protons excited by the pulse, or 2) by increasing or decreasing the strength of the gradient. Gradient changes made in multiple directions (e.g., $z$ and $y$) allow slices to be tilted away from a single axis (oblique).

Fig. 18.1 presents axial and coronal postcontrast T1-weighted slices at 3 T in a patient status post resection of a mucoepidermoid carcinoma in the parotid, now with recurrence in the lateral pterygoid muscle (black arrow) and extension into the

![Fig. 18.1](image_url) Different planes may better visualize disease, depending on anatomy.
foramen ovale (white arrows). The location of the two axial slices is depicted on the coronal scan. Slice selection was made possible by the use of the z gradient for the axial slices and the y gradient for the coronal slice. Acquisition of slices in more than one plane is the norm for a clinical MR exam today, with the specific planes employed depending on the body part and pathology in question. Relevant to Fig. 18.1, the axial plane is the standard for imaging of the skull base. However, coronal imaging is relied upon for specific additional information: for example, as shown, detection of perineural tumor spread of malignant head and neck tumors through the foramen ovale.

Fig. 18.2 depicts sagittal and axial T2-weighted images of the cervical spine acquired at 3 T, with the positioning of the axial scan indicated on the sagittal. Axial imaging of the cervical spine is typically performed with the scans tilted to be parallel to the disk space. This permits improved imaging of disk disease, such as the central disk herniation depicted (arrow). Tilting the slice in the manner illustrated required using both the z and y gradients.

Lumbar spine imaging takes particular advantage of the ability to tilt slices in MRI, with two different approaches used for axial imaging. At some sites, individual blocks of images (all part of a single axial acquisition) are tilted to the individual disk spaces being examined, whereas, at other sites, a single block of images is acquired tilted to match the orientation of one level. Difficulty in patient positioning can be corrected by tilting the images acquired—for example, by the use of a coronal scout with subsequent correction of the sagittal scan setup so that true (anatomic) sagittal images are acquired (depicting the spinal canal in its entirety on one image). Automated image selection in many regions of the body, for example the brain, knee, and spine, tilted in two planes to be truly parallel to anatomical landmarks, is now possible on many MR systems (see Chapter 139). Cardiac imaging routinely employs double oblique images, once again made possible by use of the gradients. Taken to the extreme, it is even possible to individually position and tilt each slice in a multislice scan, with this all made possible by the gradients.
The time needed to excite and spatially encode one slice is referred to as the slice loop time (Fig. 19.1). The repetition time (TR) is defined as the time between two excitation pulses for the same slice and is a major contrast-dictating parameter in MRI. On completion of the encoding of a given slice, there is usually time prior to the next excitation pulse to excite and encode other slices. In this approach, called multislice imaging, the maximum number of slices is given by the repetition time divided by the slice loop time.

If more slices are needed to cover the anatomic region than allowed by the selected TR, then there are many options, with three subsequently discussed (simultaneous multislice technique is a recently introduced further major alternative, to provide additional slices, and is discussed in detail in Chapter 73). The user can increase TR (which changes tissue contrast and prolongs scan time), use faster gradients (see Chapter 123), or employ concatenations. In the latter approach, the number of slices is evenly split into two or more sets, specified by the number of concatenations (Fig. 19.2). In doing so, TR remains unchanged. However, scan time is directly proportional to the number of concatenations. For example, if two concatenations are chosen, scan time is doubled. Choosing to employ concatenations can be very time inefficient, in particular if only a small number of additional slices are needed.

When employing concatenations, each set of slices (each concatenation) is acquired sequentially in time. Commonly, with two concatenations, every other slice (anatomically) is acquired first, and then in the second scan (concatenation) the missing interleaved slices are acquired. This is done to diminish further slice-to-slice interference, also known as crosstalk (see Chapter 22). The degree of patient motion, and the actual position of the patient, can of course be different between the sets of slices acquired with each concatenation, due to the difference in when each set of slices was actually acquired. When the images are viewed in a continuous loop, sorted anatomically (as on PACS), this effect is often observed. In the worst-case scenario, this can even

![Fig. 19.1 Definition of slice loop time.](image-url)
lead to a small lesion being missed, if the patient moved substantially between the different concatenations.

Fig. 19.3a presents the setup for the acquisition of a T1-weighted sagittal scan sequence of the brain. For the given TR a maximum number of 19 slices could be acquired, thus not covering the entire brain. By applying two concatenations (Fig. 19.3b), the number of slices can be adjusted to 30 allowing for appropriate coverage of the brain for the given TR. Note, however, that the maximum number of slices that could be achieved applying two concatenations in this instance was 38 (Fig. 19.3c). This exceeds by far the borders of the brain and demonstrates that applying concatenations, although easy to do, may not be the most time efficient way to achieve the desired number of images.
Number of Averages

The axial T2-weighted images of the cerebral peduncles and aqueduct (acquired using fast spin echo technique at 1.5 T with a 2-mm slice thickness) presented in Fig. 20.1 are from the same volunteer and were acquired using one, two, four, and eight averages, respectively. The number of averages is the number of times each line in \( k \)-space is filled or sampled. This parameter is also known as the number of acquisitions, number of signals averaged (NSA), or number of excitations (NEX), depending on vendor. Because it represents the number of times each line in \( k \)-space is filled, it directly affects scan time. In this example, the image in Fig. 20.1a (1 NSA) was acquired in 32 sec, in Fig. 20.1b in 59 sec, in Fig. 20.1c in 1 min 53 sec, and in Fig. 20.1d in 3 min 41 sec (actual scan times). As the number of averages is increased, the voxels in each corresponding image have a higher signal-to-noise ratio (SNR), with each image thus progressively less “grainy” to the eye.

The first and perhaps most important point to note is that doubling NSA doubles scan time but does not double SNR. Although scan time is linearly related to NSA, SNR is proportional to the square root of NSA (see Chapter 14). Thus, the scan time for Fig. 20.1b is twice that of Fig. 20.1a, yet the SNR of the image is only 41% (1.41 or the square root of two) higher. To double SNR using NSA, one would have to increase NSA, and therefore the overall scan time, by a factor of 4. This is in stark distinction, for example, to doubling the slice thickness, which would increase SNR by a factor of 2 without an increase in scan time. This would result in the same increase in SNR as going from 1 NSA to 4 NSA (Fig. 20.1a vs. Fig. 20.1c).
without the resultant scan time increase. It is also important to note that one eventually reaches a point of diminishing returns in regard to the SNR increase provided by increasing NSA. With high SNR images, a further increase in SNR leads to little perceptible difference in the image or additional diagnostic information. For example, there is a marked improvement in image quality from Fig. 20.1a to Fig. 20.1c, but less perceptible change from a further doubling of NSA (and scan time) for Fig. 20.1d.

Low-contrast lesion detectability, in particular, is improved by an increase in SNR. For example, in this comparison, the substantia nigra (with subtle low signal intensity, due to iron deposition) is best identified in the higher SNR images arrows, (Fig. 20.1d). Averaging can also reduce the visual appearance of motion artifacts that originate from random or aperiodic motion. Using averaging in this way, however, has become less common in the last two decades as scan times for routine imaging became progressively shorter.

Fig. 20.2 presents coronal fast spin echo T2-weighted scans obtained through the pituitary at 3 T with a slice thickness of 2 mm. (a) was acquired with 1 average and a parallel imaging factor of 2, and (b) with 3 averages. Scan time thus differed (by a factor of 6 approximately), and SNR by a factor of the square root of 6 (or 2.4). Note the decreased graininess in (b), reflecting the higher SNR. As also illustrated in Fig. 21.1, differentiation of structures with similar tissue contrast (low contrast lesion detectability), such as gray and white matter, is difficult in Fig. 21.2a, due to the low SNR, but markedly improved in (b). Thus, detection of a pituitary microadenoma (which on T2-weighted scans is typically only slightly higher in signal intensity than the normal pituitary), the primary reason for such an exam, would be markedly improved in (b).

In the past, a large number of averages was often needed to achieve acceptable image quality. Yet today, at 3 T, averaging is generally restricted to thin-section, high-resolution imaging, due to the increased available SNR. For Fig. 20.2b, despite the high in-plane spatial resolution of 0.7 × 0.6 mm, only three averages were required to obtain a high-quality exam. In many exams at 3 T, NSA is 1, making the use of parallel imaging more a reality. The latter technique provides higher spatial resolution and/or reduced scan times at the cost of lower SNR, and is only really advantageous when there is sufficient SNR to allow 1 average.
The thickness of an MR image, specified during scan setup, has a major impact on the quality of the resultant image, as well as on the imaging parameters used. When a thicker slice is acquired, the signal from more tissue is averaged, which can lead to poor anatomic definition and, in particular, to poorer definition of tissue interfaces. This phenomenon is known as volume averaging. The effect of volume averaging becomes especially problematic when the thickness of the acquired slice exceeds the size of a structure being evaluated. Therefore, it is desirable to acquire the thinnest slice possible to resolve structures with the greatest detail. However, as the slice thickness is reduced, the signal-to-noise ratio (SNR) is reduced by a proportional amount. To maintain

Fig. 21.1 Slice thickness and SNR.
adequate SNR (on thin slices) for definition of structures with low-contrast detectability, an increase in signal averaging, and thus scan time, must be applied.

Unfortunately, SNR is proportional to the square root of the number of averages, yet, as previously noted, directly proportional to slice thickness. Thus, halving the slice thickness reduces SNR by a factor of 2. To compensate for this reduction, the number of averages has to be increased by a factor of 4. Fig. 21.1 presents axial T2-weighted images acquired with a slice thickness of 8, 4, and 2 mm, respectively. For these three scans, SNR was kept constant by changing the number of averages. Thus, Fig. 21.1a was acquired with 1 average, Fig. 21.1b with 4 averages, and Fig. 21.1c with 16 averages. This means that although Fig. 21.1c shows a substantial increase in tissue contrast (note the improved delineation of cortical gray matter anteriorly and of the middle cerebral artery [MCA] branches within the sylvian fissure), the scan time for Fig. 21.1c was 16 times that of Fig. 21.1a. From a practical viewpoint, with certain scan sequences, very thin slices are simply not feasible due to the very long scan time that would be required. Fig. 21.1d demonstrates a slice thickness reduction from 8 to 2 mm without an increase in averaging to compensate for the loss in SNR. The result is a very “noisy” or low SNR image. This has little effect on high-contrast structures such as the flow voids from MCA branch vessels (black arrows) traveling through the slice, but an appreciable effect on low-contrast structures such as the putamen, globus pallidus, caudate nucleus, and especially the gray–white matter differentiation (white arrow).

When selecting the thickness of slices to be acquired, scan time, slice coverage, SNR, and (high and low) contrast detectability must all be taken into consideration. For example, Fig. 21.2 presents axial post-contrast T1-weighted images acquired with a slice thickness of 2 mm (a) and 4 mm (b), with all other parameters held constant. The scans depict a postoperative sphenoid wing meningioma, with residual enhancing tumor (arrow) within the sphenoid sinus, cavernous sinus, and medial middle cranial fossa. The SNR in (a), despite being half that of (b) due to the 2-mm slice, is still sufficient for diagnostic purposes, with the thinner section permitting improved depiction of residual tumor.
Ideally, a slice in MRI should experience a uniform radiofrequency (RF) excitation throughout its thickness. Sharp, distinct edges should exist with no excitation extending beyond slice boundaries. However, in practice, the spatial excitation of spins is invariably a distribution ranging from the RF flip angle specified at the center of the slice to largely reduced flip angles at the ill-defined edges that excite regions well beyond the desired thickness. The resultant change in signal across the thickness of a slice is termed the slice profile.

Thus, in multislice imaging (the mainstay of clinical MR today) a slice of interest may suffer interference, or “crosstalk,” from neighboring slices caused by RF excitation that extends beyond their slice boundaries. Signal-to-noise ratio (SNR) loss (compare Fig. 22.1a, which was performed with no gap, with Fig. 22.1b, which was performed with a 100% gap) and contrast changes (compare Fig. 22.2a, which was performed with no gap, with Fig. 22.2b, which was performed with a 100% gap) may result. Fig. 22.1 illustrates a brain metastasis (arrow) on post-contrast T1-weighted scans. Fig. 22.2 illustrates a recent cortical infarct (arrow) on T2-weighted scans, with the loss of contrast best identified by comparing the signal intensity of CSF (asterisk) with that of normal brain. When the gap between slices is reduced, slice-to-slice interference becomes more likely. True contiguity is theoretically not possible in two-dimensional multislice MRI, with the worst effects seen at 0% slice gaps where interference between slices is the greatest. Conversely, the best results occur when the gap is large enough so that neighboring slice excitations do not interfere with each other. Gaps of 10 to 30% are common in current clinical practice.

Because of slice profiles, a lesion directly in the center (thickness-wise) of a slice is seen best (lesion A in Fig. 22.3), and a lesion near the edge (lesion B in Fig. 22.3) potentially less well (Fig. 22.3 shows fluid-attenuated inversion recovery [FLAIR] imaging in multiple sclerosis). This also means that small lesions (e.g., lesion C) can be missed in between slices, thus one reason for imaging in two planes in disease processes such as multiple sclerosis and (post-contrast) brain metastases.

Fig. 22.1 Loss in SNR, on T1-weighted images, due to slice-to-slice interference.
Fig. 22.2 Loss of tissue contrast, on T2-weighted images, due to “crosstalk.”

Fig. 22.3 An illustration of slice profile, and pitfalls therein.
23 Slice Excitation Order (in Fast Spin Echo Imaging)

In theory, an ideal radiofrequency pulse (RF) for MRI is uniform across the slice thickness, with sharp, distinct edges and no excitation beyond the boundaries (edges) of the slice. In practice, however, the spatial excitation of spins is invariably a distribution ranging from the nominal RF excitation flip angle at the center of the slice to largely reduced flip angles at the ill-defined edges (see Chapter 22). Stating this slightly differently, the flip angle varies across the slice, from one edge to the other. It also follows that there is partial excitation of tissue adjacent to, but beyond, the theoretical boundaries of the slice. Time-limited RF pulses are inherently imperfect, resulting in this non-rectangular slice profile. In multislice fast spin echo imaging, this limitation is accentuated, because the slice profile is defined by the overlapping RF profiles of the excitation pulses as well as the 180° refocusing pulses applied over the duration of the echo train.

The top half of Fig. 23.1 presents the pulse diagram for a typical fast spin echo sequence. Illustrated in the bottom half of the figure, every slice receives in this instance a 90° excitation pulse, followed by multiple 180° refocusing pulses (each with a different phase encoding, 1 in Fig. 23.1). Each RF pulse has an imperfect slice profile, with the effects compounded in the final resultant image. Schematically sequential slice acquisition using fast spin echo technique is illustrated by 2 in Fig. 23.1. For clarification of the figure, if the slices were axial in orientation, the slices to the left might be the more caudal sections and those to the right more cranial in location. A sequential acquisition would excite first the most caudal slice, then the adjacent more cranial slice, and so on. This would emphasize in a non-desirable fashion the effects of the non-ideal flip angle profile, due to accentuated crosstalk overlapping excited regions) between adjacent slices.

Fig. 23.1 Slice excitation order, illustrated with a pulse diagram.
The nonideal RF profiles of the 90° excitation pulse and the 180° RF refocusing pulses lead to a decrease of available longitudinal magnetization in the adjacent slice. To minimize this influence, an interleaved slice acquisition (3 in Fig. 23.1) may be performed by chronologically altering the order in which slices are excited. The first, third, fifth, and subsequent slices are excited in sequence, followed then by the second, fourth, sixth, and subsequent slices. In this way, the longitudinal magnetization within adjacent slices has additional time for recovery, while sequence acquisition is performed in the other interleaved slice block. Interleaving in this fashion is routinely employed for most multislice fast spin echo scans. Interleaving of slice excitation should not be confused with concatenation (see Chapter 19), in which two separate scans are acquired. Interleaving, as discussed, occurs within a single acquisition, and specifically does not prolong scan time.

Fig. 23.2 demonstrates two fast spin echo multislice acquisitions with images acquired sequentially (a) and in an interleaved fashion (b). The differences in signal intensity and contrast depicted are due to the difference in order of slice excitation, and specifically not due to image windowing (which has been held constant). Saturation effects due to overlapping excitation and refocusing profiles between adjacent slices lead to an overall signal loss, which is accentuated with sequential slice acquisition. Note the marked loss in SNR of normal brain in (a) compared with the image (b), the latter acquired in an interleaved fashion with less crosstalk. Nonideal excitation can also lead to increased T1-weighting (another way to view this phenomenon is that the effective TR as the edge of the slice is approached is less), although this effect is not noticeable in the example provided.

Fig. 23.2 FSE T2-weighted images with sequential vs. interleaved slice excitation.
Field of View (Overview)

The field of view (FOV) is defined as the dimensions of the exact anatomic region included in a planar (2D) image. In MR, the FOV may be square or asymmetric. Depending on the equipment manufacturer, it is specified in millimeters or centimeters. The FOV is also the mathematical product of the acquisition matrix and the pixel dimensions. For example, if 512 readout and 256 phase encoding steps are specified in a scan for which the pixel dimensions are chosen to be 0.45 \times 0.9 \text{ mm}, the FOV would be 512 \times 0.45 \text{ mm} = 230 \text{ mm} by 256 \times 0.9 \text{ mm} = 230 \text{ mm} (and thus in this instance a square FOV, despite the use of a rectangular shaped pixel). Head imaging is typically performed today with a FOV of 230 mm or less (in each dimension) to achieve high in-plane spatial resolution. Depending on body habitus, the FOV for a scan of the upper abdomen may be as large as 400 mm (in each dimension).

The choice of FOV in clinical work is somewhat complex, involving a trade-off between the signal-to-noise ratio (SNR) and spatial resolution, and depending as well on the size of the body part being scanned. Frequently, a smaller FOV is desired for the improved spatial resolution it affords. However, SNR is proportional to the square of the FOV (because the FOV specifies both dimensions of a pixel and thus its area), assuming the image matrix is held constant. More specifically, \( \text{SNR} \propto \text{FOV}(r) \times \text{FOV}(p) \), where FOV(r) and FOV(p) are the FOV in the readout and phase encoding directions, respectively. If the FOV is halved, SNR decreases by a factor of 4. Less drastic changes are the norm in clinical imaging. If the resolution in a particular scan is slightly less than desired, and SNR is not a limiting factor, then a slight reduction in FOV may be worthwhile. For example, a 20% smaller FOV provides 20% better resolution in both pixel dimensions, at the expense of a 40% reduction in SNR.

The T2-weighted images presented differ only in the choice of FOV, which was 320 (in both dimensions) versus 220 versus 120 mm (Fig. 24.1a-c). Note the extensive vasogenic edema, with abnormal high signal intensity, in the left cerebral hemisphere due to metastatic disease. The visual magnification of the images was held constant making the choice of field of view more evident. A potential problem when using a very small FOV, such as in Fig. 24.1c, is image wraparound (aliasing; see Chapter 101). Aliasing can be eliminated in the phase encoding direction by oversampling (at the cost of scan time), as employed in this example. With modern scanners, aliasing in the readout direction is no longer an issue. Fig. 24.1d compares an image from the scan series illustrated in Fig. 24.1a (but at a different anatomic level) with an image (at the same level) from the scan series illustrated in Fig. 24.1c, visually magnifying the larger FOV image to match anatomically the smaller FOV image. A brain metastasis (black arrow) with surrounding vasogenic edema is depicted. The decrease in SNR by the use of a very small FOV (120 mm) is reflected by the graininess of the image (part 2, Fig. 24.1d). However, the increased spatial resolution improves anatomic depiction of small high contrast structures, such as the cerebral sulcus (white arrow). Note the blurring of the image acquired with the 320 FOV in comparison with that using a smaller FOV, reflecting the substantially larger pixel size (and thus lower resolution) of the larger FOV scan series (part 1, Fig. 24.1d).
In patients with exceptionally large heads, employing the standard FOV can lead to wraparound, particularly in the sagittal plane, with the nose (or jaw) overlapping the posterior part of the head. As previously noted, oversampling can be employed in the phase encoding direction to eliminate wraparound, but at the cost of increased scan time. However, SNR will be improved as a result. The choice of direction for the readout gradient is typically determined by the anatomic part and the possibility of aliasing: for example, with the readout direction often chosen to be craniocaudal in sagittal and coronal imaging of the head to eliminate wraparound from the neck.

**Fig. 24.1** Changing the FOV affects both spatial resolution and SNR.
Field of View (Phase Encoding Direction)

The field of view (FOV) defines the part of the patient to be imaged. This is chosen prior to scan acquisition, and need not be square. Indeed, in many circumstances a reduced FOV along one axis can be advantageous. The topic of this chapter is the choice of the FOV in the phase encoding direction.

In Fig. 25.1, the FOV in the phase encoding direction (right to left in this instance) was changed from 100% (Fig. 25.1a) to 75% (Fig. 25.1b) to 50% (Fig. 25.1c). A small enhancing left frontal metastasis is illustrated (black arrow, Fig. 25.1a) on post-contrast T1-weighted scans. Images are displayed as acquired, without cropping or differential magnification. Because the pixel size was held constant, fewer phase encoding steps were required for Fig. 25.1b (three-fourths the number) and Fig. 25.1c (one-half the number). Scan time is directly proportional to the number of phase encoding steps, and so the scan time of Fig. 25.1b was three-fourths that of Fig. 25.1a, and that for Fig. 25.1c was one half that of Fig. 25.1a.

However, as illustrated, there are two potential problems associated with using this approach, also termed a “rectangular” FOV. The first problem is the wraparound (aliasing) artifact. If the part of the patient being scanned extends beyond the FOV in the phase encoding direction, that part will appear superimposed on the image on the other side. Thus, in Fig. 25.1c, the right part of the head appears superimposed over the (anatomic) left side of the image (white arrow), and the left part of the head over the right side of the image. The second problem is reduced SNR. Using fewer phase encoding steps (p) leads to lower SNR (SNR ∝√p). SNR decreased from (a relative value of) 1 in Fig. 25.1a to 0.87 in Fig. 25.1b to 0.71 in Fig. 25.1c. The loss in SNR due to halving the FOV in the phase encoding direction could be compensated by doubling the number of acquisitions (averages), but then the two scans would have the same acquisition time.

Fig. 25.2 compares similarly magnified parts of Fig. 25.1a (top) to Fig. 25.1c (bottom) to illustrate better the SNR loss. The increased graininess of the image on the bottom is due to its lower SNR. Note that if the area of interest is near to the center of the FOV, then some image wrap can be tolerated. In this instance, however, wraparound is still evident (white arrow) on the rectangular FOV image, despite the magnification and cropping employed.

A rectangular FOV is commonly used in axial head imaging (without changing the number of averages), decreasing scan time with only a minimal reduction in SNR. Fig. 25.1b is an example of this application, with the FOV in the phase encoding direction chosen to closely match the dimension of the head (right to left). A rectangular FOV finds similar application in imaging of other body parts that have a reduced width in one dimension (e.g., for axial imaging of the upper and lower abdomen, and for many musculoskeletal exams). It should be noted that in the preceding discussion the pixel size has been assumed to be square. This need not be the case, and the shape of the pixel can be varied together with the FOV, thus also influencing spatial resolution along one axis. With modern MR scanners, the increment by which the FOV, the number of
phase encoding steps, and the number of readout steps can be changed is almost unlimited. The advent of parallel transmission brings even greater flexibility and complexity to the choice of phase encoding steps, allowing selective excitation ("zooming") for increased spatial resolution in a region of interest (see Chapter 130).

Fig. 25.1 Changing the phase encoding FOV.

Fig. 25.2 Lower SNR due to a reduced phase encoding FOV.
The images presented demonstrate the effect of changing the acquisition matrix in the frequency encoding (readout) direction. All are T2-weighted fast spin echo sagittal images of the lumbar spine. The acquisition matrix (readout x phase encoding) was 1024 × 256 for Fig. 26.1a, 512 × 256 for Fig. 26.1b, and 256 × 256 for both Fig. 26.2a and Fig. 26.2b. The image in Fig. 26.2b was further interpolated to 512 × 512 prior to display.

When the MR signal (echo) is produced, it is sampled in the presence of a gradient magnetic field. This gradient is thus referred to as the “readout” or “read” gradient. The digital sampling of the echo produces data points along the frequency direction of $k$-space. The number of samples taken during the readout period is determined by the desired number of pixels in the frequency encoding direction. Not to confuse the issue, but this is why one may see the read gradient referred to as the “frequency encoding” gradient. If a frequency resolution of 512 pixels is desired, then...
the echo will be sampled 512 times during the readout period (Fig. 26.1b). Increasing the frequency resolution to 1024 results in 1024 samples taken during readout (Fig. 26.1a). In most clinical imaging situations, the frequency acquisition matrix is equal to or greater than the phase matrix, because the choice of the frequency matrix does not affect scan time.

Although increasing the acquisition matrix in the read direction does not affect scan time, it does affect the signal-to-noise ratio (SNR) of the image. As the pixel size is reduced, the MR signal is reduced, making the noise more obvious (leading to the “grainy” appearance of the image). One can easily see that Fig. 26.1a (1024 × 256) has higher spatial resolution but also much lower SNR than Fig. 26.1b (512 × 256). Reducing the read matrix further to 256 (Fig. 26.2a,b) increases SNR again, but results in even lower spatial resolution.

MR images are commonly interpolated to higher matrices for display purposes. This interpolation, however, merely “smooths” out the pixels one would see in the image if it were not interpolated (compare Fig. 26.2a and Fig. 26.2b). The degree of interpolation and the algorithm used depend on the MR system vendor. Note that the acquisition matrix, not the reconstructed or displayed matrix, determines the spatial resolution of an MR image.

Fig. 26.2 Use of interpolation for improved image display.
Matrix Size: Phase Encoding

**Fig. 27.1** presents sagittal T2-weighted images of the midlumbar spine demonstrating mild degenerative disk disease (loss of disk hydration, with the nucleus pulposus not being of normal high signal intensity). **Fig. 27.2** presents sagittal T1-weighted images, in a different patient, of the lower thoracic and upper lumbar spine at the level of the conus medullaris, revealing a benign chronic compression fracture of L1, with anterior wedging.

The selection of the number of phase encoding steps specifies how many different lines of $k$-space will be filled during the acquisition. This likewise determines the number of pixels along the phase encoding direction of the acquired FOV. Thus, the number of phase encoding steps directly affects scan time. Flow and/or motion artifacts also are propagated along the phase encoding direction. Additionally, assuming the display and acquired FOV are identical, if there is significant signal from excited tissue outside the FOV, it will wrap, or fold over, into the displayed FOV (see Chapter 101). This occurs in both the phase and frequency direction and is overcome by oversampling. Oversampling increases the scan time when applied in the phase encoding direction (except for single shot techniques), assuming that all other parameters and in particular the number of signals averaged are held constant.

**Fig. 27.1a** was acquired using a 256 read matrix and 128 phase matrix. **Fig. 27.1b** was acquired with a 256 read and 256 phase matrix. The phase encoding direction

![Fig. 27.1 Improved spatial resolution due to more phase encoding steps.](image)
in both examples was in the craniocaudal direction, with 100% oversampling. The scan
time for Fig. 27.1a was 2:08 and for Fig. 27.1b 4:08. Although the scan time for
Fig. 27.1b was twice that of Fig. 27.1a, with an acquired FOV of 280 mm, the pixel
dimension in the phase direction for Fig. 27.1a was 2.2 mm and that for Fig. 27.1b 1.1 mm.
Fig. 27.1b thus has higher spatial resolution. As an aside, it is worthwhile noting that
selecting the phase direction to be craniocaudal for sagittal T2-weighted fast spin
echo (FSE) imaging of the spine can substantially reduce the conspicuity of CSF pulsation
artifacts.

The image in Fig. 27.2a was acquired using a read matrix of 512 and a phase matrix
of 256. Fig. 27.2b was acquired using a read matrix of 512 and a phase matrix of 512.
The phase encoding direction was anterior to posterior in both instances. The scan time
for Fig. 27.2a was 3:38 and that for Fig. 27.2b 7:15. With the higher matrix size, the
pixel dimension in Fig. 27.2b in the phase encoding direction is reduced and spatial
resolution increased (thus, the image appears less blurred). The pixel dimension in the
phase (and read) direction for Fig. 27.2b is just over 0.5 mm. One will also note that as
the pixel size is reduced, SNR is reduced, with the result being an increase in the overall
“grainy” appearance of the image. However, one should recall that with pixel size con-
stant, increasing the phase encoding matrix actually increases SNR (with SNR \( \propto \sqrt{\text{number of phase encoding steps}} \)).

In summary, spatial resolution can be improved by increasing the number of phase
encoding steps, which results in a smaller pixel dimension along the phase FOV. How-
ever, because this increases the number of k-space lines acquired, it also increases scan
time. Reducing the pixel size, however, in either the read or phase direction, reduces
SNR, assuming all other parameters are held constant.

Fig. 27.2 Lower SNR, but higher resolution, with more phase encoding steps.
Partial Fourier

Fig. 28.1 presents full (a) and half-Fourier (b) post-contrast T1-weighted spin echo scans, in a patient with a ring enhancing metastasis (arrow) from non-small-cell lung carcinoma. One scan average or acquisition was performed for (a), whereas partial Fourier imaging was implemented for (b) (specifically a 4/8ths acquisition). The image in (b) was acquired in just over half the scan time of (a). The reduction in scan time is due to the implementation of partial Fourier imaging, with fewer phase encoding steps acquired (in clinical practice, typically 4/8 to 7/8ths). The data as acquired in k-space for (a) is presented in Fig. 28.1c, and that for (b) in Fig. 28.1d. Note that just over half of k-space is sampled in (d). Spatial resolution is preserved with partial Fourier imaging and is thus identical for (a) and (b). However, there is a resultant decrease in signal-to-noise ratio (SNR) due to the implementation of partial Fourier imaging, reflected by the subtle increase in graininess of white matter seen in (b).

As discussed earlier, in k-space (see Chapter 12) the vertical and horizontal axes represent phase encoding and frequency encoding, respectively. In a 256 × 256 (readout × phase encoding) matrix, therefore, 256 phase encoding steps are acquired, from –127 to 128. It is not absolutely necessary in MR to obtain the entire dataset from –127 to 128 because there is a type of mathematical symmetry (conjugate symmetry) that exists between steps –127 to 0 and 1 to 128. Due to this symmetry, data from a fraction (at least half) of k-space may be used to create the entire image. This process is termed partial Fourier technique. In standard practice, the size of the measured matrix may range from half to all in some given increment (e.g., four-eighths, five-eighths, etc.). Of note is that in a half-Fourier acquisition, slightly more than 50% of the phase encoding steps are measured to ensure adequate data incorporation of the center of k-space.

The true benefit of partial Fourier imaging is a percentage decrease in scan time by a factor equal to the percentage decrease in the number of measured phase encoding steps. With all other imaging parameters held constant, partial Fourier technique preserves spatial resolution (see Chapter 13). This technique, however, decreases SNR due to a decrease in the total number of phase encoding steps (SNR is directly proportional to the square root of the number of phase encoding steps, when all other factors are held constant). This reflects a loss of redundant data and thus increases the relative impact of noise on the final image. In half-Fourier technique, for example, acquisition time is approximately halved with a decrease in SNR by a factor of 1.4. Note, however, that partial Fourier may not reduce scan times when employed with fast spin echo technique, depending on the manner in which the data are acquired. Its implementation and clinical use are thus generally limited to spin echo, gradient echo, and echo planar techniques. With the advent of parallel imaging, which can similarly achieve a reduction of scan time with preservation of spatial resolution (at the cost of SNR), partial Fourier imaging has become of less importance. Parallel imaging can be employed regardless of scan technique, but of course necessitates that there be multiple coil elements in the direction along which it is applied.

With a decrease in SNR, due to the smaller number of sampled phase encoding steps, there is a predictable decrease in the visualization of low-contrast lesions with
partial Fourier imaging, if SNR was marginal to begin with. Alternatively, if high-contrast objects dominate the scan, partial Fourier imaging has potential clinical applicability, providing high-contrast lesion/border detectability with high spatial resolution and reduced scan time. A natural conclusion is that partial Fourier imaging should be employed only when SNR is not a limiting factor. A common application of partial Fourier imaging is thus with HASTE, in which scans with very heavy T2-weighting are acquired, with only half of k-space sampled (see Chapter 39). Other applications include time resolved perfusion imaging and MRA. Although the previous discussion has been confined to partial Fourier technique in the phase encoding direction in 2D imaging, the general approach can be extended to both axes in 2D imaging and all three axes in 3D imaging (with application in the frequency-encoding direction as well as in the second phase encoding direction).
Image Interpolation (Zero Filling)

For discussion purposes, consider an image display monitor with a 1024 × 1024 pixel resolution. The image acquisition matrix in MRI is often lower, and some interpolation is typically performed automatically. A straightforward approach is to place an empty pixel between pixels (step 1 in Fig. 29.1) and assign the amplitude of this pixel to be the average value of the adjacent pixels (linear interpolation, step 2 in Fig. 29.1).

A more advanced solution is the use of a bicubic spline interpolation. In this case, the value of not only the adjacent pixel is taken into account, but also that of the next pixel.

The result is smoother, masking the truly measured spatial resolution. For Fourier encoding, a low spatial resolution matrix corresponds to a small $k$-space (limited dimension in the read-out and phase encoding directions). In any $k$-space matrix, the outer data points (the high spatial frequency components) contain the information concerning the detailed structure in the image. Consider a homogeneous phantom that fills the entire field of view. There is no detailed structure, resulting in zero values for the outer data points in the $k$-space matrix. Reflecting upon this situation, doubling the raw data matrix size and filling the missing (outer) data points with a value of zero will not improve
spatial resolution, but will mimic in some ways the measurement of a higher resolution matrix. Zero filling is simply the substitution of zeroes for unmeasured data points for the purpose of increasing the matrix size of the data prior to Fourier transformation. The result is an image with an increased display matrix that mathematically has pixels that are sinc-interpolated. Fig. 29.2 illustrates Fourier interpolation as applied in MRI. In step 1, the k-space data and corresponding image for a 128 × 128 acquisition matrix is illustrated. In step 2, zero filling is used to expand k-space. The resulting interpolated 256 × 256 image is compared with a true 256 × 256 acquisition in step 3. In Fig. 29.3, a T2-weighted axial image of the brain at the level of the pons acquired using a 256 × 256 matrix (Fig. 29.3a) is compared with the same image interpolated to 512 × 512 using zero filling in k-space (Fig. 29.3b) and a true 512 × 512 acquisition (Fig. 29.3c).

So what then is the advantage of zero filling in k-space over interpolation in the image domain? In the case of small structures, for example, a small vessel in a time-of-flight MRA of a size close to the spatial resolution of the measurement, the intensity of the representing pixel depends on where the vessel is located relative to the measurement grid. If the vessel is within a single voxel, the signal will be very bright. If the vessel lies between two voxels, these two voxels will share the intensity, resulting in a less dominant vessel appearance. Theoretically, the appearance will be improved by shifting the measurement grid, which is the same as shifting the position of the voxel, eliminating the above-mentioned partial volume effect. Zero filling in k-space corresponds mathematically to a voxel shift. In other words, spatial resolution is not improved, but partial volume artifacts are significantly reduced.

Fig. 29.3 Comparison of the resultant images using different acquisition matrices: 256 × 256, zero filled 512 × 512, and true 512 × 512.
Specific Absorption Rate

The specific absorption rate (SAR) is a measure of the energy deposited per unit time in a specified tissue by the radiofrequency (RF) pulses applied in MR. When the human body is exposed to an external magnetic field, there exist two different energy levels that a proton (the hydrogen nucleus) can occupy. In quantum mechanical terms, the spin of the proton can either be parallel or antiparallel to the direction of the magnetic field. To receive a signal in MR, an RF excitation pulse is first applied that provides exactly the energy difference between the parallel and antiparallel levels. After excitation, the spins return to their original state, emitting the energy as an MR signal. The RF pulse will also interact with water molecules, accelerating their rotational motion. An increase in motion is equivalent to an increase in kinetic energy and represents an increase in temperature. The likelihood of this interaction scales with the fifth power of the patient’s circumference. Thus, SAR is of even greater importance for obese patients. It is also critical to note that doubling the main magnetic field from 1.5 to 3 T leads to a quadrupling of SAR, if RF excitation is performed in the identical manner. Considerations related to SAR, therefore, inherently limit scanner performance by limiting the rate of RF energy deposition and cumulative deposition, in particular at higher field strengths (such as 7 T). This can lead to a reduction in slices per repetition time (TR), longer scan times, and “cooling” delays between acquisitions, if not otherwise compensated, an early challenge to clinical imaging at 3 T.

Heat deposition in MR is regulated by an international standard, specifically IEC 60601–2–33 (2010). As general background information, the order of magnitude of heat/power deposition during MR imaging is close to the human basal metabolic rate. Two SAR thresholds were established relevant to clinical imaging: normal mode and first level. Exposure below the normal mode SAR level is assumed to cause no physiologic stress to the patient. For SAR values within the range of the first level, medical supervision of the patient is required. Specifically, the MR operating software must indicate that it has to switch into first level to execute the requested scan protocol, and the operator must acknowledge the note in order for the system to continue. This should only be used when the patient is awake and able to communicate any discomfort to the operator.

Normal Mode:
- Up to 2 W/kg whole-body exposure
- Up to 2 to 10 W/kg partial body exposure, depending on the ratio between exposed and unexposed patient mass
- Up to 3.2 W/kg for head exposure
- Up to 10 W/kg for local SAR within the head/trunk region
- Up to 20 W/kg for local SAR values within the extremities
- Body core temperature is not to increase beyond 0.5°C

First-Level Mode:
- Up to 4 W/kg whole-body exposure
- Up to 4 to 10 W/kg partial body exposure, depending on the ratio between exposed and unexposed patient mass
• Up to 3.2 W/kg for head exposure
• Up to 10 W/kg for local SAR within the head/trunk region
• Up to 20 W/kg for local SAR values within the extremities
• Body core temperature is not to increase beyond 1°C

Values are averaged over a 6-minute time frame. For a period of 10 seconds, the average SAR may exceed up to three times the level of the current mode. These levels are valid for a bore temperature of up to 77°F (25°C) and decrease linearly to 0 for the normal mode and 2 W/kg for the first level for a bore temperature of 91.4°F (33°C).

The software on any commercial MR system will calculate and compare all possible limits for the selected mode and will in general indicate the most critical value. If the critical value exceeds the level of the selected mode, suggestions are made to the operator regarding which scan parameters to change (and to what value) to stay within the guidelines. No MR system will allow the execution of a protocol that exceeds the guidelines of the country where the scanner is located.

Some strategies that are used to limit SAR—for example, reducing the flip angle of the refocusing pulse in fast spin echo (FSE) imaging—can affect both image contrast and SNR. Newer methods permit reduction of the flip angle with minimal effect on tissue contrast, for example by gradually increasing the flip angle when acquiring the center of k-space and using a lower flip angle during acquisition of the periphery of k-space. Despite limitations, reduction of the flip angle for the refocusing pulse is commonly used to reduce SAR. It should be noted in this regard that SAR is proportional to the square of the value of the flip angle. Another common approach in FSE imaging is to employ a TR that is longer than the minimum necessary, thus in effect building in cooling time at the expense of only a slightly longer scan time.

Parallel imaging can be an important method of reducing RF deposition by decreasing the number of phase encoding steps that are performed (per unit time) in a given scan. The trade-off in SNR (a parallel imaging factor of 2 reduces SNR by 40%) can be balanced by the higher SNR of 3 T and by the further improvement in SNR provided by multielement coils now available (see Chapter 10).

Another technique for managing RF deposition is to interleave SAR-intensive sequences with low RF deposition scans—for example, to follow a long-echo train, fat-suppressed FSE scan with a 2D gradient-echo acquisition before starting the next FSE scan. This technique has limited utility for most current applications, but is of some value, particularly for body imaging on early generation 3 T systems.

Innovative methods of reducing SAR without compromising imaging continue to be developed and applied. New short-bore magnet designs can be, when correctly designed, more SAR-efficient than earlier generation long-bore systems, as less of the body is exposed to the shorter body coil transmitter. Innovations in RF chain technology (and, in particular, in the design of the RF transmitter) have also improved the efficiency of energy deposition with a resultant substantial net reduction in SAR. Advances in pulse-sequence design, such as reshaping RF and gradient waveforms (variable rate selective excitation [VERSE]), can reduce peak RF power up to 60% compared with conventional techniques. Newer imaging techniques such as SPACE (see Chapter 63) have also been introduced that lead to substantially lower SAR deposition, due to extensive use of low flip angle refocusing pulses, in an elegant manner, without the penalty of SNR loss.
Section III

Basic Image Acquisition
MRI offers multiple variables that may be assessed to obtain distinct tissue contrasts based on T1/T2 relaxation times, proton density (PD), magnetization transfer (Chapter 38), susceptibility (Chapter 64), fat (hepatic fat quantification, Chapter 96), incoherent motion diffusion (Chapter 66), coherent macroscopic motion (blood flow, Chapter 63), blood oxygen level (Chapter 69), proton spectroscopy (Chapter 71), and temperature, amongst others.

The tissue contrast mechanisms discussed in this chapter include PD, T1 relaxation time, and T2 relaxation time. Images are usually acquired for which the contrast is weighted more toward one of these parameters. The key word here is "weighted." Tissue contrast in the image has contributions from each of the various intrinsic contrast mechanisms, but is "weighted" more toward one than the others. In this context, weighting simply means the amount of contribution made to the image contrast associated with the difference between tissues on the basis of the parameter of interest (for example, PD, T1, or T2). This weighting is accomplished by the selection of the timing parameters of the pulse sequence (set prior to scan acquisition). For spin echo sequences, these are the TR (repetition time) and the TE (echo time).

TR primarily controls the amount of T1-weighting, whereas TE primarily controls the amount of T2-weighting. If one wishes to obtain images in which the contrast is weighted more toward T1 (using spin echo or fast spin echo sequences), then a relatively short TR is selected. There is no exact “best” TR, but rather a range to produce T1-weighted images. The range depends on the tissues being imaged as well as the field strength of the MR system. T1 relaxation times lengthen (increase) as field strength increases. At 1.5 T, when acquiring 2D SE T1-weighted images of the brain, the TR is usually between 400 and 550 msec. Raising the TR will not make the image more T2-weighted, but rather simply reduce the T1-weighting.

A word of caution is warranted, given the previous statements. If scan techniques other than spin echo or fast spin echo are used to produce T1- (or T2-) weighting, the choice of TR and TE will be quite different. For example, the T1-weighted image depicted in Fig. 31.1a was performed at 3 T using a gradient echo technique, with TR = 250 msec in this instance. The remainder of the current discussion is specifically in reference to spin echo and fast spin echo techniques.

As previously mentioned, TE primarily controls the amount of T2-weighting in an MR image. If one desires a T1-weighted image, a relatively short TE is selected. Often, one selects the shortest TE possible. For spin echo images, short is 25 msec or less. If one desires a T2-weighted image, then the TR is increased to reduce the amount of T1-weighting (usually 2500 msec or higher), and a long TE is selected. For spin echo, this is usually 80 to 120 msec. Fig. 31.1b is an example of a T2-weighted FSE image (long TR/long TE).

To obtain PD-weighted images (Fig. 31.1c), one increases the TR to reduce the T1-weighting (again to ≥ 2500 msec) and reduces the TE (to 25 msec or less) to reduce the T2-weighting. Although one may choose to acquire PD-weighted images, in clinical practice T2-weighted fluid-attenuated inversion recovery (FLAIR) sequences (Fig. 31.1d) have supplanted PD-weighted scans for imaging of the brain. Indeed, there are few
remaining clinical applications, outside the musculoskeletal system, for PD-weighted imaging today.

The images presented in Fig. 31.1 also illustrate the general applicability of the different parameter weightings discussed. T1-weighted scans (Fig. 31.1a) find application for depiction of detailed anatomy and detection of contrast enhancement, the latter using a gadolinium chelate (the class of IV contrast agents used in MR). Fat is high signal intensity, CSF low signal intensity, and white matter is higher signal intensity than gray matter. Depicted well in Fig. 31.1a is a metastatic lesion (arrow) from lung carcinoma, due to enhancement on this post-contrast image. Heavily T2-weighted images, such as that illustrated in Fig. 31.1b, find application for visualization of fluid (such as CSF) and to a lesser extent edema. Anatomic detail may also be high, depending on the specific technique used. CSF is high signal intensity and the gray–white matter signal intensity ratio is reversed as compared with T1-weighted images. PD-weighted images, illustrated in Fig. 31.1c, are no longer used in the brain due to poor contrast between tissues. FLAIR, illustrated in Fig. 31.1d, depicts extremely well vasogenic edema within the brain (with high signal intensity, arrow), being specifically a T2-weighted scan with CSF suppression.

Fig. 31.1 Enhanced T1-, T2-, PD weighted, and FLAIR images of a brain metastasis.
The classification of lesions based on their hypointense or hyperintense appearance is hampered by the influence of field strength and pulse sequence parameters. An alternative approach (but not commonly used today, clinically) is to calculate tissue relaxation times, thus providing a quantitative means of lesion characterization. Inversion recovery (IR) imaging techniques are considered to be the most accurate for calculation of T1 relaxation times. A simple means of estimating T1 can be achieved by acquiring two images with different inversion times, in this example, 350 and 550 msec (Fig. 32.1). The T1 value is then calculated based on these two points measured along the recovery path of the longitudinal magnetization. The resultant "calculated" image is displayed at the bottom of Fig. 32.1. In this image, the value of each pixel corresponds to the T1 of the respective tissue (a quantitative measure), as opposed to signal intensity (a qualitative or relative measure) as with the majority of MR images.

An alternative approach to the use of inversion recovery imaging is the acquisition of spin echo images with different T1-weighting (different repetition times, 550 and 950 msec in this instance). Based on the difference in signal intensity between the two acquisitions, the T1 value of the tissue can be estimated. It is important, however, that the change in signal intensity as a function of TR is well above the noise level of the image; otherwise, the calculated T1 image will be very noisy (as illustrated in Fig. 32.2).

T2 relaxation times can be estimated from a single multi-echo spin echo measurement, where the echo time–dependent signal decay follows the T2 relaxation time of the tissue. In Fig. 32.3, three images with different echo times (TE) were acquired in a single multiecho scan. The
change in signal intensity as a function of echo time is then used to generate a T2 pixel map, the final image shown in Fig. 32.3.
Sagittal and axial T1-weighted spin echo images, acquired at 1.5 T, are shown prior to (Fig. 33.1a,b) and an axial image following (Fig. 33.1c) intravenous gadolinium chelate administration. The scans demonstrate a heterogeneously enhancing hypothalamic mass, by stereotactic biopsy a grade II astrocytoma.

In an MR pulse sequence, either a gradient magnetic field or an RF pulse can be employed to form (refocus) the observed signal (the echo). A sequence that uses a gradient to refocus the echo is referred to as a “gradient echo” pulse sequence (see Chapter 40). If there is an RF pulse prior to the echo (classically a 180° pulse), then the pulse sequence is referred to as a "spin echo" sequence. Spin echo (SE) technique was widely used historically. However, today, its application in MR is limited, largely due to the

Fig. 33.1 Hypothalamic astrocytoma on sagittal pre- and axial pre- and post-contrast spin echo T1-weighted images.
emergence of fast or turbo spin echo technique. SE technique is still employed at 1.5 T in some instances for T1-weighted imaging in the brain, as illustrated in Fig. 33.1.

In a SE pulse sequence, the 90° RF pulse (the first pulse applied) produces transverse magnetization (tipping the net vector from parallel to the main magnetic field into the transverse plane). This induces a signal in the receiver coil known as the free induction decay (FID). A 180° RF pulse is then applied and the echo formed at the time TE (the time between the initial 90° pulse and the echo). The 180° RF pulse also corrects for dephasing effects from field and local inhomogeneities (T2\* effects), as well as for phase effects that can occur (and lead to loss of signal) when fat and water both occupy a single voxel.

There are two operator-selectable timing parameters, which can be varied to control the contrast of the image when using a SE sequence (Fig. 33.2). These parameters are TR (repetition time) and TE (echo time). In general, the TR determines the T1-weighting and the TE determines the T2-weighting. As discussed in Chapter 31, the use of a relatively short TR (i.e., 500 msec or less) and a short TE (i.e., 25 msec or less) produces images in which the tissue contrast is primarily related to differences in T1 relaxation times. For example, in Fig. 33.1, TR/TE = 435/14 for the sagittal and 500/12 for the axial acquisition. Tissues with short T1 relaxation times appear bright on T1-weighted images. Gadolinium, employed as an intravenous contrast agent in the form of a stable chelate, is a paramagnetic metal. When in close proximity to a water molecule, the paramagnetic effect shortens the T1 of the water protons resulting in high signal intensity on T1-weighted images (Fig. 33.1c).

Increasing the TR while maintaining a short TE produces images that are primarily proton density-weighted. Using a long TR (≥ 2000 msec) and a long TE (≥ 80 msec) produces images that are T2-weighted. Because increasing TR also increases scan time, proton density- and T2-weighted scans are now acquired using fast SE technique, as opposed to traditional SE technique (see Chapter 34).

Given that TR in a SE sequence is much longer than TE, SE scans are performed in a multislice fashion. During the time following the echo, other slices are excited (the timing for two such slices is illustrated in Fig. 33.2). The maximum number of slices one can acquire during a given pulse sequence is dependent primarily on the ratio TR/TE. Reducing the TR or increasing the TE reduces the number of slices one can acquire for a given SE pulse sequence.

Fig. 33.2 Pulse diagram for spin echo imaging, with acquisition of two slices.
Fast Spin Echo Imaging

The images illustrated in Fig. 34.1 were acquired using (a) conventional and (b) fast spin echo techniques. The use of fast or turbo spin echo (FSE, TSE) imaging has become routine in MRI today. A spin echo sequence employs a 180° RF pulse (typically) to create the echo, which also corrects for dephasing effects from slight field inhomogeneities and chemical shift. In a conventional spin echo sequence, a phase encoding gradient of defined amplitude is applied prior to the collection of the echo during readout. The amplitude of the phase encoding gradient determines the line in $k$-space that will be filled as the echo is sampled. In a conventional spin echo sequence, one line of $k$-space is filled during each repetition (TR period) of the pulse sequence. In an FSE sequence, a series of 180° pulses produces a train of echoes during a single TR period, as illustrated in Fig. 34.2.

The number of echoes produced in a single TR period is known as the echo train length (ETL). The phase encoding gradient amplitude will vary prior to each echo in the train so that each echo will fill a different line of $k$-space. In this way, multiple lines of $k$-space are filled during a single TR period. The number of lines filled in a single TR thus also corresponds to the ETL. As an example, using an ETL of 16, 16 lines of $k$-space will be filled during a single TR period. If a phase encoding matrix of 256 is selected, rather than requiring 256 repetitions of the pulse sequence to fill all the lines of $k$-space (assuming 1 for the number of signals averaged [NSA]), only 16 repetitions would be required ($256/16 = 16$). Increasing the ETL to 32 would require only eight repetitions.
repetitions to fill all 256 lines of k-space. The use of FSE sequences has not only greatly reduced the time required to obtain MR images with a long TR, but also allows the use of longer TR times for improved tissue contrast.

To demonstrate the power of FSE, consider the image in Fig. 34.1a. This was acquired using conventional spin echo technique with a TR of 3500 msec and a TE of 85 msec. The total scan time was 10 minutes, 51 seconds. The FSE sequence (Fig. 34.1b) was acquired using the same TR and TE but it had an ETL of 19. By filling 19 lines of k-space in each TR period, the scan time for the FSE images was only 35 seconds (10:51 divided by 19). The multiple 180° pulses also help reduce pulsation and flow artifacts, on the basis of reduced voxel dephasing and phase shifts due to the short interecho spacing. Note the higher, more uniform cerebrospinal fluid (CSF) signal intensity around the pons (white arrow) and improved depiction of flow voids in the basilar and internal carotid arteries in the FSE images, together with reduced ghosting (black arrows) from CSF, vessels, and the globes.

As previously mentioned, increasing the ETL reduces scan time; however, this is not without penalty. A long ETL reduces the number of slices that can be acquired in a single scan. Also, the longer the ETL, due to intrinsic T2 decay, the greater the edge blurring if a short effective TE is chosen and the greater the (artifactual) edge enhancement if a long effective TE is chosen. The blurring/edge enhancement can be minimized by the use of higher receiver bandwidths, which typically result in a shorter “readout” period and thus reduced time (spacing) between echoes (the critical factor involved). In addition, the multiple 180° RF pulses cause fat to remain high in signal intensity even with long echo times (due, in part, to rephasing of stimulated echoes). With a spin echo T2-weighted scan, fat will be of intermediate to low signal intensity, whereas with a matching fast spin echo T2-weighted scan, fat will be high signal intensity. This effect is strikingly demonstrated in Fig. 34.1, with both scalp fat and fat within the orbit having high signal intensity on the FSE image. Thus, fat suppression (see Chapter 46) is often employed with T2-weighted FSE imaging for the evaluation of soft tissue lesions in body imaging.
Although the reduction in measurement time using fast spin echo (FSE) technique is desirable (and plays a fundamental role in clinical imaging today), the very use of multiple closely spaced refocusing pulses is associated with higher RF power deposition. Thus, an FSE sequence may approach the acceptable limits of specific absorption rate (SAR) relative to patient safety. A common solution to this SAR problem is the use of a refocusing flip angle less than 180° (such as 120°, Fig. 35.1), leading to a marked reduction in SAR (which is proportional to the square of the flip angle) at the cost of signal-to-noise ratio (SNR). A refocusing pulse of less than 180° can be considered an insufficient flipping of the transverse magnetization. The induced MR signal is diminished, because it is proportional to the part of the magnetization projected onto the transverse plane. With a 180° pulse, the entire magnetization returns to lie within the transverse plane. Tilting the net magnetization back and forth with a low flip angle refocusing pulse leads to a so-called pseudo–steady state. Other approaches to decrease power deposition at 3 T with methods like variable flip angle imaging are not to be confused with the simple reduction of the refocusing flip angle in FSE imaging.

Fig. 35.1 illustrates T2-weighted FSE (TSE) scans in a patient with a large chronic left middle cerebral artery infarct acquired at 3 T with a flip angle of (a) 180° as opposed to (b) 120°. Although there is a reduction in SNR for white matter of 20% with the lower refocusing pulse, this is barely perceptible to the average radiologist. Due to SAR constraints, less than half the number of slices could be acquired when the 180° pulse was employed as opposed to the 120° pulse, using otherwise identical scan parameters. Coverage of the entire brain was thus not possible with the 180° pulse, illustrating the marked clinical applicability of a simple reduction in refocusing flip angle.
Driven-Equilibrium Fourier Transformation (DEFT)

Driven-equilibrium Fourier transformation or DEFT is an MRI technique that incorporates the addition of RF pulses at the end of an echo train to drive residual, transverse magnetization back to the longitudinal axis instead of waiting the time required for complete T1 relaxation. This technique is especially useful in situations where the TR for a given sequence would be set higher than what is required for the selected number of slices to wait for tissue relaxation.

Routine, fast or turbo spin echo (FSE) imaging incorporates the initial application of a 90° and 180° RF pulse to form an echo. Additional 180° RF pulses are used to generate subsequent echoes associated with different lines of k-space in a given TR period. However, additional time increasing (TR) is often required at the end of the echo train to allow for the recovery of magnetization of tissues with long relaxation times, such as cerebrospinal fluid (CSF) and synovial fluids, resulting in longer acquisition times. For example, in lumbar spine imaging, if TR is decreased to a value of 2000 msec or less, the time between excitations is not sufficient to allow the longitudinal magnetization of CSF to recover. This results in a partial saturation effect and a corresponding reduction in the overall signal intensity from CSF. Increasing the TR to 4000 msec will lead to a higher signal-to-noise ratio (SNR) for CSF due to reduced saturation effects. There will be, however, little gain in SNR for other tissues, despite scan time being doubled.

DEFT sequences use an additional 180° pulse to refocus the magnetization into one coherent vector in the transverse plane and then a 90° pulse is applied to drive the entire magnetization back to the longitudinal axis. The major benefit is that additional time (increasing TR) is not needed to wait for T1 tissue relaxation, and scan time is reduced due to the decreased TR.

In Fig. 36.1, T2-weighted FSE sagittal scans of the lumbar spine are presented (a) without and (b) with the addition of DEFT pulses. On the scan with DEFT, note the higher signal intensity of CSF (arrow). This patient has a grade I anterolisthesis of L4 on L5, leading to mild central canal stenosis. The DEFT technique, also called RESTORE, DRIVE, and FRFSE, has been implemented on 2D and 3D spin echo and fast spin echo sequence types with clinical applications in neurologic, orthopedic, and abdominal MRI. Its use, however, can decrease the conspicuity of cord lesions, including specifically multiple sclerosis.

Fig. 36.1 Use of DEFT, lumbar spine.
Reordering: Phase Encoding

With multiecho imaging (e.g., that employed with fast spin echo technique), where the signal contribution changes during data acquisition, the order in which each $k$-space line is acquired is an important factor in determining image contrast. Given that the low $k$-space lines contain the information about the coarse structure (and thus contrast) of the objects within the slice, the echo time at which those $k$-space lines are acquired is called the "effective" echo time.

The image shown in Fig. 37.1 was acquired with a long effective TE and thus heavy T2-weighting, the latter reflected by the high signal intensity CSF. The low $k$-space lines were acquired in this case at the 10th echo (of the 19

Fig. 37.1 Reordering of $k$-space line acquisition, effective TE = 96.
echoes acquired, representing the whole matrix), TE = 96. The echo spacing was 16 msec. Alternatively, acquiring the low k-space lines at the very beginning of the echo train, the resulting image is proton density–weighted (Fig. 37.2). Note that CSF is only slightly higher in signal intensity than brain. Fig. 37.2 was obtained using the identical scan sequence, except with the low k-space lines acquired at the first echo, TE = 16.

The scans demonstrate an early subacute unilateral pontine infarct (black arrow), which is better depicted in Fig. 37.1 due to the heavier T2-weighting. For confirmation, a diffusion-weighted scan was obtained (see Chapter 66) (Fig. 37.3), with the infarct very high signal intensity (white arrow) due to restricted diffusion.

Fig. 37.2 Reordering, effective TE = 16.

Fig. 37.3 DWI, early subacute pontine infarct.
As illustrated in Fig. 38.1, the magnetic moment of a single hydrogen nucleus, if aligned parallel to the magnetic field, superimposes its own intrinsic field on its neighbor. Depending on the orientation of the water molecule (which contains two hydrogen nuclei) within the main magnetic field, the field strength at that location will be increased or decreased. Because the number of possible orientations is large, there is also a large range of different resonance frequencies. The latter causes the sum of all magnetizations, the transverse magnetization, to dephase rapidly. The T2 relaxation time can be as short as 10 msec. Water molecules that are frozen in their orientation are not visible due to this very short T2. Ice cubes within the iced tea thus appear hypointense (Fig. 38.1). Due to the rapid tumbling motion of water molecules in the free (“visible”) water pool, the magnetic field superimposed by adjacent hydrogen nuclei is averaged out. The resonance frequencies are found in a small range, resulting in a long T2 relaxation time. An “invisible” or bound water pool also exists, due to hydrogen that is bound to complex macromolecules and thus experiences restricted motion. MRI is unable to directly image the bound pool due to its extremely short T2 values.

Placing a frequency selective saturation pulse below the resonance frequency of free water (Fig. 38.2) results in a portion of the bound water pool being saturated. Any transferred magnetization depends on the status of the magnetization within both pools. The bound pool has short T1 relaxation times due to the effective rigid lattice condition and a very short T2 relaxation time. The short T1 relaxation time of the bound pool indirectly influences the observed T1 relaxation time of the free and visible water pool. The magnetization transfer (MT) from the bound to the free pool results not only in a decrease in magnetization, but also in a decrease in the observed relaxation times. Magnetization transfer thus provides an additional contrast mechanism influencing the signal intensity of tissue above and beyond the contrast provided by the spin density, T1, and T2 differences between tissues. Clinically, MT has been used in MR angiography and for post-contrast T1-weighted imaging. In 3D time-of-flight (TOF) MR angiography, MT provides an improvement in vessel depiction, due to greater background tissue suppression. The use of MT in this instance also leads to (relative)
higher signal intensity from fat (e.g., in the orbits), which can degrade the study unless appropriately excluded by post-processing. On post-contrast scans with MT, there is an improvement in the conspicuity of enhancing lesions due to the (relative) reduction in signal of surrounding brain.

Post-contrast axial T1-weighted images are presented from a patient with adenocarcinoma of the lung metastatic to the brain, without (Fig. 38.3a) and with (Fig. 38.3b) MT. A single large enhancing metastasis is present (black arrow). The application of MT leads to lower signal intensity for normal brain, increasing the conspicuity of contrast enhancement, which is less affected. Note the loss of gray–white matter contrast on the image with MT, together with the greater conspicuity of normal vessels (white arrow). MT is little used today, due to these factors along with the poor overall SNR, for post-contrast imaging in the brain.

**Fig. 38.2** Applying MT in MR.

**Fig. 38.3** Post-contrast images without and with MT, brain metastasis.
Advances in gradient technology and the capability of RF systems on modern MR scanners have brought new life to sequences that decrease scan time and thus minimize the impact of patient motion. One such sequence is HASTE, an acronym for half acquisition single-shot turbo spin echo. This approach combines half-Fourier technique with fast spin echo imaging. With HASTE, each slice is acquired and often reconstructed before the next slice acquisition has begun (single shot, slice sequential). This is accomplished by acquiring an echo train equal to the required number of phase encoding steps for one slice. This differs from normal fast (turbo) spin echo in which phase encoding lines from multiple slices are acquired throughout the examination (multishot, multislice). The HASTE method also employs a technique known as half-Fourier in which the inherent conjugate symmetry of the raw (k-space) data are used to synthesize ~50% of the phase encoding steps for each slice. Although there is a corresponding reduction in the signal-to-noise ratio (SNR)—because the data for half of k-space is not sampled—spatial resolution is maintained. Images with HASTE are usually acquired in 2 seconds or less per slice, making HASTE very good for reducing image quality problems associated with patient motion.

Fig. 39.1 compares at 3 T (a) a breath-hold HASTE scan with (b) a fat-suppressed, free breathing, fast spin echo (FSE) technique using navigator-echo-based motion correction. The patient has a large hemangioma in the right lobe of the liver. Scan times were 2 seconds/slice versus 58 seconds for 10 slices (FSE). HASTE today plays a primary role in liver imaging at both 1.5 and 3 T.

Sampling bandwidth has a major impact on image quality with HASTE due to its effect on the spacing of the acquired echoes. If the sampling bandwidth is set too low, resulting in high echo spacing (meaning that there is greater time between each adjacent echo in the train of 128 or so echoes typically used to acquire the image), there can be substantial image blurring (Fig. 39.2a). Selecting a high bandwidth with lower echo spacing (Fig. 39.2b) produces images with slightly lower SNR (see Chapter 112), unless otherwise compensated (as in this case by doubling the number of averages). The end result is a much higher diagnostic quality image due to a reduction in image blurring. In this example, the difference in blurring is most evident within the posterior nasopharynx (white arrow, Fig. 39.2a). Illustrated is a small mass involving the tectum, causing obstruction of the cerebral aqueduct (black arrow, Fig. 39.2b). It is remarkable

Fig. 39.1 HASTE vs. FSE, T2-weighted scans, liver hemangioma.
that the scan times in this instance were only 7 and 14 seconds (per slice, with the images acquired at 3 T), despite the use of a 2-mm slice thickness for improved depiction of this anatomic region.

HASTE also finds clinical applicability, as illustrated in Fig. 39.3, for rapid brain imaging in uncooperative patients. An early subacute anterior and middle cerebral artery territory infarct is illustrated on HASTE (a) and diffusion-weighted (b) scans. This exam was acquired at 1.5 T, with the HASTE sequence requiring 1.2 seconds/slice, for a total scan time of 26 seconds with 22 slices. The conventional FSE T2-weighted scan would have required a scan time of 1.5 minutes, for comparison. For the HASTE scan, however, the effect of motion is limited to that occurring during each slice acquisition (1.2 sec), whereas for the FSE scan any motion during the entire scan time (90 sec) would degrade image quality.

Fig. 39.2 The impact of echo spacing with HASTE on image blurring.

Fig. 39.3 HASTE, for rapid brain imaging, with DWI for comparison.
This chapter and Chapter 41, Chapter 55, and Chapter 56 discuss four different basic types of gradient echo (GRE) sequences routinely applied in clinical practice. In these chapters, the evolution from a basic GRE sequence sampling data during free induction decay (FID) to more complex pulse sequences combining both gradient echo and spin echo features is explained. **Fig. 40.1** illustrates the simplest GRE sequence structure using just the FID for image acquisition. After the application of a single radio frequency (RF) pulse, with a flip angle ($\alpha$) typically less than 90°, resulting in only a fraction of the available longitudinal magnetization tipping into the transverse plane (1 in **Fig. 40.1**), a pair of bipolar readout gradient pulses are applied to sample the gradient echo. In distinction to spin echo sequences, a 180° refocusing pulse is not employed, making this sequence type more sensitive to susceptibility artifacts and static field inhomogeneities. A major benefit, however, is the possibility to decrease repetition time (TR). As a result of the short TR, however, the longitudinal magnetization cannot fully recover. After a few initial excitation pulses, there is an equilibrium or steady-state established between longitudinal magnetization and recovery. From this point on, the signal for all subsequent Fourier lines has the same magnitude, and acquisition of data begins. Prior to steady state being achieved, fluctuations in tissue magnetization can occur that would produce variations in the signal intensity and hence artifacts. The data are sampled during a gradient echo, which is achieved by dephasing the spins with a negatively pulsed readout gradient before they are rephased by another readout gradient with opposite polarity to generate the echo (2 in **Fig. 40.1**).

In short-TR GRE imaging, phase coherences may build up in transverse magnetization. To generate T1 contrast, these coherences should be eliminated because they accentuate signal from long T2* components. After data acquisition and before the next excitation, a spoiler gradient is applied to destroy all residual coherent transverse magnetization (3 in **Fig. 40.1**). To make this more effective, a method called RF spoiling may be added to disperse residual magnetization by randomizing the phases of the subsequent RF pulses. The longitudinal magnetization will further recover (4 in **Fig. 40.1**).
until the next low flip angle excitation. Siemens terms this general type of sequence fast low-angle shot (FLASH), General Electric terms it spoiled gradient-recalled acquisition in the steady state (SPGR), and Philips terms it T1 fast field echo (T1-FFE).

*Fig. 40.1* also illustrates the change in signal intensity as a function of excitation angle. The maximum signal occurs for an angle called the Ernst angle; however, this is not necessarily the best choice for flip angle. T1-weighted (short TR and TE, large α), as illustrated (*Fig. 40.2a*), as well as T2*-weighted (long TR and TE, very small α) contrast can be achieved. For proton density-weighted images, an angle smaller than the Ernst angle is desirable. Larger flip angles than the Ernst angle and short TR produce T1-weighted images with better tissue contrast but lower total signal. However, for lesion detection, CNR is typically more important than SNR.

Gradient echoes are more sensitive to field inhomogeneities, but have low RF energy deposition (specific absorption rate or SAR), and both 2D and 3D acquisitions are feasible (the latter in part due to the short TR employed). This technique can be adapted at 3 T to produce high-quality T1-weighted brain images (2D) as shown in *Fig. 40.2a*. A derivative of the spoiled GRE technique is used in VIBE (Volume Interpolated Breath-hold Examination, see Chapter 65). *Fig. 40.2* demonstrates axial T1-weighted spoiled GRE scans in a healthy volunteer; (a) was acquired with a flip angle of 70° and (b) with a flip angle of 20°, while all other parameters remained unchanged. Note the markedly improved T1 contrast and differentiation between gray and white matter in the brain in (a), due to the application of a larger flip angle.

![Fig. 40.2](image)

*Fig. 40.2* The effect of flip angle, 70° vs. 20°, spoiled GRE technique.
Another way to utilize gradient echo technique (GRE) is shown in Fig. 41.1. A refocused, rewound, or coherent gradient echo sequence incorporates free induction decay (FID) and spin echo attributes. Excitation, phase encoding, and readout for a refocused GRE sequence are identical to that for a spoiled GRE sequence, as discussed in Chapter 40. However, rather than destroy or spoil the transverse magnetization, the objective with this approach is for the transverse magnetization to remain transverse and (partially) coherent across two TR intervals, creating a spin echo (SE) from the two successive RF pulses that contributes to the signal ultimately acquired and encoded. To accomplish this, a phase encoding gradient of the opposite sign to the first phase encoding gradient is added after readout (Fig. 41.1). Because most of the SE signal will come from tissues with a long T2, the contrast will be different than with spoiled GRE technique where no SE signal is generated. Thus, this technique creates a mixed contrast of T1- and T2-weighting, whereas the spoiled GRE technique creates tissue contrast that is primarily T1-weighted.

The improvement in contrast from spoiled to refocused GRE manifests only under the conditions of short TR and large flip angle, which ensures that adequate transverse magnetization is generated. Siemens terms this approach ‘fast imaging with steady precession’ (FISP), General Electric uses the term ‘gradient-recalled acquisition in the steady state’ (GRASS), and Philips uses the term ‘fast field echo’ (FFE). As illustrated in the plot of signal response as a function of flip angle (Fig. 41.1), tissues with a long T2* relaxation time (e.g., synovial fluid and CSF) demonstrate an enhanced signal compared with muscle and compared with a spoiled GRE sequence.

Refocused sequences are excellent for high contrast between fluid and solid structures in rapid imaging and thus can be used in myelographic and arthrographic applications. Such sequences, however, are sensitive to motion and flow, which cause phase...
shifts between the spin echo and FID portions of the signal, leading to increased artifacts compared with spoiled GRE techniques (which only use the FID portion of the signal). Although refocused sequences could be used for time-of-flight MR angiography, spoiled GRE techniques are preferred in this application. Refocused sequences lead to hyperacute and late subacute hemorrhage, specifically any fluid with a long T2, being visualized as high signal intensity, and thus potentially confused with flow. Spoiled GRE techniques, however, also visualize methemoglobin as high signal intensity, but on the basis of the short T1.

Fig. 41.2 demonstrates axial refocused GRE sequences in the same individual as shown in Chapter 40’s Fig. 40.2. The sequence displayed in (a) has been acquired with a TR of 10 msec and a flip angle of 80° whereas, for the sequence in (b), a TR of 20 msec and a flip angle of 60° were used. Note that in (a) CSF is hyperintense whereas in (b) CSF is hypointense. Only slight changes in TR and flip angle result in marked differences in tissue contrast with this type of sequence. The sensitivity of refocused GRE technique to motion is also illustrated. Note the artifactual loss of signal intensity (arrow) within the frontal horns of the lateral ventricles in Fig. 41.2a, due to the CSF jets into the ventricles from the foramina of Monro.
Echo Planar Imaging

Echo planar imaging (EPI) is one of the fastest techniques for acquiring MRI data. This method incorporates rapid changes in readout gradient polarity and amplitude to refocus the signal of a single spin excitation, producing the required echoes for an entire image. The acquisition time for one slice is one TR period, which lasts as little as 40 msec, and the process is repeated for the number of slices desired.

Echo planar imaging techniques are defined by the spin preparation method used. Gradient echo (GRE-EPI) consists of a single RF excitation creating a free induction decay (FID)-based image. A dual RF pulse train generates an RF echo, which is the basis for spin echo (SE-EPI) images. Inversion pulses (IR-EPI) can also be applied to obtain fluid-attenuated inversion recovery (FLAIR)-like echo planar images. Fig. 42.1 illustrates (a) FLAIR and (b) diffusion-weighted scans both acquired with EPI technique in a patient with an early subacute middle cerebral artery territory infarction (arrow).

Collecting enough information for a complete slice in one sampling period (single-shot EPI) requires that the MR hardware be capable of reaching the peak gradient amplitude in a short period of time (the latter is defined as the rise time). The peak amplitude divided by the rise time is a gradient performance measurement known as the slew rate. Higher slew rates allow EPI information to be collected in shorter time intervals, leading to higher quality images with less distortion caused by increased echo spacing. Spectral fat saturation is typically used to reduce the presence of chemical shift artifact induced by the extreme field sensitivity of EPI scans. In addition, a reference scan or other techniques are used to reduce the severity of ghosts due to systematic errors between the odd and even echoes generated during the readout portion of the sequence. This artifact, often referred to as N/2 or Nyquist ghosting, would otherwise obscure the water image.

Axial images are presented in Fig. 42.2, comparing conventional and EPI imaging at 1.5 T, from a patient with an anaplastic astrocytoma. Due to the size of the tumor and associated edema, there is substantial mass effect upon the brainstem. Fast spin echo T2- (Fig. 42.2a), FLAIR (Fig. 42.2c), and spin echo T1-weighted post-contrast (Fig. 42.2e)
images are compared with EPI acquisitions (Fig. 42.2b,d,f) at the same slice position with similar weightings. In each case, the EPI examples showed a reduction in scan time of 50% or greater while maintaining overall image contrast. Another major difference is that the spin echo scans are multislice in technique (with motion at any time during the scan acquisition affecting all slices), whereas single-shot EPI images are single slice and thus very robust in regard to patient motion. Note the ghosting (arrow) due to inadvertent patient motion on the post-contrast T1-weighted spin echo scan (Fig. 42.2e), which is absent on the corresponding EPI scan (Fig. 42.2f). EPI finds clinical applicability in diffusion, perfusion, and functional neurologic imaging and, with single-shot technique, can be employed for reducing artifacts associated with patient motion.

Fig. 42.2 Comparison of SE and EPI technique with various image contrasts.
Adipose tissue has a very short T1 relaxation time (~260 msec at 1.5 T), a fact that is utilized in short tau inversion recovery (STIR, also known as short T1 inversion recovery) imaging to eliminate the signal from fat. As illustrated in Fig. 43.1, the sequence starts with an inversion pulse. The relatively short inversion time (called short tau), in this case 150 msec, is chosen such that the 90° pulse is applied at a time at which the longitudinal magnetization of adipose tissue is zero and therefore no transverse magnetization is generated. The graph presented in Fig. 43.1 shows the evolution of the longitudinal magnetization following the inversion pulse. The dashed lines in the graph illustrate the theoretical evolution of the longitudinal magnetization with time. The evolution is illustrated for both phase-sensitive and magnitude reconstruction. For the former, the scale of image signal intensity ranges from negative to positive. For the
latter, magnitude reconstruction, values less than zero are reflected about the x-axis, and so values range from zero to positive.

As depicted in Fig. 43.1, for an inversion time (TI) of 150 msec at 1.5 T, the signal of fatty tissue is nulled. In addition, all tissues with short relaxation times will appear darker (e.g., muscle), if phase-sensitive image reconstruction is used. The contrast of other tissues relative to fluid-filled cavities can be significantly increased by taking advantage of the long T2 relaxation time and selecting a relatively long echo time, as indicated in the lower graph. In regard to the specific choice of imaging parameters, the inversion time used in spin echo imaging to suppress tissue with a relaxation time of 250 to 260 msec (fat) is between 170 and 180 msec. In fast spin echo imaging, a shorter inversion time has to be selected (~150 msec).

Fig. 43.2 presents portions of the MR exam of a young woman with a pelvic dermoid cyst, imaged in the (a) axial plane with a T1-weighted spin echo scan and in the (b) axial and (c) coronal planes with STIR. In regard to the latter two scans, a fast spin echo pulse sequence was employed, using a TE of 70 msec, with the 90° pulse preceded by an inversion pulse and the inversion time chosen to null the signal from fat. On the STIR images, areas with high water content (fluid) appear markedly hyperintense, with fat very low signal intensity, a finding particularly striking in comparison to the T1-weighted scan. Note the fluid-fluid level within the cyst (arrow), the solid nodular component projecting from the medial wall, and the suppression on STIR of the signal intensity from the sebaceous (oily/fatty) fluid contents of the cyst. STIR saw a resurgence in use (as compared with fat-suppressed T2-weighted fast spin echo scans) with the introduction of large, short-bore 1.5 T systems, where the extremely short bore can limit the homogeneity of spectral fat suppression at the ends of the magnet.
Inversion Recovery: Part 2

With the use of fast spin echo imaging and the resulting savings in scan time, inversion times (TI) on the order of 2.5 seconds (and TRs on the order of 10 seconds) for the suppression of the signal from cerebrospinal fluid (CSF) are possible. This approach is typically performed with magnitude reconstruction, as illustrated. Imaging starts at the time at which the longitudinal magnetization of CSF is zero (Fig. 44.1). In the resultant image (Fig. 44.2), CSF appears black, with little or no signal (arrows). Fig. 44.2 demonstrates, with good conspicuity, a high signal intensity lesion anterior to the brainstem, corresponding to a grade II (fibrillary) astrocytoma. Fluid-attenuated inversion recovery (FLAIR), as this technique is known, is actually a fast spin echo inversion recovery acquisition with a long inversion time (TR/TI/TE = 8440/2500/136 msec in this instance), which also employs magnitude reconstruction.

“True” inversion recovery acquisitions take into account the “sign” of the longitudinal magnetization (Fig. 44.3). This is also referred to in the literature as inversion recovery with phase-sensitive reconstruction, as opposed to magnitude reconstruction. In such images, “zero” is displayed as intermediate gray, “negative” signal appears hypointense, and “positive” signal is hyperintense. This technique is especially valuable in the study of brain maturation, due to the high contrast between gray and white matter. Most brain tumors, such as that illustrated, have a long T1 and long T2, and thus appear on a “true” inversion recovery scan, specifically with phase sensitive reconstruction (with TR/TI/TE = 4770/400/63 msec in this instance), as very low signal intensity (Fig. 44.4, black arrow, same patient and level as Fig. 44.2). For

Fig. 44.1 Pulse diagram and tissue contrast evolution for CSF suppressed IR.
phase-sensitive reconstruction, the scale for signal intensity extends from negative to positive (e.g., -4096 to +4096, with the maximum value vendor specific). For the majority of clinical MR scans acquired today, however, as well as with magnitude reconstructed inversion recovery images, the scale extends from zero to positive.
Fluid-attenuated inversion recovery with fat saturation (FLAIR FS) takes advantage of the specific (long) relaxation time for pure fluid (cerebrospinal fluid [CSF]) and the shift in resonance frequency for adipose tissue to suppress the signal from both. The pulse diagram for this scan technique is illustrated at the top of **Fig. 45.1**. The graph below the sequence depicts the temporal evolution, during the course of RF pulses and gradient manipulations, of longitudinal magnetization for adipose tissue, brain tissue, and CSF. The inversion time is selected by positioning the 90° RF excitation pulse so that there is no longitudinal magnetization within pure fluid to be converted to transverse magnetization. This is the basic operating principle for FLAIR, resulting in little to no observed signal from CSF. Prior to RF excitation, a frequency selective saturation pulse is issued to saturate any magnetization within adipose tissue. The resonance frequency for adipose tissue is ~3.5 ppm below that of free water, ~221 Hz on a 1.5 T system.
The graph at the bottom of Fig. 45.1 shows the frequency range for carbon-bound hydrogen (fat) and oxygen-bound hydrogen (water) to illustrate the effect of a frequency selective saturation pulse. Also illustrated in Fig. 45.1 is an axial FLAIR FS image, depicting subarachnoid hemorrhage within the sylvian fissure on the right (arrow). FLAIR - with FS - is one of the most important scan types in routine brain imaging, with high sensitivity to abnormalities both within the brain and adjacent CSF.

Fig. 45.2 presents selected images from the FLAIR scan of a patient with multiple sclerosis (MS). On the axial scans (a,b), multiple punctate periventricular high signal intensity lesions are noted, characteristic for MS, while sagittal FLAIR scans (c,d) illustrate additional, distinctive ovoid perivenular (black arrows) and callosal (white arrows) lesions. FLAIR, by suppression of CSF, improves markedly the conspicuity of lesions that are adjacent to CSF, in this instance the ventricular system. Note also the low signal intensity of scalp fat, due to the application of fat saturation.

![Fig. 45.2](image-url)
Suppressing the signal from fat in MR imaging is often very useful for improved lesion detection and delineation, as well as lesion characterization. The most common technique is spectral fat saturation (often referred to as “fat sat”). Axial post-contrast T1-weighted images of the lumbar spine at 1.5 T are presented without (Fig. 46.1a) and with (Fig. 46.1b) spectral fat saturation. Depicted is a primary bone tumor (an osteoblastoma) involving the posterior elements of the L5 vertebra. By suppression of the fat signal, abnormal contrast enhancement of the lesion, extending into the posterior paraspinal soft tissues, is much more evident. Fat saturation is commonly employed for post-contrast imaging of the soft tissues of the head and neck, spine, abdomen, and pelvis. Applications for fat saturation on pre-contrast scans are also common, notably in T2-weighted imaging of the breast and spine (the latter for improved visualization of vertebral body lesions and disk hydration) and for T1-weighted imaging of the pancreas.

In vivo, water and fat protons resonate at slightly different frequencies in a magnetic field. This frequency separation increases with field strength. The difference at 1.5 T is ~220 Hz, as shown in Fig. 46.1c (bottom). A specially designed RF pulse, Fig. 46.1c (top), is applied prior to the spin preparation excitation at the specific resonance frequency of fat, which saturates the spins at this frequency. The fat tissue within the field of view remains saturated during the spin excitation and therefore does not contribute to the resulting echo and image formation. A negative consequence of the use of spectral fat saturation is mild loss in SNR even in anatomic areas without any fat, due to how close the water and fat peaks are and constraints in how truly selective in frequency the fat saturation pulse can be made. The addition of fat saturation to a pulse sequence also reduces the maximum number of slices that can be acquired within a given TR.

![Fig. 46.1 Osteoblastoma, spectral fat saturation, specifics of the RF pulse.](image)
The presence of ferromagnetic objects, differences in magnetic susceptibility between tissues, and large variations in tissue shape (as encountered for example in the neck and chest) all can make it difficult to achieve the degree of magnetic field uniformity required for spectral fat saturation. Each factor can lead to a change in the specific resonance frequency of fat in localized areas. Advanced MR systems are equipped with special hardware, called electronic shims, which produce small changes in the spatial magnetic field to optimize the homogeneity within the field of view. The result is greater uniformity of the magnetic field with less chance for incomplete or inconsistent fat saturation. Regardless, technologists should make an extra effort to assure that all metal objects, including buttons and jewelry, are removed prior to the exam to improve the final spectral fat saturation.

Spectral fat saturation can also be applied to confirm the presence of fat within a lesion. This is useful for the differentiation of extracellular methemoglobin and fat (both demonstrate high signal intensity on T1-weighted scans), for example with intracranial and spinal lipomas. Fig. 46.2 presents axial T1-weighted images through the superior portion of the lateral ventricles at 3 T without and with spectral fat saturation. With fat saturation, signal from the two small round ventricular lesions is suppressed, confirming their fatty composition. This facilitates the diagnosis of a ruptured dermoid (with the presence of small fat globules or “droplets” within the cerebrospinal fluid space characteristic).

Spectral adiabatic inversion-recovery (SPAIR) is a hybrid spectral and inversion recovery technique. An adiabatic spectrally selective inversion pulse is used to invert the fat spins, followed by a spoiler pulse to dephase any transverse magnetization. The acquisition then begins (at the time TI) when the longitudinal magnetization of fat spins is zero. Advantages include that the technique is insensitive to $B_1$ inhomogeneities and that only fat spins are suppressed.

Fig. 46.2 Ruptured dermoid, characterization by use of spectral fat saturation.
As outlined previously, the different electron environment of water-bound and fat-bound hydrogen atoms (protons) means that their respective magnetic moments process at slightly different resonance frequencies. Specifically, the magnetic moments of hydrogen nuclei in adipose tissue have a resonance frequency that is ~3.5 ppm lower than that within water-containing tissue (which equates to ~220 Hz at 1.5 T and ~440 Hz at 3 T). A frequency selective saturation pulse can thus be used to suppress the signal from adipose tissue. The quality of that saturation is a function of the overall magnetic field homogeneity within the imaging volume. In addition, the fat-saturating RF pulse is very close to the water resonance, leading to an overall loss in signal-to-noise ratio (SNR) at lower field strengths (where the separation is less), for example 1 T.

As an alternative to fat saturation, it is possible to simply excite either water or fat. In order to be successfully implemented clinically, there must be excellent magnetic field homogeneity within the volume of interest. The fundamental advantage of water excitation (over spectral fat saturation) is that the effectiveness of the fat suppression is basically immune to RF ($B_1$) spatial inhomogeneities. This is a very important point at 3 T where RF inhomogeneities can at times be relatively severe.

Water (or fat) excitation is typically accomplished using binomial pulses (1–1, 1–2–1, or 1–3–3–1). Fig. 47.1 illustrates the use of a slice-selective 1–2–1 binomial pulse for water excitation. To achieve a 90° excitation for water, there is an initial 22.5° excitation pulse. A waiting period allows the slower rotating transverse magnetization within adipose tissue to develop a phase difference: for example, 180° with respect to the phase position of the transverse magnetization within water. The 45° excitation angle then tips the magnetization within water to 67.5° with respect to the longitudinal direction, whereas the magnetization within adipose tissue is flipped to 22.5° in the same direction. After a second waiting period, a further 22.5° excitation pulse completes the 90° excitation for water, with the magnetization of fat restored to the longitudinal position and thus not contributing to the MR signal. The net result is a “water excitation” RF pulse.

Using the same approach, but with a phase shift (of the RF pulse), a “fat
excitation” pulse can be created if desired. By exciting selectively either fat or water, only the excited tissue produces an MR signal in the final image. Nonselective RF pulses can be used if the organ is small enough or the transmit coil excites only a small region, allowing phase encoding in two orthogonal directions within an acceptable measurement time (3D acquisition). Binomial pulses that are slice or slab selective are referred to as spectral spatial pulses.

Coronal images of the knee at 3 T are presented in Fig. 47.2 from a patient with mild cartilage thinning in the medial compartment, reflecting mild degenerative disease. Fig. 47.2a was acquired using a T2-weighted fast spin echo sequence with a spectral fat saturation pulse. Fig. 47.2b was acquired with the same pulse sequence (and in the same scan time), but using a composite pulse for water excitation rather than spectral fat saturation. The image acquired with water excitation has improved uniformity of fat suppression, due to the insensitivity of this technique to RF inhomogeneities. Note the nonuniformity of fat saturation on the scan with spectral fat saturation in the cranio-caudal direction, with signal from the subcutaneous fat of the calf not suppressed to the degree of that from the thigh. In this image, susceptibility artifacts are also seen at the interface of fat and air (arrows). Water excitation performs well in the setting of high $B_0$ homogeneity and exact frequency settings (such as achieved by shimming and frequency adjustments for each slice in a large stack), resulting in good fat suppression for all slices, and is independent of type of acquisition, whether FSE, GRE or EPI.

Fig. 47.2 Coronal knee, 3 T, T2 FSE with spectral FS vs. water excitation.
In inversion recovery (IR) imaging, suppression of the signal from adipose tissue can be achieved by exploiting the short T1-relaxation time of fat (see Chapter 43). An IR sequence starts with a 180° pulse, which tips the longitudinal magnetization antiparallel to the main magnetic field direction. Immediately after this inversion, the longitudinal magnetization begins to relax back to the equilibrium orientation, parallel to the main magnetic field. The time between the center of the inversion pulse and the center of the excitation pulse of the imaging sequence is called the inversion time (TI). The longitudinal magnetization of fat, as it recovers following the 180° pulse, crosses from negative to positive, passing through zero. If the excitation pulse of the imaging sequence is applied at that point in time (as in STIR), little or no transverse magnetization is generated and little MR signal is detected from fat (Fig. 48.1).

STIR imaging is usually performed in conjunction with a fast spin echo sequence, to take advantage of the shorter measurement time possible with fast spin echo imaging.
When combined with longer repetition times and higher matrix sizes, better contrast and spatial resolution is obtained in comparison with a conventional spin echo sequence. STIR is an inversion recovery technique that generally produces magnitude images that provide no information about the sign of the longitudinal magnetization. Consequently, tissues with very short relaxation times may appear bright, similar to tissues with very long relaxation times. As indicated in Fig. 48.1, the signal from fluid is proportional to the available longitudinal magnetization and is significantly higher than the signal from muscle. Adipose tissue has no signal contribution, provided that the inversion time and amplitude of the inversion pulse have been adjusted appropriately. STIR images have intrinsically low SNR, because most of the longitudinal magnetization is still oriented antiparallel to the main magnetic field and is far from being fully relaxed. They are also more prone to motion artifacts when compared with T2-weighted FSE images.

STIR should be used with caution in conjunction with T1-shortening contrast agents (e.g., the gadolinium chelates). Contrast uptake in a lesion may paradoxically lead to lower, as opposed to higher, signal in STIR images. Caution is also advised more generally in clinical interpretation of STIR images, because of the dependency of tissue signal intensity on both T1 and T2, as opposed to a single parameter.

STIR is very sensitive for fluid, whether within soft tissue or bone, around joints, or along tendon sheaths (including specifically soft tissue and marrow edema). Note in Fig. 48.2 the excellent depiction of edema within the L1, 3, and 5 vertebral bodies (arrows) in this elderly woman with multiple acute compression fractures on the STIR scan (b), which cannot be diagnosed on the basis of the non-fat-suppressed T2-weighted FSE scan (a). Because fat suppression with STIR does not depend on local field homogeneity, it serves as an alternative to spectral fat saturation in intrinsically inhomogeneous body regions (e.g., the orbits) or at the edge of the usable homogenous main magnetic field.

Fig. 48.2 Acute benign vertebral body compression fractures, utility of STIR.
Fat Suppression: Phase Cycling

The two hydrogen-based longitudinal magnetizations used in MRI have their origins in adipose- (fat) and water-containing tissue. Due to the different electronic environment in adipose tissue, the resonance frequency of that magnetization is ~3.5 ppm lower than that of water (~ –220 Hz on a 1.5 T system, as previously noted). With time, after the initial excitation (Fig. 49.1), the transverse magnetization within adipose tissue falls behind the transverse magnetization within water. The time of the gradient echo—specifically, when the MR signal is observed—can be chosen so that the transverse magnetizations from fat and water are either opposed-phase or in-phase. This technique may also be used to generate fat-suppressed and water-suppressed images (the Dixon method) that do not suffer from field inhomogeneity effects.

The duration of the acquisition window in a pulse sequence is inversely proportional to the bandwidth of the sequence. If the bandwidth is large enough (and thus the acquisition window very short), it is possible to acquire opposed-phase and in-phase images simultaneously with a double echo gradient echo sequence (Fig. 49.1). This approach, as originally performed, was employed with TE, TR, and flip angle chosen to provide T1-weighting. Thus, voxels containing fat are high signal intensity and those containing water are low signal intensity. However, in voxels in which there is both fat and water, on opposed-phase images, there is a cancellation (loss) of signal due to the transverse magnetization from fat and water being of opposite phase. This leads, for example, to signal loss at the interface between fat- and water-containing structures (as depicted in Fig. 49.1 at the interface of the liver and spleen with adjacent intraabdominal fat). Note that fat per se is not suppressed (the subcutaneous and intra-abdominal fat remain high signal intensity in Fig. 49.1), but rather only when voxels contain both fat and water.

Fig. 49.1 Opposed- and in-phase imaging, pulse diagram and corresponding scans.
Illustrated in Fig. 49.2 is a nonhyperfunctioning adrenal adenoma, on (a) in-phase and (b) opposed-phase images. A round, sharply demarcated, homogeneous lesion of the left adrenal gland is noted (arrow). On the in-phase image, the lesion is nearly isointense with normal liver parenchyma. On the opposed-phase image, the lesion is markedly hypointense. This illustrates one of the major clinical applications of opposed-phase imaging. Eighty percent of adrenal adenomas contain sufficient lipid to show a marked signal intensity reduction on opposed-phase imaging, which is not seen for the other major lesions considered in differential diagnosis (metastasis, pheochromocytoma, and adrenal carcinoma).

Hepatocellular accumulation of fat occurs in several disease entities. A very simple, but effective, MR technique to detect focal or diffuse fatty infiltration of the liver is the comparison of opposed-phase and in-phase gradient echo images. In fatty liver, there will be a marked reduction in signal intensity on opposed-phase images, when compared with in-phase images (with the spleen used as an internal reference standard). Diffuse fatty infiltration of the liver is well depicted in Fig. 49.3 by comparison of in-phase (a) and opposed-phase (b) images, with a marked reduction in liver signal intensity on the latter. Hepatic fat quantification (see Chapter 96) is easily performed today, in clinical practice, providing a marked advance over such qualitative approaches.
The Dixon technique is a chemical shift-based fat-water separation method, originally described in 1984. Most vendor implementations today are two or three point based. Two point methods acquire two images with different TEs in order to separate the fat from the water signal in the same voxel. Because water and fat spins precess at different frequencies, between excitation and acquisition, the magnetization vectors rotate with respect to each other. Depending upon timing (TE), these become alternatively in-phase and out-of-phase. In the Dixon technique, one image is acquired with the fat and water spins in-phase, and one image with the two out-of-phase. At 1.5 T, the difference in TE between the two states is about 2.46 msec, whereas at 3 T it is about 1.23 msec. In-phase TEs at 1.5 T are 4.92, 9.84, 14.76 msec, etc.; with out-of-phase at 2.46, 7.38, 12.3 msec, etc. In-phase TEs at 3 T are 2.46, 4.92, 7.38 msec, etc.; with out-of-phase at 1.23, 3.69, 6.15, etc. From a simplistic perspective, pure water images can be obtained by adding the in- and out-of-phase images (and pure fat images by subtracting the out-of-phase image from the in-phase).

Fig. 50.1 Tongue border squamous cell carcinoma (enhancing, arrow), post-contrast T1-weighted Dixon technique: in-phase, water, opposed-phase, and fat images.
Although not always displayed for the user, the end result is four images with different tissues contrasts (Fig. 50.1). The Dixon technique is insensitive to \( B_1 \) and \( B_0 \) inhomogeneities and can be applied to a wide variety of sequence types, including FSE and GRE. Although its utility in post-contrast T1-weighted images is widely known – for increased conspicuity of abnormal contrast enhancement on water images (Fig. 50.1), it can also be applied to T2-weighted sequences. The latter application can also be of substantial value, depending on the part of the body, by increasing the conspicuity of neoplastic tissue and inflammatory changes (Fig. 50.2). The Dixon sequence is also robust for fat suppression in areas of high magnetic susceptibility, thus close to metal implants. Three-point Dixon techniques acquire a third echo (thus requiring a longer scan time) in order to decrease the sensitivity to \( B_0 \) inhomogeneity, which if not compensated for can lead to fat-water swapping.

Although illustrated in the head and neck in the figures presented, the Dixon technique is routinely used in abdominal imaging too, where it also finds great applicability.
A small high-signal-intensity abnormality (arrow) is noted within the body of the lateral ventricle on a 2D sagittal gradient echo T1-weighted scan of the brain at 3 T acquired in 1:52 min:sec using a 4-mm slice thickness (Fig. 51.1a). The lesion corresponds to one of many small fat globules scattered throughout the ventricular system and subarachnoid space in this patient with a ruptured dermoid, a pathognomonic imaging presentation. This small fat globule is equally well seen on a 1-mm sagittal image (Fig. 51.1b) from a 4:21 min:sec 3D acquisition.

There are two main approaches to acquiring MR images: 2D and 3D. In a 2D acquisition, a slice is selectively excited by use of a gradient magnetic field (slice-select gradient) and then encoded in two dimensions (phase encoding and readout). The image in Fig. 51.1a is an example of such an acquisition. In a 3D acquisition, a volume or thick slab is excited rather than a single slice. To produce slices from the slab, additional phase encodings are applied along the slice direction. The image in Fig. 51.1b was acquired in a 3D fashion at 3 T using a low SAR fast spin echo sequence known as SPACE (see Chapter 63). 3D scans, like 2D, can be acquired with any parameter weighting, for example T1, T2, or proton density.

The desired number of slices, or partitions, determines the number of phase encoding steps required along the slice direction of the slab in a 3D acquisition. As an example, if one desires 20 slices, then 20 phase encodings are performed along the slice direction. If the total size of the slab is 100 mm, then those slice encodings produce 20 slices with a slice thickness of 5 mm. If the number of slices desired is increased to 40, and the total slab size remains the same, the result would be 40 slices with a slice thickness of 2.5 mm. In a conventional (steady state, not magnetization prepared) 3D sequence, the number of slice encodings applied directly affects scan time. In the previous example, the 40-slice dataset would require twice the number of slice encodings as the 20-slice dataset. As such, the scan time of the 40-slice dataset would be twice that of the 20-slice dataset.

![Fig. 51.1 Comparison of 2D and 3D T1-weighted scans, ruptured dermoid (arrow).](image)
There are many benefits to a 3D acquisition. Slices are produced with no gap or spacing between them. Because the slices are encoded, rather than excited as in a 2D acquisition, there is no crosstalk between slices, as may be seen in a 2D acquisition (which leads to a loss of SNR and changes in image contrast in 2D). 3D acquisitions inherently have higher SNR because of the additional data sampling due to phase encoding in the third dimension. Lastly, if one were to acquire the dataset using an isotropic voxel (voxel dimensions equal in all three dimensions), or simply a small voxel with near equal dimensions in all three axes, the dataset may be reformatted with high resolution in any plane. This feature is illustrated in Fig. 51.2, which compares (a) a 2D axial FLAIR scan to (b,c,d) axial, sagittal, and coronal high-resolution images all reformatted from a single 3D SPACE scan. In this instance, the 2D acquisition was performed with fat saturation, and thus the fat globule, although well depicted (arrow) on (b,c,d) the SPACE images, is not well seen in (a). Scan times were 4:32 versus 6:32 min:sec for the 2D and 3D scans, respectively, with a 4-mm slice thickness for the 2D scan and a $1 \times 1 \times 0.9$ mm voxel dimension for the 3D scan. Due to the requirement for a small, near isotropic voxel, all contrast-enhanced MR angiographic studies are obtained using 3D technique. Potential negatives to a 3D acquisition (which apply more to 3D MR imaging and not MRA) include scan time (typically longer than for alternative clinical 2D scans) and motion artifacts, the latter often accentuated due to the long scan time and propagated along two axes (as opposed to one in a 2D scan) given that there are two phase encoding directions.

![Fig. 51.2 The advantage of 3D acquisition for multiplanar imaging (fat globule, arrow).](image-url)
Contrast Media: Gadolinium Chelates with Extracellular Distribution

The only agents approved today for use in MRI as contrast media are the gadolinium (Gd) chelates. As administered, these are clear, colorless fluids, formulated without bacteriostatic additives, with the typical route of injection being intravenous. The standard dose (excluding some specialty applications) is 0.1 mmol/kg, which corresponds to 15 mL for a 75-kg patient, for the agents formulated at 0.5 molar concentration. The distribution in the body of the agents is to the extracellular space.

Lesion enhancement occurs by one of two mechanisms: disruption of the blood-brain barrier (for intraaxial brain lesions) and lesion vascularity. The gadolinium ion is strongly paramagnetic, leading to a reduction in both T1 and T2, which is visualized on T1-weighted images as an increase in signal intensity. In Fig. 52.1, thin section T1-weighted images are illustrated at the level of the internal auditory canal, revealing a soft tissue mass (a vestibular schwannoma) on the right pre-contrast (Fig. 52.1a), which demonstrates prominent enhancement post-contrast (Fig. 52.1b, white arrow). Enhancement of normal highly vascular structures includes the nasal turbinites (Fig. 52.1b, black arrow) and choroid plexus, easily recognized markers of post-contrast scans. Clinically, contrast enhancement is used both for improved lesion detection and characterization. Contrast injection is routinely performed in the question of neoplastic disease, infection, inflammation, and vascular abnormalities, with broad overall indications. In the prior decade, the field of contrast-enhanced MR angiography developed as an additional major application of the gadolinium chelates.

The word “chelate” comes from the Greek root chelos, meaning claw. The safety basis of the gadolinium chelates rests with the ability of the chelate to hold extremely tightly the gadolinium ion and assure near 100% excretion. Gadolinium is a heavy

Fig. 52.1 Vestibular schwannoma, prior to and following Gd chelate injection.
metal, a member of the transition elements (atomic number 64), and as such is extremely toxic in elemental form (Gd\(^{3+}\)). The gadolinium chelates are 100% renally excreted, with the exception of two agents with combined renal and hepatobiliary excretion (MultiHance and Eovist/Primovist).

The gadolinium chelates currently available for clinical use can be differentiated on the basis of charge (ionic or nonionic), structure (linear or cyclic), and stability (Fig. 52.2; Table 52.1). Given that the gadolinium ion carries a +3 charge, if the ligand, for example, is HP-DO3A (that for ProHance, with a charge of -3), the metal chelate itself will carry a net charge of zero, and thus be nonionic. In the U.S. market, considering only the gadolinium chelates with 100% renal excretion, there are four nonionic agents (Gadavist, ProHance, Omniscan, and Optimark) and two ionic agents (Magnevist and Dotarem). The structure of the chelate can be linear or macrocyclic (ring-like), with the cyclic chelates demonstrating higher in vivo stability and thus improved safety. Gadavist (nonionic), ProHance (nonionic), and Dotarem (ionic) are the macrocyclic chelate agents available both in the United States and internationally.

The identification of nephrogenic systemic fibrosis (NSF) in 1997 and the subsequent, although delayed, recognition of its cause has led to a reassessment of gadolinium chelate use in MRI. NSF is an uncommon but serious acquired systemic disorder affecting patients with severe renal insufficiency, now known to be due to gadolinium chelate administration. Limb contractures and pain are prominent features, with the disease fatal in a small percent of cases. Development of the disease is due to gadolinium chelate dissociation, with deposition of the free metal, and is thus related to chelate stability, dose, and cumulative (lifetime) dose. The vast majority of documented cases followed Omniscan injection, although a substantial number of cases were also documented following injection of Optimark and Magnevist. Early in 2007, the use of Omniscan (and subsequently Optimark and Magnevist) was banned in patients with an estimated GFR less than 30 mL/min/1.73 m\(^2\) by European authorities, with the FDA adopting a similar policy after some delay. Cautious use of the macrocyclic agents, with high thermodynamic and kinetic stability, is felt acceptable even in CKD4 (CKD, chronic kidney disease) and CKD5 (< 30 and < 15 mL/min/1.73 m\(^2\) glomerular filtration rate [GFR]) patients. In terms of incidence of the disease, this has been reported to be as high as 18% in CKD5 (dialysis) patients when given Omniscan.

A further reassessment of safety is ongoing due to the recognition of accentuated accumulation of Gd in the brain (and body) in patients with normal renal function following injection of the linear Gd chelates. This was first described in 2014, being recognized initially on imaging in the dentate nucleus. Of the agents involved, Magnevist is being replaced throughout the world with Gadavist by its manufacturer Bayer, and Optimark (with its former manufacturer being Covidien) being replaced with Dotarem by Guerbet. In July 2017, the European Medicines Agency (EMA) recommended suspension of the marketing authorizations for four linear chelates – Omniscan, Optimark, MultiHance and Magnevist, with the exception of the liver indication for MultiHance and the intra-articular indication for Magnevist.

Contrary to an often-used marketing/sales approach, the extracellular, renally excreted gadolinium chelates cannot be differentiated on the basis of common major reactions. All share the same safety profile in this regard, with nausea reported in 1.5% and urticaria in 0.5% of all injections. Health care personnel should be aware of the potential (although rare) for severe anaphylactoid reactions, with treatment identical to that for an iodinated contrast reaction. Patients with asthma, prominent allergies, or known drug sensitivities (including allergy to iodinated contrast media) are at increased risk for a severe anaphylactoid reaction.
### Table 52.1 The Characteristics of the Clinically Approved Gadolinium-Based Contrast Agents

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Magnevist</th>
<th>Dotarem</th>
<th>ProHance</th>
<th>Omniscan</th>
<th>Gadovist/Gadavist</th>
<th>OptiMARK</th>
<th>MultiHance</th>
<th>Primovist/Eovist</th>
<th>Vasovist/Ablavar</th>
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<tr>
<td><strong>Generic Name</strong></td>
<td>Gadopentetate dimeglumine</td>
<td>Gadoterate meglumine</td>
<td>Gadoteridol</td>
<td>Gadodiameide</td>
<td>Gadobutrol</td>
<td>Gadoversetamide</td>
<td>Gadobenate dimeglumine</td>
<td>Gadoxetic acid disodium</td>
<td>Gadofosveset trisodium</td>
</tr>
<tr>
<td><strong>Manufacturer</strong></td>
<td>Bayer</td>
<td>Guerbet</td>
<td>Bracco</td>
<td>GE Healthcare</td>
<td>Bayer</td>
<td>Guerbet</td>
<td>Bracco</td>
<td>Bayer</td>
<td></td>
</tr>
<tr>
<td><strong>Doses (mmol/kg)</strong></td>
<td>0.1</td>
<td>0.1–0.3</td>
<td>0.1–0.3</td>
<td>0.1</td>
<td>0.1</td>
<td>0.1</td>
<td>0.05–0.1</td>
<td>0.025</td>
<td>0.03</td>
</tr>
<tr>
<td><strong>Concentration (M)</strong></td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>1.0</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>0.25</td>
</tr>
<tr>
<td><strong>Excess Chelate (mg/ml)</strong></td>
<td>0.4</td>
<td>0</td>
<td>0.2</td>
<td>12</td>
<td>0.5</td>
<td>28.4</td>
<td>0</td>
<td>1.0</td>
<td></td>
</tr>
<tr>
<td><strong>Structure</strong></td>
<td>linear</td>
<td>macrocyclic</td>
<td>macrocyclic</td>
<td>linear</td>
<td>macrocyclic</td>
<td>linear</td>
<td>linear</td>
<td>linear</td>
<td></td>
</tr>
<tr>
<td><strong>Ionicity</strong></td>
<td>ionic</td>
<td>ionic</td>
<td>nonionic</td>
<td>nonionic</td>
<td>nonionic</td>
<td>nonionic</td>
<td>ionic</td>
<td>ionic</td>
<td></td>
</tr>
<tr>
<td><strong>Osmolality (mOsm/kg H₂O, 37°C)</strong></td>
<td>1960</td>
<td>1350</td>
<td>630</td>
<td>789</td>
<td>1603</td>
<td>1110</td>
<td>1970</td>
<td>688</td>
<td>825</td>
</tr>
<tr>
<td><strong>Viscosity (mPa·s, 37°C)</strong></td>
<td>2.9</td>
<td>2.0</td>
<td>1.3</td>
<td>1.4</td>
<td>5.0</td>
<td>2.0</td>
<td>5.3</td>
<td>1.2</td>
<td>2.1</td>
</tr>
<tr>
<td><strong>log K\text{therm}</strong></td>
<td>22.1</td>
<td>25.6</td>
<td>23.8</td>
<td>16.9</td>
<td>21.8</td>
<td>16.6</td>
<td>22.6</td>
<td>23.5</td>
<td>22.1</td>
</tr>
<tr>
<td><strong>log K\text{cond}</strong></td>
<td>17.7</td>
<td>19.3</td>
<td>17.1</td>
<td>14.9</td>
<td>14.7</td>
<td>15.0</td>
<td>18.4</td>
<td>18.7</td>
<td>18.9</td>
</tr>
<tr>
<td><strong>t\text{½}</strong></td>
<td>&lt;5 s</td>
<td>338 h</td>
<td>3.9 h</td>
<td>&lt;5 s</td>
<td>43 h</td>
<td>&lt;5 s</td>
<td>&lt;5 s</td>
<td>&lt;5 s</td>
<td>&lt;5 s</td>
</tr>
<tr>
<td><strong>Relaxivity (r1/r2, 1.5 T)</strong></td>
<td>3.9–4.1 / 4.6–5.3</td>
<td>3.6 / 4.3</td>
<td>4.1 / 5.0</td>
<td>4.3 / 5.2</td>
<td>4.7–5.2 / 6.1–7.5</td>
<td>4.7 / 5.2</td>
<td>6.3–7.9 / 8.7–18.9</td>
<td>6.9 / 8.7</td>
<td>19.0 / 34.0</td>
</tr>
<tr>
<td><strong>Relaxivity (r1/r2, 3 T)</strong></td>
<td>3.7–3.9 / 5.2</td>
<td>3.5 / 4.9</td>
<td>3.7 / 5.7</td>
<td>4.0 / 5.6</td>
<td>4.5–5.0 / 6.3–7.1</td>
<td>4.5 / 5.9</td>
<td>5.5–5.9 / 11.0–17.5</td>
<td>6.2 / 11.0</td>
<td>9.9 / 60.0</td>
</tr>
<tr>
<td><strong>Clearance</strong></td>
<td>renal</td>
<td>renal</td>
<td>renal</td>
<td>renal</td>
<td>renal</td>
<td>renal</td>
<td>96% renal, 4% hepatic</td>
<td>50% renal, 50% hepatic</td>
<td>79–94% (mean 84%) renal, ~5% hepatic</td>
</tr>
</tbody>
</table>

* Specific date varies from country to country
** Approval for the highest dose indicated is dependent on country
*** Values in L mmol–1 sec–1 (plasma, 37°C)
K\text{therm} = thermodynamic stability constant
K\text{cond} = conditional stability constant
t\text{½} = dissociation half-time at pH 1.0 and 25°C
Fig. 52.2 Structural formulas for the Gd chelates developed commercially, including those now withdrawn.
53 Contrast Media: Gd Chelates with Improved Relaxivity

By slight changes in structure, agents with improved relaxivity and altered distribution were subsequently developed, with research again active in this area. Gadovist (Gadovist) intrinsically has higher relaxivity than the other extracellular agents. MultiHance, which demonstrates transient protein binding, likewise has higher relaxivity (40% at 1.5 T, approximately that of double dose) and in addition partial hepatobiliary excretion (4–6%), both due to addition of a phenyl moiety. (Fig. 53.1 illustrates the improved enhancement of a brain metastasis at 1.5 T with MultiHance Fig. 53.1b) as compared to a conventional extracellular Gd chelate (Fig. 53.1a). Fig. 53.2 illustrates, using a hepatobiliary gadolinium chelate, a liver metastasis prior to (Fig. 53.2a), immediately following (Fig. 53.2b), and in the hepatobiliary phase after (Fig. 53.2c) injection. Hepatobiliary uptake and excretion of the agent are illustrated, with sustained enhancement of normal liver, opacification of the gallbladder, and slightly improved visualization of a liver metastasis (arrow) on the delayed image. High-resolution 3D breath-hold images are now standard of care, in both the dynamic and delayed phases, providing improved detection of liver metastases, such as that seen immediately below the dome of the diaphragm (Fig. 53.2d, arrow). Agents

Fig. 53.1 Improved lesion enhancement due to higher relaxivity.

Fig. 53.2 Liver metastasis, MultiHance.
with very high hepatobiliary excretion (50%), specifically Primovist (Eovist), or prolonged residence in the bloodstream due to albumin binding, specifically Ablavar (mean half-life 16 hours), have been developed. Ablavar however is no longer clinically available. In sharp distinction is Primovist, a well accepted hepatobiliary agent, available and widely used across the world.

Some of the more recently available gadolinium chelates have advantages as well in contrast-enhanced MR angiography. For example, vessel signal intensity has been shown to be markedly improved with both Gadavist and MultiHance, the former principally due to its higher formulation concentration (1 molar) and the latter due to transient albumen binding, compared with more conventional agents. Technologic advances in equipment design have made possible high-resolution whole-body 3D contrast-enhanced MR angiography (Fig. 53.3) with the Gd chelates. Five 3D datasets were acquired in immediate succession following a single IV contrast dose, with Fig. 53.4 illustrating patient positioning and image acquisition for three such image sets.

Fig. 53.3 Whole-body CE-MRA.

Fig. 53.4 Multi-station CE-MRA technique.
Contrast Media: Other Agents (Non-Gadolinium)

Other contrast media, not based on gadolinium, have been developed for MRI, but are today no longer in general use. Superparamagnetic iron particles are selectively taken up following intravenous injection by Kupffer cells, primarily in the liver. Two such iron oxide-based agents were approved in the past: ferumoxides (Endorem, distributed in the U.S. as Feridex, and at one time in use worldwide), with a particle size range of 50 to 180 nm, and Resovist (which was previously available in Europe), with a particle size of ~60 nm. The principal relaxation effect of these large particles was on T2 (due to susceptibility effects, with the contrast agent causing a decrease in signal intensity), with scans performed in a delayed fashion postinjection, allowing time for liver uptake. Resovist was also approved for bolus injection (dynamic imaging), with T1-weighted scans and positive contrast enhancement noted in this application. The safety profile for Feridex was not comparable to that of the gadolinium chelates, with a substantially higher incidence of adverse reactions. The safety profile of Resovist was superior to that of Feridex.

In Fig. 54.1, imaging with Resovist reveals a small hypervascular lesion on Fig. 54.1a, the dynamic scan, with prominent iron uptake on Fig. 54.1b, the delayed scan (white arrow), compatible with an adenoma. In Fig. 54.2, the pre-contrast in-phase T1-weighted scan (Fig. 54.2a) reveals a cirrhotic, nodular liver. On Fig. 54.2b, the delayed post-contrast scan (using Resovist), a subcapsular hepatocellular carcinoma (HCC, white arrow) and multiple low signal intensity regenerative nodules (black arrows) can be identified. In Fig. 54.3, a delayed post-contrast scan (using Resovist), spread of neoplastic disease to the liver (from a pancreatic primary) is readily identified (white arrow). There is excellent tissue contrast between tumor and normal liver, the latter with uptake of this iron-based agent (and thus low signal intensity on T2-weighted scans). Dynamic post-contrast T1-weighted imaging with a gadolinium chelate in this instance provided poor tumor versus normal liver differentiation (image not shown).
Teslascan, a manganese (Mn)-based agent, was approved in the late 1990s for delayed liver imaging. Unlike the gadolinium chelates, this agent rapidly dissociated after IV injection, yielding free Mn. Safety concerns led to eventual withdrawal of this agent. Like Feridex, the incidence of adverse events was substantially higher as compared with the Gd chelates. T1-weighted images were employed post-contrast (with Mn having paramagnetic properties similar to Gd, but of lower magnitude).

Oral MR contrast agents are classified according to the observed signal intensity (SI): positive ("bright" lumen), negative ("dark" lumen), or biphasic. Several agents were at one time commercially available in some countries, with generally low utilization. Positive agents originally included dilutions of the gadolinium chelates, specifically formulated for oral use (these are no longer available commercially), and solutions of iron or manganese ions. Some natural substances, such as milk, vegetable oil, green tea, and blueberry juice, and some manufactured products, such as ice cream, also act as positive oral contrast agents, due to either high fat or manganese ion content. Agents containing manganese typically are biphasic in character, with high SI on T1- and low SI on T2-weighted images. Negative contrast agents, which provided a dark lumen on both T1- and T2-weighted images, included several different iron particulate preparations. Water can be used as an oral contrast agent, but its use is limited by intestinal resorption. Barium sulfate can also provide some luminal contrast, with SI (low on T1 and low to high on T2) dependent on administered concentration and subsequent dilution.
A further modification of steady-state gradient echo technique involves the acquisition of two different echoes during each repetition time. The first echo is the free induction decay (FID) gradient echo used in the spoiled GRE sequence (see Chapter 40) and the second is the RF echo (SE) used in the PSIF sequence (Chapter 58). The separation and acquisition of these two echoes are illustrated in Fig. 55.1. As noted previously, the signal decay after excitation is called the FID and can be sampled with any gradient echo technique. Any RF excitation pulse also has refocusing capabilities. An echo that is formed using a refocusing RF pulse is called a spin echo (SE). With DESS, each RF pulse generates an FID and a SE signal in the steady state. In the pulsing diagram shown in Fig. 55.1, the gradients are employed such that for the first echo, the FID is rephased and the SE is spoiled, and for the second echo, the reverse is true—the FID is spoiled and the SE is rephased.

The DESS sequence starts with a slice selective low-angle excitation pulse (1 in Fig. 55.1), with the FID sampled (2 in Fig. 55.1) as per the GRE techniques described previously. The remaining transverse magnetization, previously dephased with a phase encoding gradient for the purpose of spatial encoding, is then refocused for that direction. The transverse magnetization is prepared in the direction of slice selection to be refocused in the center of the next excitation pulse to form a spin echo (3 in Fig. 55.1). The dephasing mechanism of the slice selection gradient is at this point considered in advance. The frequency-encoding gradient is left on for the period of both echoes (4 in Fig. 55.1) and is incompletely balanced to avoid dark banding artifacts otherwise associated with fully balanced steady state sequences. Note that the effective TE for the spin echo contribution is actually longer than TR. The FID and SE components are sampled in adjacent windows and combined prior to image reconstruction.

Fig. 55.1 Pulse diagram for DESS.
Advantages to this approach include improved SNR and somewhat unique contrast (due to acquisition and subsequent combination of the more T1-weighted spoiled GRE echo and the T2-weighted RF echo). Measurement times are comparable to refocused steady-state GRE technique. It should be noted that DESS is actually not that much different from a refocused GRE sequence (FISP), but because two echoes are sampled, it will typically have better SNR. Also, DESS and FISP both have T1 and T2 components (weighting) due to the contribution of the rephased spin echo signal. Applications for DESS include high-resolution 3D imaging of the joints permitting high-quality multiplanar reformats, with high signal intensity for fluid, and excellent delineation of cartilage. Note that due to the combination of T1 and T2 contrast, the use of this technique is restricted in terms of the detection of pathologies with a diffuse distribution pattern such as bone edema or infiltrative masses.

Fig. 55.2 presents multiplanar reformats from a water-excited 3D DESS scan at 3 T, acquired with a voxel size of 0.6 × 0.5 × 0.5 mm. The small near isotropic voxel permits reconstruction of high-quality reformatted images in any desired plane. Shown in Fig. 55.2 are reformatted sagittal (a), axial (b), and coronal DESS images (c), together with a coronal STIR image for comparison (d). Note the cartilage defect overlying the medial femoral condyle (white arrow) and the osteophyte and associated thinning of cartilage involving the lateral femoral condyle (asterisk), best seen on DESS. Damage to the patellar cartilage is also well depicted on the axial reformat (black arrow).

**Fig. 55.2** Multiplanar DESS reformats vs. coronal STIR for cartilage assessment.
A further refinement of GRE technique involves adjusting the FID component to coincide with the spin echo component. A balanced GRE sequence (known by the acronyms TrueFISP, FIESTA, and b-FFE) starts out with an RF pulse of less than 90° with data sampling performed in steady state. The transverse magnetization generated induces the FID signal (1 in Fig. 56.1). The next pulse (and every pulse thereafter) not only generates transverse magnetization based on the available longitudinal magnetization, but also operates as an RF refocusing pulse for the remaining transverse magnetization of the previous Fourier line measurement (2 in Fig. 56.1). The RF echoes generated by the train of excitation pulses in the steady state are then superimposed on the gradient echo. The readout gradient is prepared in such a way that the FID and the SE are sampled at the same time (3 in Fig. 56.1). Then, before the next pulse is applied, gradients in the slice, phase, and frequency encoding directions are balanced, so their net value is zero. Now, the spins are prepared to accept the next RF pulse, and the corresponding signal becomes part of the new transverse magnetization. If the balanced gradients maintain the longitudinal and transverse magnetization, the result is that both T1 and T2 contrast are represented in the image. Fat and fluids such as bile, blood, or CSF appear bright, with the use of fat suppression providing differentiation of flowing blood from fatty tissue.

Because this sequence is extremely vulnerable to field homogeneity, proper shimming prior to the acquisition is indispensable for good image quality. Motion artifacts are few, SNR is high, and the acquisition time is short. A balanced gradient echo technique is thus useful for ultrafast acquisitions and commonly employed as a scout. Balanced GRE (b-SSFP) has also gained widespread popularity in real-time imaging of the heart and major vessels. It is an excellent sequence for cardiac imaging given its insensitivity to motion artifacts (due primarily to the speed at which the scan is acquired, with TRs typically 3 msec or less) and the high contrast between myocardium and blood in the cardiac chambers. It can also be used for breath-hold
abdominal scans. The main reason that balanced GRE technique has gained popularity in cardiac applications is because of the very high signal from blood. Tissue contrast with this scan technique is based on the ratio of T1/T2. The closer the ratio is to a value of 1, the higher the signal. Thus, the highest signal comes from CSF and fresh (unexcited) blood, even though their T1 and T2 values are so different. With bright blood, the heart chambers are extremely well visualized, allowing for excellent assessment of cardiac function. However, because the contrast is neither truly T1-nor T2-weighted, a balanced GRE scan cannot substitute for standard pulse sequences in routine clinical applications outside of the heart.

Fig. 56.2 presents a retrospectively ECG-gated balanced GRE scan of the heart in short axis orientation showing thinning of the inferior aspect of the left ventricle, with dyskinesia, corresponding to scar after myocardial infarction. In combination with ECG-or pulse-gating, balanced gradient echo sequences provide excellent contrast between blood and myocardium and, in addition, give information about ventricular function and wall motility.

Fig. 56.3a presents an axial T1-weighted spoiled GRE scan through the upper abdomen demonstrating a heterogeneous, hypointense mass within the right lobe of the liver. Fig. 56.3b presents the corresponding balanced GRE scan in which the vessels are hyperintense due to blood flow. Note that a portion of the mass demonstrates high signal intensity (arrow). The lesion corresponds to a chronic hematoma secondary to a ruptured arteriovenous malformation (AVM), with persistent high flow within the AVM. In abdominal imaging, balanced gradient echo sequences can provide, as illustrated, important information concerning whether blood flow within a vessel or a lesion is present or not.

Fig. 56.2 Chronic myocardial infarct. Courtesy of Günther Schneider.

Fig. 56.3 Blood flow within an AVM seen on TrueFISP. Courtesy of Günther Schneider.
Balanced Steady-State Free Precession (b-SSFP), another term for balanced GRE, sequences are part of the family of gradient echo-based techniques employed in a variety of MR applications. The b-SSFP technique is set apart by the fact that during the acquisition the dephasing induced by each applied gradient is compensated for by a gradient of opposite polarity resulting in a unique T2/T1 contrast. Although this contrast may not provide the same diagnostic information as a classic SE based T1-or T2-weighted image, the technique exhibits very high SNR and excellent tissue differentiation making it well suited for imaging abdominal, vascular, and cardiac structures. The examples provided in Fig. 57.1 demonstrate coronal b-SSFP (TrueFISP) imaging of the upper abdomen (a) without and (b) with fat suppression. The abdominal vasculature is especially well delineated in these ~1 second acquisitions. Fig. 57.2 illustrates the superior tissue differentiation of (b) b-SSFP technique over (a) standard RF-spoiled gradient technique in the left ventricle, where myocardial-blood pool differentiation is important for the assessment of myocardial structure and function (see Chapter 89 and Chapter 90).

Balancing tissue magnetization with gradient changes requires a very homogenous $B_0$ field to minimize susceptibility related artifacts. These artifacts, depicted in Fig. 57.3a, appear as repeatedly arising dark bands or lines in the image, with the distance between being inversely proportional to $B_0$ and TR. Improved shimming techniques, providing better magnetic field homogeneity, and more accurate frequency adjustments today limit the presence of such bands. However, in rare instances the artifact may still be present and can simply be shifted out of the region of interest through a frequency offset to the nominal resonance frequency of the magnetic field, on the order of 50 to 250 Hz. This method incorporates a specialized sequence known as a frequency scout and is used to visualize a series of images with different frequency offsets (Fig. 57.4). The frequency offset (shift) of the image that represents the least artifact in the area of interest is selected and used in the subsequent b-SSFP acquisition to improve image quality, with the result illustrated in Fig. 57.3b. To further reduce the effects of magnetic susceptibility variations on b-SSFP techniques, the TE and TR are maintained at the lowest possible values.

Fig. 57.1  Coronal abdominal TrueFISP, without and with fat saturation.
B-SSFP techniques are marketed in several variations with a variety of names including TrueFISP (Siemens), FIESTA (GE), and balanced FFE (Philips). Although the implementations vary, the clinical utility of the sequence is more a product of field homogeneity than variations in pulse scheme making it critical that all ferrous objects, including metal rings of all types, be removed when these techniques are employed.

Fig. 57.2 Spoiled GRE vs TrueFISP for the heart.

Fig. 57.3 Improved imaging with TrueFISP using a frequency offset.

Fig. 57.4 The effect of different frequency offsets on image artifacts.
It is possible to collect only the spin echo component of the previously described balanced gradient echo. Interestingly, a backward-running fast imaging with steady precession (FISP) generates just the spin echo. The basic sequence loop is illustrated in Fig. 58.1, indicated between two vertical dashed lines. The excitation pulse (1) for a given acquisition cycle generates transverse magnetization, which is dephased with the gradient arrangement that follows, refocused by the subsequent RF excitation pulse (2), and read out as the spin echo signal (3). Because the timing is identical to a backward-running FISP, the acronym FISP has been reversed to form PSIF, which has been adopted for this sequence.

Starting with the first excitation pulse, the transverse magnetization dephases in the immediate repetition time interval due to the application of the phase encoding and frequency-encoding gradients. No signal is returned during the data acquisition window. In the second cycle, the RF excitation pulse refocuses the dephased magnetization. For a PSIF sequence, the echo time is actually longer than the repetition time. The PSIF sequence, although a backward-running FISP, is actually a spin echo sequence. PSIF isolates the spin echo contribution of a balanced gradient echo sequence like TrueFISP (or, equivalently, FIESTA or b-FFE). PSIF is a type of SSFP technique, although the term SSFP is usually equated with TrueFISP, the truest form of steady-state free precession.

Fig. 58.2 demonstrates the difference between PSIF (a) and constructive interference in a steady state or CISS (b) using images at the level of the internal auditory canal (see Chapter 59). CISS is a phase cycled, balanced gradient echo technique with very low sensitivity to flow. The PSIF signal represents the isolated spin echo of that sequence and is very sensitive to flow. The flow sensitivity of PSIF is readily evident in Fig. 58.2, with signal loss in areas of cerebrospinal fluid (CSF) motion, in particular.

Fig. 58.1 Pulse diagram for PSIF.
within the prepontine cistern. PSIF, therefore, is not an alternative to display cranial nerves surrounded by CSF, but is a helpful adjunct in cases where the documentation of CSF flow based on the signal void is of diagnostic relevance.

PSIF in recent years has found clinical application for acquisition of diffusion-weighted images of the spine, as well as of the peripheral nerves. Fig. 58.3 presents T1-weighted spin echo, STIR, and PSIF diffusion-weighted sagittal images of the lumbar spine in an elderly patient with an L4 compression fracture and known lung carcinoma. The L4 vertebral body demonstrates abnormal low signal intensity on the T1-weighted scan and abnormal high signal intensity on STIR; however, this appearance does not allow for differentiation between an acute benign compression fracture and a metastasis. Metastases generally demonstrate high signal intensity on diffusion-weighted images, in comparison to normal bone marrow, thus favoring the diagnosis of a benign acute compression fracture in this instance (with the L4 vertebral body, white arrow, being low signal intensity on the diffusion-weighted image). An important caveat however is sclerotic metastases, such as from prostate carcinoma, which may not demonstrate hyperintensity on PSIF diffusion-weighted scans.

Fig. 58.2 Comparison of PSIF and CISS at the level of the pons.

Fig. 58.3 T1-weighted, STIR, and PSIF DWI images of a benign, acute fracture. Courtesy of Andrea Baur-Melnyk.
Constructive Interference in a Steady State (CISS)

When certain echo paths of a balanced-SSFP (b-SSFP) sequence (see Chapter 57) are out of phase during the data acquisition, a destructive interference pattern is seen. The resultant signal voids occur predominantly in areas, such as air-tissue interfaces, where tissues display relatively large differences in magnetic susceptibility. These differences degrade the homogeneity of the local magnetic field, leading to a rapid dephasing of encoded echoes. The inner ear is a good example of a complex tissue with areas of very different magnetic susceptibility in close proximity, being composed of air, bone, nerves, and vessels. In order for high-resolution b-SSFP techniques to have clinical utility in such areas of highly variable magnetic susceptibility, the CISS technique (Constructive Interference in a Steady State) was developed.

The CISS technique is comprised of two 3D b-SSFP acquisitions measured in succession. By slightly varying the excitation (a) pulses, the destructive interference patterns (signal voids) will be spatially shifted from each other in the images. A complex algorithm is used to add the two acquisitions together. Signal voids due to interference in one acquisition are filled in with data from the second acquisition, with SNR improved as well by the square root of two.

Images are presented in Fig. 59.1 from a patient with a very small left-sided intracanicular vestibular schwannoma. Axial fast spin echo (FSE) T2-weighted (a), contrast-enhanced, spin echo T1-weighted (b), and CISS sequences (c) were acquired through the internal auditory canals. The lesion is difficult to detect on the FSE T2-weighted image, acquired with a 3-mm slice thickness. However, the CISS sequence displays well the focal enlargement of the nerve (arrow, c), thus enabling detection of this small tumor. Indeed, the lesion was visualized on two adjacent slices from the CISS scan, due to acquisition of 1-mm contiguous sections. The lesion is also well visualized on (b) the T1-weighted scan due to contrast enhancement.

Fig. 59.1 T2 FSE, enhanced T1, and CISS images of a small tumor on cranial nerve VIII.
The heavily T2-weighted, 3D CISS sequence is typically acquired with high spatial resolution and sub-millimeter partitions offering detailed delineation of small structures of the inner ear and cerebellopontine angle (Fig. 59.2a). 3D CISS scans can be processed using maximum intensity and surface rendering algorithms, with the result being, when acquired with a very small, isotropic voxel, high-quality 3D projections of the cochlea, vestibule, and semicircular canals (Fig. 59.2b). An additional benefit of an isotropic acquisition, in the area of the inner ear, is the ability to reconstruct images in any orientation to better examine the 7th and 8th cranial nerves. Fig. 59.3a demonstrates the parallel ranges used to reconstruct the image in Fig. 59.3b, with the inferior vestibular nerve (part of cranial nerve VIII, arrows) depicted for the full length of its intracisternal course.

Improvements in magnet homogeneity have allowed this imaging approach to be explored in areas such as the spine. Fig. 59.4 presents an axial CISS section in the cervical spine at 1.5 T. The excellent CSF to soft tissue contrast provides improved depiction of the dorsal and ventral cervical nerves as they traverse the thecal sac. The CISS sequence is just one of many 3D techniques that have gained popularity as newer hardware and software promoted a paradigm shift in MR imaging from 2D slices to 3D volumes, similar to the change observed in CT with the transition to multidetector scanners.
TurboFLASH (FSPGR, TFE) is a special, ultrafast spoiled gradient echo technique with very short TR, TE, and small excitation flip angle (note the comments at the end of this case, in respect to the definition of TR). A major consequence of the short TR and the small flip angle is that this type of sequence, in principle, has low SNR and poor T1 contrast. However, image quality can be markedly improved by using a magnetization preparation pulse, commonly a 180° inversion pulse once at the beginning of the acquisition (instead of prior to every acquisition of each Fourier line, which would imply a very long measurement time), as illustrated in Fig. 60.1. A different approach to magnetization preparation would be the use of a “saturation recovery” pulse (SR), specifically a 90° RF pulse, saturating all tissues prior to the start of the imaging sequence. The preparation pulse and the short sequence parameters allow one to center the subsequent ultrafast gradient echo data acquisition at an inversion time TI, chosen to null signal from a selected tissue.

One application of this technique is the visualization of cardiac perfusion. TurboFLASH techniques provide excellent depiction of the first pass of the contrast agent through the cardiac chambers, followed by contrast uptake in normal myocardium. In the same way, perfusion studies may be performed in the liver and the kidneys. For perfusion studies, TurboFLASH techniques are usually applied with a single shot approach (as illustrated in Fig. 60.1). By using a segmented k-space data sampling approach, TurboFLASH techniques may be used for abdominal imaging (with T1-weighting), for in-phase and opposed-phase imaging or for contrast-enhanced MR angiography during breath-holding. TurboFLASH can be used as well for T1-weighted imaging of the brain, with 2D application making possible the implementation of BLADE (see Chapter 106) and 3D application (MP-RAGE; see Chapter 62) providing a high-resolution dataset with high inherent tissue contrast and image quality.

Fig. 60.2 presents images acquired during the first pass of a contrast bolus through the left ventricle employing a TurboFLASH sequence (one image per heartbeat, with every fourth image illustrated). A nonenhancing mass (corresponding to thrombus) with associated thinning of the
adjacent apical myocardium is depicted in this patient with the history of an apical myocardial infarction. Fig. 60.3 presents a comparison of T1-weighted post-contrast sagittal images acquired at 3 T with (a) short TE FLASH and (b) BLADE TurboFLASH techniques in a patient with a large enhancing meningioma (black arrow). Note the improved gray-white matter contrast of the TurboFLASH scan, together with the absence of ghost artifacts (a, white arrow) due to the use of BLADE.

One point of potential confusion with TurboFLASH is the definition of TR. There are actually two TRs in this type of methodology. One is the TR for the rapid gradient echo readout, which is typically on the order of 6 to 12 msec. The other TR is the repeat rate of the magnetization preparation, which is typically on the order of 3000 msec or more. Because the tissue contrast is dictated by the magnetization preparation, it has become generally accepted that the “real” TR is the one defining the repetition rate of the magnetization preparation. The real TR is in fact this value, if the scan is performed using k-space segmentation, or in 3D mode, or as a BLADE application, or with multiple averages, or as a perfusion technique.

Fig. 60.2 Cardiac imaging with TurboFLASH, revealing left ventricular thrombus.

Fig. 60.3 FLASH vs. BLADE TurboFLASH, the latter with reduced motion artifacts.
PETRA (UTE)

Sequences with ultrashort TEs enable new MR applications, including specifically imaging of bones, tendons, ligaments and teeth. In standard SE sequences, the minimum TE is in the range of a few msec, and in GRE sequences down to 1 msec. Tissue with a very short T2, well below 1 msec, will not be visible (and will appear dark) due to signal decay by the time of acquisition. In addition to the imaging of tissue with very short T2s, ultrashort TE sequences have other applications. Dephasing of spins between excitation and acquisition is much less, enabling improved imaging for areas with large changes in susceptibility (such as close to metal prostheses). If imaging is performed with a receive-only coil, which is within the FOV, then the coil itself can be visualized. Imaging of bone also enables tissue segmentation (water, bone, air) needed for attenuation correction in MR-PET.

A challenge with the common implementation, an ultrashort echo time (UTE) sequence, is that acquiring data during gradient ramping can lead to major image distortions in part due to eddy currents. PETRA (pointwise encoding time reduction with radial acquisition) combines radial and Cartesian single point acquisition. Outer and inner k-space are acquired separately, the low-resolution image in the Cartesian part, and the high resolution (high spatial frequency data) in the radial part. Each k-space point is measured with the smallest possible encoding time. Due to the gradients being turned on during excitation, PETRA can only be acquired as a 3D scan with nonselective excitation. The sequence can be applied with fat saturation. Higher resolution and SNR can be achieved, when compared to UTE sequences, with one limitation being that the sequence is 3D (both 2D slice selective and 3D acquisitions are possible with UTE).

One major application for PETRA is for quiet sequences, specifically for sedated patients and pediatrics (as illustrated in Fig. 61.1). Fast gradient switching, used in

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**Fig. 61.1** MP-RAGE and T1-PETRA acquired at 3 T in a 2-year-old child. Courtesy of Noriko Aida, Kanagawa Children’s Medical Center, Yokohama, Japan.
conventional scans, leads to high acoustic noise. PETRA has such low demands on gradient switching and ramping that no acoustic noise is generated by the gradient coil.

Visualization of arterial calcifications can be important for clinical prognosis and patient management. For example, in the peripheral arteries, the presence of dense calcifications decreases success rates for balloon angioplasty. These can also alter the choice of access sites for patients undergoing percutaneous revascularization or aortic valve replacement. CTA unlike MRA is able to reliably detect and display vascular calcifications in a large FOV projection format, a major reason that this technique is preferred by vascular surgeons. MR can however be used for the detection of calcifications and display in projection formats similar to CT. In order to display calcifications, which have lower signal intensity (SI) than all other tissues, images are processed with a minimum intensity projection algorithm, followed by grayscale inversion. Of importance is to distinguish other tissue types which have low SI including specifically ligaments and tendons. With PETRA, these soft tissues all have SI well above background levels. These can thus be distinguished from vascular calcifications, which although recovering some signal, maintain sufficiently low SI.

As shown in Fig. 61.2, depicted on grayscale inversion using PETRA, vascular calcifications appear bright. These can thus be distinguished from the vessel lumen and surrounding soft tissue. StarVIBE can also be used in a similar fashion and in preliminary analysis outperforms PETRA. In this application of StarVIBE, the echo time is adjusted so that fat and water are in phase, and a very small flip angle is used for RF excitation. The result is a homogeneous SI across the field of view for most tissues, with calcifications being dark due to the very short T2*. StarVIBE also benefits due to its relative insensitivity to motion.

![Fig. 61.2 CTA, PETRA, StarVIBE – focal calcification (arrow) with patent lumen, along the common femoral artery. Courtesy of Robert R. Edelman, NorthShore University HealthSystem, Evanston, IL.](image)
The challenge of extending TurboFLASH (FSPGR, TFE) from a 2D to a 3D acquisition is due to the fact that the effect of an inversion pulse applied at the very beginning of the acquisition is lost during the acquisition. This is due to the high number of excitation pulses and the relatively long measurement time associated with 3D acquisitions. This may be overcome by repeating the inversion pulse during the measurement, an approach called magnetization-prepared rapid gradient echo (MP-RAGE). Three-dimensional image acquisition requires a repetition of all phase encoding steps in the slab (partition) direction for every phase encoding step within the imaging plane. The inversion pulse is thus placed just prior to the partition encoding loop (in most cases). Within the partition encoding loop, the signal is changing due to recovery of the longitudinal magnetization.

There are several advantages to using MP-RAGE for T1-weighted imaging. Unlike multislice 2D imaging, the partition encoding used in 3D acquisitions, such as in MP-RAGE, allows for continuous coverage with thin slices in a reasonable measurement time (5 minutes or less at 3 T). Inversion of the magnetization allows better control over T1-weighting, permitting greater T1 contrast compared with spin echo and 2D gradient echo imaging. Fig. 62.1 presents post-contrast axial 3 T T1-weighted 2D GRE (Fig. 62.1a) and MP-RAGE (Fig. 62.1b) images of a 28-year-old, severely disabled woman with a 7-year history of relapsing/remitting multiple sclerosis (MS). There are a large number of MS plaques (many confluent), with abnormal low signal intensity, noted in the periventricular and more peripheral white matter. Note the greater T1 contrast with (b) MP-RAGE, reflected by the better gray-white matter differentiation (small black arrow) and the improved MS plaque conspicuity (white arrows), the latter relative to normal adjacent white matter. Scan times were 1:52 (2D GRE) and 3:52 (MP-RAGE) min:sec, respectively. The slice thickness for the 2D acquisition was 4 mm, with an in-plane resolution of 0.86 × 0.86 mm. The 3D scan was acquired with a voxel dimension of 1 × 1 × 1 mm. The axial scan was reformatted using a slice thickness of 1.5 mm. The axial MP-RAGE image appears slightly blurred when compared with the 2D scan, principally due to the slightly poorer in-plane resolution. A major advantage of this type of isotropic acquisition is the ability to reformat images in other planes with high resolution, as illustrated in (b) and (c). 1.5 mm coronal and sagittal reformatted images. Thus, high-resolution images in all planes are available from a single 3:52 min:sec acquisition, whereas the 2D GRE scan—obtained in about half the acquisition time—provided only axial images. Note the active MS plaque (large black arrow) demonstrating abnormal contrast enhancement, which is well visualized in all three planes.

MP-RAGE has been used both at 1.5 and 3 T for high-resolution, structural brain imaging. However, multiple caveats exist in regard to its use. MP-RAGE is a gradient echo technique, and thus the artifact generated due to metal can be greater than with fast spin echo technique. Depending upon the specific choice of pulse parameters, visualization of lesion enhancement (particularly weakly enhancing lesions) may be poor with MP-RAGE when compared with a conventional spin echo pulse sequence or a 3D fast spin echo sequence such as SPACE, discussed subsequently. In routine clinical imaging at 1.5 T, the relatively long scan times have also prevented widespread use in screening brain exams. At 3 T, however, high-quality T1-weighted scans can be generated in a reasonable time with MP-RAGE, leading to its broad use in routine exams.
Fig. 62.1 Multiple sclerosis, comparison of 2D GRE and MP-RAGE post-contrast.
When MR was in its infancy, 2D spin echo (SE) imaging was the most commonly used scan technique. SE imaging was robust to artifacts from magnetic field inhomogeneities, an advantage with early magnets. 3D SE imaging was possible, but failed to gain widespread acceptance due to extremely long scan times. A potential advantage of such an acquisition would be the possibility of retrospectively reconstructing high-spatial resolution images in any orientation from a single scan acquisition. With the advent of turbo or fast spin echo imaging (these two terms are interchangeable, and are abbreviated TSE and FSE), 3D imaging became feasible. However, power deposition with 3D FSE is high and indeed prohibitive in some instances, particularly at 3 T. Several variants of FSE technique were subsequently introduced, with one objective being to reduce the power absorbed by the patient. All achieve a decrease in SAR by reducing in some fashion the flip angle of the refocusing pulse in the FSE sequence. One such technique, termed TRAPS (transition between pseudo steady states), employs a train of variable refocusing flip angles in an arrangement such that the echoes close to the center of k-space are fully refocused. Modifying this approach by using a flip angle series that yields a constant signal for the majority of the spatially encoding echo train, taking into account the relaxation during data sampling, will further improve image quality. This optimization requires knowledge of the relaxation values for the tissue to be imaged.

This complex approach, with application-specific flip angle evolution, achieves the desired contrast using ultralong echo trains in combination with a 3D FSE imaging acquisition scheme and has been termed SPACE (sampling perfection with application optimized contrasts by using different flip angle evolutions). This method provides for the necessary reduction in SAR, taking advantage of a volume acquisition and optimizing the achievable CNR for a specific application. Images with proton density, T1-, or T2-weighting can be achieved, as well as FLAIR-like contrast for brain and cord imaging.

Fig. 63.1 illustrates a T2-weighted SPACE acquisition in the cervical spine. The scan time was 6 minutes (using a parallel imaging factor of 3) with a spatial resolution of 0.9 × 0.8 × 0.8 mm. Near isotropic resolution allows high-resolution reformatted images.
in any arbitrary plane, with 1.5-mm thick sagittal and axial scans illustrated (the latter parallel to the disk space) in this example of a central disk herniation at C5–6. **Fig. 63.2** illustrates a SPACE brain acquisition with FLAIR-like contrast in a patient with a large chronic MCA infarct, with the abnormal high and low signal intensity areas reflecting respectively gliosis and tissue loss (cystic encephalomalacia). Axial, sagittal, and coronal 1.5 mm images are illustrated, all reformatted from a single 6.5-minute acquisition (using a parallel imaging factor of 2) with $1 \times 1 \times 0.9$ mm voxel dimensions.

Looking at the SPACE technique in more detail, nonselective refocusing pulses are used to achieve ultrashort echo spacing (as low as 3 msec). This permits in part echo train lengths > 300, with the result (when used in combination with parallel imaging) being a high-resolution 3D image acquisition within an acceptable scan time. This approach is clinically advantageous in particular in areas of complex anatomic detail, including applications in the spine, musculoskeletal system (knee, meniscus), and brain (internal auditory canal). For detection of small lesions, such as MS plaques, SPACE has been shown to be superior to thin section (3 mm) 2D imaging.

**Fig. 63.2** FLAIR SPACE, chronic MCA infarct.
Susceptibility-weighted imaging (SWI) employs a high-resolution, velocity-compensated (for flow in all three axes), spoiled 3D GRE scan. Of special importance, SWI utilizes the phase difference information (commonly discarded on most imaging techniques) to emphasize susceptibility differences between tissues. In the SWI implementation, two types of images are reconstructed: a high pass filtered phase image and a minimum intensity projection, the latter being with phase-weighted magnitude information. SWI has high sensitivity to normal venous blood (in particular small veins), blood flow, acute hemorrhage (deoxyhemoglobin), chronic hemorrhage (hemosiderin), and iron in the form of ferritin. In common with most MR applications, this technique is further improved at 3 T, due to both increased SNR and more prominent susceptibility-related effects. The latter allows TE and thus TR to be halved, markedly reducing scan time.

Primary applications for SWI include traumatic brain injury (both acute and chronic) and cavernous malformations. In regard to trauma, SWI offers improved depiction of the multiple punctate hemorrhages seen in diffuse axonal injury (DAI), relative to 2D T2*–weighted GRE scans. The lesions in DAI, which are due to shear injury, occur anatomically at the corticomedullary junction and within the corpus callosum, deep gray nuclei, and brainstem. Also improved with SWI is the visualization of intraventricular and subarachnoid hemorrhage, both depicted with low signal intensity within otherwise normal CSF-containing spaces. Use of SWI thus complements that of FLAIR in the detection of acute subarachnoid hemorrhage. In regard to patients with cavernous malformations (Fig. 64.1, acquired at 3 T), due to the improved sensitivity to hemosiderin, these lesions will in general appear larger on SWI (arrow, b) when compared with 2D T2*–weighted GRE scans (a). SWI offers improved detectability of very small cavernous malformations, an important point given that 10 to 30% of patients have the familial

Fig. 64.1 2D T2* GRE vs. SWI in visualization of a cavernous malformation (arrow).
type of disease in which multiple lesions are seen, many very small. Venous angiomas and the venous drainage within neoplastic tissue are also well depicted by SWI.

The images in Fig. 64.2 present a confusing radiologic picture, with SWI facilitating scan interpretation. On higher sections (not shown), the patient had an easily recognized, early subacute, anterior cerebral artery territory infarct. At the anatomic level illustrated, FLAIR demonstrates a mass lesion with surrounding high signal intensity (a). No abnormal enhancement is seen post-contrast on the T1-weighted image (b). On DWI (c), the abnormal high signal intensity on FLAIR is demonstrated to be composed both of cytotoxic edema (high signal intensity on DWI, white arrows, due to the infarct) and vasogenic edema (intermediate signal intensity on DWI, black arrow, due to the mass). SWI (d) simplifies the diagnosis, making the large cerebral hematoma (the central mass, with marked hypointensity) readily evident. Subsequent to the onset of ischemia, the patient had suffered the complication of parenchymal hemorrhage within the area of the infarct. These scans were acquired at 3 T, with the SWI scan (d, in common with Fig. 64.1b) being the projection image.

Fig. 64.2 Depiction of an acute hemorrhage on FLAIR, enhanced T1, DWI and SWI.
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The outer lines of $k$-space contain information about the details within the image, the high spatial frequencies. For a heterogeneous object, there is some signal in this portion of $k$-space, although it is very small. Omitting data collection from this part of $k$-space causes truncation artifacts (see Chapter 102). Filling these data lines with zero values (as illustrated in Fig. 65.1b), will greatly improve the image appearance. Fig. 65.1a illustrates the situation in which all $k$-space lines are acquired in a 3D scan. In Fig. 65.1b, only the more central lines along the slice gradient axis are acquired. It would be incorrect to believe that images interpolated in this way are identical to what would have been obtained if the object had been scanned with a full $k$-space acquisition. On the other hand, zero filling, as implemented with the volume interpolated breath-hold examination (VIBE) technique, does improve partial volume effects. In other words, zero filling does not improve spatial or volume resolution, but it does reduce the artifacts caused by the shape and size of pixels in the image.

Fig. 65.2 presents images from a 1.5 T VIBE acquisition in a patient with multiple liver hemangiomas. With VIBE, contiguous thin section (2 to 3 mm) imaging is possible.
with a single 20-second breath-hold, providing coverage of a large field of view with an in-plane resolution of \(\sim 2 \times 2\) mm (at 3 T, voxel dimensions under 2 mm in all axes can be achieved). In clinical practice, VIBE is used for acquisition of dynamic post-contrast scans of the upper abdomen and MR colonography. It is typically implemented with fat suppression, using a 3D fast low-angle shot (FLASH) pulse sequence.

**Fig. 65.3** compares breath-hold post-contrast (a) 3.5-mm (slice thickness) VIBE (using a conventional gadolinium chelate) and (b) 6-mm 2D FLASH (using a small iron particle, Resovist, see Chapter 54) scans. A large hepatocellular carcinoma (HCC) is noted in the left lobe of the liver. Advantages of VIBE over 2D GRE sequences include the ability to provide thinner sections, fat saturation, and higher SNR all within a single breath-hold, and with the absence of gaps between slices due to nature of the 3D acquisition.
Diffusion-Weighted Imaging

An early subacute right middle cerebral artery (MCA) infarct (arrow) is illustrated in Fig. 66.1. Images were acquired with diffusion gradients applied in the phase, read, and slice directions and combined to form a single composite diffusion-weighted image (Fig. 66.1a), also known as a trace-weighted image. This image has both diffusion and T2 weighting. A parameter specific to diffusion-weighted imaging (DWI) is the \( b \)-value, which defines the strength of the applied diffusion gradient. Increasing the \( b \)-value increases the diffusion weighting and, thus, up to a point, the conspicuity of early infarcts, but with a concomitant decrease in signal-to-noise ratio. A typical \( b \)-value for modern 1.5 and 3 T scanners, for brain imaging, is 1000 sec/mm\(^2\). To obtain an image with only diffusion information, at least one additional scan has to be acquired, specifically using the same technique but without the application of diffusion gradients (\( b = 0 \)). This scan is then used in conjunction with the diffusion-weighted scan to calculate an apparent diffusion coefficient (ADC) map (Fig. 66.1b), in which there is no T2-weighting. A lesion with restricted diffusion, such as an early infarct (due to the presence of cytotoxic edema), has high signal intensity (SI) on diffusion-weighted images and low SI on the ADC map, the latter reflecting the low diffusion coefficient. The fluid-attenuated inversion recovery (FLAIR) scan (Fig. 66.1c) in this instance also visualizes the MCA infarct well, but for a different reason, the prolongation of T2 due to the presence of vasogenic edema.

Vasogenic edema only begins to occur sufficiently for detection by MR 8 hours after ictus, with 90% of patients demonstrating an abnormality at 24 hours. Thus, T2-weighted scans can miss very early infarcts. Illustrated in Fig. 66.2 is an infarct less than 24 hours in age, located in the posterior limb of the internal capsule (black arrow). It is well seen on the DWI scan (a), but poorly visualized on the FLAIR scan (b) due to the lack of vasogenic edema. DWI scans can also be very useful for identification of early infarcts upon a background of chronic ischemic changes, which also appear as abnormal high signal intensity on T2-weighted scans such as FLAIR.

**Fig. 66.1** DWI, ADC, and FLAIR images in an early subacute infarct.
DWI is a much more sensitive means of detecting acute brain ischemia than conventional MR techniques or CT. In the standard Stejskal-Tanner pulsing scheme, additional strong, pulsed gradients are applied on both sides of the 180° pulse in the three orthogonal axes using an echo planar type spin echo sequence. Alternatively, a double-echo 90–180–180 pulsing scheme can be employed, which reduces the influence of eddy currents. Sensitization of the spin echo signal to diffusion motion occurs during the diffusion gradient pulsing. The longer the wait is between pulsing, the greater the signal loss due to diffusion. In normal brain tissue, random water diffusion (Brownian motion) decreases the amplitude of the signal due to phase incoherence. With cytotoxic edema, there is no net change in water, only a change in its distribution. Within minutes following an ischemic insult, the amount of water within the intracellular compartment, where Brownian motion is restricted, increases. Phase coherence here is preserved and more signal recovered during the diffusion observation period, producing high SI.

One caveat in the common, current implementation of DWI scans is the use of echo planar imaging. This provides rapid scans with relative insensitivity to patient motion; however, spatial distortion and image degradation occur due to differences in the magnetic susceptibility of tissues. This is particularly evident at the interface between brain and the air-filled sinuses. For example, note the spatial distortion anteriorly in Fig. 66.2a, along with the artifactual high signal intensity (white arrows), due to the frontal sinuses. Techniques used to reduce these artifacts are discussed in Chapter 67. Furthermore, until the advent of 3 T, DWI scans were typically acquired with a low-resolution (128^2) matrix (due to both gradient limitations and poor SNR), with resultant visual blurring when compared with higher matrix size conventional scans. At 3 T, brain DWI scans are usually acquired using higher matrices, such as 192^2 or 256^2.

Hyperintensity on DWI, in and of itself, does not always correspond to a low diffusion value (as seen with cytotoxic edema, in the setting of acute infarction). Hyperintensity on DWI may be caused by “T2 shine through,” with the change in signal intensity reflecting the T2, rather than diffusion, weighting of the image. To clarify the ambiguity inherent in DWI, an ADC map, in which there is no T2 contribution, is utilized. Acute (< 24 hours) and early subacute (1 to 7 days) infarctions are hypointense on an ADC map (decreased diffusion) and hyperintense on DWI. Hyperacute (< 6 hours) infarcts are also hyperintense on DWI (and hypointense on the ADC map), but isointense on T2-weighted scans (due to the presence of cytotoxic edema, without sufficient time for vasogenic edema to occur). After about a week (late subacute), normalization on DWI and ADC maps may be seen, with only T2 hyperintensity present as evidence of an interval infarct.

Fig. 66.2  DWI vs. FLAIR, sensitivity and image artifacts.
Artifacts arising from motion-induced phase errors pose a particular challenge in diffusion-weighted imaging (DWI; see Chapter 66). These can be caused by coherent macroscopic or bulk motion such as CSF pulsation. This challenge has resulted in the use of rapid acquisition techniques, most commonly single-shot echo planar imaging (ss-EPI), to overcome ghosting artifacts from such phase variations in multi-shot sequences. However, the image quality of ss-EPI DWI suffers from relatively low spatial resolution, blurring and bulk susceptibility artifacts generated by tissue interfaces and metal implants. With increasing field strength, susceptibility artifacts and degradation of spatial resolution become more pronounced. Susceptibility artifacts with ss-EPI can be attributed to the inherently long readout time resulting from acquisition of the entirety of k-space for a given slice in a single shot (i.e., using a single RF excitation pulse). Different strategies exist for reducing readout time with ss-EPI, and thus the bulk susceptibility artifacts and image blur, in particular the use of parallel imaging.

To overcome the limitations of ss-EPI, several diffusion-weighted, multi-shot EPI (ms-EPI) sequences have been introduced that incorporate a phase correction to avoid artifacts caused by shot-to-shot, motion-induced phase errors. This correction is applied using data from the central region of k-space, which is acquired at each shot using a navigator echo. Utilization of navigator data to perform 2D nonlinear phase corrections is particularly effective. Sampling only a subset of k-space at each excitation with ms-EPI leads to shorter readout times and reduces artifacts from bulk susceptibility relative to ss-EPI. Conventional ms-EPI sequences acquire data by interleaving lines of k-space in the ky direction from different excitations. Although this approach is effective at reducing susceptibility artifacts, it presents some challenges for 2D phase correction procedures. These challenges relate to the failure of traditional ms-EPI acquisition schemes to fulfill the Nyquist sampling condition, causing some of the data to be aliased in the image domain. New techniques, such as readout-segmented EPI (rs-EPI) (also called RESOLVE, or REadout Segmentation Of Long Variable Echo trains), have been developed to overcome this problem. Rs-EPI uses a modified k-space sampling scheme in which the readout (kx) direction is divided into “readout segments” that cover the full extent of k-space in the ky direction (Fig. 67.1).

Each readout segment is acquired as a separate shot
and, in each case, an additional 180° pulse is used to generate a second spin echo. This additional spin echo is used to sample 2D navigator data from the center of k-space using a second EPI readout. The reduced number of columns in the readout (kx) direction allows fast gradient reversals, thereby reducing echo spacing, overall readout time, and susceptibility artifacts. The data points from each shot of the rs-EPI sequence are contiguous in k-space thus fulfilling the Nyquist sampling condition, and making it possible to use the navigator data to apply a non-linear, 2D phase correction in the image domain without complications from aliased signal contributions.

Compared with ss-EPI, rs-EPI increases scan time due to the larger number of RF excitations (or shots) that are required to sample the k-space data required for each image. However, this increase in scan time is accompanied by a decrease in image blur, bulk susceptibility artifacts (arrows, Fig. 67.2), and geometric distortion (Fig. 67.3).

Fig. 67.4 shows multiple left MCA territory embolic infarcts as depicted on ss-EPI (a) and rs-EPI (b) acquired at equivalent nominal spatial resolutions. Image blur is markedly decreased in the rs-EPI image. Note the improved sulcal definition (arrow, Fig. 67.4b) with the rs-EPI sequence. At 3 T, where it is commonly employed clinically, rs-EPI provides improved image quality when compared to ss-EPI in brain, soft tissue neck, breast and prostate exams.
Diffusion-weighted imaging (DWI) is a critical tool for the MR assessment of ischemic changes of the brain and in the differential diagnosis of cystic brain lesions. Diffusion tensor imaging (DTI), which provides additional information beyond that available from conventional DWI, is an important, noninvasive method for studying white matter structure of the human brain in vivo. A variant of DTI is diffusion spectrum imaging (DSI), which is sensitive to intravoxel heterogeneities caused by crossing fiber tracts, and allows improved mapping of axonal trajectories.

The basis of the signal measured in a diffusion-weighted MRI experiment is the Brownian motion of water protons, specifically the random movement of particles based on their thermal energy. This movement is also referred to as diffusion; hence, the technique is named diffusion-weighted imaging. In a static, homogeneous magnetic field, all protons precess with the same frequency. If a gradient (a temporary magnetic field) is turned on, this homogeneity can be disturbed in a known fashion and the signal from the object in the scanner decreases. This first step of diffusion preparation is called dephasing. If this gradient is followed by a gradient with opposite direction, the signal can be restored, the so-called rephasing. If protons move between dephasing and rephasing, the restoration of the signal is incomplete. The more movement, the more signal loss. In this fashion, MRI can be sensitized to diffusion. The strength of diffusion weighting is dependent on the gradient strength and the duration of and time between the dephasing and rephasing gradients, and is summarized in the so-called b-value. The higher the b-value, the stronger is the diffusion weighting. Typical b-values for DTI of the brain vary between 600 and 1000 sec/mm², with clinical DWI performed usually with a b-value of 1000.

As can be seen in Fig. 68.1 (b-values of 0, 300, 600, and 900 are illustrated in a normal volunteer), the stronger the diffusion weighting, the greater the signal loss in areas with strong free motion (e.g., the ventricle). The apparent diffusion coefficient (ADC) is calculated by measuring the signal loss when comparing the non-diffusion-weighted image with the weighted image. To calculate a direction invariant ADC, the diffusion weighting is performed in three main axes. Thus, to obtain an ADC map, a minimum of four datasets (one non-weighted and three weighted) are required.
In healthy parenchyma of many organs and in the gray matter of the brain, diffusion is non-direction dependent. This means that diffusion, measured in any direction, yields the same value. This direction independent diffusion is also called isotropic diffusion. In white matter and muscle, however, the measured diffusion is direction dependent. This is due to the longitudinal intrinsic structure within the tissue. In white matter, axonal tracts restrain the movement of water protons perpendicular to the axonal direction but enable increased movement along the axonal tracts. This can be seen by inspecting individual diffusion-weighted images. If a gradient is applied along a certain tract, signal loss will be large in this area, whereas when applied in a direction perpendicular to this tract, signal loss will be minimal. This effect is readily evident in large white matter tracts, such as the corpus callosum and internal capsule, as illustrated in Fig. 68.2 in a patient with a large, early subacute, left middle cerebral artery territory infarction. The diffusion gradient has been applied in the craniocaudal direction (a), left-right direction (b) (note the low signal intensity of the genu and splenium of the corpus callosum [arrows]), and in the anteroposterior direction (c). The trace image, the sum of the diffusion (x, y, and z) tensors is illustrated in (d).

Fig. 68.2 MCA infarct. Separate application of a diffusion gradient (in each of the three orthogonal directions), together with the trace geometric average image.
To describe this anisotropic diffusion, a mathematical model for the description of the water displacement, a so-called tensor, is used. To fully determine this tensor, a minimum of seven independent datasets (compared with four in DWI) are needed. This includes one non-weighted and six diffusion-weighted images, the latter in non-collinear directions.

Imagine a densely-packed fiber bundle like the corpus callosum. Here, the movement perpendicular to the tract is almost zero whereas the diffusion along the tract is large. The signal loss in the direction of the tract is large (free motion) and perpendicular to the tract it is minimal (restricted diffusion). Thus, the proportion of water displacement along, to the proportion of displacement perpendicular to, the tract is large. This proportion is called fractional anisotropy (FA) and ranges from 0 to 1. In gray matter, water displacement in all directions is equal and the FA is 0. In a highly-organized tract like the corpus callosum, the FA can reach values as high as 0.85. As can be seen in Fig. 68.3a, FA maps show highest intensity in highly organized fibers whereas lower intensity can be found in subcortical C fibers. In gray matter or CSF, the value approaches 0. The FA can be used as a measure of fiber integrity and is reduced in neurodegenerative diseases and due to infiltration by brain tumors.

Because the direction of largest signal loss is equivalent to the predominant axonal direction in a voxel, this direction can be used to visualize fiber direction. In color maps (Fig. 68.3b), the direction of fibers is coded on a 2D map. By convention, fibers running left-right are coded red, fibers running anteroposterior are coded green, and fibers running craniocaudal are coded blue. Thus, in one image, fiber integrity and direction can be appreciated. To obtain a 3D visualization of the fiber tracts, vectors can be connected on a voxel by voxel basis. This process usually involves the definition of a start ROI and a target ROI and fibers running through both ROIs will be included in the final result. This method enables the visualization of known fiber tracts (Fig. 68.4) and their displacement or diminution due to tissue or structural abnormalities (Fig. 68.5).

![Image](image-url)
In conclusion, DTI provides a measure of fiber integrity and enables the 3D visualization of fiber tracts. It is applied in neuroscience studies of brain connectivity, often in combination with fMRI. Its clinical application is limited in part due to long processing times and user dependency of fiber tracking results, with ongoing research and development in these areas. With current automated software packages, DTI has been applied clinically in surgical planning and may have potential as well in therapy monitoring.

Fig. 68.4 Normal volunteer. High resolution, whole brain, tractography.

Fig. 68.5 Sturge-Weber syndrome. Diffusion tractography reveals unilateral degeneration of the arcuate (blue) and pyramidal (green) tracts, with the inferior fronto-occipital fasciculus (orange) largely intact. Subsequent hemispherotomy for intractable epilepsy was successful without aphasic deficits, coronal post-contrast T1-weighted scan. Courtesy of Andreas J. Bartsch, Radiologie Bamberg, Germany.
Blood oxygen level-dependent (BOLD) imaging, a type of functional MR imaging, exploits the decrease in magnetic susceptibility caused by small changes in the volume of oxygenated blood supplied to a specific region of the brain during and following increased neuronal activity. To understand the BOLD effect, a discussion of the fundamental aspects of susceptibility imaging as well as an explanation of the physiologic changes in the brain during activity is required.

Magnetic susceptibility (see Chapter 113) relates to the ability of a material to become magnetized within an externally applied magnetic field and is measured by the magnetization of the material divided by the field strength. Materials with a strong magnetic susceptibility are referred to as ferromagnetic, those with a weak magnetic susceptibility are known as paramagnetic materials, and diamagnetic materials have little to no effect on the localized magnetic field with respect to surrounding tissues.

During an MRI exam, ferromagnetic and paramagnetic materials within the body take on magnetic characteristics causing localized changes or inhomogeneity within the field. Protons in the vicinity are affected by this change and experience a greater phase shift after a spin excitation, increasing T2* decay and leading to a reduction in localized signal. In the presence of strongly ferrous materials, a complete decay of signal takes place leading to a signal void in the image.

Hemoglobin, a constituent of red blood cells that carries oxygen, displays varying levels of magnetic susceptibility based on the amount of exposed iron found within its molecular structure. When oxygen is present, the iron in hemoglobin is shielded, reducing its magnetic susceptibility effect and leading to oxyhemoglobin being classified as diamagnetic. As oxygen is consumed, the iron is exposed and takes on magnetic properties leading to deoxyhemoglobin’s paramagnetic effect. The presence of hemoglobin in both states throughout the microvasculature of the brain leads to a small, intrinsic level of magnetic susceptibility during an MR exam. The magnetic susceptibility effect of deoxygenated blood is well demonstrated by susceptibility-weighted imaging (Fig. 69.1). This technique (see Chapter 64) is designed specifically to exploit the paramagnetic properties of unshielded hemoglobin, highlighting for example normal draining veins.

Fig. 69.1 Deoxygenated blood, SWI.
The activation of neurons during sensory-motor activity increases the consumption of oxygen within a specific area, decreasing the level of oxyhemoglobin. However, the reduced oxygen triggers adjacent capillaries to transport surplus levels of oxygenated blood to the area to facilitate further activity, increasing regional cerebral blood flow (Fig. 69.2). The increased blood flow floods localized veins with oxyhemoglobin, leading to a small, localized decrease in susceptibility and increasing the MR signal. Both states, at rest (a) and during activation (b), are illustrated, with the activity in this example being finger tapping (for functional localization of primary motor cortex). With finger tapping, blood flow and oxygenated hemoglobin increase in the corresponding area of the motor cortex, leading to a signal change (albeit weak) that can be detected by MRI.

To visualize the subtle susceptibility change, sequences displaying a high level of sensitivity to magnetic susceptibility are used. Gradient echo-based sequences, especially those used for echo planar imaging (EPI) (see Chapter 42), use gradient polarity changes instead of RF pulses to rephase transverse magnetization and thus form an echo. This method leads to an increase in the viewable susceptibility changes by allowing the T2* effects to evolve during the pulse sequence. Magnetic susceptibility scales with field strength so higher field strengths (e.g., 3 vs. 1.5 T) improve the visualization of the BOLD phenomenon by up to a factor of 2. However, the negative effects of magnetic susceptibility in areas such as air-tissue interfaces increase as well and induce geometric distortions in the anatomy during EPI measurements. It is therefore important that the gradient hardware of MRI systems used to acquire BOLD data is efficient and powerful enough to produce high-quality images with limited geometric distortion. When multichannel coils are used, further reductions in geometric distortions can be made by incorporating parallel imaging techniques to reduce the negative effects of phase evolution in EPI based sequences.

The complexity of BOLD imaging techniques continues to increase, making it possible to interrogate new areas of cognitive processing never before seen. In the next chapter we discuss the application of BOLD imaging paradigms.
Blood Oxygen Level-Dependent (BOLD) Imaging: Applications

Variations in magnetic susceptibility induced by an increase in oxygenation within a localized area are very minute, as is the resulting SNR increase. To visualize the effect during an MRI measurement, the patient is asked to perform a series of tasks during the sequence acquisition to induce an alternating increase and decrease in the levels of oxyhemoglobin to a specific area of the brain. This increase and decrease follows a well-known pattern called the hemodynamic response function (Fig. 70.1).

The tasks are performed in a cycle called a paradigm specified in the sequence parameters prior to the sequence acquisition. By synchronizing the data acquisition and activity paradigm to the hemodynamic response within the brain, images can be acquired and sorted to depict the area of activation. For example, Fig. 70.2 illustrates finger tapping in a patient with an oligodendroglioma, WHO grade 2. The tumor lies just anterior to the precentral gyrus (primary motor area) in this instance. The fMRI exam provides direct visualization of the primary motor cortex, providing a roadmap for presurgical planning.

Following completion of data acquisition, complex further processing is required to portray the results of an fMRI study in a meaningful way. The difference in the mean signal of the activation and rest portions of the EPI measurement is evaluated statistically. The analysis assigns each pixel a level of significance based on the degree of real signal change. The results are then overlaid onto a T1-(or T2-) weighted image to provide the anatomic reference for analysis.

To provide an example of a common paradigm, bilateral finger tapping is subsequently described. This paradigm can be employed to display activation within the cerebral motor cortex as well as ipsilateral activation in the motor preparatory and timing area of the cerebellum. The paradigm consists of 10 right followed by 10 left hand activations performed over 60 measurements for a total scan time of approximately 3 minutes. The patient is instructed to touch the thumb to the fingers in a repeated series (e.g., 1–2–3–4–3–2–1, with 1 being the index finger and 4 being the fifth digit). The task is initiated with the start of the imaging sequence and begins with the right hand for the first 10 activations and left hand for the second 10 and so forth.

Fig. 70.1 The brain's hemodynamic response with activation.
In Fig. 70.3, activation of primary motor cortex (thin arrow) just superior to an oligodendroglioma (arrow) during finger tapping (movement of the right fingers in red, left in green) is depicted in the first image. Activation of the motor (Broca) and sensory (Wernicke) speech areas (thin arrows) are illustrated in the middle image, anterior and posterior to but distant from the tumor (arrow). DTI, the third image, shows deviated and thinned arcuate (U) fibers (thin arrows), which connect the Broca and Wernicke areas, indicating infiltration.

Image acquisition at 3 T, as opposed to 1.5 T, improves visualization of the BOLD effect. BOLD imaging has proven to be a valuable clinical tool for analysis and mapping of the sensory-motor portions of the cerebral cortex.

Fig. 70.2 Finger tapping, BOLD.

Fig. 70.3 Oligodendroglioma, activation studies (fMRI) and tractography (DTI). Courtesy of Bernhard Schuknecht, Bethanien Hospital, Switzerland.
Proton Spectroscopy (Theory)

The visual assessment of mass lesions is accomplished in a variety of ways following an MR exam. A few examples of these techniques include the displacement of normal anatomy, the difference in molecular mobility affecting the relaxation times and thus image contrast, and finally the enhancement pattern following contrast administration. In addition to viewing anatomic structure, MR offers the possibility to “visualize” the chemical environment via MR spectroscopy, examining the metabolism of areas in question. Fig. 71.1 depicts a lesion (a) examined with a routine imaging sequence, in this instance FLAIR (b), followed by a spectroscopic assessment. The results of the spectroscopy measurement (c, d) demonstrate the elevated level of choline and decreased level of N-acetylaspartate (NAA) typical of malignant lesions. This information can be used to further clarify ambiguous imaging findings.

The two common approaches in MR spectroscopy, to sample tissue, are single voxel spectroscopy (SVS) and chemical shift imaging (CSI). For SVS techniques, the two main acquisition schemes are spin echo (SE) and stimulated echo acquisition method (STEAM).
**Fig. 71.2** illustrates the basic concept of SVS using a spin echo acquisition technique. The 90° pulse excites a slice. The slice selective 180° pulse refocuses the transverse magnetization only in a row of tissue within the slice. A second 180° pulse then refocuses only the magnetization within a column of the row. Thus, only the signal originating in a single voxel remains. For proton spectroscopy, it is essential to suppress the water signal and, for regions outside the central nervous system, the lipid signal ([1] in **Fig. 71.2** marks the binomial RF pulse used to suppress either the water or lipid signal, see Chapter 46).

Spectroscopic results are displayed using a parts-per-million scale that removes field strength dependence. However, increasing the field strength (e.g., changing from 1.5 to 3 T) improves the spectral separation of the metabolic signals and, therefore, improves the results of these measurements. Additionally, the metabolites in question are orders of magnitude lower in concentration than water resulting in very low SNR and long acquisition times. The higher SNR of 3 T systems further improves the results of these measurements by reducing acquisition time.

**Fig. 71.3** demonstrates the ratio of the water signal to the metabolites in question (a). As mentioned, the water signal is suppressed during spectral measurements and the metabolite signals are scaled to aid the analysis (b). There are a variety of benefits to spectroscopy including lesion characterization, evaluation of tumor borders, and assessment of the effectiveness of treatment such as radiation therapy.

**Fig. 71.2** Pulse diagram for SVS.

**Fig. 71.3** Water suppression, for enhanced detection of metabolites.
The major healthy brain metabolite peaks that are seen on long TE spectra include N-acetylaspartate (NAA) at 2.02 ppm, choline (Cho) at 3.20 ppm, and creatine (Cr) at 3.02 ppm and 3.9 ppm (see Chapter 71). Variable TE values in the sequence provide the ability to control the T2 “contrast” of the spectral peaks in the same way tissue T2 contrast is controlled in 2D FSE MR imaging. Short TE measurements are important for the detection of metabolite signals that have a short T2 decay and are not visible on long TE spectra. These include myo-inositol (ml) at 3.56 ppm, glutamine and glutamate (Glx) between 2.05 and 2.5 ppm and 3.65–3.8 ppm, and glucose at 3.43 ppm. These short TE metabolites are also critical in the assessment of the developing brain in pediatric imaging.

A major limitation of SVS (see Chapter 71) is that variations in metabolic ratios within the single acquired voxel cannot be assessed. For a complete evaluation, the SVS measurement must be repeated numerous times to properly interrogate the lesion, its surrounding tissue, and contralateral regions of normal tissue. Because these measurements can take from 5 to 7 min each to acquire, the SVS technique can, in some cases, be prohibitively long.

Multivoxel techniques, referred to as chemical shift imaging (CSI), offer the significant benefit of a large segmented voxel acquired in one 5- to 7-minute measurement. This segmented voxel is normally large enough to cover the entire lesion as well as surrounding tissues and tissues outside the lesion area for a complete spectral assessment, in an acceptable acquisition time. Fig. 72.1 depicts the normal coverage of a CSI acquisition as well as the individual encoded partitions of the voxel represented by the black box. The additional grid lines outside the black box represent oversampling to eliminate aliasing similar to the oversampling used in routine imaging techniques.

Fig. 72.1 Specification of anatomic coverage for 2D chemical shift imaging.
The frequency-encoding gradient is omitted in localized spectroscopy, but the phase encoding gradient can be utilized, as in imaging, to encode spatial information into the signal. Fig. 72.2 is an illustration of a simple 2D chemical shift imaging (CSI) acquisition scheme sampling the free induction decay. A selective 90° RF pulse, in the presence of a magnetic field gradient, generates transverse magnetization within the slice. Orthogonal magnetic field gradients of short duration will encode spatial information into the signal to be acquired.

Fig. 72.3 presents the spectra of a low-grade brainstem glioma. Data were acquired with a SE 2D CSI acquisition scheme, providing two spectra with echo times of 30 msec (second column) and 144 msec (third column). The lesion spectra (first row) demonstrate decreased NAA (a marker of neuronal integrity) and increased choline (a marker of myelin breakdown). The second row image and spectra depict an adjacent area of tissue presumed to be unaffected by the lesion. The spectral results confirm the higher NAA and decreased choline signal characteristic of normal tissue. Spectroscopic results can also be displayed as metabolite or metabolite ratio maps, superimposed upon a conventional T1- or T2-weighted image (e.g., the displayed choline map).

Fig. 72.2 Pulse diagram for 2D CSI.

Fig. 72.3 Short and long TE voxel spectra, and the choline map.
Simultaneous Multislice

When MR was initially introduced in the early 1980s, acquisition of only a single slice was possible. Rapidly thereafter, 3D techniques were developed, and subsequently 2D multislice imaging. The latter represented a major advance for clinical application, providing multiple images in the same time previously required for a single slice. Multislice imaging has since then become a mainstay for clinical MR, being used in most clinical 2D acquisitions. The name “multislice,” however, can be somewhat misleading. With this technique images are not actually acquired simultaneously, but rather each slice is acquired sequentially within a very short span of time. True simultaneous multislice (SMS) imaging was conceptualized much later, with a viable approach for clinical implementation presented only recently (Fig. 73.1).

SMS represents likely the single most significant advance for clinical MR in the current decade, making possible in its initial implementation scan time to be further reduced by a factor of 2 to 3. It should be noted that implementation of SMS would not be possible without substantial prior advances in both hardware and software, including multicoil arrays, CAIPIRINHA and slice-GRAPPA reconstruction.

The interest in the use of SMS is based upon the increase in temporal efficiency that can be achieved with this parallel imaging-based sequence. The availability of SMS has already fundamentally changed the scope of studies that can be realistically acquired clinically both with EPI and FSE sequences. Equally or perhaps more important in the current clinical environment is the application of SMS to shorten scan times and improve patient throughput.

For diffusion-weighted scans, rapid single-shot 2D SE EPI sequences are commonly used. As conventionally implemented, 2D SE EPI is highly inefficient time-wise, because diffusion encoding, which represents a significant portion of the acquisition time, is performed for the whole imaging volume with each single 2D imaging slice.

Fig. 73.1 Slice acquisition and reconstruction using SMS.
excitation. Parallel imaging can be, and typically is, used. However, the application of parallel imaging in this instance does not provide significant acceleration to the scan, because it only shortens the EPI encoding period. The use of parallel imaging in this application is important, regardless, for reducing image distortion and blur. Simultaneous multislice on the other hand allows for concurrent acquisition of multiple slices, with scan time reduced by the slice acceleration factor (mitigated to a small extent by the time needed for the fast reference scan, the latter required for slice separation during image reconstruction). Unlike with parallel imaging by itself, there is no SNR penalty related to the acceleration, which with parallel imaging occurs due to acquisition of a reduced number of phase encoding lines (SNR being proportional to $1/\sqrt{R}$, where $R$ is the acceleration factor). Data from multiple slices are acquired simultaneously with SMS. With fewer slice excitations needed to cover the anatomic volume, TR can be reduced and thus the scan accelerated.

Critical to the implementation of SMS is the unaliasing of the simultaneously acquired yet closely spaced slices. Use of a blipped CAIPIRINHA approach allows sufficient interslice image shifts between aliased voxels in the phase encoding direction, avoiding the high $g$-factor (SNR) penalty and blurring (“tilted voxel”) artifact associated with previous approaches. Important as well has been the implementation of specialized reconstruction techniques to reduce the leakage signal contamination between simultaneously acquired slices.

Of particular importance for the application of SMS is the design of the multiband RF pulse, which specifically allows for simultaneous excitation and refocusing of multiple slices. Application of such multiband RF pulses increases Specific Absorption Rate (SAR). At 3 T, use of low SAR variable-rate selective excitation (VERSE) pulses—introduced in the early 2000s and subsequently widely used at 3 T—provides adequate SAR reduction.

Subsequent to the development of SMS for ss-EPI, attention turned to the implementation of this technique for diffusion-weighted readout-segmented EPI (rs-EPI, RESOLVE). Although ss-EPI continues to be widely used, this technique suffers from geometric distortion and blurring, the latter due to $T2^*$ decay. These two factors limit the use of ss-EPI at high field strengths, where improved spatial resolution would be a major goal. Geometric distortion and blurring are markedly reduced with rs-EPI compared with ss-EPI, due to the acquisition of subsets of $k$-space during multiple repetition times (TRs) rather than encoding the whole of $k$-space in a single shot. Visualization of areas prone to bulk susceptibility artifacts, particularly the interfaces between the paranasal sinuses and brain, is markedly improved.

An important limitation of rs-EPI, however, is that scan time is increased compared with ss-EPI. Thus, techniques to accelerate the sequence are important for clinical use, not only for routine screening examinations but also for advanced applications such as DTI and tractography, with a major role for SMS.

Regardless of whether ss-EPI or rs-EPI is used, SMS can be used in clinical DWI exams to reduce acquisition time, increase slice coverage, increase the number of diffusion encoding directions or make possible higher resolution scans not previously feasible (due to long scan times). Applications include all body regions where DWI is employed clinically today. Diffusion tensor imaging (DTI) and fMRI also benefit from the implementation of SMS.

With SMS, because multiple slices are excited simultaneously, the TR for the desired spatial coverage can be reduced. For example, a typical TR at 3 T using ss-EPI DWI for coverage of the whole brain in the axial plane (34 slices using a 4-mm slice thickness) is 6.3 sec. This can be reduced to 3.5 sec, without any substantial loss in SNR and while
maintaining slice coverage, using an SMS acceleration factor of 2. The decrease in scan time is directly proportional to the reduction in TR (not considering the small amount of time required for the additional reference scan). In the application with ss-EPI, the resultant time savings is small given the short scan times involved (1 to 2 minutes), when used in a screening fashion for the brain at 3 T with thick (4 mm) sections. However, in application for high-resolution, thin section imaging, the time savings can be substantial (Fig. 73.2).

In Fig. 73.2 high spatial resolution (0.6 × 0.6 × 1 mm) conventional (a) and SMS (b) acquisitions are compared in a patient with multiple, small, bilateral subacute infarcts,

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**Fig. 73.2** Increased slice coverage using SMS with thin section ss-EPI DWI.

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**Fig. 73.3** Breast cancer on T2 FSE, zoomed ss-DWI, and SMS rs-DWI.
with equivalent scan times. Both scans depict a tiny right caudate head lacunar infarct and a small right parietal watershed infarct (arrows). Coronal reformatted images (c: conventional, d: SMS) demonstrate the increased slice coverage with SMS, allowing depiction of the cerebellum and two small lacunar infarcts (asterisk) therein.

SMS EPI—with an acceleration factor of 2—today represents the standard implementation of DWI for 3 T of the brain, whether used as a screening examination with thicker (4 mm) sections, achieving scan times on the order of 2 minutes (with rs-EPI), or applied for whole-brain thin section imaging, with scan times of 4 to 6 minutes (for ss-EPI).

DWI today is also performed routinely for body imaging with SMS. Although thinner sections or higher resolution could be achieved, given the effects of motion the focus has been to decrease scan time. For example, in Fig. 73.3 a primary breast cancer (black arrow, a) is well depicted on (c) SMS rs-DWI, together with detection of a metastatic node (white arrow), with scan time nearly halved compared to (b) ss-DWI.

SMS can also be applied to FSE techniques (whether PD-, T1-, or T2-weighted). Immediate applicability is to scans in which two or more concatenations are used for slice coverage. Major applications exist in brain, spine, and musculoskeletal imaging. In Fig. 73.4 depiction of an ACL rupture (arrow) and marrow edema (asterisks) is equivalent without and with SMS, despite scan time being halved. As with other areas of the body, the decision tree becomes whether to use SMS simply to decrease scan time, or to provide higher resolution images within a more acceptable scan time.

![Fig. 73.4 Comparison of conventional and SMS scans of the knee in a trauma patient.](image)
Section V

Flow
Flow Effects: Fast and Slow Flow

Fig. 74.1 presents images from two separate scans in the same patient, varying only the number of slices acquired. (a) Images were acquired on both sides of the slice illustrated. In (b) fewer slices were acquired, more specifically the scan illustrates the outermost slice. Thus, despite identical scan technique in terms of T1-weighting, Fig. 74.1b illustrates inflow effects (flow-related enhancement) in the cortical veins (arrows) draining into the sagittal sinus, which are absent in Fig. 74.1a.

Flow is an intrinsic contrast mechanism in MR, analogous to some degree to T1, T2, and proton density. The signal intensity of flow depends on the pulse sequence selected, the velocity, the direction of flow (and whether it is in-plane or through-plane), and the presence or absence of turbulence, as well as the method in which the slice or volume of data are acquired. A spin echo pulse sequence consists of a 90° RF pulse followed by a 180° pulse. Both pulses are slice selective and thus any tissue or substance within the slice that receives both pulses can produce an MR signal (given the proper TR and TE). Blood is no exception. If the blood flowing into the slice has not been presaturated by an RF pulse outside the imaging volume, and it is flowing slowly enough to receive both the 90° and 180° pulse, then it can have high signal intensity (flow-related enhancement). This is well demonstrated by the high signal intensity seen in the cortical veins in Fig. 74.1b. It should also be noted that flow-related enhancement is more prevalent when blood flows perpendicular to the slice. On the other hand, if the blood is flowing fairly rapidly so that it does not receive both RF pulses, or if it has become saturated as it flows through a multislice data-set or volume, then it will have little (Fig. 74.1a) to no signal intensity, the latter being a flow “void.”

Fig. 74.2 is from a pediatric patient with stenosis (obstruction) of the cerebral aqueduct, resulting in longstanding enlargement of the third and lateral ventricles. Presented in Fig. 74.2a and Fig. 74.2b are sagittal T1-weighted spin echo scans. Flow (signal) voids were seen in the basilar artery (arrow, Fig. 74.2a), cavernous carotid artery (arrow, Fig. 74.2b), and in the superior sagittal sinus and draining cortical veins (on an axial T2-weighted fast spin echo scan, not shown).

As illustrated, flowing blood (arterial or venous) can produce either a signal...
void or high signal depending on (in particular) flow velocity, vessel orientation relative to the slice, and, most importantly, the pulse sequence selected. In general, spin echo sequences depict flowing blood as a signal void (with some exceptions as noted). Vascular imaging using spin echo-based scans is sometimes referred to as “black blood” or “dark blood” technique. Gradient echo sequences depict flowing blood as high signal, particularly when acquired in a single-slice fashion (or as a single “slab” as in 3D time-of-flight magnetic resonance angiography [TOF MRA]) to take advantage of flow-related enhancement. Images of blood flow acquired in this manner are referred to as “bright blood” technique and are the basis for TOF MRA.

Recognition and appropriate interpretation of flow phenomena are important in clinical practice. An example is provided in Fig. 74.3, which illustrates axial T2-weighted fast spin echo scans in a young woman who presented with headache, nausea, and vomiting, due to thrombosis of the superior sagittal sinus (with the etiology being oral contraceptive use). Major cerebral veins typically demonstrate flow voids on fast spin echo T2-weighted scans, with the absence of a normal flow void (arrow) in Fig. 74.3a due to clot (which is also slightly expanding the sinus). Following anticoagulation, there was recanalization of the sinus, with the MR scan one month later (Fig. 74.3b) demonstrating a normal flow void. Flow phenomena can be very confusing, with one suggestion—to determine if flow is present (on nonangiographic sequences)—being imaging in two orthogonal planes with the same exact scan sequence. If the signal intensity within the vessel of interest is the same on both scans (and not a flow void), then occlusion is likely.
Phase imaging, as applied to visualize flow, permits the differentiation of stationary spins from moving spins. Recall that the Larmor frequency of spinning protons depends on the strength of the applied magnetic field and that the frequency increases with increasing field strength. Assuming that a proton is traveling along a magnetic field gradient, it is continuously exposed to different field strengths and subsequently the frequency of the spin changes. In contrast to stationary protons, which precess with a constant frequency, moving protons thus experience a phase shift while traveling along the applied magnetic field gradient. This phase shift increases linearly with the velocity of the protons, so that the amount of phase shift indirectly encodes the velocity of the protons. The measurement of these phase shifts can be used to visualize flow, to calculate flow velocities, or to display an MR angiogram.

Phase imaging is performed by applying velocity (flow)-encoding gradients, which can be adjusted in strength and duration according to the velocity of the traveling protons (i.e., blood or CSF). This so-called velocity-encoding (VENC) factor should always be slightly larger than the flow velocity being measured. Due to local magnetic field inhomogeneities, however, stationary protons may also experience phase shifts. To overcome this error, two sequences are acquired, one with flow encoding and one with flow compensation, the latter being a technique to suppress flow effects by applying additional nulling gradients. The subsequent subtraction results in a zero phase shift for stationary protons and allows for background suppression. This is also advantageous for observation of the phase shift of flowing spins, as there is no alteration by background noise, so that only net phase shifts contribute to velocity encoding.

Fig. 75.1 presents the pulse diagram of a flow quantification or phase contrast acquisition scheme. Two sequences, one flow compensated and the other flow sensitized, are executed in an interleaved fashion. For tissues or fluids moving with a constant velocity, a three-lobe gradient structure will ensure that the phase is null at the echo for
both moving and stationary tissue (1). Such a sequence is termed flow-insensitive or flow compensated. “Detuning” the gradient arrangement (2) for the slice selective gradient (GS) results in through-plane “flow sensitivity.” The transverse magnetization within fluids that are moving will have a different phase position (2) compared with that measured with flow compensation (1). The difference in phase, Δφ, is directly proportional to the velocity of the flow and can therefore be used for quantification. The magnitude of the “vector” ΔM between the two magnetizations (1 and 2 in Fig. 75.1) demonstrates the presence of flow regardless of the direction and velocity and is used in phase-contrast MR angiography.

Fig. 75.2 presents images from the MR exam of a 42-year-old woman suffering from Takayasu arteritis. The contrast-enhanced MRA exam in this patient is presented using both (a) maximum intensity projection (MIP) reconstruction and (b) volume rendering technique (VRT). Note the distinct irregularities of the aortic wall together with a high-grade stenosis of the proximal left subclavian artery (arrow, a). Fig. 75.2c,d presents axial phase-contrast images at the level of the lower neck depicting the cervical vessels: (c) corresponds to the magnitude image in which any vessel with flow is depicted as high signal intensity, while (d) is the phase image in which additional information regarding flow direction is presented. In this “through-plane” phase contrast sequence, different flow directions are depicted with different pixel intensities. Vessels with flow toward the brain (e.g., the carotid arteries) are hyperintense (bright), whereas the right jugular vein with blood flowing in the opposite direction is hypointense (dark). Note that the right vertebral artery is hyperintense and the left vertebral artery hypointense (arrow, d), corresponding to a reversal of flow direction in the latter vessel. This is a case of subclavian steal secondary to severe arteritis. Although the magnitude image in (c) only provides information regarding whether flow is present or not, the phase image in (d) provides additional information regarding the direction of flow.

Fig. 75.2 CE-MRA and phase imaging in subclavian steal.
A two-dimensional (2D) spoiled gradient echo pulse sequence is typically used for 2D time-of-flight (TOF) MR angiography (MRA). Slices are acquired sequentially, in a single slice mode, one slice at a time. A short repetition time (TR) is employed together with a high flip angle. This results in reduced signal from background (stationary) tissue, as there is insufficient time for recovery of longitudinal magnetization (T1 relaxation). On the other hand, protons moving into the slice (flow) have full longitudinal magnetization, as they have yet to be exposed to an RF pulse. The net result is that blood flowing into the slice has high signal intensity. The high signal resulting from inflow of unsaturated blood protons is referred to as “flow-related enhancement” and is the basic contrast mechanism in a TOF acquisition. If arterial flow is principally along one axis, and venous return in the opposite direction (e.g., in the neck, with flow to the brain via the carotid arteries and return via the jugular veins), saturation pulses can be used to eliminate the signal from flow in one direction. For example, to eliminate the signal from the jugular veins, an additional RF (saturation) pulse is applied superior to the axial slice. This presaturation pulse (see Chapter 118) moves together with the axial slice as each subsequent slice is acquired. Blood flowing in the craniocaudal direction (in the jugular veins) is thus saturated as it flows into the imaging slice and produces no MR signal. This approach was commonly used prior to the advent of contrast-enhanced MRA to image the carotid bifurcations.

The axial images from a 2D TOF acquisition (source images) in a patient with a severe stenosis just distal to the origin of the left internal carotid artery are shown in Fig. 76.1a. To view the images in a more convenient and familiar fashion, the images are “stacked”
and a postprocessing technique known as maximum intensity projection (MIP) employed to produce “angiographic-like” projection images (Fig. 76.1b). It is important to note that the MIP images are not an “angiogram” but rather a projection produced from the axial dataset. Any tissue or substance with a short T1-relaxation time (e.g., fat or methemoglobin), depending on the exact scan technique used, may be hyperintense on the axial dataset, and thus on MIP images potentially confused for flow. Note the two areas of signal misregistration (arrows) in Fig. 76.1b. This is due to patient motion during or between slice acquisitions in the sequential 2D scan. Although each individual slice is acquired in only seconds, the total scan time is typically 5 minutes or more. When interpreting scans, it is important to review both the MIP and source images. The latter provide a means of assessing surrounding tissues and anatomy, and often make easier identification of artifacts due to motion, signal loss (metal clips), and the presence of fat or methemoglobin (in the setting of intra-plaque hemorrhage).

A major drawback of 2D TOF MRA is that complex flow, such as that seen with turbulence, is not easily compensated for and can result in artifactual signal loss. Complex flow patterns specifically lead to signal loss in the region of a stenotic lesion, and thus overestimation of the degree of stenosis. Although the slice thickness of 2D TOF sequences is usually less than 2 mm, 3D acquisitions can be used to produce images with thinner slices, which together with the use of shorter TEs (as generally employed in 3D TOF MRA), greatly reduces the artifactual signal loss. 3D TOF techniques also do not demonstrate the misregistration artifact previously described.

Although no longer employed for carotid imaging, 2D TOF remains useful clinically to highlight flow, and continues to play a role in venography. In most applications, however, contrast-enhanced MRA (CE-MRA) has replaced 2D TOF due to markedly decreased scan times (minimizing the effects of patient motion) and relative insensitivity to complex flow patterns. The MIP image in Fig. 76.1c is from a CE-MRA acquisition. There is less artifactual signal loss in the area of stenosis, and identification of an ulcerated plaque (asterisk) is possible just proximal to the stenosis.

The clinical utility, as well as a pitfall, of using 2D TOF effects is illustrated at 3 T in Fig. 76.2. Flow-related enhancement in (a) a fat-suppressed, axial T1-weighted GRE scan just above the level of the carotid bifurcation leads to depiction of the internal and external carotid arteries (small black arrows) with high signal intensity. However, when a saturation pulse is placed inferiorly (b), the signal in the right internal carotid artery is not suppressed (white arrow), demonstrating this to be a methemoglobin thrombus (not flow) due to recent occlusion. (c) The source image from a dedicated 2D TOF MRA scan at the same level, due to the use of a different sequence type, does not depict the clot as high signal intensity, displaying only true inflow.

![Fig. 76.2 Methemoglobin clot mimicking flow on a 2D T1-weighted GRE scan.](image)
A 7 × 9 mm saccular aneurysm (arrow) is noted arising at the left middle cerebral artery (MCA) bifurcation on a maximum intensity projection (MIP) image from a 3D time-of-flight (TOF) MRA of the circle of Willis at 3 T (Fig. 77.1). An axial source image from the MRA scan is illustrated in Fig. 77.2a, together with (b) coronal and (c) sagittal images reformatted from the original axial acquisition, all depicting the aneurysm (white arrow, a). Note that the aneurysm incorporates multiple MCA branches, and thus is not amenable to endovascular treatment (coiling).

Flow is an intrinsic contrast mechanism that can be imaged with MRI in several ways. Spin echo (SE) sequences utilize a 90° RF pulse followed by a 180° RF pulse to produce an echo (the observed signal). In standard imaging sequences, both the 90° and 180° pulses are slice selective. In typical vessels with normal flow velocities, blood is not in the slice to receive both RF pulses, and thus flow is seen as a signal void. Gradient echo (GRE) sequences utilize only a single RF excitation pulse and the echo is formed by a gradient magnetic field reversal. In this situation, blood that flows into the slice is hyperintense. This is particularly true when a short TE is used.

TOF MRA uses GRE sequences to depict flowing blood as bright signal. TOF sequences can be acquired in either a 2D (slice-by-slice) or 3D (volume excitation) fashion. A 3D sequence allows for thinner slices (higher spatial resolution). Small voxel size

Fig. 77.1 3D TOF MRA, MIP image demonstrating a MCA bifurcation aneurysm.
and a short TE are important for producing MRA scans with minimal signal loss due to voxel dephasing (from complex or turbulent flow). Additionally, 3D acquisitions can produce images with very high signal-to-noise ratio (SNR) despite the use of very thin slices. High SNR and spatial resolution are critical when imaging flow within the intracranial vasculature. In routine clinical practice today, the major use of 3D TOF MRA is for imaging of the circle of Willis (for aneurysms and vascular stenoses/occlusions). In an MRA sequence, the slices, or in this case the volume, are acquired with a fairly short TR. The background tissue becomes saturated and thus yields very little MR signal. Blood, on the other hand, when flowing at normal velocities, passes through the slice or volume and is not given time to become saturated. As fresh, unsaturated blood continues to flow into the slice or volume, it produces a strong MR signal. This is known as flow-related enhancement.

The source images acquired during a 3D TOF MRA scan (Fig. 77.2a) are then processed using one of several techniques, the most common today being MIP. Surface or volume rendering is also possible on most MR systems and represents an alternative postprocessing technique. Using MIP (or surface rendering), multiple projections, or views, are produced. Fig. 77.1 is one such projection. It is important to realize that the MIP projections are not the actual acquired images and may not always clearly delineate the pathology. As such, the source images should be reviewed as well. Given that the source images are acquired with a very small voxel size, these can be reformatted if desired in alternative planes, as illustrated in Fig. 77.2b,c. The band-like artifacts (black arrows, b, c) are the result of the scan actually being a multislab acquisition. Three thinner slabs are acquired as opposed to a single thick slab. This yields higher vessel contrast in the resultant final MRA, with good image quality and vessel visualization also achieved for a broader range of flow velocities (from fast flow in the young to slow flow in the elderly). The scan time for a 3D TOF MRA of the circle of Willis is typically 4 to 6 minutes, with a voxel resolution of 0.7 × 0.7 × 0.7 mm or better. At 3 T, the longer T1 times of tissue lead to improved background suppression, which in combination with the intrinsically higher SNR of 3 T lead to improved vessel depiction on 3D TOF MRA when compared with 1.5 T.

Fig. 77.2 Source image (axial), 3D TOF MRA, together with multiplanar reformats.
Three-dimensional time-of-flight (3D TOF) magnetic resonance angiography (MRA) continues to be the dominant MR technique for imaging the intracranial arterial vasculature. As discussed in Chapter 77, the contrast between flow and stationary tissue relies upon preferential saturation of the latter. This is more difficult to achieve when acquiring a 3D volume or slab, and is highly dependent on flip angle and TR. The longer blood remains in the imaging volume, the lower its signal intensity due to repeated excitation and saturation effects. At the point of entry, the blood is fully magnetized. Higher flip angles generate much larger signal and contrast than lower flip angles. But the rate at which the signal from flowing blood approaches a suppressed steady state also is much greater. Therefore, there is less depth penetration with higher flip angles, but flow will have greater signal at the point of entry. Conversely, with lower flip angles, the vessel signal and contrast at the slab entry are much less, but likewise approach a suppressed steady state much more slowly, thereby allowing for better depth penetration. With a longer TR, there is increased contrast at entry, but less depth penetration. With a shorter TR, there is decreased contrast, but better depth penetration. These are major tradeoffs in the selection of imaging parameters for 3D TOF MRA.

Fig. 78.1 presents MIP images from two circle of Willis 3D TOF MRA exams, comparing a flip angle of (a) 75° to that of (b) 25°. Saturation effects (arrows, Fig. 78.1a) are noted at the distal edges of the slabs (in this three-part multislab exam) when a too high flip angle is employed. To further complicate matters, techniques also exist in which the flip angle is varied across the imaging volume to more fully utilize the available magnetization and lessen saturation effects.

Magnetization transfer (MT) is a technique that can be employed to further increase the contrast between background tissue and flowing blood. A short explanation of MT follows (see also Chapter 38). There are two basic pools of hydrogen nuclei (protons) in the body. The “free” pool is that made up by water molecules. The “bound” pool is composed of hydrogen nuclei bound to macromolecules. These tightly bound protons tumble slowly and have an extremely short T2, and are thus not visualized by conventional MR because the signal decays too rapidly. Additionally, they resonate over a broad
frequency range. MT techniques use a special pulse to excite some of the “bound” pool, which then transfer their magnetization to the “free” pool. The net result is a reduction in signal of the latter. The amount of signal reduction depends on the makeup of the tissue with respect to the “bound” and “free” pool, as well as the method used to apply the MT pulse. The more macromolecules and long-chain protein molecules there are, the greater the effect of MT in that voxel. For brain, applying an MT pulse can reduce the signal intensity by up to 40%. Reducing the MR signal from background tissue can greatly improve vessel contrast on an MRA study. Fig. 78.2a was acquired without MT and Fig. 78.2b with MT (source images from the 3D TOF scans are presented). Note the reduction in signal intensity from brain due to the application of MT, leading to higher relative contrast for arterial vessels. One caveat for MT techniques is the relative lack of reduction in signal intensity for fat, making this more conspicuous and degrading MIP projections unless care is taken to carefully exclude fat (e.g., from the orbits) from the targeted reconstruction.

3 T provides a further major improvement in image quality for TOF MRA. This is due both to the reduced signal of stationary tissue, due to the longer T1 of brain, and better depth penetration, due to the longer T1 of blood. Without the need for MT, shorter TRs can be employed together with lower flip angles, the latter to increase depth penetration. Fig. 78.3 presents targeted MIP images of a multilobed MCA aneurysm acquired at 1.5 (a) and 3 T (b). In this case, the higher CNR available at 3 T has been used to improve spatial resolution. Two small branch vessels (arrows) arise from the aneurysm with the origin of the larger vessel not depicted and the smaller vessel itself not evident on the 1.5 T scan. The resolution along each axis was improved by a factor of two at 3 T, with equivalent scan times.
Phase contrast (PC) MRA techniques rely on flow-induced phase shifts to distinguish between flowing and stationary protons. Flow-encoding gradients are used to sensitize the image to flow. PC sequences can be acquired in either a 2D or 3D fashion. As blood flows along the direction of a gradient magnetic field from a lower to higher field, it increases in frequency (or gains phase) relative to stationary tissue. If flowing in the opposite direction, it decreases in frequency or loses phase. The amount of phase shift depends on the velocity and direction of flow as well as the amplitude and duration of the flow-encoding gradients. Because blood flow can be in any direction, depending on the orientation of the vessel, flow-encoding gradients are applied in either one or all three axes (right-left, anteroposterior, and/or craniocaudal). The choice of one or all directions is typically user selectable. Stationary tissue does not have any phase shift, and thus is not depicted on PC images. Two types of images are typically generated with phase contrast sequences: phase images, in which the pixel intensity (i.e., bright or dark) relates to the direction and velocity of flow (Fig. 79.1a,b) and magnitude images, in which the pixel intensity relates only to flow velocity (Fig. 79.1c). The images depicted are from a 2D PC acquisition, with a single slab of 20 mm acquired in 54 seconds.

Fig. 79.1 Phase vs. magnitude images.
One of the major advantages of PC techniques is the ability to choose the velocity to which the acquisition will be sensitive (i.e., slow or fast flow). The parameter selected by the operator is the velocity encoding (commonly referred to as VENC). The images displayed in Fig. 79.1 were acquired using a VENC of 40 cm/sec. With that VENC, blood protons flowing 40 cm/sec accumulate the maximum phase shift (–180° to +180°) and therefore are depicted with the brightest and darkest pixel intensity. Protons flowing slower than the VENC exhibit less pixel intensity. The magnitude images from the different flow-encoding directions can be combined to form a single image, the “magnitude sum.” The image in Fig. 79.1c is, in fact, a magnitude sum.

In certain situations, one may wish to ascertain the direction of flow. In phase images, blood protons flowing along the direction of the flow-encoding gradient are reconstructed as white, and those flowing counter to the encoding gradient are reconstructed as black. In Fig. 79.1a, the flow-encoding gradient is inferior to superior. As such, blood flowing within the anterior portion of superior sagittal sinus is bright. Blood flowing in the posterior portion is black because it is flowing counter to the direction of encoding. In Fig. 79.1b, the flow-encoding gradient is anterior to posterior. Thus, flow in the superior sagittal sinus, internal cerebral veins, and straight sinus (white arrows) is white (flowing posteriorly), whereas flow in the basilar artery (black arrow) is black (flowing anteriorly). It is important to note that if blood is flowing faster than the selected VENC, then it will experience > 180° of phase shift and will appear to the reconstruction algorithm to be flowing in the opposite direction (phase aliasing).

Fig. 79.2 displays images from a 3D PC scan acquired in slightly < 6 minutes, with sub-mm voxel dimensions and a VENC of 75 cm/sec in all three axes. Presented are 0.9-mm axial sections from this scan. In a PC scan, both flow-compensated and flow-encoded datasets are acquired, with the PC being the difference between the two. The flow compensated image (a) serves as a reference, with all spins rephased, and is thus also called a rephased image. Images (b) and (c) are the respective phase contrast images for two of the three velocity-encoding directions, with the flow-encoding gradient left to right in (b) and anterior to posterior in (c). Note that this leads to differentiation of flow direction in the right versus the left middle cerebral arteries (arrows) in (b) and differentiation of flow, principally anteriorly in the middle cerebral arteries (black arrow) in (c) from the posterior cerebral arteries (white arrow), in which flow is principally posteriorly. An alternative approach to acquiring a PC scan is to obtain two datasets with different flow-encoding, and then subtract the two, obviating the need for a flow compensated image.

Fig. 79.2 Rephased axial image (3D PC), together with LR and AP flow encoding.
4D flow MRI enables depiction of time-resolved 3D flow in-vivo. Using 3D phase contrast (PC) technique, velocity is encoded in all three dimensions throughout the cardiac cycle, providing time-resolved 3D data. The result is full volumetric coverage of the region of interest, with both spatial and temporal information (4D = 3D + time). Until recently, scan times were long with 4D flow MRI, up to 20 minutes. The integration of data sparsity techniques together with advanced respiration control today allows acquisitions within clinically acceptable scan times of 5 to 10 minutes. Ongoing methodologic improvements hold the potential for acquisition in a single breath-hold, together with coverage of a higher dynamic range for arterial and venous flow velocities.

3D streamlines, illustrated in the figures, are commonly used for the visualization of flow patterns with 4D flow technique. Benefits of the use of a 4D flow acquisition include the possibility for retrospective analysis of blood flow without being limited to the specific 2D planes acquired in a 2D cine PC scan. With 4D flow MRI, there is a single, easily prescribed data acquisition, covering the entire region of interest as opposed to multiple 2D planes that may be difficult to position in the presence of complex vascular anatomy, such as with congenital heart disease. 4D flow assessment also allows improved assessment of peak velocities, that may be underestimated with 2D cine PC.

In the thoracic aorta, 4D flow analysis is important to determine the impact of focal abnormalities, including those involving the aortic valve, coarctation, and aneurysmal dilatation - that can affect flow within the entire aorta (Fig. 80.1). For example, although changes in flow due to aortic valve stenosis or a bicuspid aortic valve predominantly affect the ascending aorta, these can extend to involve the arch and descending aorta. Otherwise, if 2D cine phase contrast is employed, hemodynamic changes distal to the area of anatomic abnormality may be overlooked.

After segmentation of a vascular region of interest (e.g. the ascending aorta), peak blood velocity can be quantified automatically, enabling estimation of pressure gradients. New pathophysiologic hemodynamic parameters can also be calculated, such as wall shear stress, pulse wave velocity and pressure difference maps.

Fig. 80.1 Bicuspid aortic valve, dilated ascending aorta with pronounced helical flow (arrows). Courtesy of Michael Markl, Northwestern University, Chicago, IL.
Whole heart 4D flow facilitates systematic assessment of the heart and great vessels, enabling retrospective analysis of any region within the imaging volume (Fig. 80.2). This is particularly valuable in the pre- and post-surgical assessment of flow in congenital heart disease, in particular for the right heart and pulmonary arteries. Here analysis by 2D cine PC may be difficult, with 4D providing critical information regarding possible re-stenosis or other postsurgical sequelae that might require re-intervention.

Applications for 4D flow MRI in the brain include hemodynamic evaluation of aneurysms, arteriovenous and other vascular malformations (Fig. 80.3), venous flow (as well as the effects of venous thrombosis) and atherosclerotic disease. The technique may in the future provide important information regarding the impact of an AVM on flow redistribution, and affect treatment planning by identifying feeding vessels with the highest flow for targeted embolization.

Fig. 80.2 Tricuspid atresia with Fontan procedure, flow from the superior and inferior vena cava is directed into the left and right pulmonary arteries. Courtesy of Michael Markl.

Fig. 80.3 Temporal lobe AVM (yellow arrow), 4D flow depiction of the arterial supply and venous drainage (white arrow). Courtesy of Michael Markl.
Advanced Non-Contrast MRA Techniques

Contrast-enhanced MRA remains the technique of choice for vascular imaging due to its high spatial resolution and SNR. However, when imaging patients with impaired renal function, and in certain other uncommon settings, gadolinium chelate administration may not be advised. Non-contrast MRA techniques have been developed for this use.

The mechanism for image contrast with NATIVE TrueFISP results from preparation of the imaging volume with a spatially selective inversion pulse. This suppresses signal from both stationary and mobile (i.e., intravascular blood) tissues. However, blood flowing into the imaging volume during the inversion time demonstrates high signal intensity, typical of TrueFISP sequences. Contrast between vascular structures and background tissue is further increased by suppression of background signal with the inversion pulse. This pulse can be positioned independently of the selected imaging volume, and NATIVE TrueFISP can thus be utilized for arterial or venous imaging. One major application is visualization of renal artery stenosis. NATIVE can be combined with SPACE sequences—a modified variable flip angle 3D FSE technique. With SPACE, visualization of intravascular blood flow is based on the difference in intravascular signal during periods of maximum and minimum vascular flow within the cardiac cycle. ECG monitoring is advised/required for these two NATIVE techniques.

Many unenhanced MRA techniques are based on the identification of moving protons. In this instance, any type of patient motion (respiratory or otherwise) will generate substantial artifacts. Fig. 81.1 presents a NATIVE coronal MIP of the peripheral circulation obtained using

Fig. 81.1 Non-contrast MRA from the pelvis to the calves, NATIVE TrueFISP.
PACE (Prospective Acquisition Correction, monitoring the patient’s position by use of a navigator echo to permit motion correction). The NATIVE TrueFISP sequence accommodates 3D, 2D, breath-hold, PACE navigated, and respiratory triggered approaches depending on the clinical environment.

An alternative to NATIVE is Quiescent-Interval Single-Shot MRA (QISS). For the peripheral vasculature, QISS is preferred, but in other applications such as the renal arteries NATIVE is preferred. Both techniques well visualize flow in heavily calcified arteries, in distinction to CTA. QISS is an ECG-triggered technique that employs initial saturation pulses (to suppress both background and venous blood) followed by a 2D single-shot balanced SSFP readout. A quiescent interval prior to the readout allows inflow of unsaturated arterial spins. Use of single shot technique makes the technique relatively insensitive to patient motion.

With black blood sequences, a large coverage 180° inversion pulse is first applied. This is followed immediately by a second 180° inversion in the slice/slab region of interest. In-flowing blood will only experience the initial inversion pulse, while the imaging volume will experience a 180+180=360=0° result. This is then followed by excitation and readout. Optimal selection of TI yields black blood at the null point of the relaxation recovery of blood.

Black blood techniques can be used for assessment of carotid bulb atherosclerosis. This approach provides excellent image contrast between the low signal vascular lumen and its wall, allowing for measurement of plaque volume, identification of plaque components including intraplaque hemorrhage and lipid, and evaluation of plaque surface morphology specifically in the setting of ulcerations or fibrous caps. These imaging characteristics have been utilized to identify “vulnerable plaque” possessing a greater embolic risk. Fig. 81.2 presents TOF, T1, and black blood images of the left carotid bulb revealing hemorrhagic plaque (arrow, high signal on black blood images) that was responsible for an embolic infarction of the left MCA territory. Of note, this plaque carries no hemodynamically significant stenosis. Further potential clinical applications for black blood imaging include imaging of arterial dissection and vasculitis.

A known pitfall of TOF MRA is that short T1 tissues such as fat, methemoglobin, or proteinaceous material also exhibit high signal on TOF MRA and can thus mimic normal intravascular signal, potentially resulting in an erroneous diagnosis. As black blood MRA exhibits signal voids corresponding to moving blood, the sequence may be utilized to distinguish thrombus from normal intravascular signal and for evaluating vessels immediately adjacent to such pathology.

![Fig. 81.2 Black blood imaging of carotid plaque. Courtesy of Tobias Saam.](image)
Contrast-Enhanced MRA: Basics; Renal, Abdomen

The contrast-enhanced MRA maximum intensity projection (MIP) image displayed in Fig. 82.1a demonstrates the abdominal aorta and common iliac arteries, with moderate to severe stenosis noted at the origin of the left renal artery (arrow). Fig. 82.1b demonstrates extensive atherosclerotic disease involving the aorta, with severe stenosis at the origin of the left renal artery. Both studies were performed at 1.5 T.

The study presented in Fig. 82.2 illustrates the feasibility of high spatial resolution contrast-enhanced MRA at 3 T, providing a further improvement in evaluation of the renal artery and its branches. Early branching is demonstrated, involving both renal arteries, in this potential living related kidney donor. The voxel size was 1 × 1 × 1 mm, the contrast dose 0.1 mmol/kg injected at 2 mL/sec, and the scan time 16 sec (with a parallel imaging factor of 3). An additional advantage of this type of acquisition is the ability to reconstruct high-resolution images in any desired plane, given the high spatial resolution and isotropic voxel dimensions. Such reformatted images are similarly advantageous for the evaluation of renal artery stenosis. 3 T in this application offers substantial advantages as compared with 1.5 T, largely due to the inherent increase in SNR. Improved suppression of background tissue, due to the prolongation of T1 at 3 T, aids as well by further increasing CNR.

Contrast-enhanced MRA (CE-MRA) has become the exam of choice for evaluation of the abdominal aorta and renal arteries. The standard imaging sequence is a fast 3D spoiled gradient echo scan. Thin sections (on the order of 2 mm or less) are acquired in the coronal plane within a breath-hold (typically 20 seconds or less), using a very short
TR and TE. Acquiring a 3D volume in such a fashion results in a high degree of saturation (low MR signal) of the background tissues. Bolus injection of a gadolinium chelate leads to a substantial reduction in the T1 relaxation time of blood, producing images with very high signal intensity vascular structures, due to the gadolinium chelate “enhanced” blood within. Typically 20 to 40 mL of contrast media is injected at a rate of 1.5 to 3 mL/sec. The bolus of contrast agent is immediately followed by a bolus of normal saline, typically 20 to 30 mL injected at the same rate. The purpose of the saline is to maintain the contrast in a tight bolus as it travels through the vascular system.

In every MR acquisition, and of particular relevance for CE-MRA, the raw data (as sampled) occupies $k$-space, the coordinates of which are frequency and phase as opposed to $x$ and $y$. High spatial frequency data, found in the periphery of $k$-space, contains information regarding predominantly image detail. Low spatial frequency data, found in the center of $k$-space, contains information regarding predominantly image (tissue) contrast. The position of data in $k$-space is determined by the amplitude of the phase encoding gradient applied prior to sampling of the echo (MR signal). Echoes acquired during the application of high-amplitude gradients contain principally information regarding spatial resolution. Echoes acquired during the application of low-amplitude gradients contain principally information regarding tissue contrast (as well as containing most of the observed signal). CE-MRA acquisitions are timed such that the acquisition of the central lines of $k$-space coincides with the maximum concentration of contrast media (gadolinium chelate) in the area of interest. It is therefore very important to know both the circulation time to the area of interest and the order in which the $k$-space data are collected. The first may be determined by the use of a test bolus, or bolus-tracking techniques can be employed. In regard to the second, most systems allow the operator to select the order of $k$-space filling. Typical terminology used refers to the filling of the central portion first as “centric” and the filling of the outer lines first as “linear.”

![Fig. 82.2](image)

**Fig. 82.2** High spatial resolution CE-MRA of the renal arteries. Used with permission from Kramer U, Thiel C, Seeger A, et al. Preoperative evaluation of potential living related kidney donors with high-spatial-resolution magnetic resonance angiography at 3 tesla: comparison with intraoperative findings. *Invest Radiol.* 2007;42:747-55.
A maximum intensity projection (MIP) of the entire dataset (Fig. 83.1a) from a contrast-enhanced MRA (CE-MRA) of the carotids at 3 T reveals the aortic arch, origin of the great vessels, carotid arteries, and vertebrobasilar system. There is nonvisualization of the right internal carotid artery (ICA), with occlusion of this vessel just subsequent to the carotid bulb (arrow). The MIP image is created following completion of the scan from source images, which are usually acquired in the coronal plane. In many instances, a dataset is acquired both immediately prior to and during contrast injection, with the first set of images serving as a mask (and the MIP image created from the subtraction of the two datasets). A single source image (from the 80 images that constituted the coronal slab) is also illustrated (Fig. 83.1b). The internal carotid occlusion in this instance was secondary to crystal methamphetamine use, with both this illegal drug and cocaine implicated in acute carotid dissection with subsequent occlusion and ischemic stroke.

For carotid CE-MRA, 3 T offers a substantial improvement in SNR as compared with 1.5 T. This allows, in combination with parallel imaging, higher spatial resolution, resulting in diagnostic quality comparable to CT angiography and digital subtraction angiography for
detection of arterial stenoses. The voxel dimensions for the scan depicted in Fig. 83.1 were 0.8 × 0.8 × 0.8 mm. Use of small isotropic voxels permits reconstruction of images in any arbitrary plane with high image quality, permitting improved assessment of arterial stenoses. Fig. 83.2 illustrates coronal, axial, and sagittal reformatted images from the original dataset acquired for Fig. 83.1. The axial image (b) is just distal to the carotid bulb, depicting the occluded internal carotid artery (arrow). The sagittal image (c) was obliqued slightly to permit visualization of the distal common carotid artery, the carotid bulb (and subsequent occlusion), and the external carotid artery origin. Volume rendering of the data, using a 5-mm thickness and the plane of section from (c), leads to the image in (d), which well depicts the occlusion just subsequent to the carotid bulb (arrow).

CE-MRA of the carotid arteries represents a substantial advance over conventional 2D or 3D time-of-flight (TOF) techniques, which were employed for this purpose in the 1990s. In TOF acquisitions, a thin slice (2D) or a slab (3D) is excited using a relatively short TR. This results in saturation of (reduced signal from) background tissue. "Fresh" unsaturated blood flowing into the slice or slab has not been exposed to an RF pulse and thus has high signal intensity. TOF techniques produce images based on this flow-related enhancement. Problems arise when there is turbulent flow, reversal of flow, higher orders of motion (acceleration), or very slow flow. Any of these conditions can result in loss of signal within a vessel. These complex flow patterns occur in both normal as well as diseased vessels, which historically resulted in overestimation of the degree of stenoses. With CE-MRA techniques, the image contrast is based on T1 differences between background tissue and the gadolinium chelate "enhanced" blood. This, together with the very short echo time (TE) typically used, greatly reduces the artifactual signal loss due to complex flow patterns and allows for excellent visualization of the arterial system. The use of a coronal 3D slab permits improved craniocaudal coverage without impacting scan time. Another advantage of CE-MRA is the short scan time. In the example illustrated, the scan time was 23 sec compared with several minutes for TOF technique. In addition, 2D TOF sequences acquire slices in a sequential fashion, with swallowing motion producing severe misregistration artifacts. One caveat with CE-MRA is that the order of k-space sampling and the timing of image acquisition are crucial. The transit time of the contrast bolus from the common carotid arteries to the jugular veins can be as fast as 6 to 7 seconds, with little margin for error in timing the acquisition to avoid venous contamination.
**Contrast-Enhanced MRA: Peripheral Circulation**

*Fig. 84.1a* illustrates 3D contrast-enhanced MR angiography (CE-MRA) of the lower extremities, in a patient with no significant stenoses or occlusions. The adjacent 3D CE-MRA examination of the femoropopliteal distribution (*Fig. 84.1b*) reveals, in a different patient, bilateral superficial femoral artery occlusion with development of profunda femoral artery collaterals. *Fig. 84.1c,d* are multiphase CE-MRA images of the tibioperoneal distribution obtained during early arterial enhancement and (with a slight time delay) after substantial venous filling. In the latter image, the large vascular malformation in the left gastrocnemius is more completely visualized, due to opacification of the venous component.

Peripheral MRA may be performed with time-of-flight (TOF), phase contrast, NATIVE TrueFISP and contrast enhanced techniques. The latter dominates in clinical practice, due to the short scan time and, more importantly, because its sensitivity and specificity approach that of traditional X-ray angiography for peripheral vascular disease. Peripheral 3D CE-MRA typically utilizes short TR/short TE 3D gradient echo sequences, obtained in three to four stations in the coronal plane.

Automated table positioning is incorporated with image acquisition at each station. Timing is set so that the scan is acquired during passage of the gadolinium chelate bolus, or equivalently during maximal contrast concentration within the vessel of interest. Detection of T1 shortening and therefore the start of image acquisition occur via one of four methods: (1) A test bolus may be used to approximate the timing of bolus arrival during the actual examination. (2) In MR fluoroscopy, rapid 2D scans allow the technologist to monitor for arrival of the contrast bolus and thereby manually initiate the 3D CE-MRA scan (used for *Fig. 84.1a,b*). (3) In multiphase CE-MRA (*Fig. 84.1c,d*), rapid time sequential 3D scans are acquired, permitting dynamic imaging of the passage of contrast through the arterial and venous circulation. (4) Automated bolus detection algorithms involve computer detection of bolus arrival and initiation of image acquisition. The center of $k$-space (see Chapter 12), which is the major determinant of image contrast, must be obtained when arterial enhancement is at its peak (not on the up-slope as this increases ring artifact) and venous enhancement is at a minimum. At station 1 (aortoiliac), the center of $k$-space is obtained near either the midpoint or end of the scan to ensure that data acquisition occurs during arterial enhancement (with the scan initiated when contrast is first visualized in the proximal aorta). This order of acquisition is commonly “reversed” in station 2 (femoropopliteal) and station 3 (tibioperoneal) or station 3 alone. The center of $k$-space in the latter station(s) is obtained during the beginning of the scan via centric phase reordering, assuring that this portion of the scan is acquired during peak arterial enhancement, with minimum venous contamination.

Recent innovations continue to improve the diagnostic quality of peripheral CE-MRA. Simplifying patient setup, image acquisition can be performed during continuous table movement (see Chapter 136), as opposed to a multistation approach. For the calf and foot, pulse sequences developed to provide time-resolved high-resolution MRA (see Chapter 85) offer improved distal arterial vessel visualization and decreased venous contamination.
Fig. 84.1 Normal CE-MRA, atherosclerotic disease, vascular malformation.
One of the major disadvantages of contrast-enhanced MR angiography (MRA), early in its development, was the limitation in temporal resolution. Acquisition time could be substantially reduced using conventional MRA techniques, but at the expense of a marked reduction in spatial resolution. On the other hand, acquiring MRA images with a spatial resolution close to conventional digital subtraction angiography was feasible, but acquisition time increased and thus temporal resolution was impaired: a major

Fig. 85.1 TWIST, delayed flow in the vessels of the left calf.
By the evolution that occurred in hardware (strong and fast gradients) and software, acquisition times can now be markedly decreased for contrast-enhanced MRA. Based in part on the use of parallel imaging in combination with the improved SNR of 3 T systems, high temporal resolution MRA studies became feasible without diminishing image quality in terms of spatial resolution. For example, high-resolution contrast-enhanced MRA of the carotid arteries can be acquired with a temporal resolution as high as 1.5 sec per scan (for a 3D volume acquisition).

Strategies to reduce acquisition time include the reduction of TR, a rectangular FOV, partial Fourier data sampling, and parallel imaging (applied both in-plane and through-plane). In this setting, the main technical advance to substantially further reduce acquisition time consisted of the application of highly accelerated, parallel imaging in combination with temporal echo sharing. These fast, time-resolved MRA techniques are now available from all vendors, with one acronym for this approach being TWIST (Time-resolved angiography With Interleaved Stochastic Trajectories).

These imaging techniques are based on special k-space sampling. In this setting, k-space is divided into two regions. Although the centrally located region of k-space provides information regarding image contrast, the peripherally located aspect of k-space contributes principally to high spatial resolution. The main factor contributing to the acceleration of the sequence acquisition in this particular imaging approach is the fact that k-space lines in the center are more frequently sampled than the k-space lines in the periphery during the passage of the contrast medium bolus through the covered 3D volume. For that reason, the frame rate with this technique is much higher than in conventional full k-space acquisition. The acceleration depends principally on the undersampling of peripheral k-space lines.

Major advantages of time resolved MRA protocols include 1) a substantial reduction in gadolinium chelate dose (e.g., a quarter or less of that required with conventional techniques); 2) the ability to obtain multiple phases, enabling demonstration of arterial and venous flow, directionality of flow, and visualization of delayed filling; and 3) the lack of a need for a test bolus. Decreased arteriovenous overlay, reduction in motion artifacts, and the additional diagnostic information available from the temporal assessment of contrast enhancement (of the vessels and tissue perfusion) are important features clinically of this technique.

**Fig. 85.1** presents four consecutive MIP reconstructions of a dynamic contrast-enhanced MRA study of the lower legs after IV application of 2 mL MultiHance. Temporal resolution in this sequence acquisition was 2 sec per scan series. The dynamic exam reveals the delayed contrast bolus arrival in the left lower leg as compared to the contralateral limb without confounding venous overlay. Single phase CE-MRA of the upper legs (**Fig. 85.2**) depicts a short-segment occlusion (arrow) of the left superficial femoral artery (SFA), the culprit lesion.
A hyperacute middle cerebral artery (MCA) infarct demonstrates only a small area of restricted diffusion (arrow) on DWI (**Fig. 86.1a**). There is a paucity of left MCA branches on the TOF MRA, reflecting either occlusion or slow flow (**Fig. 86.1b**). The mean transit time (MTT) map (**Fig. 86.1c**) reveals that there is increased transit time slower arrival of the contrast agent bolus) within a large portion of the MCA territory (arrows). Perfusion is however reduced in a much smaller area, as visualized on the relative cerebral blood volume (rCBV) map (**Fig. 86.1d**). These mismatches (CBV-MTT, CBV-DWI) provide a measure of tissue at risk and are used to guide the choice of therapy.

**Fig. 86.1** DWI, TOF MRA, MTT and rCBV in a hyperacute MCA infarct.
Cerebral perfusion imaging is the visualization of changes or delays in microvascular blood flow in the brain. This imaging technique can be used to facilitate the evaluation of strokes, tumors, and the differentiation of radiation necrosis versus recurrent tumor. Perfusion imaging is made possible through the acquisition of multiple, time sequential, single-shot echo planar imaging (EPI) slices (see Chapter 42) measured with a temporal resolution of 1 sec or less during the rapid administration of a gadolinium chelate. The use of echo planar technique enables rapid image acquisition with high sensitivity to the T2* effect of the contrast agent.

Today's high-end MR scanners with advanced gradient technology are able to accommodate coverage of the entire brain with slices acquired in a dynamic fashion immediately prior to, during, and following passage of the contrast bolus through the brain. The transit of a gadolinium chelate (as a concentrated, compact bolus) through the brain causes a decrease in tissue signal intensity on echo planar images due to the T2* or magnetic susceptibility effect of the agent. A graph of signal intensity versus time during bolus contrast injection for two arbitrary regions of interest is illustrated in Fig. 86.2. Using this acquired data, calculations are made to demonstrate the rate of change in the MR signal as well as the relative volume and flow of blood to the visualized area. Calculated results are displayed in the form of images or maps where each image encompasses information from the entire dynamic dataset for that slice position.

The MTT map (Fig. 86.1c) depicts the time required for fresh blood to completely replace that in the volume of interest. The time to peak (TTP) map is a simpler, related quantity that measures the arrival time of the bolus. Depicted in green and red on the MTT map are areas of reduced MTT, in which red blood cells spend less time within the capillaries and thus less oxygen can be extracted. The rCBV map (Fig. 86.1d) is calculated based on changes in the intensity of pixels over time and conveys information regarding tissue blood volume within the displayed slice. Darker areas on this map correspond to regions with lower blood volume. Note that normal gray matter and white matter are well-differentiated on the rCBV image due to the higher blood volume of gray matter. Relative cerebral blood flow maps (rCBF) (not shown) can also be calculated, but, like MTT, they require measurement of the signal intensity within an artery supplying the tissue of interest (the arterial input function).
Arterial spin labeling (ASL) perfusion MR emerged in the 1990s as a noninvasive technique (employing EPI) for the evaluation of cerebral blood flow, specifically without the need for intravenous contrast media injection. ASL uses magnetically labeled arterial blood water as an endogenous tracer for measuring blood flow. This magnetic labeling is performed using radiofrequency pulses, which alter the magnetization of the blood flowing into the brain. The extent of this alteration is determined by comparing the ASL image with a reference (control) image obtained without blood water labeling. Quantitative regional maps of cerebral blood flow can be obtained in units of mL/min/100g of tissue if the kinetic parameters of perfusion are known. Several ASL methods have been proposed, in the following only two methods will be described in detail: pulsed ASL (PASL) and pseudo-continuous ASL (PCASL).

**PASL:** With pulsed labeling techniques, a volume of blood is labeled upstream of the region of interest by a short RF pulse (Fig. 87.1). Signal acquisition is performed after a delay TI (of 1 to 2 seconds). The difference image (obtained by subtracting an image acquired without labeling) reflects the amount of labeled blood arriving in the volume of interest during the delay TI and is proportional to CBF. A fundamental trade-off to

![Fig. 87.1 Technique for image acquisition, ASL (specifically pulsed labeling).](image)

![Fig. 87.2 Hypoperfusion, ASL and FLAIR, cortical hamartoma (tuberous sclerosis).](image)
any implementation of ASL is that there is not complete delivery of labeled blood to the area of interest with a short delay, while a long delay results in T1 decay and thus reduced SNR. PASL was initially implemented using 2D acquisition, and subsequently extended to 3D.

**PCASL:** With pseudo-continuous labeling techniques (the new clinical standard), a long labeling pulse or pulse train is applied to the blood upstream of the slice using a continuous radiofrequency of weak intensity associated with a gradient applied in the direction of flow. As a result of pseudo-continuous spin labeling, the signal of the labeled slice of interest will reach steady state. Subtracting a control image (acquired without spin labeling) from the spin labeled image yields an image with signal proportional to cerebral blood flow (CBF).

A limitation of ASL is low SNR, in particular when compared with perfusion imaging using a gadolinium chelate. ASL is markedly improved at 3 T, due to the higher CNR and improved labeling efficiency, the latter on the basis of T1 prolongation, and with the implementation of 3D acquisition. **Fig. 87.2** presents 2D PASL images of a patient with tuberous sclerosis, in the axial and sagittal planes, demonstrating reduced blood flow in a large cortical tuber (black arrows), which is seen with slight hyperintensity on the FLAIR image. Regardless of approach used, spatial resolution for ASL will be inferior to that for dynamic susceptibility contrast technique (observing the first pass of a gadolinium chelate through the brain; see Chapter 86).

In clinical application, ASL provides reproducible and reliable quantitative CBF measurements for a spectrum of brain abnormalities, including neoplastic disease, ischemia, degenerative diseases (Alzheimer’s disease and frontotemporal dementia) and arteriovenous abnormalities. The repeatability and high temporal resolution of ASL also makes this technique suitable for functional MRI (fMRI) studies, specifically localization of task activation. Outside of the brain, ASL can be used to assess, non-invasively, renal function (e.g. in renal donors) and is well suited for repeated longitudinal use (e.g. in renal transplant recipients). **Fig. 87.3** presents a sagittal T1-weighted image and the corresponding ASL perfusion map in a normal renal donor.

**Fig. 87.3** Normal renal perfusion, ASL. Used with permission from Niles DJ, Artz NS, Djamali A, et al. Longitudinal assessment of renal perfusion and oxygenation in transplant donor-recipient pairs using arterial spin labeling and blood oxygen level-dependent magnetic resonance imaging. *Invest Radiol.* 2016;51:113-20.
Section VI

Tissue-Specific Techniques
Brain Segmentation, Quantitative MR Imaging

Rapid simultaneous quantification of T1, T2, proton density (PD) and the B$_1$ field (for correction of B$_1$ inhomogeneity) can today be achieved for the brain by the use of a single multislice, multi-echo, multi-delay acquisition. The imaging approach is illustrated in Fig. 88.1, together with the resultant maps of R1 ($R1 = 1/T1$), R2 ($R2 = 1/T2$) and PD. The results obtained agree well with established literature values for these parameters. Use of a single scan avoids both the long scan times necessary to acquire independently each reference value as well as the associated potential for misregistration. From such an acquisition (typically 5 minutes in duration, covering the entire brain with 5 mm sections), using synthetic MR, images with conventional like tissue contrast can be generated. Further quantification and segmentation into tissue types is also readily performed, a process termed synthetic tissue mapping. In principle, the use of absolute quantification can remove the scanner dependency from such measurements, and allow reliable automatic segmentation by which a patient and the disease process being examined (e.g. MS in the brain) can be accurately assessed, and temporally followed. Advantages to the synthetic MR images generated from such a data set include perfect image registration and removal of scanner and image sequence dependencies, including for example TR and TE. Using synthetic MR, these variables can be specified and customized (to produce optimal contrast for the pathology in question or according to the preference of the radiologist reading the exam), since they are simply used to reconstruct images to be viewed, from the original single acquisition.

Using the R1, R2 and proton density values so acquired, different tissue types can be segmented, for example intracranial volume (ICV), and the volume for gray matter, white matter and CSF. From these segmentations, brain parenchymal fraction (BPF), which is (ICV – CSF volume)/ICV, can also be calculated. The latter is considered to be a robust measure for monitoring relative brain and CSF volumes in disease processes, e.g. MS, eliminating the effect of head size differences.

Fig. 88.1 Overview of the MR technique for brain tissue quantification. Courtesy of Marcel Warntjes, Synthetic MR, Linköping, Sweden.
Amongst the many disease entities that can be quantitatively assessed and followed temporally with this approach are multiple sclerosis, mild cognitive impairment (MCI) and Alzheimer’s disease, and hydrocephalus. The amount of myelin, as well as the amount of vasogenic edema, can also be estimated from such a quantitative acquisition. With collection of a healthy pediatric reference database, the latter would also enable the progression of myelination to be monitored in the pediatric population, which is of importance in inborn errors of metabolism.

The assessment/calculation of myelin is somewhat more complex than that of the other quantitative parameters, and is based upon the effect that myelin has on surrounding cellular water (Fig. 88.2). In the vicinity of myelin, the relaxation rates of cellular water increase (due to magnetization exchange), while the observable proton density decreases (since the signal from myelin water decays faster than can be observed with conventional imaging). These effects can be used, with data from the multi-parametric acquisition, to estimate myelin partial volume. Normal brain (each voxel) is modeled into three compartments: myelin partial volume ($V_{MY}$), cellular partial volume ($V_{CL}$), and free water partial volume ($V_{FW}$, at the interface with CSF). In pathologic conditions, there may be demyelination, leading to a decrease in $V_{MY}$. There may also be an increase in water, specifically parenchymal edema, modeled by a fourth parameter/compartment, excess parenchymal water partial volume ($V_{EPW}$).

Summarizing the approach for quantitative MR imaging of the brain, a scan is first acquired from which R1, R2 and PD values can be calculated. All combinations of these parameters that can be assigned to gray or white matter and CSF are put together, defining the intracranial cavity. The edge of the latter is assigned using the PD map. CSF is segmented and removed, leaving the brain. This process also provides the brain parenchymal fraction (BPF), defined as the ratio of brain to intracranial volume, which provides an excellent measure to monitor brain atrophy. Finally, each brain voxel can be decomposed into the four myelin model compartments. Summing the myelin partial volume of all voxels gives the total myelin volume for the patient.

Fig. 88.2 Myelination quantification in a MS patient: synthetic FLAIR, color overlay, and myelin map. Courtesy of Marcel Warntjes.
Cardiac Morphology

High-quality MR images require the object being imaged to remain still during the acquisition. Substantial object motion during data sampling results in image blurring and/or degradation by ghosting. To obtain diagnostic images of objects exposed to continuous motion, such as the heart, careful selection of imaging strategies is of utmost importance.

Fig. 89.1 presents a T1-weighted, morphologic image of the heart. Notice that the heart has been captured without blurring (motion is ‘frozen’) and the blood signal has been ‘nulled’ for a better delineation of the cardiac structures. Generally, cardiac imaging requires ECG synchronization for acquisition purposes. A method used to assist in imaging of the heart without motion is \( k \)-space segmentation; the phase encoding steps required to create an image are split into multiple segments and acquired over multiple cardiac cycles. Furthermore, data is typically acquired during the relatively motionless resting phase (mid-diastole) of the cardiac cycle. Data acquisition within this phase can last up to 140–160 msec without notifiable motion artifacts in the resulting image. The number of phase encoding lines that can be acquired within such a window depend on the sequence repetition time (TR). The total acquisition time per image is related to the total number of phase encoding steps required and the number of phase encoding lines that can be acquired per

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**Fig. 89.1** T1-weighted Fast Spin-Echo (Turbo Spin-Echo) image with black-blood preparation acquired in short axis orientation.

**Fig. 89.2** T2-weighted HASTE image in transverse orientation of the heart.
cardiac cycle. At faster heart rates image acquisition is shorter; a shorter resting window, however, may require a reduction of phase encoding lines acquired per cardiac cycle.

Another technique used to freeze cardiac motion is single-shot imaging. In single-shot imaging, all phase encoding steps necessary to create an image are acquired in one heart beat (one shot). As with k-space segmentation, the phase encoding lines are typically acquired during mid-diastole. To achieve images without motion, the number of total phase encoding steps, hence spatial resolution, is limited because of the short acquisition window within the cardiac cycle. HASTE (see Chapter 39) is a pulse sequence technique that can be employed in this fashion, with an example presented in Fig. 89.2. Note that the edges of the myocardium are not quite as sharp (with slight blurring), when compared with Fig. 89.1, due to the reduced spatial resolution.

A highly important aspect of morphologic imaging of the heart is the elimination (‘nulling’) of the blood signal. This is accomplished through a series of nonselective and selective inversion pulses (black-blood preparation), with the steps depicted in Fig. 89.3. An initial nonselective RF inversion pulse (180 degree) is applied to invert all spins (tissue and blood) within the imaging field (1). This is immediately followed by a selective RF inversion pulse (180 degree) (2) designed to invert the imaging slice into the transverse plane preparing for the standard imaging part of the sequence (3). The blood encoded during the slice acquisition is being replaced by inverted blood from outside the imaging plane due to flow (4), and as a result the blood within the acquired slice does not provide a signal (5).

The utility of black-blood (dark-blood) imaging is generally not limited to the heart but may also improve visualization of the pulsatile great vessels such as the aorta, as shown in Fig. 89.4 (an aortic coarctation). Challenges for black-blood imaging include arrhythmias and post-contrast imaging due to shortened blood T1 times.
Cardiac Function

As discussed in Chapter 89, MRI readily provides static morphologic information concerning the heart, with high tissue contrast and high spatial resolution. However, comprehensive analysis of the heart also requires the assessment of ventricular function which is of high importance regarding differential diagnosis and therapy decisions/monitoring. To achieve this goal, specifically adapted sequences and dedicated postprocessing software algorithms are required.

In static morphologic imaging, required $k$-space lines are split into small segments and are acquired during a single phase of diastole in consecutive cardiac cycles until data for an image is completed. Imaging of cardiac function is accomplished in a similar fashion except that $k$-space lines are acquired throughout the entire cardiac cycle but binned as smaller segments into $k$-space for different phases. In this way, information is gathered about each phase of the cardiac cycle in consecutive heartbeats until a complete image of each phase is acquired. The most commonly applied techniques employ retrospective ECG gating; rarely prospective ECG triggering is used today (possibly beneficial in arrhythmia). After

Fig. 90.1 Selected images, single mid-ventricular cine slice, cardiac cycle.

Fig. 90.2 (a) 3D representation of short axis slices (every 2nd is shown) covering the long axis of the heart. (b) Semi-automated post-processing with contouring of individual phases of the slice stack.
reconstruction, individual images are combined into a cine loop covering all aspects/phases of the cardiac cycle (Fig. 90.1). Data acquisition for a single slice typically requires a breath-hold of 8–12 heartbeats achieving a temporal resolution of <40–50 ms.

For quantification of ventricular size and function, a stack of parallel slices covering the entirety of the ventricles from base to apex (~5–8min total scan time) is acquired (Fig. 90.2a). Dedicated post-processing software allows for semi-automated contouring (segmentation) of the blood volume (endocardial border) and myocardial mass (epicardial border) on each slice at end-systole (ES, minimal volume) and end-diastole (ED, maximum volume) (Fig. 90.2b). Based on planimetry of individual slices, the total ventricular volumes can be calculated without assumption of geometric models using Simpsons’ rule (slice summation). With known end-diastolic volume (EDV) and end-systolic volume (ESV) various indicators of cardiac function and performance can be derived (Table 90.1). Related to the high blood/myocardium contrast, the full anatomical coverage, and the spatial resolution, resulting global volumetric and functional information in MRI is superior to other imaging modalities (specifically echocardiography and radionuclide ventriculography) (Fig. 90.3).

In recent years, a variety of new techniques has been developed to further improve the spatial and temporal resolution of cardiac MR techniques as well as ease of workflow. One such technique is echo sharing, in which a portion of the acquired phase encoding steps from one segment of one phase of the cardiac cycle is shared with an adjacent segment. This results in the ability to acquire images with higher spatial resolution for finer functional detail. Another approach uses extremely rapid, low resolution, single-shot techniques to collect data in a nearly real-time fashion. The breath-hold requirements are greatly reduced or eliminated and a quick assessment of the entire ventricle can be completed in ~15 to 20 heartbeats, including patients with arrhythmia.

Finally, it should be noted that robust parallel imaging capabilities (see Chapter 127) and recent developments with data sparsity techniques (see Chapter 131 and Chapter 132) allow further speed-up of high-quality and multi-contrast cardiac imaging.

<table>
<thead>
<tr>
<th>Measured Parameters</th>
<th>Units</th>
<th>Units (indexed to BSA)</th>
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<tbody>
<tr>
<td>End-Diastolic Volume (EDV)</td>
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<td>mL/m²</td>
</tr>
<tr>
<td>End-Systolic Volume (ESV)</td>
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<td>mL/m²</td>
</tr>
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<td>Myocardial Mass</td>
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<td>g/m²</td>
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<table>
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<tr>
<th>Derived Parameters</th>
<th>Units</th>
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<tbody>
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<tr>
<td>Ejection Fraction (EF) = SV/EDV</td>
<td>%</td>
<td>[not applicable]</td>
</tr>
<tr>
<td>Cardiac Output (CO) = SV × heart rate</td>
<td>L/min</td>
<td>Cardiac Index (Cl) L/min/m²</td>
</tr>
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Fig. 90.1 Measured and derived parameters of cardiac size and function; for most, indexation to the patient’s body surface area (BSA) allows comparison with published reference values.

Fig. 90.3 High-contrast cine imaging at ED and ES of a ventricular aneurysm (upper row) and dilated cardiomyopathy (lower row) demonstrate independence of volumetric assessment on ventricular geometry.
Luminal narrowing of the epicardial coronary arteries may lead to a reduction or cessation of the flow of oxygenated blood and nutrients to the myocardium required for cellular activity and normal cardiac function. The requirements in myocardial blood flow change with activity and while a certain level might be sufficient at rest it may be insufficient at peak stress level.

The most common MR approach used to assess the extent and uniformity of blood supply to the myocardium is referred to as first pass myocardial perfusion imaging. Typically, saturation recovery T1-weighted scans are employed with acquisition of multiple levels (typically 3 or 4) in short axis orientation covering the left ventricle with each slice being acquired during every cardiac cycle after bolus injection of a gadolinium chelate (0.05–0.10 mmol/kg). This approach provides dynamic coverage of the contrast agent influx into the cardiac chambers and the myocardium, ultimately reflecting the level of myocardial blood supply.

Data acquisition is commonly performed over a period of approximately 40 to 60 heart beats during shallow respiration. In territories of reduced myocardial perfusion, delayed and less prominent enhancement occurs typically commencing in the subendocardial myocardium (Fig. 91.1). Assessment of hemodynamically relevant coronary artery stenosis requires the combination of first pass perfusion at rest and during pharmacologically induced vasodilatation (stress) (Fig. 91.2).

Data analysis is typically performed by visual assessment and comparison of stress and rest data. Available non-rigid motion correction and registration algorithms correct for possible displacement of the heart related to respiration and enable automated semi-quantitative perfusion maps (Fig. 91.3). Alternatively, region-of-interest analysis may be performed aiming at various semi-quantitative parameters; absolute
quantification approaches are labor intensive and require sophisticated modeling of the arterial input function and tissue response.

The comparability of stress vs. rest data is of outmost important for identification of possible perfusion deficits. The administration of vasodilator stress agents typically results in higher heart rates potentially limiting the number of slices that can be acquired within the RR interval. When performing the stress acquisition first, adjusted to the potentially high heart rate, identical slices and slice locations can then easily be acquired at rest after the heart rate has returned to normal.

‘Dark rim’ artifacts may occur at the subendocardial level and suggest perfusion deficits in areas of normal perfusion. They are typically related to the following aspects: susceptibility due to high contrast differences between the LV cavity and myocardium, subtle cardiac motion during the acquisition of each slice, and Gibbs ringing (see Chapter 102). To reduce such artifacts the acquisition time per slice should be limited to <150 ms.

Fig. 91.2 Same patient as in Fig 91.1.; the left column (pre CABG) again demonstrates the reversible perfusion deficit at stress (upper row) while the right column (post CABG) demonstrates homogenous perfusion as a result of revascularization therapy.

Fig. 91.3 Visualization of perfusion abnormalities using parametric mapping (non-rigid motion correction). The left column demonstrates single images of a perfusion data set at stress (upper row) and rest (lower row) with visualization of non-transmural perfusion deficits at stress (arrows). The right column presents the respective color-coded SI upslope maps of the respective stress perfusion data set (upper row) and rest perfusion data set (lower row), also with identification of the stress perfusion deficits (arrows).
Ischemia may affect the myocardium by limiting flow of oxygenated blood with a resulting mismatch of demand and supply in various ways. The first category, stunned myocardium, is living, viable tissue that has been subjected to short durations of reduced blood flow. In most cases, stunned myocardium will recover to normal function with time. The second category, hibernating myocardium, is viable tissue which has been subjected to chronically reduced blood flow requiring interventional (percutaneous or surgical) measures to restore adequate blood flow with subsequent normalization of cardiac function. The third category refers to infarcted/nonviable myocardium, tissue that no longer contains functional cardiomyocytes and is therefore unable to be returned to normal function. Patients with known ischemic cardiomyopathy resulting from decreased myocardium blood supply may benefit greatly from therapeutic interventions if the area in question is viable. However, if the tissue is no longer viable, there is little benefit in subjecting the patient to the possible risks of such a procedure. The differentiation between viable and nonviable myocardium is the focus of cardiac viability MR imaging.

Following IV injection, like in most tissues, the gadolinium chelates are distributed to the extracellular space. Furthermore, after an early peak the concentration in normal myocardium rapidly diminishes due to renal excretion and redistribution (see Chapter 91). However, in nonviable myocardium, despite an initially slower uptake of the contrast agent, there is also delayed clearance from the expanded contrast agent distribution volume. Thus, contrast media accumulates in nonviable myocardium, resulting in differences in tissue contrast agent concentration. Such changes are seen in both acute and chronic infarction, but not in reversible ischemic injury. In acute infarction, the changes in the contrast distribution volume relate to myocyte necrosis while in chronic infarction this is attributed to collagenous scar.

To visualize such changes in local myocardial contrast agent concentration, late gadolinium enhancement (LGE) imaging techniques employ strongly T1-weighted segmented inversion recovery (IR) gradient echo sequences. LGE imaging is typically performed ~8–15min after contrast injection and requires user adjustment of the TI. Based on the higher contrast agent concentration in infarcted areas (and areas of increased fibrosis in non-ischemic cardiomyopathies), the longitudinal magnetization recovers faster than normal myocardium based on the shorter TI. Adjustment of the TI aims to ‘null’ the signal of normal myocardium and as such highlights ‘hyperenhancement’ in pathologies (Fig. 92.1). The use of fast Look-Locker sequences (‘TI scout’) help to adequately adjust the TI (Fig. 92.2), with minor re-adjustments.
sometimes necessary, reflecting the continuously changing contrast agent concentrations. Phase sensitive (PS) reconstruction techniques enable sufficient nulling of normal myocardium over a wide range of TI values. This technique has a slightly different image appearance but has been shown to provide similar diagnostic value to standard techniques (Fig. 92.3).

LGE imaging techniques are also commonly used in the assessment of non-ischemic cardiomyopathies, storage diseases and inflammatory changes. As in ischemic changes any expansion (fibrosis, edema, etc.) of the extracellular space will lead to differences in local contrast concentration.
In recent years, parametric mapping techniques have become increasingly important for assessment of a wide range of cardiac diseases. Such techniques not only reduce the subjectivity encountered with traditional qualitative techniques, but also help improve the detection of subtle diffuse pathological changes within the myocardium. Quantitative parametric mapping techniques rely on the signal evolution across serial images rather than the signal intensity of a single image, and as such are largely independent of surface array-coil profiles and other sequence details. Related to the constraints of ECG triggering and patient breath-hold limitations, various specific techniques have been developed for parametric mapping all of which provide an estimate of the specific tissue parameter under investigation (e.g. T1, T2, T2*) rather than an exact value. Based on the specific technique, accuracy and precision of the techniques may vary.

Quantification of T1 relaxation, with and without the use of gadolinium-based contrast agents (GBCA), is of great interest for myocardial characterization in nonischemic cardiomyopathies. Without GBCA administration, elevated myocardial T1 values may relate to changes such as myocardial fibrosis or tissue edema. In myocardial infiltration, such as amyloid, T1 values may be substantially above normal. In late gadolinium-enhanced (LGE) MRI (see Chapter 92), the increased extracellular space or extracellular volume (ECV) in fibrotic scar, diffuse fibrosis or also extracellular edema, results in a shorter T1 than normal myocardium (corresponding to increased GBCA concentration). While late gadolinium-enhanced (LGE) MRI has proven accurate in focal pathologies, it is inferior to post-contrast T1 mapping in subtle diffuse changes (e.g. fibrosis, edema). By incorporating both pre- and post-contrast T1 maps, and a measure of blood hematocrit, quantitative ECV maps can be derived. These ECV maps have the potential to reliably identify diffuse fibrosis.

Cardiac T1 mapping is performed by sampling the T1 recovery curve using ECG-triggered single shot acquisitions, following an inversion recovery (IR) or saturation recovery (SR) radiofrequency pulse. The magnetization is sampled over multiple heartbeats, and recovery periods are typically required to allow for sufficient T1 recovery before the next preparation pulse. Several IR or SR acquisitions with different recovery times are performed to obtain enough images (each with a different contrast) within a single breath-hold for precise T1 quantification. For quantitative map generation using pixel-wise curve fitting to estimate the T1 recovery, non-rigid registration algorithms are employed to align individual images and compensate for subtle respiratory motion. Fig. 93.1 highlights the use of T1 mapping and subsequent ECV mapping in a patient with cardiac amyloidosis.

Quantitative T2 mapping has been shown to increase the accuracy and reliability of edema assessment in pathologies which alter myocardial water content and consequently prolong tissue T2. Sampling T2 decay in the heart may be performed using either a segmented multi-echo spin echo sequence or single shot acquisitions with variable T2 preparation RF pulses. Images are acquired at intervals of several heartbeats, to allow for sufficient magnetization recovery in between acquisitions. As for T1 mapping co-registration algorithms are employed for acquired images and pixel-wise
T2 fitting is done assuming mono-exponential signal decay. **Fig. 93.2** presents an example of a patient with acute myocarditis and regional differences in T2 relaxation times.

Myocardial T2* assessment entered clinical use in the non-invasive monitoring of transfusion dependent anemia patients (e.g. Thalassemia major). The technique allows tailoring of potential chelation therapies and has demonstrated significant impact on patient outcome. T2* mapping is based on a segmented ECG-triggered multi-echo gradient echo sequence with data acquired over several heartbeats and sampling at multiple echo-times (~7–20 ms) during T2* decay. Black blood preparation can be applied for artifact reduction. Other than in T1/T2 mapping, pixel based mapping is less commonly performed and region-of-interest analysis typically preferred. Data fitting is performed assuming a mono-exponential decay and optimized algorithms help to reduce the effect of noise in images with longer TEs.
There is little difference in the values of T1 and T2 between malignant and benign breast tissue. To visualize breast cancer, contrast enhanced dynamic acquisitions are needed. Tumors create specialized rather permeable (“leaky”) blood vessels to survive and grow. This angiogenesis takes up contrast agents more quickly than normal or benign breast tissue. Multiple identical T1-weighted, usually 3D, series are acquired with one pre-contrast as reference. Fat signal is strong in T1-weighted scans but does not contribute to the cancer evaluation. To get rid of this signal either the pre-contrast series is subtracted from the post-contrast ones, leaving only the enhancing tissues visible and/or a fat suppression algorithm is applied. Observing the contrast uptake and kinetic behavior allows further differentiation of the lesions: malignancies often enhance quickly and show a decrease in signal intensity over time (wash-out) whereas benign lesions enhance more slowly and over a longer time (persistent). Evaluating the signal curves over a time period of 5 to 10 minutes therefore provides powerful characteristic information. Many of these individual series are acquired in a relatively short time, typically 1 to 3 minutes each, in some cases even less than one minute. Of almost equal importance is the morphological presentation of the lesions. Malignancies often show irregular contours, a fingering into the adjacent tissue, called spiculation, whereas many benign lesions, for example fibroadenomas usually have smooth edges. This requires a high-resolution series in addition to the time-limited dynamic ones.

Normal breast tissue can also enhance over time. This “background” signal can be rather strong if hormonally stimulated and can either obscure or mimic malignancies. To reduce this effect, women should be scanned 5 to 14 days after onset of menstruation if pre-menopausal. If post-menopausal, hormonal replacement therapy should be avoided for at least 6 months prior to the MR scan.
In Fig. 94.1, four dynamic scans are displayed (immediate, 1, 3, and 5 minutes post-contrast). In the right breast (arrows), there are areas with prominent, fast enhancement, demonstrating some plateau, and thus suspicious for malignancy (invasive ductal carcinoma, IDC, by surgical pathology). In Fig. 94.2, three dynamic scans (immediate, 1, and 3 min post-contrast) through a large malignant lymph node (arrow) in the same patient depict several typical signs of malignancy: a focal lesion with fast, prominent enhancement and early washout. Region of interest (quantitative) analysis is typically performed to evaluate the time course of enhancement (Fig. 94.3). Abridged (abbreviated) dynamic scans (Fig. 94.4, MIP of baseline MR subtracted from that immediately post-contrast) allow for an MR system table time of as little as 3 minutes, potentially increasing access to screening breast MRI.

Fig. 94.2 Dynamic MR imaging of a large malignant lymph node.

Fig. 94.3 Breast malignancy, dynamic MR imaging time curves.

Fig. 94.4 Negative mammogram, dense breast tissue, invasive breast cancer (MR). Reprinted with permission from Kuhl CK. The changing world of breast cancer: a radiologist's perspective. Invest Radiol. 2015;50:615-28.
Fig. 95.1 presents multiple axial images from a silicone breast implant MR exam, together with a single sagittal image through the left breast. The imaging sequence employed both water and fat suppression, with silicone depicted as high signal intensity. Both implants demonstrate infolding of the silicone envelope, with silicone outside the envelope but contained within the fibrous capsule (a bilateral encapsulated leak, also known as an intracapsular rupture). The most reliable indicator of implant rupture on MR is the wavy line or “linguine” sign. Well seen in the left breast, this sign...
is characterized by multiple thin curvilinear lines within the implant, corresponding to the collapsed implant shell surrounded by silicone.

The main indication for MR of breast implants is for evaluation of possible rupture involving a silicone gel-filled implant. The prevalence of implant rupture is high, likely > 50%, with many asymptomatic. If rupture of an implant is detected, the amount and location of silicone within the soft tissues should be reported. In the imaging of implants, 2D inversion recovery sequences (using fast spin echo technique, see Chapter 34) are typically employed. The inversion time TI can be set for either fat suppression (TI ≈ 150 msec at 1.5 T and about 230 msec at 3 T) or for silicone suppression (TI ≈ 400 msec at 1.5 T). To depict silicone only (as in Fig. 95.1), the signal from both fat and water must be suppressed. This is accomplished by employing an inversion recovery sequence (with TI set for fat suppression, see Chapter 48) in combination with a spectral water suppression pulse like that employed for spectral fat saturation (see Chapter 46), but with the RF pulse applied prior to the spin preparation excitation set at the specific resonant frequency of water, thus saturating the spins at this frequency. Excellent homogeneity of the main magnetic field is a prerequisite for this approach. The presence of a high concentration of fat or silicone can lead to an incorrect adjustment of the MR system. It should therefore be confirmed that the resonant frequency is centered on water. The pulse sequence, with spectral saturation thus set to the resonance frequency of water, is then initiated.

In a plot of amplitude versus frequency (Fig. 95.2), when displaying the MR spectra, three peaks will be seen (in the order of increasing frequency from left to right silicone, fat, and water). The fat peak is ≈ 3.5 ppm and the silicone peak (which normally slightly overlaps the fat peak) 4.5 ppm lower than that of the water peak (independent of field strength). This means that the silicon peak is about 300 Hz lower than the one from water at 1.5 T, the fat peak about 220 Hz. Investigating the spectral peaks, one can therefore easily determine the composition of the implant whether it is silicone (peak 300 Hz from water) or saline (peak at water).
Today, hepatic fat can be quantified by several different MR techniques. Two well-established methods are fat-selective imaging using spectral-spatial excitation and in-phase/opposed-phase gradient echo imaging (the Dixon technique). Illustrated in Fig. 96.1 are regional differences in intrahepatic lipid (IHL) revealed by the (a, b) in-phase/opposed-phase gradient echo technique. The IHL map calculated by this technique is illustrated in (c), as an overlay (on the opposed-phase image), and was corrected for signal decay due to both transverse and longitudinal relaxation. Note the increased lipid content in the lateral right lobe. Both methods are reliable with some inherent limitations: spectral-spatial excitation requires a very homogenous static magnetic field, necessitating time-consuming shimming in particular with obese patients, but directly visualizes IHL. In distinction, the in-phase/opposed-phase technique requires correction for both T1- and T2*-relaxation effects, which necessitates additional measurements and leads to more complex post-processing. However, both techniques provide a quantitative measurement of percentage of intrahepatic lipid, and correlate well with each other.

In the case of GRE-based Dixon techniques, additional T1 correction techniques may not be necessary if the influence of longitudinal relaxation on hepatic fat quantification is minimized by using low excitation flip angles (≤ 10 degrees). Furthermore, T2* relaxation effects can also be estimated by the acquisition of an additional in-phase echo within the same sequence, which is known as the three-point (echo) Dixon method. The effective transverse relaxation (T2*) can then be relatively simply calculated from the two in-phase images and used either for the correction of IHL quantification or as an estimate of hepatic iron deposition.
The three-point Dixon approach has excellent correlation with liver biopsy for hepatic fat content and is not limited by fibrosis or cirrhosis. Illustrated in Fig. 96.2 are results in a different patient also depicting regional differences in hepatic fat content, but using a three-echo Dixon technique (with correction for T2* effects). (a) is a fat fraction map that covers a fat range of 0–100%, with (b, c) demonstrating on histology hepatic fat content of 5% and 95%, with these two specimens obtained by ultrasound-guided biopsy from the two regions of interest noted in (a).

Another recent approach is based on single voxel spectroscopy, in which fat quantification is achieved by extrapolating the fat and water integrals for TE = 0 from an exponential fit of points acquired at multiple TEs. This spectroscopy-based method also allows estimation of liver iron deposition. Although restricted to a single voxel, the scan can be acquired in a single breath-hold (15-second acquisition time). This approach potentially is the most accurate for quantifying hepatic fat, although results are comparable with multiecho methods. Technical complexity and spatial coverage are limiting factors for spectroscopy-based techniques.

Fig. 96.2 Three-point Dixon quantification of hepatic fat content, with histologic correlation. Used with permission from Kühn JP, Evert M, Friedrich N, et al. Noninvasive quantification of hepatic fat content using three-echo dixon magnetic resonance imaging with correction for T2* relaxation effects. Invest Radiol. 2011;46:783-89.
Liver iron content is an important clinical marker, with biopsy still considered to be the gold standard for its determination. The MR determination of iron offers advantages clinically due to the invasive nature of biopsy and the large sampling variability of the latter.

Iron stored in the liver alters the MR signal in characteristic ways: by increasing the magnetic susceptibility it alters the Larmor frequency, and by microscopic effects increases the transverse relaxation rates, $R_2$ and $R_2^*$. The latter effect is most commonly exploited for MR-based techniques. Standardized multi-spin-echo imaging ($R_2$) relaxometry is the base of a commercial iron quantification service with regulatory approval: FerriScan (Resonance Health, Claremont, Australia), but only for a per-scan fee and at 1.5 T. Iron-induced signal intensity changes between liver and a reference tissue can be exploited. $R_2^*$ itself can be estimated on most modern scanners, but iron calibration equations vary with the details of acquisition and post-processing, and these have not been standardized. Recently, however, single breath-hold 3D multi-gradient-echo sequences with short initial echo times and short echo spacing, both on the order of 1 msec, have gained considerable interest. In combination with an $R_2^*$ estimation which corrects for the major confounding factors causing variability, fat signal effects and noise bias, the matching iron calibration equations seem to converge. Also, they provide simultaneous estimation of the liver fat fraction which is corrected for relaxation effects.

Fig. 97.1 Imaging of fat and iron, single breath-hold multi-echo VIBE Dixon.
In the approach just described, a VIBE Dixon scan is acquired with six echoes and whole liver coverage in a single breath-hold. \textbf{Fig. 97.1} and \textbf{Fig. 97.2} show the inline results in a subject with geographic fatty liver deposition scanned at 3 T.

In \textbf{Fig. 97.1}, the third and fourth echoes are first presented. Underneath these are the subsequently calculated water and fat images. Note the low signal intensity areas (asterisk) on the out of phase gradient echo scan (TE = 3.7) both anteriorly and posteriorly in the right lobe of the liver, reflecting fatty deposition. These are high signal intensity on the fat image. The third column of \textbf{Fig. 97.1} presents the fat fraction image and the R2* map. Note that the liver is homogeneous in the latter, reflecting a uniform distribution of iron.

\textbf{Fig. 97.2} presents the results from additional inline processing based on a previous step of liver segmentation and a ROI prescribed during sequence planning. The upper part of the figure gives the mean fat fraction and the lower part the mean R2* over the segmentation region and ROI, respectively. A 8.5% fat fraction (approximate) represents an elevated value (mild steatosis), while an R2* of about 60 sec$^{-1}$ (at 3 T) is in the normal range.

There are many caveats to the estimation of liver iron, with potential pitfalls for the multi-echo Dixon approach just described. Studies exist using calibration equations to convert R2* to a quantitative iron value, however caution is advised due to the dependence of these on the specific acquisition parameters and details of signal model and data fitting procedure. It should also be noted that that R2* scales linearly with magnetic field strength. The measurement protocols require echo times as short as possible due to the rapid signal decay. However, the initial water/fat separation step in the adaptive fitting approach used here works best when the first two echo times are near opposed- and in-phase. These two requirements are best met at 3 T (as opposed to 1.5 T) where the first opposed-phase echo time is 1.2 msec. SNR should be as high as possible to reduce noise bias, mandating larger voxel sizes. If liver iron is very high, it can also be advantageous to use only four as opposed to six echoes. The reader is referred to published liver iron concentration calibration curves for R2* (e.g. JMRI 2014;40:1003). Approximate conversion values for R2* (in units of sec$^{-1}$) to liver iron in units of mg/g dry weight is 0.032 for 1.5 T and 0.017 mg/g Fe$_{dw}$/Sec$^{-1}$ for 3 T.
Elastography

MR elastography (MRE) noninvasively and quantitatively measures tissue stiffness. Although this technique has been demonstrated in the brain, breast, heart, and kidneys, its primary clinical application is for the liver. Three steps are required to obtain an image: 1) generating mechanical shear waves; 2) imaging the propagating waves; and 3) processing the information to calculate shear stiffness. Step 1 is accomplished by the use of an active driver outside the scan room to produce vibrations (at 60 Hz, with multi-frequency MRE applying a range, 40–70 Hz) in combination with a passive driver, typically a flat disk-shaped vibration source, placed against the body in the region of interest. The latter generates the acoustic waves. During imaging, motion sensitizing gradients are used in combination with a phase contrast pulse sequence, acquiring images of the propagating waves. This data is then processed to generate cross-sectional images depicting the tissue’s mechanical properties (typically the magnitude of the shear modulus, \(|G^*|\), in units of kilopascal) using a color scale (Fig. 98.1). In addition to the shear stiffness maps, “confidence maps” are provided, as are stiffness maps with the “unreliable” regions marked. Images depicting the wave shape and amplitude are also provided to allow additional assessment of the scan quality. The commercially available 2D techniques are MRE-encoded only in one direction (through-plane for axial scans), and it is recommended to use only a single stiffness value for liver characterization, averaged over the largest possible ROI, excluding vessels. Localized stiffness measurements are not recommended. 3D-encoded (research) applications allow the characterization of other, smaller organs, and localized stiffness measures.

Fig. 98.1 MRE magnitude and \(|G^*|\) maps, pre- and post-transjugular intrahepatic portosystemic shunt implantation (TIPS). Used with permission from Guo J, Büning C, Schott E, et al. In vivo abdominal magnetic resonance elastography for the assessment of portal hypertension before and after transjugular intrahepatic portosystemic shunt implantation. *Invest Radiol.* 2015; 50:347-51.
For the liver, MRE offers an alternative to liver biopsy for the diagnosis of hepatic fibrosis. Advanced fibrosis can result in cirrhosis, portal hypertension, and liver failure, with transplantation one of the few treatment options. Imaging of the liver is performed with single-slice techniques, acquired in a single breath-hold of approximately 20 seconds at end expiration. Multiple slices may be acquired in additional breath-holds for more reliable results. Confounding factors include the presence of inflammation, either acute or chronic, portal hypertension, venous congestion and malignant cellular infiltrates: all can elevate liver stiffness independent of fibrosis. If the liver iron content is very high, the signal intensity may also be too low to visualize the mechanical waves, resulting in a failed exam (about 4% of clinical exams). Spin echo EPI based (as opposed to GRE) MRE sequences, recently introduced, are much less sensitive to iron overload, resulting in few if any failed exams.

More recently, liver and spleen stiffness have been measured by 3D multi-frequency MRE in order to detect and assess the severity of portal hypertension. MRE in this application can be used to assess therapeutic success, as well as for long term monitoring, after TIPS intervention for portal hypertension. With this specific approach, |G*| for both the liver and spleen are sensitive for portal decompression. Values in the spleen before and after TIPS correlate linearly with the hepatic venous pressure gradient, an invasive measurement, which currently is the gold standard for assessing the hemodynamic outcome of this procedure.

Potential applications in the brain include assessing tumor stiffness for meningiomas, pituitary macroadenomas, and vestibular schwannomas, to discriminate between firm tumors and more easily resected soft tumors. Tumor-brain adhesion can also be assessed, for example with vestibular schwannomas, providing information regarding surgical difficulty and risk. In the heart, MRE can be used to assess myocardial relaxation abnormalities (diastolic functional disorders), which otherwise can only be assessed by ventricular filling on echocardiography.

Renal transplant fibrosis has been shown to be a predictor of progression in chronic renal disease. As with the liver, MRE offers a more complete anatomical assessment of fibrosis, which is important for the kidney where fibrosis is distributed heterogeneously. Multi-frequency MRE can effectively assess renal allograft function and detect impaired graft function. Stiffness also correlates with GFR and resistive index. Interestingly, renal allograft insufficiency leads to lower |G*| (Fig. 98.2), unlike hepatic fibrosis which results in higher values. Fibrosis in a kidney transplant does not translate to increased tissue stiffness, likely due to multiple factors, with renal stiffness higher in functioning versus nonfunctioning allografts.

![Fig. 98.2 MRE parameter map |G*|, functioning and non-functioning pelvic renal transplants. Used with permission from Marticorena Garcia SR, Fischer T, Durr M. Multifrequency magnetic resonance elastography for the assessment of renal allograft function. Invest Radiol. 2016; 51:591-95.](image-url)
Magnetic Resonance Cholangiopancreatography (MRCP)

Obstruction of the biliary or pancreatic ducts (Fig. 99.1) can lead to a variety of complications and symptoms, including severe pain. Examining these areas to rule out the presence of an obstructing stone historically required invasive procedures such as endoscopic retrograde cholangiopancreatography (ERCP). Magnetic resonance cholangiopancreatography (MRCP) is a simple, noninvasive MR technique used to investigate the biliary and pancreatic systems as well as surrounding organ systems.

MRCP studies begin with a series of T1- or T2-weighted acquisitions used for localization as well as visualization of tissues surrounding the ducts in question. This is demonstrated in Fig. 99.2a where a 2D spoiled gradient echo sequence, acquired in a breath-hold fashion, depicts the distended common bile duct (CBD, arrow) and normal pancreatic duct. Additional coronal acquisitions are occasionally used to further specify ductal locations or the positioning of the field-of-view, as presented in Fig. 99.2b. Once localization is complete, two MRCP methods are then routinely employed.

Following conventional imaging, a series of single-slice (thick slab), single-shot, 2D acquisitions with a long echo time on the order of 150 milliseconds or higher are acquired.
at various angles in an effort to best display the anatomy without super-imposition of vessels or bowel. Fig. 99.3a demonstrates the localizer used to acquire Fig. 99.3b where overlay from the common bile duct impairs visualization of the pancreatic duct. By slightly angling the localizer (Fig. 99.3c) and repeating the acquisition, the overlap is avoided and the ducts are displayed clearly (Fig. 99.3d). This 2D approach offers the advantage of rapid acquisitions (< 5 sec), allowing for repeated variations.

In addition, or as an alternative to the 2D acquisitions, a 3D scan can be acquired with isotropic, high spatial resolution, to make the most benefit of post-processing techniques. This scan in the past required 4 to 6 minutes, with navigator echoes used to track respiration, allowing the patient to breathe freely while the data are acquired. Fig. 99.4 demonstrates the resulting maximum intensity projection (MIP) image. The MIP removes tissues exhibiting lower SNR, resulting in a clear visualization of the ductal system. The data are acquired isotropically allowing reconstructions with various rotations so that the ductal system can be viewed from different angles. Taking advantage of data sparsity, this high-resolution 3D scan can now be acquired in a single breath-hold.
Articular cartilage is made up of hyaline cartilage, which has little capacity to recover once injured. It consists of collagen fibers, glycosaminoglycans, extracellular water, and a small number of chondrocytes: glycosaminoglycans and collagen being the two most abundant components in healthy cartilage. Cartilage mapping enables detection and quantification of changes in cartilage biochemical composition noninvasively. The cartilage mapping techniques most widely used include delayed gadolinium-enhanced MR of cartilage (dGEMRIC) together with T2 and T2* mapping. While some authors have proposed to use these techniques routinely in clinical exams, they remain mainly in the research domain, e.g. for quantifying cartilage quality in patients for a clinical trial.

dGEMRIC is an MR technique that allows evaluation of glycosaminoglycan content in articular cartilage by assessment of the T1 relaxation time. Glycosaminoglycans are large macromolecules with negatively charged side chains. Some gadolinium chelates are negatively charged (the ionic agents), and after IV administration show a distribution within the cartilage that is inversely proportional to the concentration of glycosaminoglycans, thus permitting the indirect evaluation of its content. dGEMRIC uses a 2D fast inversion recovery technique or 3D VIBE pulse sequence with variable flip angles to generate T1 maps. Following IV administration of a double dose of an ionic gadolinium chelate, the patient performs physical exercises such as using a treadmill or walking on stairs for 10 to 20 minutes. Lower values (on T1 maps) are commonly observed with dGEMRIC in joints with osteoarthritis, due to T1 shortening in areas with glycosaminoglycan loss. The dGEMRIC technique can also be used to detect early osteoarthritis. An alternative to IV contrast administration, for dGEMRIC, is direct intraarticular injection of gadolinium. Fig. 100.1 shows a sagittal T1 dGEMRIC image of the hip in a 31-year old male patient, with lower T1 values in the periphery of the acetabular cartilage at the anterosuperior portion of the joint compared to that more centrally where normal cartilage is present (mean measurement 229 vs. 507 ms, inset).

T2 mapping is a technique to assess the interaction between water molecules and the collagen fiber network in articular cartilage. T2 mapping typically uses a 2D multislice multiecho Carr-Purcell-Meiboom-Gill (CPMG) sequence. T2 values of hyaline cartilage are highly sensitive to changes in collagen concentration and anisotropic orientation of the collagen fibers. Increased T2 values are commonly related to cartilage damage. Fig. 100.2 shows a macroscopic defect (arrow) at the lateral femoral condyle seen both on sagittal T2 mapping of the knee and the sagittal PD fat sat at the corresponding position.
Elevated T2 values (arrowheads) on T2 mapping correspond to areas with deterioration of the integrity of the collagen network. This cannot be detected on the PD fat sat sequence. T2 mapping is mainly used to detect early osteoarthritis and breakdown of cartilage architecture and to monitor, following surgery, cartilage repair.

T2* mapping (Fig. 100.3) is an alternative technique to T2 mapping. T2* mapping uses a 3D multi-echo gradient-echo pulse sequence. T2* relaxation reflects both T2 relaxation and microscopic susceptibility related dephasing effects within the net T2* decay. The regional variation in T2* also reflects regional variation in T2 and the micro-structure of the cartilage. The lower T2* mapping values reflect the low T2 values with the additional contribution of microscopic susceptibility fields. Fig. 100.4 demonstrates T2* mapping of the knee with normal areas of cartilage in dark blue at the anterior portion of the lateral tibia (arrowhead). Directly posterior to the normal cartilage there is an area of degenerated cartilage with higher T2* signal (arrow). T2* mapping offers shorter imaging times and higher spatial resolution, using 3D acquisitions to assess the ultrastructure of articular cartilage. Secondary reconstructions of the acquired 3D data set can be performed for better evaluation of the different regions of the whole joint.

Cartilage mapping can be implemented on most clinical MR scanners and provides quantitative techniques to assess the biochemical properties of cartilage. The use of cartilage mapping may improve detection of early cartilage lesions as well as providing for monitoring, via imaging, of the surgical outcome of cartilage repair.

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Fig. 100.2 T2 mapping, cartilage defect, associated femoral and tibial cartilage damage.

Fig. 100.3 T2* mapping, normal patellar cartilage. Used with permission from Andreisek G, Weiger M. T2* mapping of articular cartilage: current status of research and first clinical applications. Invest Radiol. 2014;49:57-62.

Fig. 100.4 T2* mapping, tibial cartilage.
Section VII

Artifacts, Including Those Due to Motion, and the Reduction Thereof
Fig. 101.1 and Fig. 101.2 present examples of aliasing, defined as wraparound of structures outside a specified field of view (FOV) to the opposite side of the image. The phase encoding direction is along the horizontal axis in Fig. 101.1 and along the vertical axis in Fig. 101.2. In Fig. 101.1a, a coronal post-contrast T1-weighted scan at 3 T (in a 5-year-old patient), the left side of the head is aliased to overlie the right side of the head, and vice versa. Aliasing from the ears (white arrows) obscures both the low signal intensity portion (on the patient’s right) and the enhancing portion (on the patient’s left, black arrow) of this pontine glioma. In Fig. 101.2a, the neck is aliased to the top of the image (arrow). In Fig. 101.2c, the top of the spine is aliased to the bottom of the image and vice versa. Increasing the FOV by oversampling eliminates aliasing (Fig. 101.1b; Fig. 101.2b,d).

Aliasing can occur along any axis, including the frequency-encoding, phase encoding, and, in 3D imaging, along the slice selection axis. In the frequency-encoding axis, sampling of an echo occurs at a rate known as the sampling bandwidth. The range of frequencies sampled increases with higher sampling bandwidths. The highest frequency that can be sampled unambiguously is a value termed the Nyquist frequency. Frequencies higher than the Nyquist frequency, corresponding to tissue outside the specified FOV, falsely appear to originate from a lower frequency. The end result, in terms of the image, is that signal from such tissue wraps around to the opposite side along the frequency-encoding axis. With modern MR systems, however, aliasing along the frequency-encoding axis has been largely eliminated by automatic oversampling methods. Oversampling involves increasing the FOV, and thus increasing the number of sampled frequencies. This is typically done by increasing the sampling rate, without any change in overall acquisition time.

In the phase encoding axis, aliasing may also occur due to the same principles as noted previously for the frequency-encoding direction (i.e., the signal is not sampled

Fig. 101.1 Aliasing (wraparound artifact) in the coronal plane, pontine glioma.
“finely” enough to logically represent the signal distribution in the raw data). As with the frequency-encoding axis, oversampling can eliminate wraparound. Oversampling in the phase encoding axis has the disadvantage of increasing scan time by a factor equal to the percent of oversampling (with all other factors held constant). However, it has the advantage of increasing the signal-to-noise ratio (see Chapter 27).

In 3D imaging, aliasing may also occur along the slice selection axis if the slab excitation includes structures beyond what is phase encoded. Again, oversampling by increasing the number of phase encoding steps so that all of an excited slab is encoded can remove aliasing. Use of saturation pulses and removing extraneous slices from each end of an excitation slab are alternate methods of removing wraparound artifact.

Fig. 101.2 Aliasing in the phase encoding direction, sagittal imaging.
Truncation Artifacts

Fig. 102.1a suffers from prominent truncation artifacts (black arrows) in the form of ringing (or “ring-down”), seen as alternating light and dark bands within the periphery of the brain. Fig. 102.2a illustrates a syrinx-like artifact (white arrows) within the center of the thoracic cord on a T1-weighted sagittal scan, a classic truncation artifact that appears as a result of alternating dark and light signal bands overlying the spinal cord. This artifact is easily mistaken for hydromyelia (dilatation of the central canal, more commonly but incorrectly referred to as a “syrinx”), with the scan obtained in this instance in a normal, healthy volunteer. The artifacts in both scans result from an inappropriate choice of pixel size (too large). Truncation artifacts are reduced by the use of a smaller field of view (FOV) or a larger matrix size, the latter approach is illustrated in both Fig. 102.1b and Fig. 102.2b.

The data acquired in k-space (see Chapter 12) for an MR image represents a finite amount of digital information due to the fact that the MR signal is sampled over a limited period of time. This inevitably leaves out (truncates) waveforms of data. A Fourier transform is used to reconstruct the MR image from the phase-encoded and frequency-encoded signal (in k-space), but is necessarily applied to this incomplete or truncated dataset. The end result, in the image, is deviation from the ideal signal intensity for any given image pixel. This phenomenon causes artifacts in particular in areas where the signal changes abruptly, because Fourier transform analysis poorly accounts for such changes (it is better suited for gradual changes in signal intensity).

Truncation artifact, also referred to as “Gibbs’ ringing” or edge-ringing artifact, appears in an MR image as “ringing,” which involves alternating light and dark bands.
As a result, there may also be edge enhancement between areas of low and high signal intensity. Truncation artifacts are often also associated with image blurring and distortion of the size and shape of structures, both results of the use of a low-resolution matrix. Truncation artifacts can be diminished by several mechanisms. Decreasing the size of the pixels relative to the geometrical structures in the image, achieved by decreasing the FOV or increasing the matrix size, reduces the representation of the artifact on the final image. However, increasing the matrix size increases scan time and decreases the signal-to-noise ratio. Alternatively, decreasing the FOV may cause aliasing. These detrimental effects on image quality must be weighed against the possible benefit of removing the truncation artifact. Use of a filter that discards artifactual frequencies is another means of decreasing truncation artifact.

![Fig. 102.2 Truncation artifacts in sagittal spine imaging, mimicking a syrinx.](image-url)
103  Motion: Ghosting and Smearing

Ghosting typically occurs in the phase encoding direction, which is along the vertical axis in Fig. 103.1a and the horizontal axis in Fig. 103.1b. Note that wraparound artifact (see Chapter 101) is present in both images also along the respective phase encoding axes. Fig. 103.1 and Fig. 103.2 display ghosts (arrows) caused by motion during acquisition of data that encodes edge information (high spatial frequency at the periphery of k-space). Thus, motion artifact appears as edge-related ghosts of structures of high signal intensity. Fig. 103.2a presents a T1-weighted axial image of the brain without motion artifact. With a small amount of motion (Fig. 103.2b), a few ghosts appear, but with more substantial motion (Fig. 103.2c), increased number and better defined ghosts appear. Discrete ghosts of the entire spatial information (not illustrated) are caused by cyclic alteration in the MR signal from motion occurring throughout the period of data acquisition, often seen in pulsatile artifact (see Chapter 121). Blurring (not illustrated), another form of ghosting, is the loss of boundary definition due to slow modulation of the signal throughout data acquisition. Smearing (also not illustrated) occurs when signal is acutely disturbed during acquisition of the center lines of k-space. It appears as smudging across a major portion of an axis (usually the phase encoding axis) from any structure that produces signal, not just those at the edges of the image.

Ghosting or ghost artifacts are mis-mapped signals of a part or all of a structure within a given image. The appearance of ghosts depends further on when and how alteration of signal occurs during data acquisition. Structures with high signal intensity (e.g., fat on a T1-weighted scan) have the highest propensity to cause ghosting. Although ghosting theoretically may occur in either the frequency-encoding (readout) or phase encoding direction, it almost always occurs in the phase encoding direction. Data in the phase encoding direction takes a relatively long time to obtain (hundreds of milliseconds to minutes), and thus motion has ample opportunity to alter the MR signal and cause artifact. This is in contradistinction to data acquisition in
the frequency-encoding direction, which takes only several milliseconds to occur. In such a short period of time, motion is not able to cause any significant artifact. Theoretically, if data acquisition in the frequency-encoding direction were to occur over a longer period of time, motion artifact might be seen in that direction. This, however, does not typically occur in clinical MR. Ghost artifacts may obscure true abnormalities or even mimic pathology. In addition, such artifact removes signal intensity from structures of interest (lowers the signal-to-noise ratio), and thus decreases one’s ability to distinguish between adjacent tissues.

Improper encoding due to error induced by the MR unit or the environment causes ghosts. Motion either during or between sampling of echoes also causes ghosts. When motion occurs between MR echo sampling, data from a single point in the image may actually come from a different structure or different area within the same structure. Each view of the data point, therefore, will contain an MR signal with a different intensity. This results in a form of ghosting. Motion between sampling of echoes may be due to macroscopic physical motion. Macroscopic physical motion during a scan may be from patient movement, breathing, the cardiac cycle, bowel peristalsis, or pulsatile motion of large vessels (see Chapter 121). Large pulsatile vessels may also cause artifact due to motion that occurs during echo sampling, leading to phase shifts and saturation effects.

Fig. 103.2 Severity of ghosting, dependence upon degree of motion.
One of the major drawbacks in MRI has been, and still is, artifacts due to motion of the patient, which may be secondary to discomfort, pain, unconsciousness, or lack of compliance. In addition, cardiac motion, respiratory motion, and peristalsis of the bowel may markedly degrade image quality. Although the development of modern sequences with motion correction (e.g., BLADE; see Chapter 106) allows for diagnostic imaging in uncooperative patients, MR evaluation of the thorax and abdomen in uncooperative patients generally necessitates special triggering and gating techniques to obtain diagnostic images.

◆ Cardiac Triggering

Despite the availability of MR sequences with an acquisition time below 1 sec, cardiac MR (Chapter 89 through Chapter 93) in general requires special triggering techniques to provide diagnostic images with sufficient spatial resolution and SNR without motion artifact. Cardiac triggering requires monitoring of a biological signal corresponding to the heartbeat. This is feasible by either monitoring the ECG or the pulse. For ECG monitoring, MRI-compatible ECG leads are necessary to prevent burns at the sites of application. Typically, R waves are utilized to trigger the sequence acquisition as they commonly provide the highest voltage and are thus easiest to detect. Alternatively, pulse signals may be obtained for triggering of the sequence acquisition by applying a pulse monitor to a finger or a toe. Although its ease of use is an advantage, major drawbacks include inconsistencies due to finger motion, weak pulses due to peripheral artery disease or cold extremities, and a slight delay in signal triggering compared with R wave detection by ECG.

In cardiac MR, one must decide whether to apply a single-phase or a multiphase (cine) imaging approach. Note, however, that in this context phase refers to different times within the cardiac cycle and not to a certain phase of proton spins. Cine imaging is a multiphase technique that is used to obtain information about ventricular function and the morphology and function of the cardiac valves, and to quantify flow and describe the flow pattern within the cardiac chambers, across the valves, and within the great arteries.

The techniques used to perform cine imaging (e.g., SSFP) employ a very short TR. This allows the same slice to be acquired at different times during the cardiac cycle. SSFP sequences can employ either prospective triggering or retrospective gating, monitoring either pulse or ECG signals. In the latter, data acquisition is performed continuously during the cardiac cycle while the ECG signal is monitored at the same time. The acquired k-space data are thus given a time stamp relative to the cardiac phase. After complete acquisition of all Fourier lines, in a second step the acquired data are sorted into different k-space matrices encoding for different images displaying different times in the cardiac cycle. Using a prospectively triggered approach, cine SSFP
sequences allow the option of arrhythmia rejection based on the average duration of the cardiac cycle. In the case of an abnormally long or short R-R interval, the acquired data during the corresponding cardiac phase is retrospectively rejected. In addition, this application, known as arrhythmia rejection, automatically extends the scan time to sample the rejected data in further prospectively triggered phase encoding steps.

The method of choice for single-phase imaging is prospective triggering. With this imaging approach, one slice is acquired during a single breath-hold (or during free breathing with a navigator echo) providing so-called dark-blood images, with excellent depiction of myocardial morphology. In prospective ECG triggering, image acquisition starts with the detection of the QRS complex. After a trigger delay, which is user dependent but should be between 60 and 80% of the R-R interval, a so-called segmental data acquisition will be performed during the end-diastolic cardiac phase. At this phase of the cardiac cycle, the cardiac chambers are not actively contracting, diastolic filling is completed, and there is no significant flow through the chambers; therefore, the images are less vulnerable to artifacts based on cardiac motion and blood flow. With each trigger pulse, a certain number of phase encoding steps are acquired, contributing to the information needed to reconstruct the slice. This means that with each heartbeat a certain segment of k-space is acquired. Assuming for example that the fast spin echo sequence employed has a matrix size of 256 × 160 with an echo train length (ETL) of 32, then 5 (160/32) trigger pulses are necessary for full k-space acquisition. With a TR covering two R-R intervals, the entire k-space sampling will require 10 heartbeats.

Images are illustrated in Fig. 104.1 from the cardiac MR exam of a 51-year-old woman who presented with acute pulmonary edema. Four-chamber cine SSFP scans are illustrated, acquired during (a) systole and (b) diastole using retrospective ECG gating. Note the large inhomogeneous hypointense mass within the left atrium that

Fig. 104.1 Left atrial myxoma, cardiac triggering.
prolapses into the left ventricle, across the mitral valve, during diastole. On (c) the corresponding two-chamber cine SSFP scan, the attachment of the mass to the posterior left atrial wall is clearly depicted. Multiphase dynamic SSFP sequences in this instance clearly demonstrate the location of the mass (a large left atrial myxoma), its attachment at the left atrial wall, and the prolapse across the mitral valve leading to pulmonary venous congestion with subsequent pulmonary edema. Fig. 104.1 illustrates a T1-weighted inversion recovery sequence acquired following contrast administration using a prospective ECG-triggered technique. On this single-phase image acquired during diastole, the mass is hypointense compared with the surrounding enhancing blood.

◆ Respiratory Gating

MR imaging of the chest and abdomen in particular is very vulnerable to respiratory motion. In newborns and children, MRI rather than CT is the preferred examination due to the lack of ionizing radiation and the superior soft tissue contrast. Respiratory artifacts represent a major drawback, particularly in this population, because of the inability of the patient to cooperate with breath-holding. In addition, in adult patients who are not able to perform an adequate breath-hold, the quality of MRI studies may be markedly limited due to breathing artifacts. Nevertheless, there are several possibilities to overcome these problems.

One of the simplest options to reduce respiratory artifacts during sequence acquisition is to reduce the scan time per acquisition or to split up the examination of a certain anatomical section into several shorter acquisitions. This approach may be successful in semi-compliant patients who are able to perform a short breath-hold.

In uncooperative patients, one alternative to breath-hold scans is to increase the number of signal averages (NSA). Increasing the NSA increases the signal of the static anatomy image while the SNR attributes of the ghosts due to motion are diminished. However, limitations of this technique, which is today rarely employed, include an increase in scan time and that artifacts are not avoided or eliminated.

Further imaging techniques to limit patient motion are based on monitoring of respiration. Respiratory gating and respiratory compensation (or reordering) are two possible techniques to improve image quality. With either technique, respiratory motion can be monitored using a bellows or breathing belt, typically strapped around the thorax. Inspiration and expiration cause the bellows to first distend and then contract, which is registered and subsequently transformed into an electric signal.

In respiratory gating, sequence acquisition is performed during only a portion of the respiratory cycle, usually end-expiration. During end-expiration, the chest and abdominal wall are nearly motionless, limiting ghost artifacts that degrade image quality. TR is defined at least in part by the respiratory rate. Due to inconsistent results, and the time needed for setup and scan acquisition, respiratory gating is now little used.

Respiratory compensation or phase reordering, also referred to as respiratory-ordered phase encoding (ROPE), is another technique to reduce respiratory artifacts. With this technique, image acquisition, or more precisely phase encoding, is performed throughout the respiratory cycle. The most positive values of phase encoding are sampled during inspiration whereas the most negative values are acquired during expiration. In contrast to respiratory gating, in which sequence acquisition is performed during expiration, the intermediate phases of the respiratory cycle are also used for consecutive k-space sampling. This makes the technique more time efficient. Further
improvements of this technique use the longer expiratory cycle for central k-space sampling, leading to improved image quality. Like respiratory gating, ROPE is little used today.

◆ Navigator Echoes

Respiration can also be monitored, in a very sophisticated way, by adding an extra "navigator" echo to the scan acquisition. The additional information from the navigator echo is used to improve image quality either by retrospective correction or in a prospective manner. The most common clinical application, which has gained widespread acceptance, is the monitoring of diaphragm motion in abdominal imaging. In this context, a navigator echo is acquired that samples a small column of tissue in the craniocaudal direction across the diaphragm. The result is a one-dimensional image of the tissue boundary between the thorax and abdomen, with the temporal change in signal intensity providing a reference for the position of the diaphragm. One such technique is termed Prospective Acquisition CorrEction (PACE).

Fig. 104.2a presents a coronal scout image of the lower thorax and upper abdomen. The vertical column covering the right dome of the liver represents the navigator echo. The temporal (one-dimensional) data acquired from the navigator echo during free breathing is displayed in Fig. 104.2b. Fig. 104.2c presents an axial abdominal image acquired during normal breathing using a navigator echo, with data acquisition only during the user defined acceptance window relative to the position of the diaphragm. Fig. 104.2d illustrates the same scan technique, acquired during normal breathing, but without the use of a navigator echo. The sequence acquired with a navigator echo provides excellent depiction of the abdominal organs, while the scan acquired without a navigator echo is markedly degraded by motion artifacts and can be considered nondiagnostic.

Fig. 104.2 Use of a navigator echo for improved image quality in the upper abdomen.
Physiologic motion (e.g., respiration and cardiac pulsation) can lead to artifacts, reducing the diagnostic quality of MR images. Most MR systems offer hardware, software, and sequence-based options designed to minimize or eliminate the effects of physiologic motion during scan acquisition. Motion artifacts in MR images are caused by translations or shifts in the position of the imaged structure during the acquisition of data. Routine spin echo and gradient echo sequences encode and acquire data to fill k-space in a Cartesian fashion, with the collection of each line separated by a certain amount of time (TR). In fast spin echo imaging, several, but not all, lines are collected with each TR. Because all the data for an entire slice is not acquired simultaneously, changes in the location of anatomic structures between TR periods can lead to misalignment in spatial encoding. Reconstruction of the data results in blurring or ghosting (see Chapter 103) in the final image when compared to that without motion (Fig. 105.1).

The simplest and most effective method for eliminating respiratory motion is to have the patient hold their breath. Rapidly acquired, multislice, T1- and T2-weighted scans are used to collect data during suspended respiration, thereby minimizing motion artifacts. However, this approach (until recently) required the scan time to be 25 seconds or less for the average patient (vs. 1 minute in 1988), leading to limitations in spatial resolution and/or number of slices. The use of this approach is also restricted by the patient’s health and mental status, as well as their age. Single slice techniques such as HASTE (see Chapter 39), TrueFISP (see Chapter 56, Chapter 57), and echo planar imaging (EPI) acquire the data for an entire slice before beginning the next slice. In most cases, the acquisition is rapid enough to freeze respiratory motion, greatly minimizing artifacts within each slice. These methods do not, however, account for changes in anatomic position between slices. The addition of advanced techniques, in particular software-based tracking, is necessary to assure consistent and complete anatomic coverage.

Hardware-based gating techniques made use of a bellows-like device placed around the patient’s chest to track respiratory motion. Data was acquired during a specific portion of the respiratory cycle defined during setup. This method did greatly reduce

![Fig. 105.1 Breathing vs. breath-hold, SE technique, 1 min acquisition from 1988.](image-url)
respiratory artifacts; however, scan times were lengthened by a factor of 2 or more, with this approach no longer in general use.

Software-based gating/tracking, such as the implementation of navigator echoes, are an important alternative—to breath-hold techniques and ultrafast imaging—for the reduction of respiratory-related image artifacts. These methods do not require additional hardware or patient cooperation. The simplest, one-dimensional navigator (see Chapter 104) incorporates additional RF pulses in the sequence to track superior to inferior translational motion of abdominal and thoracic structures based on the position of the diaphragm. Information from the navigator echo can be used to trigger data acquisition during a specified portion of the respiratory cycle or to adjust the anatomic (craniocaudal) position of a slice group to follow the change occurring during respiration.

In summary, there are currently four major methods available to compensate for motion and organ displacement during respiration: (1) Simply instructing the patient to hold his or her breath (2) Tracing the respiratory cycle using a “navigator” echo: selecting/rejecting acquired data and/or adjusting the slice position following the respiratory organ displacement (3) Software-based tools to correct for anatomic displacement in a post-processing fashion for repeated dynamic acquisitions (4) Novel approaches incorporating data sparsity, with continuous radial \( k \)-space data acquisition, such as XD-GRASP (see Chapter 109).

A more in-depth explanation of the use of a navigator echo, which traces the motion of the liver/lung interface to achieve improved abdominal image quality, is warranted, given the importance and relative sophistication of this approach (see also Chapter 104). For the navigator echo, a “rod of tissue” is excited, placed through the dome of the liver with a one-dimensional craniocaudal extension. The one-dimensional information is read out in parallel to the imaging sequence with the motion of the liver–lung interface serving as an indicator for breathing. The user can define whether data acquisition should be activated close to the inspiratory or expiratory portion of the respiratory cycle. Usually a tolerance in millimeters of liver excursion is taken as indication whether the data acquisition is to be switched on or off.

The use of software-based navigator echoes to reduce the time required for breath-holding (as opposed to the approach just described, which is employed with free breathing) involves breaking up large, multi-slice measurements into smaller groups of slices, reducing the breath-hold duration of each group by a corresponding amount. Differences in the diaphragmatic position between measurements due to varying levels of inspiratory volume are corrected by gradient system adjustments, reducing the chance for overlapping of slices or large gaps in slice coverage.

Some types of clinical studies—e.g., the evaluation of a liver lesion—benefit from the use of dynamic imaging. Such an exam is often performed by acquiring multiple, rapidly collected 3D datasets during IV contrast injection, requiring a breath-hold for each dataset. Frequently, the patient is unable to hold their breath with the same inspiratory volume for each acquisition, making side-by-side comparisons and subtractions difficult. Recently, respiratory motion-resolved compressed sensing reconstruction of free-breathing radially acquired scans has been advocated in this scenario, enabling diagnostic quality multiphase liver imaging following bolus IV contrast injection.
Despite the faster scans available today, motion artifacts can still substantially degrade clinical images. With BLADE (PROPELLER), which was developed to reduce motion artifacts, k-space is sampled by multiple echo trains in a rotating, partially overlapping fashion (like the rotation of a propeller), rather than in the standard rectilinear (Cartesian) fashion. Motion artifacts are diminished primarily due to the k-space trajectory, which is substantially more robust in terms of ghosting, although correction for gross in-plane rotation and translational motion can also be performed.

BLADE was first developed for T2-weighted fast spin echo (FSE) imaging and subsequently extended to FLAIR. Fig. 106.1 illustrates heavily T2-weighted scans in a patient with a large acute middle cerebral artery infarct (arrow) acquired with (a) conventional and (b) BLADE FSE technique. Marked ghosting degrades the conventional scan. Fig. 106.2 illustrates FLAIR scans acquired with (a) conventional and (b) BLADE technique. An acute pontine infarct (black arrow) is more clearly delineated in the BLADE scan, with the conventional scan degraded by gross motion artifacts and a prominent pulsation artifact (white arrow). If all other parameters are held constant, a BLADE scan

Fig. 106.1 FSE T2-weighted scans of an MCA infarct without and with BLADE.

Fig. 106.2 Reduced ghosting on FLAIR with BLADE, pontine infarct.
will be ~50% longer, but also has improved SNR - due to the inherent oversampling of the center of k-space.

T1-weighted BLADE techniques, in general, are not clinically used or available. The two primary approaches to T1-weighted brain imaging, spin echo and FLASH, cannot be adapted to BLADE due to the lack of an echo train. Fig. 106.3 compares post-contrast images acquired at 3 T with (a) FLASH and (b) TurboFLASH BLADE in a patient with numerous enhancing multiple sclerosis plaques. The GRE image is substantially degraded due to inadvertent patient motion—despite the scan time being under 1 minute—with both blurring and ghosting (arrows) evident. The BLADE scan is artifact free.

BLADE could theoretically be applied in DWI, providing an alternative to EPI approaches and associated artifacts—principally geometrical distortion and bulk susceptibility. The disadvantages in this application are low SNR and long scan time. In Fig. 106.4, axial diffusion-weighted (a) echo planar and (b) BLADE FSE scans at 3 T are compared in a patient with an acute left thalamic infarct (white arrow). Despite the use of a parallel imaging factor of 4, the susceptibility artifact from the frontal sinus (black arrow) is prominent on the EPI DWI scan. This is eliminated with BLADE. Such artifacts, particularly in the posterior fossa, can make image interpretation difficult.
Dynamic imaging of the liver allows for improved detection and characterization of focal lesions including specifically liver metastases and hepatocellular carcinoma. The advent of parallel imaging allowed the acquisition of such images during an acceptable breath-hold time, ≤20 seconds. The subsequent introduction of 2d acceleration, in particular of CAIPIRINHA, facilitated the use of higher acceleration factors. This allowed a reduction in scan time to ~10 sec, with complete coverage of the abdomen.

First past imaging with higher temporal resolution can yield additional valuable diagnostic information. During this time interval, the primary change is the extent of enhancement, with image acquisition thus amenable to time resolved techniques using echo (view) sharing such as TWIST. In such an approach, the central portion of $k$-space, which controls image contrast, is sampled more frequently than the periphery of $k$-space, which contributes principally to high spatial resolution. Although TWIST and other similar techniques were initially employed for CE-MRA, they can be combined with 3D GRE sequences such as VIBE. Combining this further with CAIPIRINHA allows for significant scan acceleration, making possible first pass dynamic contrast enhanced imaging within a single breath-hold. In addition, the Dixon technique is employed to provide robust fat suppression with high SNR and no sensitivity to $B_0$ inhomogeneities. In the initial implementation of TWIST VIBE, 14 3D data sets were acquired in an acquisition time of 29 seconds (Fig. 107.1). Disadvantages to this approach include the long breath-hold time as well as the large number of images acquired. However, some hypervascular metastases could only be visualized in clinical trials using the TWIST VIBE technique.

**Fig. 107.1** Colorectal cancer, following radiotherapy, with avascular and avidly enhancing (arrow) metastases. Fourteen dynamic frames acquired with TWIST VIBE in a single 29-second breath-hold are illustrated. A second faintly enhancing metastasis is noted (box) in only a few of the dynamic frames. Used with permission from Michaely HJ, Morelli JN, Budjan J. CAIPIRINHA-Dixon-TWIST-volume-interpolated breath-hold examination: a new technique for fast time-resolved dynamic 3-dimensional imaging of the abdomen with high spatial resolution. *Invest Radiol.* 2013;48:590-97.
In addition to improved detection of small hypervascular lesions in the arterial phase (including both metastases and hepatocellular carcinomas), TWIST VIBE provides improved depiction of arterial anatomy together with visualization of the contrast kinetics for pathologic lesions. This is in distinction to conventional imaging approaches capturing a single hepatic arterial phase. The two major disadvantages of the latter are non-optimal timing of the acquisition, commonly, and the lack of a wide window for observation of arterial hypervascular enhancement.

Subsequent implementations of TWIST VIBE have employed five or six hepatic arterial sub-phases only (Fig. 107.2), demonstrating as with the original work robust, optimized timing of the arterial phase and improved detection of the small hypervascular lesions in comparison to the standard single arterial phase acquisition. TWIST VIBE used in this fashion permits coverage of the abdomen (and specifically the entire liver) with 3-mm sections in a single breathhold, for example of 19 seconds with five acquired sub-phases.

Fig. 107.2 Hypervascular liver metastases (arrows) from thyroid carcinoma. Three (of five acquired) TWIST VIBE hepatic arterial phase images, after intravenous administration of Gd EOB-DTPA, are illustrated. The total scan time was 19 seconds, with a temporal resolution of 2.6 seconds. Note the progressive enhancement of the lesions. Used with permission from Kazmierczak PM, Theisen D, Thierfelder KM, et al. Improved detection of hypervascular liver lesions with CAIPIRINHA–Dixon–TWIST–volume-interpolated breath-hold examination. *Invest Radiol*. 2015;50:153-60.
Radial trajectories have significantly lower sensitivity to motion, when compared to conventional Cartesian schemes, with potential major advantages in instances where patient motion is problematic, for example abdominal and pediatric imaging. Radial VIBE is a 3D GRE sequence with stack-of-stars sampling, which can be employed with either a nonselective or slab-selective RF excitation. Use of this scan has proven more robust in comparison to conventional sequences for patient studies in which motion cannot be otherwise prevented.

Data in the plane of imaging (x, y) are acquired with radial sampling (radial spokes rotating around the center), while standard Cartesian phase encoding is used in the third dimension (slice direction). This results in cylindrical coverage of $k$-space, specifically a stack of discs (thus the term “stack-of-stars”). Changing the data sampling pattern from parallel lines (Cartesian) to overlapping spokes (radial) introduces unique imaging properties, both advantageous and disadvantageous. A key advantage is less sensitivity to object motion. This is due to an inherent data averaging at the overlap of the spokes in the center of $k$-space, as well as the varying readout direction in $k$-space, which prevents appearance of the typical MR ghost artifacts in the phase-encoding direction. Instead, streak artifacts can appear that radiate from the area of motion but that are usually less severe.

3D imaging with submillimeter spatial resolution can be critical for evaluation of head and neck cancer, with post contrast fat-saturated 3D VIBE playing an important role. However, motion due to patient movement, swallowing and breathing degrades a high number of studies. VIBE is particularly prone to motion artifact due to its long scan time and Cartesian filling of $k$-space. In uncooperative head and neck patients, radial VIBE is advocated due to the reduction of motion artifacts, yet still enabling submillimeter resolution (although not matching that of conventional VIBE when scan time is limited) with excellent sensitivity to the injected contrast due to the sampling of the center of $k$-space with every acquired line (Fig. 108.1).

Rotating the readout direction also allows readout oversampling in both the read and phase directions, without an effect on scan time, as occurs with Cartesian oversampling in the phase direction. Thus, a FOV can be chosen smaller than the object size without aliasing. A disadvantage of radial acquisition is higher sampling requirements due to less...
efficient $k$-space coverage (and thus longer acquisition time, Fig. 108.2). This can limit its use for breath-hold scans. In general, the number of spokes should be at least $\pi/2 \times$ matrix size to fulfill the Nyquist criterion. If the object is larger than the FOV, more spokes are needed to completely avoid streak artifacts, whereas if the object is smaller than the FOV, fewer spokes are sufficient.

Faster scans are possible in instances where some streaking is tolerable (Fig. 108.3). When imaging speed is not critical, it is advantageous to acquire more spokes than strictly necessary because this both reduces residual motion artifacts further and improves SNR. Intensities in the image reflect the probability of localization during acquisition, with motion affected areas thus seen with smudging. Due to higher sensitivity to off-resonance effects (which cause blurring instead of the chemical shift artifact seen with Cartesian sampling), radial VIBE should generally be used with fat suppression. Disabling coil elements distant from the ROI is important, as these elements tend to cause strong streak artifacts that can propagate across the image and degrade the overall quality.

**Fig. 108.2** Breath-hold 19-second CAIPIRINHA-VIBE vs. 2-minute non-breath-hold radial VIBE, delayed imaging following gadoxetate disodium injection, small hepatocellular carcinoma (arrow) in liver cirrhosis. Note the reduction in ghost artifacts.

**Fig. 108.3** Breath-hold Dixon VIBE (4.8 sec) vs. non-breath-hold radial VIBE (2 min vs. 21 sec – with differing degree of streak artifacts), liver (arrow) and renal cysts.
Dynamic contrast-enhanced T1-weighted imaging is an integral part of diagnostic abdominal-pelvic MR exams, also playing a role in other anatomic areas, for identifying and characterizing lesions. Yet such exams remain challenging with a high failure rate, especially in the abdomen. Challenges include acquiring the exam with the correct timing post-contrast, synchronization with breath-holding, and temporal/spatial resolution. Golden-angle radial sparse parallel MR imaging (GRASP) uses a T1-weighted fat-suppressed 3D spoiled GRE pulse sequence with a “stack-of-stars” radial k-space acquisition, yet instead of performing individual acquisitions at specific time points, data is acquired continuously following contrast administration. Images for the needed time points are then calculated, retrospectively, using iterative reconstruction. In addition to other advantages, this significantly simplifies the clinical workflow. For example, it is no longer necessary to precisely time the contrast injection relative to image acquisition, and the exam can be acquired during continuous breathing.

For GRASP, data acquisition is based on the radial golden angle ordering scheme. Radial data is acquired with a constant angular increment of 111.25 degrees, with successively sampled radial spokes adding complementary k-space information. Any number of grouped spokes cover k-space relatively uniformly, and thus can be used to reconstruct an individual image. Both the temporal resolution and the specific time points can be determined retrospectively. When only a few spokes are used to create an image, to achieve high temporal resolution, conventionally this would lead to severe streak artifacts. These are suppressed with GRASP by the synergistic use of compressed sensing and parallel imaging. If viewed temporally, the streak artifacts lead to a strong flickering pattern, this being artificial relative to the actual smooth temporal pattern of contrast enhancement. The total variation of each pixel over time is used as regularization during compressed sensing reconstruction iterations to obtain a solution with low total variation and, thus, reduced artifacts. Parallel imaging contributes as well to the suppression of streak artifacts by the exploitation of local coil sensitivities. Computational demands are high, both due to the application of compressed sensing as well as the enormous amount of acquired data due to the continuous acquisition.

GRASP can be used for dynamic pituitary imaging (with improved SNR and spatial resolution when compared to conventional scans, Fig. 109.1), free breathing.

Fig. 109.1 Dynamic imaging of a presumed microadenoma (arrow) of the pituitary, conventional 2D GRE vs. GRASP. Courtesy of Kai Tobias Block, NYU Langone Medical Center, New York, NY.
dynamic post-contrast imaging of the liver and kidney (Fig. 109.2; Fig. 109.3), and dynamic imaging of the prostate (here providing higher spatial and temporal resolution, Fig. 109.4, and thus improved detectability and characterization of small tumors). Given that the same data set can be retrospectively reconstructed with different temporal resolutions, quantitative tissue perfusion analysis can also be performed simultaneously.

Although GRASP works well with free-breathing in the abdomen, in some patients results are suboptimal, due to coughing or deep respiration. Extracting a respiration curve from the $k$-space center, it is possible to perform respiratory gating of the raw data prior to reconstruction. Alternatively, by sorting the data according to respiratory state, the latter can be treated as an extra dimension, allowing freezing of respiration in reconstructed images (XD-GRASP).

![Free breathing, dynamic liver GRASP pre-contrast and venous phase.](image)

**Fig. 109.2** Free breathing, dynamic liver GRASP pre-contrast and venous phase.

![Papillary renal cell carcinoma, T2 HASTE vs. GRASP: arterial phase and renal plasma flow map.](image)

**Fig. 109.3** Papillary renal cell carcinoma, T2 HASTE vs. GRASP: arterial phase and renal plasma flow map. Used with permission from Riffel P, Zoellner FG, Budjan J, et al. “One-stop shop”: free-breathing dynamic contrast-enhanced magnetic resonance imaging of the kidney using iterative reconstruction and continuous golden-angle radial sampling. *Invest Radiol.* 2016;51:714-19.

![Dynamic prostate GRASP, early enhancement suspicious for tumor (arrow).](image)

**Fig. 109.4** Dynamic prostate GRASP, early enhancement suspicious for tumor (arrow).
Filtering Images (to Reduce Artifacts)

A variety of filters are available for use on most MR scanners to improve the appearance of images and minimize artifacts. Although the specific characteristics of each filter vary from vendor to vendor, there are inherent similarities. Filters can be k-space-based ("raw data filtering") or image-based ("image data filtering"). Described here are several examples of filters designed to reduce artifacts associated with image acquisition and hardware limitations.

The first filter addresses artifacts associated with an incomplete digitization or truncation of the MR echo (see Chapter 102). This artifact, known as Gibbs’ ringing or simply truncation artifact, is visible as additional lines or ringing near the sharp changes in intensity at the edge of a volume of tissue or at an interface (e.g., air/head, cord/CSF). Truncation artifacts arise due to the discrete sampling of the signal (the abrupt starting and stopping). This type of artifact is more conspicuous in lower resolution images, specifically along an axis with a large pixel dimension, and for that reason also often along the phase encoding direction. Filtering in k-space for Gibbs’ ringing is done using a bell-shaped filter such as a Hanning or Gaussian filter. The result is a dampening of the signal toward the edges of k-space creating a smooth (nontruncated) transition.

The second example of a filter (for illustrations and further detail, see Chapter 111) deals with spatial distortion caused by imaging obtained at or near the edge of the usable field of view of the MR hardware. This distortion can be caused by a change in the gradient linearity or falloff of the main $B_0$ field and can lead to a warping (distortion) of the ends of the image or signal loss at the edge of the field of view. The image-based filter applied in this instance attempts to correct for these changes, thus leading to a more accurate spatial representation of tissues: through calculations that measure the extent of and correct for spatial distortion. Although helpful, distortion correction filters do have limitations when the extent of distortion becomes severe. For example, spatial resolution is not maintained, and is often poor, in areas of distortion. Users, therefore, should make every effort to position the body part being examined as close to the center of the main magnetic field as possible. Most MR manufacturers have distortion correction for gradient linearity, but not for the $B_0$ field, which is much more complex.

The third example of a filter is one that deals with variations in signal intensity across the field of view. This class of filters has assumed increased importance with the advent of multichannel phased array surface coils (see Chapter 6) and is routinely employed. Variations in signal may be especially noticeable in the abdomen, with multielement/multichannel coils, where tissues close to the coil elements in the anterior and posterior aspect of the body are bright but the center of the body far from the coil is dark due to insufficient coil element penetration. Signal normalization filters balance the pixel intensities between the bright pixels adjacent to the coil and darker pixels in the center of the image to create a more uniform intensity across the entire image. Fig. 110.1 depicts fat-suppressed fluid-attenuated inversion recovery (FLAIR) images acquired using a multichannel coil without (Fig. 110.1a) and with (Fig. 110.1b) application of a normalization filter. Note the artificially increased signal intensity along the periphery of
the brain in the unfiltered image, which makes windowing of the postoperative gliosis in the right occipital lobe difficult. The normalized image (Fig. 110.1b) exhibits more uniform signal intensity across the field of view. Another example is presented in Fig. 110.2, which illustrates axial post-contrast T1-weighted scans at the L5-S1 level in a patient following surgery for a left paracentral disk herniation, with enhancing scar tissue seen surrounding the exiting left S1 nerve root (arrow). Depiction of tissue close to a surface coil—for example, in thoracic and lumbar spine imaging—can be substantially improved with application of a normalization filter. This makes it possible to examine the entire field of view with a single window/level setting. In Fig. 110.2, subcutaneous fat has markedly high signal intensity in (a) the unfiltered image. Following (b) application of a normalization filter, the signal intensity of tissues is more uniform across the field of view, making it possible to examine the tissue close to the coil for any possible abnormality.
Geometric Distortion

Fig. 111.1 and Fig. 111.2a have substantial image distortion, the first at the bottom of the image and the second at the top. This results in the femur being artificially curved medially (arrow) in Fig. 111.1a and the vertebrae of the upper lumbar spine appearing progressively smaller (arrow) in Fig. 111.2a. In each case, the image distortion was due to nonlinear gradients. Virtually all MR systems suffer from some gradient nonlinearity, oftentimes due to coil design that seeks to optimize other aspects of gradient performance. It may be difficult on any one MR system to actually obtain an image depicting such spatial distortion, because postprocessing techniques may be employed unbeknownst to the user to correct the appearance of the image (Fig. 111.1b; Fig. 111.2b).

Geometric distortion occurs commonly in MR. All forms of spatial distortion have to do with mismapping of signal data due to errors in phase or frequency. Distortion can be due to a variety of factors, some intrinsic, others extrinsic, to the subject being studied. During production of a magnet, despite attempts to create a very uniform magnetic field, small inhomogeneities inevitably occur. Gradient nonlinearity may also exist, due to gradient coil design. With a nonlinear gradient, the magnetic field produced by the gradient coil does not increase in a linear fashion. Correction for image distortion due to inhomogeneities in $B_0$, the main magnetic field, is difficult. However, geometric distortion due to gradient nonlinearity is predictable and will not vary from patient to patient, with correction thus possible. For this latter case, postprocessing methods are employed by most manufacturers to correct for this source of distortion, remapping the incorrect spatial positions so that the final image more accurately reflects the actual spatial positions. Of importance, however, is that despite being able to correct for errors in spatial positioning, this technique cannot correct for the loss of spatial resolution that occurs due to the initial error in spatial encoding in regions of nonlinearity. For example,
in Fig. 111.2, spatial resolution is degraded in the upper part of the image. In the case of geometric distortion due to gradient nonlinearity, the severity of the problem is typically greatest at the ends of the usable magnet bore (along the z axis).

Unpredictable factors that cause distortion are inherently difficult to correct for. There are many such factors, including 1) internal distortions due to differences in magnetic susceptibility between tissues (see Chapter 113); 2) motion; 3) metal objects in and around the patient (see Chapter 115); 4) eddy currents, currents created by rapidly turning on and off the gradients; and 5) incorrect gradient adjustments and calibrations. The latter are sometimes not corrected for simply because the user is unaware of such problems.

Geometric distortion is a significant issue that all radiologists should be aware of as it potentially can alter image interpretation. For example, in a patient exam with spatial distortion, a specified axial slice may in fact be a curved plane (even if postprocessing methods make the final image appear without distortion). Thus, there can be a distinct mismatch between a specified slice and what the MR machine offers as being that slice. This could result in missing a small lesion that actually occurs in a slice other than the one designated. Realization of this common phenomenon is extremely important. Fig. 111.3 illustrates one of the several ways that a slice can be distorted, by visualization of the effect of geometric distortion on an anterior saturation band (see Chapter 118). Note that distortion of the anatomy of the lower lumbar spine is not readily evident, with the pars interarticularis defect of L5 (black arrow) well visualized in both the uncorrected (a) and corrected (b) images. Yet, the saturation band itself in the corrected image is not of uniform thickness, but flared at the end (white arrow).

Fig. 111.3 Warping of a saturation pulse due to correction of geometric distortion.
Chemical shift, a common source of artifacts, occurs in the readout direction. The artifact due to chemical shift is most evident in MR images at the interface of tissues with significantly different fat and water contents. In Fig. 112.1 and Fig. 112.2, the readout direction was top to bottom. The sampling bandwidth of a pulse sequence (given in hertz/pixel on each image), specified prior to scan acquisition, controls the magnitude of the effect. The lower the bandwidth, the more prominent the artifact. The artifact from chemical shift appears on images as artifactual bright (black arrows) or dark (white arrows) lines at fat–water interfaces (Fig. 112.1) or by displacement of the water image relative to the fat image (Fig. 112.2). In Fig. 112.2b, the brain appears displaced posteriorly relative to the fat of the scalp and diploic space, with the inner table of the

![Image](image-url)

Fig. 112.1 Chemical shift artifact in coronal abdominal images.
skull (curved arrow, Fig. 112.2a) no longer visible posteriorly on the low bandwidth image (Fig. 112.2b). Depending on MR vendor, bandwidth may be specified in hertz/pixel, or by the total frequency range that is uniquely encoded (the Nyquist bandwidth). For example, when an image is acquired with a 130 Hz/pixel bandwidth, a 256 image has a full image bandwidth of ±16 kHz (128 pixels × 130 Hz/pixel).

Chemical shift can be reduced to the degree that it is no longer apparent by using a high bandwidth. The downside is that high bandwidth images have a lower signal-to-noise ratio (SNR), with a high bandwidth image appearing “grainy,” as in both examples. Thus, the choice of bandwidth is a compromise between the chemical shift that can be tolerated and SNR. The concept itself is also not that difficult to understand. When an individual echo (or signal) in MR is observed, typically 256 to 512 samples are taken (defining the resolution in the readout direction), with the Nyquist bandwidth being the reciprocal of the time between consecutive data samples. In most applications at 1.5 T, the bandwidth is chosen to be between 130 and 195 Hz/pixel (with bandwidths at 3 T being higher). The artifact due to chemical shift is less conspicuous, and indeed may not be noticeable, if fat saturation is employed.

Fig. 112.2 Chemical shift artifact, comparison of high and low bandwidths.
The relation between the magnetization of a material and the applied external magnetic field is given by the physical property known as magnetic susceptibility. The difference in magnetic susceptibility between adjacent tissues or structures is termed a susceptibility gradient and may cause local magnetic field inhomogeneities, resulting in a nonlinear distribution of resonance frequencies. Because magnetic field gradients are used for spatial encoding, any nonlinear distribution causes image distortion and artificial signal variations. **Fig. 113.1a** shows the normal image appearance with minor variations in magnetic susceptibility within the tissue and a linear relationship between location and local magnetic field gradient. **Fig. 113.1b** shows the effect of a paper clip placed within the field of view. The magnetic dipole formed by the ferromagnetic material changes the magnetic field far beyond the paper clip itself. Said in a different way, the field is strengthened far beyond and around the ferromagnetic object. The image reconstruction assumes a linear relationship between magnetic field and location, leading to displaced anatomy according to the strengthened (higher) magnetic field. Because less tissue is available with the expected resonance frequency in the immediate vicinity of the paper clip, the signal for this region is significantly reduced (a signal void). Further away from the ferromagnetic object, this effect vanishes and finally the signal of the tissue within the area of a strengthened magnetic field constructively interferes with signal from tissue with the same resonance frequency, resulting in a hyperintense appearance at that location.

**Fig. 113.2** illustrates this effect in a postoperative patient with an anterior metal plate and screw fusion. A disk herniation (black arrow) is noted at C6–7, the level below the fusion. Signal voids with surrounding artifactual hyperintensity occur at the locations of the metal screws placed anteriorly in the C5 and C6 (white arrow) vertebral bodies.

**Fig. 113.1** Consequences of a substantial change in magnetic susceptibility.
The nonlinearity of the magnetic field gradient caused by any ferromagnetic material has consequences beyond its effect on frequency encoding. The field gradient responsible for slice selection is influenced as well, causing a significantly distorted slice profile. In gradient echo imaging, the substantial differences in resonance frequencies cause rapid phase dispersion leading to a much larger signal void. The sensitivity of gradient echo sequences for demonstrating susceptibility gradients (e.g., with hemorrhagic lesions) can be further enhanced by allowing more time for the dephasing process (using a longer echo time). This can be combined with an approach to increase the signal-to-noise ratio (SNR), because a low bandwidth sequence (which has higher SNR) requires a prolonged acquisition window, which results in a prolonged echo time.

Fig. 113.3 illustrates the effect of changing the bandwidth using pulse sequence diagrams. For a given slice selection gradient (GS), RF pulse, and phase encoding gradient (GP), lowering the bandwidth leads to an increase in the acquisition window and a longer TE. The frequency range across the field of view (FOV) in the readout direction is termed the sequence or image bandwidth. The span of frequencies across a single pixel is also commonly referred to as the bandwidth, which can be confusing, given in hertz/pixel. The noise in an image is distributed evenly across all frequencies. It thus follows that if the frequency range of the measurement is smaller (as with low bandwidth imaging), the noise is less (and thus the SNR higher).
Maximizing Magnetic Susceptibility

Fig. 114.1 presents scans from a patient with multiple cavernous angiomas using (a,c) fast spin echo (FSE) and (b,d) gradient echo (GRE) T2-weighted techniques at (a,b) 1.5 and (c,d) 3 T. The lesions, seen as small, focal, low signal intensity abnormalities, are much more evident on the GRE scans due to phase dispersion from local gradient (susceptibility) effects. The result is a clearly demarcated area of signal loss, increasing lesion sensitivity. Susceptibility-related effects are also more prominent at higher fields, leading to highest sensitivity on (d) the GRE scan at 3 T. Advanced techniques to image susceptibility–related effects are presented in Chapter 64.

The appearance of hemorrhage on MR depends on the age of the blood, with striking changes occurring from the hyperacute (< 24 hours) to chronic (> 14 days) stages. Visuali-zation of hemorrhage varies even further depending on the anatomic compartment within which it occurs. For brain hemorrhages, the blood therein evolves temporally from oxyhemoglobin to deoxyhemoglobin, intracellular methemoglobin, extracellular methemoglobin, and eventually hemosiderin. Hemorrhage on T1-weighted scans in the hyperacute and acute stages is isointense to hypointense due to the long T1 imposed by relatively high water content. As deoxyhemoglobin is transformed into methemoglobin, T1 is shortened (leading to high signal intensity) due to water’s ability to interact with the paramagnetic ferric iron within methemoglobin.

The explanation for the appearance of hemorrhage on T2-weighted scans is more complicated. In a region of hemorrhage, the presence of paramagnetic blood
breakdown products causes areas of varying magnetic susceptibilities and thus magnetic field nonuniformity. Magnetic susceptibility (χ) is a measure of the ability of an object or substance to become magnetized in a magnetic field. In mathematical terms, it is defined as the ratio of induced magnetic field (B₀) to applied magnetic field (H₀) (strictly, B₀ ∝ [1+ χ]H₀). Due to variability in magnetic susceptibility, there are static local field gradients that cause loss of phase coherence. Along with clot retraction, these phenomena effectively reduce T2 to a value known as T2* (T2 star), causing signal loss in regions of acute (deoxyhemoglobin), early subacute (intracellular methemoglobin), and chronic hemorrhage (hemosiderin and ferritin).

As with the visualization of metal artifacts (see Chapter 115), the extent of signal loss is dependent on scan technique, imaging parameters, and field strength. In imaging hemorrhage, it is at times useful to increase signal loss (T2* contrast) to improve lesion detection. FSE scans correct for phase dispersion due to local field gradients because of the use of a 180° (or, for the more general case, a refocusing) RF pulse that causes phase reversal. In distinction, gradient echo scans do not correct for such phase dispersion. This results in increased signal loss and therefore increased sensitivity to detecting hemorrhage (better T2* contrast). T2* contrast may also be exaggerated by the use of a long TE, larger voxel size, and higher field strength. This explains the improved depiction of iron in the globus pallidus, substantia nigra, red nucleus, and dentate nucleus at 3 T, with these nuclei often diffusely low signal intensity on diffusion and FLAIR scans at that field strength. With optimization of T2* contrast, 2D GRE scans may reveal hemorrhages (or iron) not seen on FSE scans. Fig. 114.2 presents T2-weighted FSE, FLAIR, and GRE scans at 1.5 T in a patient with scattered subarachnoid hemorrhage (deoxy- and intracellular methemoglobin). Although subtle abnormal high signal intensity is present in the subarachnoid space on the FLAIR scan, the hemorrhage is best seen on the GRE scan, with abnormal low signal intensity noted within several sulci and the sylvian fissure (arrows). However, signal loss from T2* effects may also obscure adjacent brain structures, may be mistaken for other pathology, and may even make characterization of the hemorrhage itself difficult. In clinical practice, therefore, FSE and 2D gradient echo scans complement one another in the visualization of hemorrhage.

![Fig. 114.2 Subarachnoid hemorrhage, FSE vs. FLAIR vs. GRE.](image-url)
The images in Fig. 115.1 show the artifact (white arrows) from a nonferromagnetic aneurysm clip at 1.5 T. Fast spin echo sequences (a) have the least distortion due to the presence of multiple RF refocusing pulses. A diffusion-weighted single shot EPI image (b; see also Chapter 42), of all illustrated techniques, should show the greatest effect. In this instance, however, parallel imaging has been employed with the result being substantially reduced artifact (but still greater than fast spin echo). Gradient echo scans (c) have very prominent artifact, due to the absence of a RF refocusing pulse. 3D time-of-flight MRA (see Chapter 77) utilizes a short TE and small voxels that decrease the extent of the artifact, viewed on (d) the source image, when compared with (c) the 2D gradient echo scan. The distal right internal carotid artery (black arrow, d) is visualized; however, the clip artifact obliterates the corresponding vessel on the left. This resulted in nonvisualization of the left middle cerebral artery on MIP reconstructions.

Although metal artifacts are generally focal and centered on the object causing the distortion, there can be effects anatomically distant from the piece of metal. This case also demonstrates the importance of viewing source images for MR angiography.

Ferromagnetic and nonferromagnetic materials can cause significant artifacts in MRI, as illustrated. The severity of the artifact correlates with the degree of magnetic susceptibility (see Chapter 113). Metal artifact may appear on MR in three ways: geometrical image distortion, especially at the boundaries of the object; gradual or distinct
signal void around the object; and areas of sharply defined high signal intensity (signal pile-up) adjacent to the object. Of note, geometric distortion will also propagate in the slice-encoding direction generating curved slices (like a potato chip).

Ferromagnetic materials, by definition, exhibit a large, positive magnetic susceptibility. The difference in magnetic susceptibility between the metal and the surrounding tissue causes a local magnetic field gradient, changing the magnetic field in the area of the object. Thus, the unique resonant frequency created by the gradients to encode spatial data is lost. This results in spatial misrepresentation and thus artifact. In addition, turning on and off gradient magnetic fields causes currents to flow in the metal object (eddy currents) that further change the local magnetic field. Nonferromagnetic metals, which exhibit a smaller, positive magnetic susceptibility, cause image distortion by the same two mechanisms, albeit to a lesser degree.

It is important to realize that metal artifact can be accentuated or diminished depending on the choice of imaging technique. Selecting a scan with small voxels, a short TE, a short interecho interval, and/or a high sampling bandwidth can markedly decrease the severity of the artifact. In addition, a common misconception is that metal artifacts are necessarily worse at 3 T as compared with 1.5 T. If identical scans are compared, this is true. However, identical scan techniques are rarely used between 1.5 and 3 T. **Fig. 115.2** presents sagittal T1-weighted (a,b) and axial T2-weighted (c,d) cropped scans of the brain at 1.5 T (a,c) and 3 T (b,d). The artifact in this case is due to a very small piece of metal in the subcutaneous tissues. On the T1-weighted scans, the artifact is substantially increased in size at 3 T (**black arrow**), but not intrinsically due to the field strength but rather to the fact that a gradient echo scan was used at 3 T as opposed to a spin echo scan at 1.5 T. On the T2-weighted scans the artifact is, conversely, much less evident at 3 T (**white arrow**), with the primary reason being the change in bandwidth from 81 to 355 Hz/pixel.

**Fig. 115.2** Metal artifact on 1.5 vs. 3 T, importance of imaging technique.
Minimizing Metal Artifacts

With metallic implants becoming common, techniques that reduce metallic susceptibility artifacts and allow imaging close to metal are desirable. Historically, high bandwidth has been the method of choice to reduce metallic artifacts in MRI. However, increasing bandwidth has drawbacks, including a decrease in SNR and a higher specific absorption rate (SAR). Recently, two new techniques have been developed for further metal-related artifact reduction. View Angle Tilting (VAT) and Slice Encoding for Metal Artifact Correction (SEMAC) provide a unique combination of metallic artifact reduction methods. Both VAT and SEMAC can be used separately or together (SEMAC-VAT) for 2D imaging of metal implants. Fig. 116.1 shows proton density–weighted images of a stainless steel screw imaged with a conventional FSE sequence (a), VAT (b), and SEMAC-VAT (c). The small inset shows a multiplanar reconstruction (MPR) in the slice direction to visualize through-plane distortion. Fig. 116.1 clearly demonstrates decreased in-plane geometric distortion, signal void and pile-up artifacts with VAT (b), with further artifact reduction including through plane distortion correction with SEMAC-VAT.

VAT (Fig. 116.1b) corrects for geometric in-plane distortion by applying a compensation gradient in the slice-encoding direction during readout. A major advantage is that this does not require an increase in scan time. Drawbacks however include image blurring and the lack of through-plane distortion correction. SEMAC uses additional z-phase encoding steps in the slice-encoding direction to correct for geometric through-plane distortion. Its drawbacks include an increase in SAR and scan time, the latter being dependent on the number of additional phase encoding steps used. Additional phase encoding steps also increase SEMAC’s ability to reduce metal artifacts in general. SEMAC is often combined with parallel imaging to reduce the otherwise substantially prolonged acquisition time. This is also an excellent area for the application of data.
sparsity (see Chapter 131), making SEMAC clinically viable and likely leading to its extensive use in the future. The combination of both VAT and SEMAC techniques, called SEMAC-VAT (Fig. 116.1c), provides both geometric in-plane and through-plane correction of metal artifacts.

Both strategies have proven to be effective in clinical application for metal artifact correction. Fig. 116.2 shows T2-weighted images of an anterior cervical spine fusion using a conventional 2D sequence (a) and VAT (b). The conventional image shows large signal voids with areas of signal pile-ups and geometric distortion (arrows, Fig. 116.2a). From this image alone, it is unclear whether the high signal within the cord is due to a syrinx or artifacts. The VAT image shows decreased signal void artifact, with almost no signal pile-up artifact and negligible geometric distortion. Although the VAT technique is not artifact free, it considerably reduces metal artifacts thus making pathology more evident.

Fig. 116.3 shows STIR images of a total knee replacement acquired using a conventional 2D technique (a) and SEMAC-VAT (b). SEMAC-VAT allows for markedly reduced metal artifact when compared with conventional 2D images. Note the irregular line of abnormal signal intensity (arrow, Fig. 116.3b), demonstrated only with the SEMAC-VAT technique, consistent with a healing peri-prosthetic fracture not seen on conventional imaging (a).

**Fig. 116.2** Cord syrinx, improved visualization with VAT. Courtesy of Yair Safriel.

**Fig. 116.3** Knee implant comparing conventional MR images to SEMAC-VAT.
**Fig. 117.1a** and **Fig. 117.2a** present spin echo contrast-enhanced T1- and non-contrast T2-weighted images, respectively, with prominent motion artifacts (arrows) in the phase encoding direction. With the use of gradient moment nulling (GMN), artifact is markedly reduced and signal intensity is returned to dynamic structures of interest such as blood vessels (**Fig. 117.1b**) and cerebrospinal fluid (CSF) (**Fig. 117.2b**). The enhancing lesion (black arrow) in **Fig. 117.1b**, a benign nerve tumor (trigeminal schwannoma), is substantially more apparent on the image with GMN, due to reduced ghosting from the carotid arteries and transverse sinuses. Note that the use of GMN markedly improves depiction of CSF in **Fig. 117.2b**, which is now seen with uniform high signal intensity both anterior and posterior to the cervical cord.

In scans of static objects, the rephasing gradient pulse successfully corrects for dephasing that occurs during the initial gradient pulse. With flow, however, the rephasing gradient pulse cannot correct for the additional phase changes, resulting in dephasing and therefore artifact in the phase encoding direction (see Chapter 121). The phase shifts that occur during the application of gradients relate to motion and gradient moments. Flow-related motion has three main dynamic components that contribute to artifact: constant velocity (first order), changing velocity or acceleration (second order), and changing acceleration, jerk or pulsatility (third order). During the application of a gradient, each order of motion creates a phase component over the period of time the gradient is applied. Each phase component, termed a gradient moment, is referred to with respect to its associated order of motion (first-order motion is associated with first-order gradient moment, etc.). The total phase of a spin is then the sum of all the gradient moments.

Gradient moment nulling (GMN)—also referred to as flow compensation, gradient moment rephasing (GMR), and motion artifact suppression technique (MAST)—represents a common way to reduce errors caused by various orders of flow-related motion and the gradient moments created by such motion. This technique decreases the number and prominence of ghosts and returns signal intensity to the primary structure from which ghosting occurred. GMN involves the use of additional positive and negative gradients to null the net gradient moments of moving and stationary spins at

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**Fig. 117.1** Decreased vessel pulsation artifacts with gradient moment nulling.
the time data are collected. If stationary and moving spins have the same phase, signal will be correctly mapped. GMN may be used to correct for different orders of motion. In clinical practice, correction for first-order motion is usually adequate for diagnostic purposes. This is due to the fact that the time between applications of gradient pulses during which phase shifts occur is so short that flow velocity is nearly constant within this period. For certain scan techniques (e.g., in cardiac imaging), correction of higher order motion may be advantageous. However, the complexity and number of gradients applied increases for higher order motion.

Although motion is random in the x, y, and z axes, GMN is effective when employed in the frequency-encoding and slice-selection directions. It is rarely utilized in the phase encoding direction because gradients in this direction are short and weak, and thus contribute little to overall phase shifts.

GMN does not increase the examination time. However, the lowest possible TE that can be used increases, especially for complex gradient pulsing. T1-weighted exams are especially sensitive to an increase in TE as they rely on the use of a very short TE. This limits the ability of GMN to decrease flow artifact in T1-weighted scans. GMN is especially useful for T2-weighted scans and more so in the cervical and thoracic spine where CSF flow (pulsation) is greatest. Note that GMN increases the signal intensity of flowing fluids, which must be kept in mind when viewing the resultant images.

![Fig. 117.2 Reduced CSF pulsation artifacts and restoration of CSF signal intensity, with gradient moment nulling, on T2-weighted images.](image)
Spatial saturation is an important technique for the reduction of motion artifacts, specifically ghosting (displaced false images of a body region). In this approach, an additional radiofrequency (RF) pulse is applied at the beginning of a pulse sequence to eliminate the signal from unwanted tissue. Magnetization in the area of the designated slab does not have sufficient time to recover (prior to the actual imaging part of the pulse sequence) and thus does not contribute to the observed signal.

A presaturation pulse can be applied either within (Fig. 118.1) or parallel to (Fig. 118.2) the imaging plane. The only difference between the two images in Fig. 118.1 is that a presaturation pulse (graphically shown by the label) has been applied during the acquisition of (b). This negates the signal from soft tissue anterior to the cervical spine, preventing ghosting from motion therein propagating across the image. In this instance, ghosts primarily arise from swallowing, breathing, and motion of the mouth. Note how much clearer the depiction is of the vertebral bodies and spinal cord with spatial saturation. For an in-plane presaturation pulse to be effective, the direction of phase encoding must be such that the ghosts propagate and overlie the tissue of

![Image](image_url)

**Fig. 118.1** Improved cervical spine imaging by use of a presaturation pulse.
interest. Common applications include the spine (in sagittal or axial imaging if the phase encoding direction is anterior to posterior), the abdomen (to eliminate ghosting from the anterior abdominal wall), and the chest (to eliminate ghosting from the heart).

A less well appreciated application of spatial saturation is that illustrated in Fig. 118.2, in which the presaturation pulse is applied parallel to the plane of imaging in (b). Note the pulsation artifacts (white arrows) from the transverse sinus that degrade visualization of the cerebellum in the image without the parallel presaturation pulse (a), but are absent with the presaturation pulse applied (b). In this comparison, the depiction of the carotid arteries is also slightly improved. Blood that flows into a slice can have high signal intensity, with this phenomenon being the basis for time-of-flight MR angiography. Whether the flow is arterial or venous, this can produce substantial pulsation artifacts that degrade the image. By application of spatial saturation above and below the block of slices to be acquired, pulsation artifacts from arteries and veins can be markedly reduced. Common applications include imaging of the sella, internal auditory canals, knee, and abdomen and pelvis (used for pre-contrast scans, and in some instances for post-contrast scans, depending upon the application).

The addition of a presaturation pulse, because it does require time to perform within the pulse sequence, may reduce the number of slices that can be acquired for a given TR. Another possible disadvantage is the additional heat deposition, if specific absorption rate (SAR) limits are a constraint.

![Fig. 118.2 Application of spatial saturation parallel to the imaging plane.](image-url)
Shaped Saturation

Anatomically shaped saturation pulses (signal suppression) are possible with the application of multidimensional spatially selective RF pulses. The latter can excite and saturate arbitrarily shaped areas, with parallel transmit technology beneficial, as these otherwise require long RF pulse durations and are sensitive to off resonance effects. Due to the required high flip angle for the RF pulse, their design has been challenging, with both RF hardware and SAR constraints. This has been largely solved using a k-space trajectory method, offering an inherent power-efficient RF pulse design.

In clinical MR, suppression (saturation) of unwanted signal is important and commonly applied. A major example is for sagittal spine imaging where application of an anterior saturation slab suppresses (when the phase encoding is AP) the signal and motion artifacts therein from anterior anatomical structures.

Fig. 119.1 was acquired on a 3 T scanner equipped with a two channel transmit array. Target patterns were chosen to illustrate, although only partially, the spectrum of saturation regions that can be achieved. A conventional saturation slab, applied in the AP dimension, is illustrated together with 2D rectangular, circular, and oval shaped saturation regions. Although not shown, these saturation regions extend anatomically in the craniocaudal dimension (in this instance) maintaining their shape. Large curved saturation regions, not illustrated, are also possible, with applicability for example for sagittal thoracic spine imaging (to account for kyphotic deformities). There is no specific constraint on the potential shape of the saturation region using the described approach. In addition to the improved anatomical accuracy (and thus improved saturation of unwanted signal), anatomically shaped saturation pulses offer potentially an improvement in SAR when compared to the application of multiple saturation pulses that would otherwise be required to achieve the same effect.

Fig. 119.1 Conventional saturation slab compared to spatially shaped saturation regions.
Advanced Slice/Sub-Volume Shimming

Multislice 2D imaging in regions of the body with considerable field inhomogeneities, for example the neck, can be improved by dynamically adjusting parameters during measurement. The main magnetic field, $B_0$, can be optimized for a slice or sub-volume by adjusting first-order shims (gradient offsets), and the excitation (RF) field $B_1$ optimized by frequency adjustments. Sequence wise, this has wide applicability, and can be implemented with echo planar, gradient echo, and fast spin echo techniques. For EPI, specifically, such an approach can improve SNR (eliminating areas of signal loss) and decrease spatial displacement of signal (geometric distortion). This is particularly important when the area to be scanned covers the neck and part of the shoulders. Here $B_0$ varies rapidly in the head-foot direction due to the sudden change in body shape. Also, since whole-body MR images are acquired with multiple table positions, and then composed (for a whole-body view), signal homogeneity across different body sections is very important, with slice/sub-volume adjustments here providing smoother signal transition. Another general area of application for dynamic shimming is improvement in the quality of fat suppression for techniques employing water excitation. Regional saturation pulses can also benefit, with improved consistency and efficiency of saturation.

In Fig. 120.1 axial FSE T2-weighted neck images are presented, without (a) and with (b,c) fat suppression. In (b) spectral adiabatic inversion recovery (SPAIR), a hybrid technique for fat suppression, specifically utilizing a spectrally selective inversion pulse to achieve chemical shift-selective fat suppression, was employed. For (c), which employed water excitation, $B_0$ and $B_1$ were optimized (using dynamic shimming) for the relevant sub-volume, with resultant marked improved uniformity (in [b] the subcutaneous fat is poorly suppressed, arrow) and effectiveness of fat suppression (note the greater suppression of fat within the mandible, asterisk) when compared to (b).

![Fig. 120.1 Improved fat suppression with dynamic, integrated, sub-volume shimming.](image)
Flow Artifacts

A T1-weighted axial gradient echo scan of the abdomen is illustrated in Fig. 121.1a. The addition of spatial saturation pulses, above and below the level of the slice, leads to loss of flow-related enhancement within the major blood vessels (Fig. 121.1b). Prominent ghosts are noted from the inferior vena cava and aorta in the phase encoding direction on an axial T1-weighted fast spin echo image (Fig. 121.1c). Placement of a saturation pulse above the slice eliminates the ghosts from the aorta, but not the vena cava (arrow, Fig. 121.1d), whereas a saturation pulse placed below the slice eliminates the ghosts from the vena cava but not the aorta (arrow, Fig. 121.1e). Using saturation pulses both above and below the slice largely eliminates these vascular ghosts (Fig. 121.1f).

Saturation pulses cause the signal of in-flowing blood to be markedly reduced. The ghosts from blood vessels are caused by phase dispersion, and the less the signal (due to the use of the saturation pulses), the less the ghosts.

Fig. 121.1 Flow artifacts on GRE and FSE images, effects of spatial saturation.
Flow may cause ghosting via multiple mechanisms, including phase shifts, time-of-flight (TOF), and saturation effects. In large vessels with rapid flow, macroscopic motion due to physiologic pulsation may also contribute to ghost artifacts. Ghosts may be bright or dark depending on their phase relative to the background in which they occur. In pulsatile flow, the distance between such ghosts is dependent on the difference between TR and RR (the time between heartbeats). In this instance, if TR and RR are completely synchronous, ghosting will not occur.

For a nonmoving object, the rephasing gradient pulse in a basic gradient or spin echo sequence successfully corrects for dephasing. However, with flow, further phase changes occur. When uncorrected, these lead to ghosting. As discussed in Chapter 117, however, additional rephasing gradients can be applied along the frequency encoding and slice-selection axes, thereby reducing the additional phase shifts due to flow and thus reducing ghosting.

Time-of-flight and saturation effects may also contribute to flow artifact. With slow flow, protons present during the initial radiofrequency pulse are partially saturated and thus have lower signal intensity. However, nonexcited protons entering an already excited slice are unsaturated (fully magnetized) and have high signal intensity. This phenomenon is called inflow enhancement and can be seen in both spin echo and gradient echo scans. With fast flow, in spin echo imaging, protons are exposed only to a 90° excitation pulse, but not the 180° refocusing pulse, resulting in low signal intensity (a flow void). In gradient echo scans, as in 2D TOF angiography, flow-related enhancement occurs due to unsaturated blood entering the imaging plane and can be seen with both fast and slow flow. The signal intensity of flow may thus be high or low depending on the scan technique and flow velocity. In reality, flow does not strictly adhere to any expected signal intensity. But, when flow leads to high signal intensity within the imaging plane, whether due to blood flow or even CSF motion, without proper flow compensation to correct for phase shifts, ghosting occurs.

Flow effects have real world consequences. Fig. 121.2 depicts an intrasellar meningioma on coronal post-contrast imaging at 3 T. Prominent pulsation artifacts (white arrow, Fig. 121.2a) limit the utility of a spin echo scan at 3 T. For this reason, a short TE gradient echo scan is used, with the inadvertent consequence that some arteries now have high signal intensity (black arrow, Fig. 121.2b).
Section VIII

Further Improving Diagnostic Quality, Technologic Innovation
The two main factors describing the strength of a gradient system are the maximum amplitude of the gradient and the rise time to achieve this maximum gradient. **Fig. 122.1** illustrates the consequences of stronger and faster gradient systems on the imaging capabilities of a gradient echo sequence. The frequency range for the excited slice is usually untouched, dictating the amplitude for the slice select gradient (GS).

**Fig. 122.1** GRE pulse diagram: the impact of faster and stronger gradients.
The same situation applies to the selected bandwidth per pixel, dictating the amplitude for the readout gradient (GR) during readout. The most important part, therefore, is how fast the gradients can achieve their normal value, so that the excitation can begin and the data can then be sampled. The faster the gradient (Fig. 122.1b), the shorter the time between tasks, the shorter the echo times (achievable), the shorter the slice loop, and the shorter the repetition time (achievable). With fast spin echo technique, a shorter echo time or echo spacing also contributes to improved SNR. A stronger gradient also enables some of the preparation pulses to become shorter, permitting a further reduction of TR and TE (Fig. 122.1c). In addition, by allowing the same bandwidth for a smaller field of view, stronger gradients enable high-resolution imaging if feasible in terms of SNR. Likewise stronger gradients allow thinner slices. A stronger gradient system can also be beneficial for some specific applications like diffusion-weighted imaging (DWI).

Fig. 122.2 is a comparison of TrueFISP or balanced gradient echo imaging in a normal volunteer using different gradient strengths. Fig. 122.2a has been acquired using “strong” gradients whereas Fig. 122.2b has been acquired using weak gradients. Applying the weak gradient results in an increase of minimum echo time achievable from 2.15 msec to 2.59 msec. Because the repetition time also depends on the echo time, the latter increased from 4.3 msec to 5.18 msec. Although these changes appear to be minor, it has to be kept in mind that balanced gradient echo imaging relies on a short TR to gain additional signal from tissues with long T2 relaxation times. In the presented case, the decrease in repetition time applying a stronger gradient (Fig. 122.2a) is only 0.88 msec, but the gain in SNR for tissues with long T2 relaxation time is visually evident, with cerebrospinal fluid of substantially higher signal intensity (Fig. 122.2a, arrow) as compared with the sequence acquired using a weaker gradient (Fig. 122.2b).

Fig. 122.2  Improved SNR with TrueFISP due to stronger gradients.
This chapter illustrates the effect of faster and stronger gradient systems on fast spin echo techniques (Fig. 123.1). The benefits applicable for sequences with a single echo, as explained in Chapter 122, also contribute to an improvement in multiecho techniques such as fast spin echo. The most important factor for multiecho techniques is “faster” gradients: that is, minimizing the time for the gradients to reach their nominal value. Once the gradients achieve their nominal value, excitation or refocusing pulses can be applied or data sampling can be initiated.

The strength of a gradient has less impact on the speed of the acquisition. Specifically, for the readout gradient preparation pulse, only the time-amplitude integral is of importance. Applying a pulse with increased amplitude, the duration of the application may be reduced while leading to the same preparatory dephasing of the transverse magnetization. The same holds true for the phase encoding gradient, with a shorter gradient duration possible with use of higher gradient amplitude.

As illustrated in Fig. 123.1, a faster and stronger gradient system permits shorter echo spacing in multiecho imaging. Compare the upper pulse diagram with the lower, which employs a faster and stronger gradient system. The improved gradient performance allows more echoes to be sampled within the same time. This may either be used to decrease scan time or to increase the number of slices acquired for the same given repetition time. In regard to the latter, the gain in number of slices is typically small. Image quality using fast spin echo technique will only be minimally affected by the speed and strength of the gradient system.

Fig. 123.2 demonstrates the slice positioning (a,b) for two T2-weighted fast spin echo sequence acquisitions (c,d) with different gradient settings. Applying weak gradients a total number of 21 slices could be acquired for the given TR (a,c). By changing the
gradient properties from weak to strong (b,d) a total number of 23 slices could be obtained with all other basic sequence parameters left unchanged. Increasing the gradient strength decreases the echo spacing resulting in a reduced echo train length. Subsequently, the slice loop time is also decreased. Therefore, more slices can be acquired per given repetition time without changing imaging quality (c,d).

Fig. 123.2 Increased slice coverage due to better gradient performance.
Gradient system performance is the horsepower of an MR System, and similar in a sense to the horsepower of a car (horsepower is of limited relevance in city traffic): a strong and fast gradient system is of advantage for specific applications, whereas no visible benefit is seen in others. There are basically two items characterizing the performance of a gradient system, that is the maximal amplitude in mT/m and the time needed to ramp to the desired magnetic field gradient in μs. The simplifying term of a slew rate has been introduced, documented in T/m/s, referring to the ratio between maximum possible gradient amplitude and the ramp time needed to get there.

For routine clinical imaging, the required amplitude in generating a magnetic field gradient in the direction of slice selection is given by the desired slice thickness (and the bandwidth of the applied RF pulse). High gradient amplitudes allow the selective excitation of thin slices. The amplitude of the magnetic field gradient during data acquisition is given by the desired imaging bandwidth and the selected spatial resolution. The bandwidth is usually selected based on application restrictions. A high bandwidth will also include a broad frequency range of patient noise; thus, the noise level is increased. A high bandwidth is also correlated with short data acquisition windows making the acquisition less sensitive to motion and causing echo trains to be of short durations (allowing more slices for a selected TR). For T1- and T2-weighted brain imaging, a bandwidth around 90–250 Hz/pixel is considered optimal. For EPI applications a bandwidth around 1200–2600 Hz/pixel is considered optimal. The selection of the

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**Fig. 124.1** One frame of the cardiac cycle measured with a TrueFisp sequence. In comparing a 33 mT/m system (125 T/m/s) and a 45 mT/m system (200 T/m/s), the improvement in image quality is minor (mainly based on the reduction of echo time from 1.4 to 1.2 ms), improving the temporal resolution about 20%, from 42.1 to 35.5 msec.

**Fig. 124.2** Axial abdominal imaging with VIBE. Compared to a 33 mT/m system, a 45 mT/m system provides a shorter TE (2.2 versus 2.4 ms), with a marginal improvement in image quality. As the TR changed from 4.8 to 4.5 ms, the time required for breath-holding decreased from 14.75 to 12.63 sec.
bandwidth is not based on the performance capabilities of the gradient system, but rather on an SNR optimization and the reduction of artifacts by selecting a short data acquisition window. What remains as the advantage of a strong gradient system is access to high resolution imaging, with the restriction that the signal will scale with the voxel size and you need to be able to afford the correlated signal drop in selecting high spatial resolution.

The phase encoding is based on providing a frequency difference for an arbitrary time period. Selecting high magnetic field gradient amplitudes will allow a shorter phase encoding period, but, in SE imaging symmetry has to be maintained between excitation, refocusing and data acquisition, usually dominated by the length of the data acquisition window (depending on the selected imaging bandwidth). Thus, reducing the duration of phase encoding is not helpful in SE imaging.

From a simplified perspective the benefit of a fast and strong gradient system comes into play in dynamic cardiac imaging (Fig. 124.1), where it allows higher temporal resolution and shorter breath-hold times. Similar effects can be seen in breath-hold abdominal imaging (Fig. 124.2). For the central nervous system a faster and stronger gradient system demonstrates its main advantage in diffusion weighted imaging (with additional application in body imaging, Fig. 124.3), where the shorter echo train length will result in an improved signal-to-noise ratio (Fig. 124.4; Fig. 124.5). Scaling with the demand, there are systems offered with e.g. 30 mT/m (100 T/m/s), 33 mT/m (125 T/m/s), 45 mT/m (200 T/m/s) and 80 mT/m/s.

Fig. 124.3 Depicted is one level of an axial abdominal body diffusion study. Compared to the 33 mT/m system, the 45 mT/m system provided an improvement in spatial resolution of 14% with a reduction in measurement time about 5%.

Fig. 124.4 Brain diffusion, axial image with a b-value of 1000 s/mm\(^2\). Compared to a 45 mT/m system, the 80 mT/m system provided an improvement in SNR of almost a factor of 2. When it comes to diffusion spectrum imaging, a fast and strong gradient system is a mandatory requirement. For example, the MGH-USC connectome scanner works with a gradient system capable of 300 mT/m, providing b-values of up to 30,000 s/mm\(^2\).

Fig. 124.5 Multidirectional diffusion spectrum imaging following tractography, acquired on an 80 mT/m system (514 directions, b max = 5000 s/mm\(^2\)).
Image composing represents a postprocessing application that allows for combination of coherent MRI datasets acquired at different times and displaying different parts of an anatomical system. Due to a limited field of view (FOV), it is not always feasible to display an entire anatomic system (e.g., the whole spine) with one sequence acquisition. However, technological advances have made it possible today to perform whole-body MR imaging within a reasonable scan time. This opens the door to new diagnostic fields, such as screening for metastases with whole-body diffusion and short tau inversion recovery (STIR) and whole-body contrast-enhanced magnetic resonance angiography (MRA). In addition, imaging of the entire spine is feasible, with high-resolution cervical, thoracic, and lumbar scans all acquired in a single setting.

These imaging approaches share in common the consecutive acquisition of multiple exams or segments to achieve coverage for the extended anatomical region to be evaluated. A requirement for efficient workflow is the presence of coils or a matrix of coils covering the entire region to be evaluated, in essence in many cases the entire body. In a second step, with modern software, the acquired data can be composed to a single dataset to enhance image evaluation/diagnostic interpretation.

However, there are certain restrictions regarding sequence parameters that one has to consider to make possible image composing. Different MR images cannot be composed arbitrarily, with the requirement being that some basic sequence parameters must conform to each other. This means that one must know before the sequence acquisition whether the image data are to be composed or not. Put simply, to do image composition implies some restrictions to the datasets to be composed. For example, differing slice thickness and pixel size may prevent image composition, depending on the software available. Limitations exist regarding the relative tilt of the two image sets to be composed. Furthermore, if correction is made for geometrical distortion, then corrected and noncorrected images cannot be mixed.

**Fig. 125.1** presents a contrast-enhanced MR angiogram of the abdominal, pelvic, and lower extremity vessels in a 76-year-old man with advanced atherosclerosis. After intravenous gadolinium chelate administration using a power injector, three coronal 3D FLASH scans have been acquired consecutively at different anatomical levels, starting with the lower abdomen and pelvis, then covering the upper legs and finally the lower legs and feet. With this approach, the MR scans follow the contrast bolus from the distal aorta through the vessels of the lower legs. **Fig. 125.1a,b,c** presents maximum intensity projection reconstructions of the corresponding three subtracted raw datasets in a coronal view. Note that at the upper and lower borders of the respective fields of view, anatomical structures are displayed twice: once in the upper image and once in the adjacent lower image (arrows). This is due to an overlap in the acquisition of the corresponding MRA sequences, performed to ensure complete coverage of the anatomy in question. However, in this setting, the radiologist interpreting the exam has to be well aware of the normal anatomy in order not to mistake a single lesion for two, due to it appearing twice, and to be able to integrate the scans.

**Fig. 125.1d** demonstrates the same consecutively acquired and subtracted datasets after image composing. Due to slight differences between the three respective datasets, one can still detect that the MRA images were obtained with three consecutive
Fig. 125.1 Contrast enhanced MRA (advanced atherosclerosis). Depicted are the original three stations acquired in a rapid temporal sequence, covering from the abdomen to the feet, together with the final resultant composed image.

acquisitions. However, the gaps (arrowheads) at the borders of the field of views are no longer seen and the double display (arrows) of adjacent anatomical structures at the interfaces between the three scans, due to the overlap during sequence acquisition, is no longer present.
Filters also exist with no other purpose but to improve the overall signal-to-noise ratio (SNR) of an image. This class of filter is normally $k$-space-based, with one approach being to remove data from the corners of $k$-space where a large portion of the noise contribution to an image originates. Fig. 126.1a illustrates the $k$-space data for a single slice acquisition, prior to application of a filter. Fig. 126.1b is following the application of an “elliptical” filter (so designated by one manufacturer), designed to reduce noise with minimal compromise in edge definition. The simple application of this or similar filters can increase SNR by 20%. Depending on vendor, the operator may or may not have control over the application of any filters, whether they be $k$-space- or image-based. Most user-selectable, $k$-space-based filters must be specified prior to image acquisition (as they are applied to the raw data).

Post-processing (image-based) filters that reduce noise and improve the overall aesthetic appearance of images are applied after image acquisition and reconstruction, leading—if properly applied—to an improvement in SNR and overall image quality. Fig. 126.2 illustrates the application of a noise-reducing filter that assesses the randomness of conspicuous structures within the image to eliminate noise, thereby increasing SNR and improving the overall appearance of the image. The software takes into account the tissue structure continuity and signal intensity distribution to increase the accuracy in noise reduction.

Axial post-contrast T1-weighted images of the brain are illustrated in Fig. 126.2: (a,b,c) are from the same image acquisition and differ only in post-processing. No filter was applied in (a), a medium filter in (b), and a heavy filter in (c). Image (d) is from a separate scan, with twice the number of scan acquisitions and thus twice the scan time. Simply by processing the original image (a) with a medium filter (b), an improvement in SNR and image quality is achieved similar to that obtained by doubling the scan acquisitions (d), but without the corresponding increase in acquisition time.
Note the reduction in graininess of the image in (b) as compared with (a). With a heavy filter (c), there is a further reduction in noise, but also an artificial smoothing of the image. The use of post-acquisition or post-processing filters, as with all filters, varies by MR manufacturer, in type, degree, and transparency to the user. Some vendors allow the user to choose whether a filter is applied and its degree, whereas other vendors apply a standard filter without the knowledge of the user. Employed judiciously, post-processing filters can improve overall image quality. It should be noted that increasing SNR by increasing the number of scan acquisitions has negatives other than simply scan time. The longer the scan, the greater is the chance for ghosting and blurring from inadvertent patient motion.

Fig. 126.2 Image based filter, results compared to doubling the scan time.
Although there are other applications of parallel imaging that will be discussed, parallel imaging is typically used to reduce MR acquisition time, albeit with a resulting loss in signal-to-noise ratio (SNR). Since the clinical introduction of MR in the early 1980s, software and hardware advances have been consistently made, permitting faster and higher quality (improved spatial resolution and SNR) exams. In order to avoid picking up the electromagnetic noise from the whole patient’s body and to be close to the origin of the signal, surface coils were introduced as a major early advance. Surface coils offered improved SNR, but limited anatomic coverage, which was subsequently overcome by the use of carefully engineered combinations of surface coils, called phased array coils. Severe coupling between surface coils makes multiple coils operate as one large coil and the benefits of a phased array is lost. To reduce the effects of coupling between coils, mutual inductance has to be minimized, for example, by using appropriate coil overlap, or alternatively by coil design with orthogonal modes (CP mode), and by isolation via separate preamplifiers.

**Fig. 127.1** Using multiple coils for improved coverage and SNR.

**Fig. 127.2** The basics of parallel imaging, illustrated schematically with two coils.
this illustration by a depiction of the data in $k$-space in the center between the two images acquired (one by each coil). In parallel imaging, these so-called phased-array coils are used to reduce scan time while obtaining the same field of view (FOV) and the same spatial resolution, or alternatively (a common clinical application of parallel imaging) to increase resolution while maintaining the same scan time.

In conventional MR imaging, the scan time and the resolution in the phase encoding direction are determined in part by the number of phase encoding steps (e.g., 512 steps or Fourier lines for a $512 \times 512$ matrix). With parallel imaging, the number of sampled $k$-space lines and thus scan time are reduced by the acceleration factor $R$, ($R \leq$ number of independent coils in the phase encoding direction). In the simplest situation, $R = 2$, every other line of $k$-space is sampled. Spatial resolution remains the same because the entire spectrum of $k$-space is still sampled (including in particular the periphery of $k$-space).

If phase encoding steps are skipped, or in other words $k$-space is undersampled, the resulting Fast Fourier Transformation (FFT) is an aliased or “folded” image with a rectangular FOV (an image usually never shown to the user) (Fig. 127.2a). There are two principal ways to create a full FOV. Either the missing $k$-space lines are calculated and filled in before the FFT and the reconstruction of the image is done as described below ($k$-space-based reconstruction), or the image is “unfolded” after the FFT (image-based reconstruction, described subsequently).

The underlying principle of all $k$-space-based reconstruction algorithms is to use the spatial information of multiple surface coils to calculate the lines in $k$-space that were skipped to realize a decrease in imaging time. At the moment, GRAPPA (GeneRalized Autocalibrating Partially Parallel Acquisition) and ARC (Autocalibrating Reconstruction for Cartesian sampling) are the only commercially available and clinically used algorithms to reconstruct the missing lines in $k$-space for parallel imaging. They use multiple acquired lines from all coils to reconstruct one line in the $k$-space measured by one single coil. This procedure is repeated for every coil and every missing line, so the results after FFT are uncombined single coil images, which are combined in a second step to produce the final image.
Image-based reconstructions operate in a similar fashion but take place after under-sampled \( k \)-space data has been converted to an image. As discussed previously, missing lines of \( k \)-space result in an aliased reduced FOV image (Fig. 127.2a). Due to the different coil sensitivities, the aliased and superimposed pixels are weighted by that particular coil's sensitivity pattern. In the reduced FOV, every pixel contains information from multiple (R) equidistantly distributed pixels in the full FOV. The SENSE-algorithm (SENSitivity Encoding) is employed to "unfold" the images by using the spatial information contained in the array of coils. The reconstruction is done in a pixel-by-pixel manner separating the signal contributions of all superimposed pixels according to the weights of the coil sensitivities. The result is a full FOV and unaliased image (Fig. 127.2b).

One important drawback of SENSE encoding is that this algorithm is not able to reconstruct a full FOV without aliasing when the FOV chosen is smaller than the object.

Coil sensitivity profiles are the key to reconstructing missing lines of \( k \)-space or unfold an image. There are two main calibration strategies to obtain the coil sensitivities: external and internal calibration. For external calibration, a low-resolution scan is acquired for each coil prior to the parallel acquisition. This generally prolongs the total scan time and must be re-acquired if there is coil or patient movement. Internal calibration is done during the scan by encoding a few extra lines near the center of \( k \)-space. Compared with external calibration, total scan time is typically less and image quality is improved, if the lines are integrated into the final image.

Some conditions must be fulfilled for successful parallel imaging:

1. Multiple coils, each with its own receiver pathway and a distinct coil sensitivity profile over the FOV, are required.
2. The exact sensitivity profile of each coil must be known to reconstruct the final image.
3. At least two coils with different sensitivities distributed in the phase encoding direction are needed.
It is important to note that in general, the SNR in parallel imaging is reduced by at least a factor of $\sqrt{R}$. With the advent of clinical imaging at 3 T, and the higher SNR so provided, parallel imaging became more important. From one point of view, 3 T makes parallel imaging practical because, with many scan techniques, there is sufficient SNR to allow its use. Parallel imaging techniques can be employed to either shorten measurement time or increase the spatial resolution in a given measurement time or a combination of the two. If SAR (specific absorption rate) is an issue (recall that SAR is proportional to the square of the field strength as well as proportional to the number of RF pulses applied), parallel imaging can be used to reduce SAR by decreasing the number of RF pulses (e.g., phase encoding steps). The loss in SNR in this situation is compensated by the increase due to the higher field strength.

Parallel imaging has indeed at 3 T become ubiquitous, with its use in a surprising number of common applications. **Fig. 127.3** illustrates a 3D FLAIR SPACE acquisition, in a patient with a large middle cerebral artery infarct, acquired using a parallel imaging acceleration factor (R) of 2. **Fig. 127.4** illustrates a 3D T2 SPACE acquisition, in a patient with a cervical disk herniation, acquired using a parallel imaging R of 3: making possible a $0.8 \times 0.8 \times 0.9$ mm voxel resolution. **Fig. 127.5** illustrates a 3D T2 SPACE acquisition, in a patient with a small intrathecal lumbar schwannoma, acquired using a parallel imaging R of 3: making possible a $1.2 \times 0.9 \times 1.0$ mm voxel resolution. **Fig. 127.6** illustrates a short axis CINE study of the heart, acquired using a parallel imaging R of 4, which permitted image acquisition in two breath-holds of 15 heartbeats each. **Fig. 127.7** illustrates a 3D heavily T2-weighted SPACE acquisition, providing excellent detail of the hepatic ducts (arrows), acquired using a parallel imaging factor of 3 in under 4 min, with excellent liver background suppression and adequate SNR. **Fig. 127.8** illustrates a contrast-enhanced MRA exam of the carotid arteries, with a severe stenosis at the origin of the left internal carotid artery well delineated due to the intrinsic high spatial resolution of the scan, made possible by the use of a parallel imaging R of 4. Today, the use of parallel imaging is central to clinical MRI and, as is discussed in the next chapter, continues to expand.

**Fig. 127.8** Carotid CE-MRA.
As discussed previously, parallel imaging exploits the properties of phased array coils. In the previous chapter, a simple case of two elements was presented. In current clinical scanners, many element phased array coils are commonly used. Phased array coils are often operated in different coil modes, with the primary goal being to maximize the SNR for the selected coil setup, or to enable distinct coil sensitivity profiles to be utilized in parallel imaging for the purpose of reducing measurement time, or to optimize regional coverage at some expense in SNR. In order to avoid picking up electromagnetic patient related noise, coil sensitivity tends to be limited to the adjacent region. As a consequence, the detected signal is greater close to the coil as compared with areas further away. To deal with this problem, a normalization filter (see Chapter 110) can be applied to the images to reduce the brightness of the areas in the near vicinity of the coil and increase the brightness in areas further away from the coil. Signal intensity appears more uniform across the image after application of such a normalization filter. However, the filter does not improve SNR, and more specifically in areas of low SNR further from the coil noise is quite evident and annoyingly scaled. The variation in signal intensity due to each small coil element, however, can be used to identify the position of the coils, a prerequisite for parallel acquisition techniques.

To take advantage of parallel imaging, coils need to be designed so that there are at least two elements in the direction of the intended imaging plane (and, specifically, in the phase encoding direction). Clinical MR systems are designed to make this possible for all applications, specifically allowing parallel acquisition in all three planes. It should also be kept in mind that although parallel imaging usually implies a reduction in SNR together with a decrease in scan time, this is not always the case. Parallel imaging when applied to single shot echo planar imaging (e.g., for diffusion-weighted scans) leads to reduced bulk susceptibility artifacts (and thus improved image quality), its major utility in this application. However, there is not a substantial SNR penalty because TE can be shortened, gaining back most of the lost SNR. However, the scan time is also not reduced, because all the phase encoding steps are acquired in a single shot (with scan time dependent on TR and not in this instance on the number of phase encoding steps).

**Fig. 128.1** is presented to illustrate how parallel imaging can be combined with a multichannel coil to great advantage, using a 32-channel head coil at 3 T. Coronal images of the brain in a normal volunteer are presented, all diffusion-weighted, with (a,b,c) being echo planar in type with (a) no parallel imaging and parallel imaging acceleration factors (R) of 2 (b) and 4 (c). The advanced coil design makes possible an R = 4 with only a small loss in SNR, providing a...
marked reduction in bulk susceptibility artifacts (a, arrow). A diffusion-weighted scan acquired with BLADE FSE is shown in (d), where the high SNR of the coil is used in combination with parallel imaging (R = 2) in this instance to reduce scan time.

Parallel imaging can also be done in two dimensions, a technique which has been employed clinically now for many years. In a 3D acquisition scheme, phase encoding lines can be under-sampled along both phase encoding directions. This can substantially reduce scan time. For example, if half the lines are sampled in both directions, this results in a reduction in scan time of 4 (2 × 2 = 4).

One issue with increasing R is that there is often some residual aliasing that remains in the image and degrades image quality. An acquisition and postprocessing technique known as “Controlled Aliasing In Parallel Imaging Results IN Higher Acceleration” (CAIPIRINHA) further reduces residual aliasing artifacts. This technique functions by shifting aliasing artifact across slices either in a 3D acquisition or in a simultaneous multislice acquisition. This allows for a more efficient unfolding of the image and potential gains in SNR.

Fig. 128.2 displays a 3D VIBE liver protocol with traditional parallel imaging techniques having acquisition times of 17 sec (a) mSENSE R = 4; (b) GRAPPA R = 4; (c) with acceleration in phase- and slice-encoding directions using CAIPIRINHA R = 2 × 2, acquisition time 11.4 sec; and (d) no parallel imaging, acquisition time 43 sec. The arrows point to prominent residual aliasing artifact with GRAPPA and mSENSE due to the high acceleration factor.
Parallel imaging is intrinsically associated with a SNR loss when compared to a fully encoded image. There are two parts to this loss, the first being the square root of the acceleration factor, simply due to the fact that less data is acquired. The second is the geometry (g)-factor, reflecting the noise amplification after the parallel imaging reconstruction process. The g-factor related SNR loss depends mainly on the encoding capability of the receiver array but in almost all cases only a small portion of the overall encoding power can be employed. One potential way to reduce the g-factor noise is optimization of the receiver array geometry (hardware) towards the application at hand. Alternatively, new acquisition techniques (software), such as CAIPIRINHA, can be used to improve the yield of encoding power given a certain coil geometry. CAIPIRINHA aims at modifying the appearance of aliasing artifacts during the acquisition to improve the subsequent parallel imaging reconstruction process. This approach has been employed in two applications where parallel encoding can be performed in two encoding directions, specifically with 3D (2D-CAIPIRINHA) and simultaneous multislice imaging (MS-CAIPIRINHA).

In 3D imaging, acceleration is applied in both phase encoding directions at the same time, with the total parallel imaging factor being the product of the two. The improvement in SNR with 2D-CAIPIRINHA comes by modification of the appearance of aliasing in a way that sensitivity variations provided by the receiver array are employed more efficiently. The same idea can be applied to simultaneous multislice imaging (SMS). Parallel imaging applied to SMS results in overlapping slices (aliasing) that needs to be untangled by the sensitivity variations of the receiver array. Without CAIPIRINHA, this can be extremely difficult (high g-factor loss) especially when the distance between the slices is small (no or little sensitivity variations along the slice direction). With CAIPIRINHA the individual slices can be shifted with respect to each other such that the

Fig. 129.1 Pre-contrast, 1st and 3rd arterial, and 20 min hepatobiliary phases, new small metastasis (arrow), 4 × 2 CAIPIRINHA acceleration, 4.8 sec breath-hold acquisitions.
sensitivity variations available along other directions can be employed to improve the g-factor. In SMS these shifts can be accomplished by alternating multiband RF pulses in combination with phase cycling along the phase encoding direction or alternatively by switching additional gradient blips along the slice direction (blipped CAIPI).

Applying CAIPIRINHA to 3D VIBE, acceleration factors (R) of 2 × 2 are routinely achieved clinically, providing fast and robust imaging comparable to 2-fold acceleration with GRAPPA. This has led to acquisition times on the order of 10 seconds for imaging of the upper abdomen, a marked advantage for breath-hold scans, in particular for elderly patients and indeed for the patient population in general. With advanced receiver coil designs, for example the 60-channel body array as used in Fig. 129.1 and Fig. 129.2, with R = 4 × 2, a scan time of 5 sec can be achieved. Multiple arterial sub-phases can thus be acquired, or alternatively the technique can simply be applied to improve hepatic arterial phase image quality, with a lower incidence of breath-hold difficulty, particularly during IV bolus contrast administration of liver specific contrast agents.

Another variation, for liver imaging, is to combine view sharing with CAIPIRINHA and VIBE, specifically the technique CAIPIRINHA-Dixon-TWIST-VIBE. Multiple high resolution 3D data sets can be acquired during a single breath-hold, with a temporal resolution of 2 sec, providing 5 to 6 arterial sub-phases with improved detection of hypervascular liver lesions.

Applying 2D CAIPIRINHA to 3D SPACE (Fig. 129.3) enables high-resolution imaging of joints such as the knee, with visualization of anatomic structures, SNR and CNR comparable to conventional 3D SPACE and 2D FSE but with scan time halved.

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Fig. 129.2 Failed 16 sec breath-hold vs. 4.8 sec CAIPIRINHA exam, arterial phase, large partially necrotic HCC following embolization.

Fig. 129.3 3D SPACE with 2D CAIPIRINHA, as compared to 2-fold acceleration and 2D FSE, normal knee. Used with permission from Fritz J, Fritz B, Thawait GG, Meyer H, Gilson WD, Raithel E. Three-dimensional CAIPIRINHA SPACE TSE for 5-minute high-resolution MRI of the knee. Invest Radiol. 2016; 51:609-17.
Excitation of a reduced FOV in the phase-encoding direction has many advantages, including foremost that the encoding time can be decreased while maintaining spatial resolution, or that spatial resolution can be increased. Zoomed imaging can in principle be performed on conventional scanners; however, it is substantially benefited by the implementation of parallel transmission. The latter allows with zoomed imaging local $B_1$ shimming (increasing the accuracy of the shim and $B_1$ homogeneity), higher flip angle homogeneity, and improved $B_0$ homogeneity (by improved higher order shimming) and additional reduction in encoding time and shorter echo times using Transmit SENSE. The result with DWI is overall improved image quality, which is subsequently described in greater detail.

Regardless of anatomical area, single-shot (ss) EPI suffers from low in-plane spatial resolution. Many applications require higher spatial resolution in order to make possible detection of small lesions. ss-EPI scans are also prone to susceptibility artifacts. Using a reduced FOV in the phase-encoding direction can mitigate both problems. By exciting only the part of the volume that will be used for imaging, the echo train length can be shortened, improving resolution and reducing geometric distortion. MR scanners are available today with fully dynamic, 2-channel transmit arrays, enabling spatially selective excitation (parallel transmission). In clinical comparisons, zoomed parallel transmit EPI has been shown to be superior to conventional ss-EPI in terms of overall image quality, differentiability of tissues and reduction of susceptibility artifacts (Fig. 130.1). Geometric distortion can be additionally reduced in EPI by the application of parallel transmission (Fig. 130.2), i.e. Transmit SENSE.
The images presented in Fig. 130.1 and Fig. 130.2 (with and without zooming) are matched parameter wise as closely as possible, with the differentiating feature being excitation of a conventional versus a reduced FOV. Note in the images of the prostate, obtained without a reduced FOV, the marked foreshortening of the gland in the AP dimension together with the bulk susceptibility (high signal intensity) artifact posteriorly at the prostate-air (colon) interface. In the axial images of the cervical cord, the foreshortening of the cord in the AP dimension (geometric distortion) with conventional ss-EPI, in comparison with the zoomed image, is evident. Note also the distortion of the shape of the thecal sac on the corresponding ADC map.

Zoomed parallel transmit EPI should be viewed as a tool for improved detection and localization of small lesions in specific anatomic areas. In addition to its utility for the prostate and cervical spine, zoomed EPI is beneficial in imaging of the middle ear, specifically for the detection and localization of cholesteatomas.

An important additional feature that can be implemented for zoomed EPI is partial Fourier excitation. Employing a partial Fourier factor down to 6/8, the duration of the RF excitation pulse can be reduced, leading to a reduction in TE and an increase in SNR, with excitation quality maintained. It should also be noted that zoomed EPI can be applied to BOLD, with similar improvements in overall image quality as with its application to 2D diffusion EPI.
Compressed sensing is an important, innovative imaging approach that with time will likely be implemented widely across MR, with the end result being for those specific applications a decrease in the time required for data acquisition. Compressed sensing has some similarities to parallel imaging in that both techniques obtain a reduced number of initial measurements and then utilize advanced mathematics to reconstruct a final image (Fig. 131.1). To date, limited imaging sequences are available for clinical use. Regardless, this technology represents an extremely promising development.

Outside of medical applications, compressed sensing techniques are routinely utilized to compress image data. Probably the best-known example is the JPEG compression of digital images. Typically, when a digital image is compressed into JPEG format, there is a 10-fold reduction in storage requirements with only minimal loss in image quality. The JPEG technique compresses redundant data within the original image matrix. Greater degrees of compression result in smaller image file sizes but a higher probability that the exclusion of data will adversely affect image quality. The ideal is that with the image optimally compressed, and image quality maintained.

Several major types of MR acquisitions are well-suited for the application of compressed sensing. The condition of sparsity (specifically that the signal is sparse in some domain) is the most difficult prerequisite for MR datasets to fulfill. Dynamic MR imaging acquisitions easily meet the sparsity criteria, with such methods (e.g. CE-MRA – Fig. 131.2, and cardiac imaging) being optimal candidates for initial implementation of compressed sensing. However, the application of this technique is not necessarily limited to dynamic images and can be extended to most MR acquisitions, as all to some extent...

**Fig. 131.1** Compressed sensing - data acquisition and reconstruction.
extent contain redundant data. Another early, easily identifiable application is metal artifact reduction, used in musculoskeletal imaging (MSK), specifically slice encoding for metal artifact correction (SEMAC).

Sparse reconstruction techniques, of which there are many, require that the data be sparse spatially and/or temporally and that a specialized data sampling pattern be used resulting in non-regular noise like aliasing artifacts – with the actual image distinguishable from these. The data reconstruction process itself, to recover the un-aliased, sparse image, is complex regardless of the specific technique used (Fig. 131.1), with long reconstruction times initially slowing development of this field. Applications demonstrated in clinical studies, in addition to those already noted, include dynamic contrast enhanced exams of any type, such as the pituitary and liver, 3D TOF MRA (Fig. 131.3), fMRI, DTI and high-resolution large FOV 3D imaging (such as that with SPACE). Applicability exists in all of the major body regions – brain, abdomen, MSK and heart.

Fig. 131.2 Time sequential MIPs, CE-MRA, compressed sensing. Courtesy of Mark Griswold.

Fig. 131.3 Conventional vs. compressed sensing TOF MRA (2x acceleration), vertebral artery aneurysm. Courtesy of Kaori Togashi, Kyoto University, Japan.
Cardiovascular MR imaging commonly requires repeated lengthy suspension of respiration for high spatial and temporal resolution. Furthermore, the acquisition of high resolution data sets requires sampling over multiple heart beats, with possible image quality deterioration in the presence of cardiac arrhythmia. Parallel imaging methods (see Chapter 127, Chapter 128, and Chapter 129) have led to an improvement in image quality, but techniques such as real-time/single-shot imaging or dynamic contrast-enhanced (CE) MR angiography to exploit rapid physiologic changes are still limited.

Recent implementations of compressed Sensing (CS) techniques exploit the ‘compressibility’ and ‘sparsity’ of information required for medical images and aim at significant acceleration of data sampling by exploring the redundancy of temporal and spatial information. Successful implementation of CS requires an optimized combination of incoherent sampling patterns and iterative image reconstruction approaches. Based on the dynamic imaging approach, cardiac cine imaging (see Chapter 90) and dynamic CE-MRA (see Chapter 85) are particularly well suited for compressed sensing.

Keyhole techniques in dynamic CE-MRA are employed to allow for a higher frequency of central k-space sampling to improve contrast dynamics with phase-sharing of peripheral k-space data to reduce peripheral data sampling as much as possible. As a result, the temporal footprint of the scan is much longer than the apparent temporal resolution of the data. While this reduction of the peripheral sampling density improves the apparent temporal resolution, it is detrimental to the temporal footprint (Fig. 132.1). The application of CS in this scenario allows for the calculation of individual time frames based on a single central k-space segment with the addition of an incoherently sampled

![Fig. 132.1](image.png)
single peripheral \( k \)-space segment that is changing from time frame to time frame. As a result, the temporal footprint matches the apparent temporal resolution of the scan and blurring of structures in motion sensitive areas is much reduced (Fig. 132.2).

While the overall approach to cine imaging is quite different to keyhole imaging in dynamic CE-MRA, the explicit ‘sparsity’ in variation of information from frame-to-frame allows CS algorithms to be successfully employed. Temporal regularization allows not only significant acceleration in 2D imaging but also 3D approaches with the latter requiring substantially longer data reconstruction intervals. As a benefit of this possible further acceleration beyond the use of parallel imaging alone, cine imaging can be performed without a loss in spatial and temporal resolution in only 2–3 heartbeats/slice or, if desired, in real-time cine imaging at high temporal resolution (Fig. 132.3). Early concepts have also explored the approach of CS techniques in dynamic volume scanning of the heart with simultaneous acquisition of multiple contrasts.

![Fig. 132.2 Thoracic dynamic CE-MRA data set reconstructed with standard and CS based algorithms (identical k-space data). CS based reconstruction provides substantially better details of the smaller pulmonary vascular branches and lower noise levels.](image)

![Fig. 132.3 Cine SSFP acquired with standard segmented techniques over 8–10 heart beats/slice (left), segmented CS techniques in 3 heart beats/slice (center) and real-time CS techniques (right) at identical slice locations.](image)
Tumor ablation can be done under MR guidance with radiofrequency (RF), focused ultrasound, the use of a laser, or by freezing (cryoablation). Radiofrequency ablation (RFA) is routinely performed for treatment of hepatocellular carcinoma and nonoperable liver metastases. Whether ultrasound, CT, or MR is used as the imaging modality for this procedure, visualization and differentiation of tumor from normal parenchyma, as well as identification of residual tumor (incomplete tumor ablation), is critical. Contrast enhancement, irrespective of modality, allows reliable identification of nonvital parenchyma, but is typically limited to a single use (during the interventional procedure). Monitoring of temperature is now possible by MR, with critical utility for monitoring treatment application. In terms of identification of the ablation zone on MR, contrast-enhanced scans performed immediately following treatment serve as the gold standard, with unenhanced T1-weighted scans useful for monitoring the ablation. T2-weighted scans, however, tend to underestimate the ablation zone, in the acute phase following intervention. Fig. 133.1 illustrates an 8-mm cholangiocellular carcinoma metastasis (a) prior to treatment, and postintervention (b) enhanced T1-, (c) unenhanced T1-, and (d) T2-weighted images. MR offers the advantages of higher sensitivity for liver lesions than either CT or ultrasound, with improved therapy monitoring due to multisequence capability.

Transcutaneous high-intensity focused ultrasound tumor ablation under MR guidance (MRI-guided focused ultrasound, or MRgFUS) is approved both in the European Union and in the United States for the treatment of uterine fibroids, prostate cancer and palliation of pain from bone metastases. Approval in Europe is broader, extending beyond these areas. MRgFUS is currently being evaluated as well for treatment of brain, breast, liver and pancreas tumors. This thermal ablation method can be employed, like RF and laser ablation, with temperature monitoring. Normal tissue along the beam path, proximal to the lesion, is preserved. As with RF ablation, definition of target volume, real time guidance, thermal monitoring of the ablation, and postprocedure treatment evaluation can all be accomplished by MR. Coagulative necrosis with accompanying denervation provides relief of symptoms from painful bone metastases.

Other alternatives include laser and cryoablation. A major advantage for laser ablation is the ability to perform real-time MR thermometry. One can monitor the heating by phase-difference imaging with a gradient echo acquisition, using the temperature sensitivity of the proton resonance frequency of water. Changes in phase are converted to estimated changes in tissue temperature on a pixel-by-pixel basis. There is limited clinical experience currently with this technique, but the technology is promising.

Excellent outcomes have been demonstrated in initial clinical trials of MRI-guided cryoablation for small renal tumors. Despite widespread use, CT is not an ideal imaging modality in this application, with these tumors often isodense to normal renal parenchyma. Real-time CT-fluoroscopy is limited to a single axial plane, and the procedure inherently involves a substantial dose of ionizing radiation. As with RF ablation and MRgFUS, MR offers superior soft tissue contrast, fast real-time imaging in any plane, and real-time monitoring of the ablation. Fig. 133.2 presents sagittal T2-weighted images (a) prior to (bSSFP) and (b) following (HASTE) cryoablation of a small renal cell
Fig. 133.1 Liver metastasis (arrow) prior to and immediately following MR-guided RFA. Used with permission from Rempp H, Unterberg J, Hoffmann R, et al. Therapy monitoring of magnetic resonance-guided radiofrequency ablation using T1- and T2-weighted sequences at 1.5 T: reliability of estimated ablation zones. *Invest Radiol*. 2013;48:429-36.

carcinoma in the upper pole of the kidney. The tumor is hyperintense prior to treatment, with a HASTE sequence subsequently depicting the ice ball, which is hypointense due to the lack of mobile protons, covering the lesion with adequate margins.

Tissue biopsy for the diagnosis of malignant disease is today routinely performed under MR guidance, with many indications, in particular for the breast and prostate. MRI-guided biopsy is also an excellent alternative to CT or ultrasound for sampling of small lesions in the liver that may not be easily seen with CT or ultrasound, lesions in the dome of the liver, isodense tumors in the kidney, and musculoskeletal lesions. **Fig. 133.3** presents images from a patient with a lesion suspicious for malignancy in segment IV of the liver. (a) T2-weighted MR, (b) enhanced CT, and images from the MR biopsy—specifically (c) axial and (d) oblique coronal bSSFP scans—are illustrated. The lesion (arrow, a) is difficult to identify on CT, thus biopsy was performed under MR guidance, with tissue pathology revealing a sclerosed hemangioma. **Fig. 133.4**
presents (a) fat-saturated bSSFP and (b) enhanced T1-weighted images from a patient with knee pain and suspected pigmented villonodular synovitis (PVNS). The biopsy needle is visualized in (b), with tissue pathology revealing acute and chronic inflammation, but without PVNS.

MR-guided breast biopsy (vacuum-assisted), preoperative wire localization, and clip placement are indicated in patients with focal lesions (specifically BI-RADS 4 or 5) on MR mammography that cannot be visualized by X-ray mammography and ultrasonography. These procedures are well established, with high clinical success and low complication rates. Contrast-enhanced breast MR is the most sensitive imaging study for the detection of breast cancer, with sensitivity > 90% and specificity comparable to mammography (70%). More than 40% of BI-RADS 4 or 5 lesions identified on MR cannot be visualized by mammography or ultrasound, with consideration of biopsy recommended. For histological clarification, MR-guided biopsy or lesion localization followed by surgical biopsy can be performed. MR-guided vacuum-assisted biopsy is the preferred interventional approach, avoiding unnecessary surgical procedures.

The current gold standard for confirmation of prostate cancer is transrectal ultrasound-guided biopsy (TRUSBx), which is characterized by low sensitivity and high specificity. The incidence of missed prostate cancer on first TRUSBx is ~25%, necessitating repeated biopsies. MR-guided biopsy of the prostate is today performed routinely at high field (3 T) in patients with elevated PSA, negative TRUSBx, and lesions suspicious for prostate cancer on MR imaging.

Two procedures, specifically nerve root injection and abscess drainage, previously performed under X-ray guidance, are now being conducted in some centers using MR. Percutaneous periradicular nerve root injection with corticosteroids and/or anesthetics, for therapy of lumbosacral radicular pain, has to date been predominantly performed under fluoroscopy or CT guidance. These X-ray-based modalities offer high temporal resolution and good bone–tissue contrast. Soft tissue visualization, even with CT, is however less than satisfactory, with the use of ionizing radiation particularly in younger patients and with the possibility of repeated procedures a concern. Interventional MR, which has been shown in this context to be safe, accurate, and time efficient, offers excellent soft tissue contrast, rapid multiplanar imaging, and the possibility of monitoring the injection (without the use of a contrast agent). Outcomes are comparable to studies using fluoroscopy or CT guidance, but without the use of ionizing radiation.

Image-guided percutaneous drainage, using either CT or ultrasound, is employed routinely for abdominal and pelvic abscesses in selected patients where surgical intervention is not indicated. Treatment success rates are > 90%. Limitations of ultrasound include the acoustic window, along with reduced image quality in obese patients. CT fluoroscopy guidance is limited to the transverse plane and can entail a substantial radiation dose, both to the patient and the interventionalist. MR offers excellent soft tissue contrast and depiction of fluid collections, with the capability to angulate without restriction the imaging plane for appropriate guidance. Drainage catheters can be visualized, using fast dynamic T1- and T2-weighted sequences, by instillation of either saline or diluted contrast media. Initial clinical experience has shown this approach to be feasible, effective, and safe.
MRI at 7 T offers improved diagnostic confidence when compared to 3 T in demyelinating disease, temporal lobe epilepsy, cerebrovascular disease and neoplastic disease, the four most frequently encountered neurologic disorders. Small structures and subtle pathologies are better depicted. Examples include cortical lesions in ischemic disease and multiple sclerosis (MS), structural epileptogenic lesions in the hippocampus, and depiction of a central vein and iron accumulation within a MS plaque. This is the result of increased SNR available at 7 T, which translates to higher spatial resolution, and improved contrast resolution for methods such as SWI.

Fig. 134.1 is from a 2016 study in which scan times and techniques were matched with 40 patients being evaluated at both 3 and 7 T. Protocols were adjusted to achieve higher spatial resolution at 7 T while maintaining similar, clinically acceptable, CNR and

Fig. 134.1 T2-weighted and SWI images at 3 T vs 7 T in MS, demonstrating the higher resolution achievable at 7 T. Used with permission from Springer E, Dymerska B, Cardoso PL, et al. Comparison of routine brain imaging at 3T and 7T. *Invest Radiol.* 2016; 51: 469-82.
SNR. T2 weighted images are shown on the left, SWI on the right, with the upper row 3 T and the lower 7 T.

The strength of 7 T in MS is better characterization of white matter lesions, specifically those with central veins and iron content, with higher detection rates for cortical MS plaques. Due both to findings on MP2RAGE and 3D double inversion recovery (DIR), MS lesions can be more correctly localized as cortical, subcortical, or mixed, with 7 T also having the potential to detect cortical microinfarcts. For mesial temporal sclerosis, 7 T offers higher diagnostic yield for diagnosis and improved assessment of histologic subtypes, due to higher spatial resolution.

MP2RAGE is being employed at 7 T for high-resolution 3D T1-weighted scans, in order to overcome the large spatial inhomogeneity in $B_1$ (transmit field) that otherwise severely impairs image quality. This approach yields uniform image contrast, with a strong reduction in proton density and T2* weighting, leading to excellent gray–white matter contrast. Images with $0.58 \times 0.58 \times 0.58$ mm and $0.75 \times 0.75 \times 0.90$ mm resolution at 7 T are compared in Fig. 134.2 with $1.0 \times 1.0 \times 1.2$ mm resolution at 3 T in a patient with early MS. Scan times were $2\times10:27$ min:sec vs. $9:33$ vs. $8:22$. Only at 7 T, and with the highest spatial resolution, was the plaque visualized (arrow) correctly classified as purely white matter in location. This scan approach provides a more accurate depiction of white and gray matter, due to improved spatial resolution, higher CNR, and lower partial volume effects.

Most clinical applications at 7 T in the brain are being driven by the need for higher spatial resolution and/or higher gray–white matter contrast, as discussed. Although not yet evaluated in depth in a comparative manner with 3 T, most non-cross-sectional MR specialty techniques will also benefit by the improved SNR. These include fMRI and proton as well as other nuclei (such as phosphorus) spectroscopy.

For the implementation of some pulse sequences, 7 T offers challenges. Image blurring, geometrical distortion and signal loss due to bulk susceptibility effects are inherent with echo planar imaging (EPI) and limit the use of single-shot EPI for diffusion weighted scans of the brain. Readout-segmented EPI offers benefits in terms of SNR, image quality and image distortion, albeit with a longer scan time, with great potential for clinical diagnosis at this field strength.

Fig. 134.2 T1-weighted sagittal images of the cerebellum in an MS patient, 7 T high vs standard resolution vs 3 T. Used with permission from Fartaria MJ, O’Brien K, Sorega A, et al. An ultra-high field study of cerebellar pathology in early relapsing-remitting multiple sclerosis using MP2RAGE. Invest Radiol. 2017;52:265-73.
When comparing 7 T to 3 T, the higher achievable spatial resolution at 7 T leads to higher diagnostic confidence—overall—for diagnosis and exclusion of pathologic findings in the knee. As in the brain, for knee imaging the higher SNR achieved at 7 T can be used to provide higher spatial resolution, when compared to 3 T, in the same scan time. Small joint structures and subtle lesions are more clearly depicted. Care must be taken however, in design of imaging protocols, due to continued challenges involving chemical shift, RF power deposition, and fat saturation. Adverse effects, seen with all 7 T exams, include dizziness, metallic taste, nausea and flashes of light in a small number of patients.

In a study published in 2017, employing a transmit-receive coil, all MR sequences conventionally used for imaging of the knee provided higher voxel-volume-adjusted SNR at 7 T when compared to 3 T. This included the evaluation of both fast spin echo and gradient echo sequences. In terms of the patient exams, the use of 7 T led to a higher number of detected lesions as well as higher diagnostic confidence scores.

Diagnosis or exclusion of bone marrow edema, subchondral cysts and cystic changes at the fibro-osseous junction were all improved at 7 T. Anatomic localization of cystic changes was also improved, together with improved differentiation of cystic changes from bone marrow edema.

In terms of the larger ligaments and lesions therein, little difference is seen diagnostically between 3 and 7 T. However, for the assessment of the anterior and posterior cruciate ligaments, the higher spatial resolution and potential for thinner slices should lead to improved delineation (Fig. 135.1). This was confirmed in the 2017 study by higher diagnostic constant scores.

Fig. 135.1 Partial tear of the anterior cruciate ligament (arrow), 3 vs 7 T. Used with permission from Springer E, Bohndorf K, Juras V, et al. Comparison of routine knee magnetic resonance imaging at 3T and 7T. *Invest Radiol.* 2017;52(1):42-54.
The higher spatial resolution of 7 T plays a direct role in improved diagnosis of meniscal tears, together with improved differentiation from mucoid degeneration (Fig. 135.2). CNR between cartilage and synovial fluid is also higher at 7 T, enhancing visualization of morphologic changes and lesion characterization. Evaluation of early and mild cartilage damage is improved, together with recognition of areas of involvement.

Sodium imaging has the potential to determine glycosaminoglycan (GAG) content in cartilage and tendons, where the sodium SNR correlates directly with GAG concentration. An important clinical application would be detection of early osteoarthritis and tendinopathy, when disease is potentially reversible (Fig. 135.3) prior to morphologic changes and biochemical involvement of joint structures in systemic diseases such as diabetes mellitus. 7 T, due to increased SNR, has the potential for evaluation of nuclei other than hydrogen, including specifically sodium and phosphorus.

Fig. 135.2 Small posterior horn medial meniscus tear diagnosed only at 7 T (arrow). Used with permission from Springer E, Bohndorf K, Juras V, et al. Comparison of routine knee magnetic resonance imaging at 3T and 7T. Invest Radiol. 2017;52(1):42-54.

Fig. 135.3 Sodium 7 T MR, patellar tendon, type 1 diabetes mellitus. Used with permission from Marik W, Nemec SF, Zbyn S, et al. Changes in cartilage and tendon composition of patients with type 1 diabetes mellitus: identification by quantitative sodium magnetic resonance imaging at 7T. Invest Radiol. 2016;51(4):266-72.
Continuous Moving Table

MR imaging with a continuous moving table allows for large field-of-view and whole-body imaging to be completed concurrently. Thus, multiple organ systems or anatomic areas can be evaluated efficiently. This is a key and necessary feature for simultaneous, integrated MR-PET, and also facilitates the performance of long segment angiography and metastatic workups. It is further uniquely suited for performance of localizer scans.

In employing conventional scan techniques, there are limitations due to the z-extent of the homogeneous region of a magnet, which inherently limits the size of the field of view. The only remedy for a field of view larger than the homogeneous region of the magnet is to stop imaging and move the patient table during the examination, which is cumbersome and prone to errors. Maximizing the z-extent of each station is typically performed, but can lead to large differences in image quality depending on distance from the isocenter of the magnet. Distortion and blurring are often seen, as well as inhomogeneous fat saturation. The image fusion step can lead to situations where anatomy of interest or pathology happens to be in a transition region between stations and thus portrayed. For imaging of dynamic processes, especially involving first pass of a gadolinium chelate, the individual stations depict temporal phases that are separated by comparatively long time periods.

The concept of Total Imaging Matrix (TIM) involves the ability of multiple coils to be electronically activated, selectively combined, and deactivated remotely. The need to manually change coils for evaluation of multiple organ systems is eliminated. This technology overrides any intrinsic limitation by sliding the patient through the magnet and always acquiring images at the center of the magnet (isocenter). This type of data acquisition takes advantage of the optimal magnetic field homogeneity and gradient linearity at the center of the magnet.

Various techniques may be performed including 2D axial sequential scans, 2D axial multislice scans, 3D axial scans, 2D and 3D radial axial scans, and 3D coronal scans. The sequential mode makes sense with fast sequences like TurboFLASH, HASTE, and single shot EPI where the complete phase encoding for a slice can be applied without pauses. Each axial slice is acquired completely before proceeding to the next slice, thus scanning every slice near isocenter and allowing optimal image quality.

This approach is one option for screening the body in oncological imaging as part of a metastatic disease workup. Fig. 136.1 presents an example of extended anatomic coverage using sequential 2D imaging with a coronal reformatted T1 fat-saturated and axial T1 in-phase, T1 fat-saturated, T1 opposed-phase, and STIR scans.

Respiratory motion still poses a challenge, with special solutions necessary. Some examinations work well in free-breathing, others require breath-holding. For multiple breath-holds, the continuous moving table has added the flexibility of selecting the breath-hold duration independent of the z-coverage, as well as the ability to flexibly prescribe the ranges of imaging where breath-holding is performed. An important future goal is to extend the applicability of free breathing acquisitions and to develop techniques that are insensitive to, or compensate for, respiratory motion, with GRASP (see Chapter 109) an important such innovation.
3D coronal scans are especially suitable for MR angiography where the area of interest has a large extent in the craniocaudal (z) and left-right directions, but not in the anteroposterior direction. Scanning with the continuous moving table is performed as portrayed in Fig. 136.2 using sequential increases in the coronal FOV of the subsets. These subsets are “sheared” in hybrid k-space, the echoes in z-direction are Fourier-transformed and combined allowing a single larger field of view image of the total scan range. This method has no scanning pauses; maximizing scan efficiency, the total scan range is a single entity and is “intrinsically composed” producing less boundary artifacts.

It is now possible to perform whole-body imaging in a few minutes, with some scans taking as little as a minute to complete. Streamlined tumor staging for detection of lymph node involvement and metastases can be done using either the continuous moving table or step-by-step acquisitions.
The combination of high soft tissue contrast provided by MRI and functional information obtained by PET can improve diagnostic confidence in a variety of indications, namely those where MRI outperforms CT in morphologic imaging. As a new technology, the full impact of MR-PET in the clinical and investigational world is still unclear. The integration of PET technology in an MR unit is technically demanding, in part because standard PET detectors, consisting of scintillation crystal blocks read out by photomultiplier tubes, are highly susceptible to magnetic fields. These had to be replaced by MR-compatible PET detectors such as Avalanche photo diodes (APD), which can tolerate high magnetic fields, up to 9.4 T.

PET data needs to be attenuation corrected (AC) to allow quantification of tracer activity. Two 511-keV photons are emitted when a PET tracer emits a positron that annihilates with an electron. These photons can be attenuated when travelling through tissue, but only photons that reach the detector contribute to PET signal. Attenuation is determined by tissue characteristics (electron density, tissue thickness) and photon energy and can be as high as 90% in the center of the body. An attenuation map, representing the different attenuation coefficients throughout the body, is required to correct the reconstructed PET data.

Due to its high accuracy and short acquisition time, PET attenuation correction using CT is regarded as the clinical reference standard. An attenuation map of density differences within the body is constructed from the CT scan, and then used to correct for the absorption of photons. Because the MR-PET hybrid system cannot measure linear attenuation in any individual patient, AC here needs to be performed differently.

The PET signal attenuation of rigid and stationary equipment such as the RF spine array and the RF head/neck coil can be compensated for by straightforward AC methods. This equipment can be scanned with CT, and a three-dimensional (3D) map of attenuation values can be computed (the so called μ-map). The flexible body matrix RF array is more problematic, because the geometry and position of the coils are dependent on the patient’s individual anatomy and thus attenuation cannot be precisely predicted, so the coil design has to be as “PET transparent” as possible.

The most complicated step is the generation of an accurate patient specific attenuation map, because no linear attenuation coefficient-based CT information is available. Tissue-specific AC has to be derived from MR information, which is based on proton density and relaxation properties, rather than on the attenuation of X-rays in tissue. Air, lung tissue, and solid bone all have very low signal in MRI, thus these fundamentally different tissue classes are difficult to separate. In the current implementation of integrated clinical MR-PET systems, tissue attenuation and scatter correction are performed using a two-point 3D Dixon-VIBE (see Chapter 50 and Chapter 65) technique, providing two sets of images where water and fat are “in-phase” and “out-of-phase.” This allows reconstruction of fat-only, water-only, and fat-water images and results in tissue segmentation of air, fat, muscle, and lungs. Cortical bone is ignored in this Dixon-based MR-AC approach, as bone is classified as soft tissue.

Fig. 137.1 presents coronal images obtained with continuous moving table technology (see Chapter 136), specifically the (a) nonattenuation corrected PET image,
(b) μ-map for MR-based attenuation correction, (c) attenuation corrected PET image, and (d) fused T1-weighted FSE and AC-PET data. An alternative approach to the segmentation based method just described is the use of an atlas of bone masks and aligned MR images. A third approach, which has been proposed but is still not feasible in clinical routine, requires a time of flight MR-PET scanner and uses PET emission data to determine attenuation correction.

Patient positioning in PET/CT is usually head-first, supine with the arms elevated. Data acquisition time is on the order of 3 minutes per bed position with seven bed positions covering the region from the vertex to below the pelvic floor. Typical voxel size in PET data is $\sim 4 \times 4 \times 2$ mm. Patient positioning in MR-PET is head-first, supine, with the arms in a resting position along the body. Simultaneous acquisition of MR-PET data is performed in expiration in the thoracic and abdominal region. For the Siemens Biograph mMR, the patient is covered with a rigid (16-channel) head/neck coil, a rigid (24-channel) spine array coil, and up to four flexible (6-channel) body matrix RF coils. The 3 T integrated whole-body MR-PET system has a field-of-view of 50 × 50 × 45 cm. Eight detector rings, consisting of 56 lutetium oxyorthosilicate-APD detector blocks per ring and 64 crystal elements per block (block area of 32 × 32 mm) constitute the PET detector unit, spanning a field of view of 25.8 cm in the z-direction. Typical voxel size in PET data from an MR-PET unit is similar to PET/CT: $\sim 4 \times 4 \times 2$ mm.

Fig. 137.1 Attenuation correction in MR-PET, liver cancer.
Scout scans for each bed position in three image orientations are performed followed by a multistation 3D Dixon VIBE sequence in the coronal orientation for attenuation correction. Dixon-VIBE and morphologic high-resolution T1- and T2-weighted sequences can be performed while PET data are acquired. Further MR sequences including contrast-enhanced T1-weighted, FLAIR (for brain imaging), and diffusion-weighted imaging can be added as indicated, as well as perfusion weighted scans, BOLD, and MR spectroscopy.

Fig. 137.2 Simultaneous PET and MR acquisition, with excellent co-registration.

In the normal workflow for a MR-PET acquisition, T1- and T2-weighted FSE images are acquired simultaneously with the PET data acquisition. Contrast-enhanced sequences and additional sequences can be added either for the whole body (e.g., DWI) or for specific locations (FLAIR for the CNS, multiphasic CE-T1-weighted GRE for abdominal imaging, etc.). Because the PET and MR data acquisitions are performed simultaneously, the acquisition time is mostly determined by the MR sequences. PET data acquisition time in MR-PET hybrid systems is similar to that of PET/CT (~3 min per bed position), although slightly longer if DWI is performed. Simultaneous data acquisition compensates for physiologic shifts in tracer accumulation and excretion. Additional motion correction can be performed to compensate for respiratory motion, thus improving quantification and image interpretation. Fig. 137.2 shows excellent alignment between the PET and MR data due to simultaneous acquisition. A left upper lobe lung mass with increased FDG uptake is noted.

Despite the limited data to date, $^{18}$F-FDG MR-PET has already proven to have higher sensitivity than $^{18}$F-FDG PET/CT in several patient populations. For lymphoma patients, this is the result of the increased sensitivity of DWI. In a second major application, MR-PET enables improved breast cancer staging in comparison to PET/CT, also due to the additional information from the MR component. For oncologic imaging in young children (Fig. 137.3; Fig. 137.4), the superior soft tissue contrast of MR provides higher confidence in lesion interpretation. Important in this population as well is the substantial savings in radiation exposure that can be achieved, due to the elimination of the CT component.
Fig. 137.3 Rhabdomyosarcoma of the orbit (arrow). MR/PET better delineates the lesion, as compared to PET/CT, due to the superior soft tissue contrast of MR. Used with permission from Gatidis S, Schmidt H, Gücke B, et al. Comprehensive oncologic imaging in infants and preschool children with substantially reduced radiation exposure using combined simultaneous 18f-fluorodeoxyglucose positron emission tomography/magnetic resonance imaging. Invest Radiol. 2016;51:7-14.

Fig. 137.4 Residual disease in post-transplant lymphoproliferative disorder. The T2-weighted MR reveals an involved interenteric lymph node, not seen by CT, with the PET/CT thus incorrectly interpreted as nonspecific enteric uptake. Used with permission from Gatidis S, Schmidt H, Gücke B, et al. Comprehensive oncologic imaging in infants and preschool children with substantially reduced radiation exposure using combined simultaneous 18f-fluorodeoxyglucose positron emission tomography/magnetic resonance imaging. Invest Radiol. 2016;51:7-14.
Three-dimensional (3D) imaging techniques in MR provide image datasets that can be further processed to create additional representations of anatomy or pathology.

◆ **Multiplanar Reconstruction (MPR)**

With MPR, images can be calculated for an infinite number of orientations within the acquired volume. The reconstruction is not necessarily restricted to planes, but can also be performed along curved surfaces. Fig. 138.1 illustrates (a) an oblique sagittal reconstruction through the anterior cruciate ligament of the knee, from a 3 T 3D acquisition (using the scan technique SPACE) with 0.5 × 0.5 × 0.6 mm voxel dimensions, and (b) an oblique coronal reconstruction from the reference line defined in (a).

![Fig. 138.1 Multiplanar reconstruction, anterior cruciate ligament.](image)

◆ **Maximum Intensity Projection (MIP)**

In an MIP image, by definition, each pixel is assigned the highest observed signal from along the specified trajectory through the 3D dataset. The user specifies the trajectory or perspective from which the 3D dataset is thus viewed in 2D. Typically, the dataset is viewed from multiple angles, meaning that a series of MIPs are created for scan interpretation. A targeted MIP is one in which only part of the volume, typically containing the vessels of interest (when processing a 3D MRA dataset), is used. The projection created from such a targeted volume contains less noise and fewer extraneous
structures, substantially improving image quality. **Fig. 138.2** illustrates a targeted sagittal MIP of the pulmonary circulation obtained at 3 T in a 20 sec breath-hold with $1 \times 1 \times 1$ mm voxel dimensions, depicting up to fifth order arterial branches.

#### Surface Rendering

With surface rendering, the surfaces of relevant anatomic structures are depicted. The user defines lower and upper gray scale limits, and the first value found along the trajectory from the user’s view through the 3D image dataset is declared the “surface.” A virtual light source is then used to modify the gray scale of the surface to create a 3D impression.

#### Volume Rendering

Volume rendering is a complex and extremely powerful processing technique that evolved from developments in special effects by the movie industry. A major difference from surface rendering is the assignment to each voxel of an opacity value, instead of simply selecting the single brightest voxel value along a viewing ray. An advantage of volume rendering is that the process is integrative, with multiple voxels contributing to the final gray level of each pixel, suppressing noise. Information about the original gray level differences is also preserved, along with the depiction of surface shapes. A lighting model is applied, as with surface rendering, to achieve the 3D effect, with the brightness of each voxel related to the amount of artificial light reflected from it. A color scale is assigned to the original signal intensity values from the source data, preserving this information. Areas of interest such as bone and blood vessels can thus be emphasized interactively by assigning appropriate color and transparency values. **Fig. 138.3** illustrates a volume-rendered MRA of the pulmonary circulation.

![Fig. 138.2 Targeted MIP, pulmonary circulation. Used with permission from Nael K, Fenchel M, Krishnam M, Finn JP, Laub G, Ruehm SG. 3.0 tesla high spatial resolution CE-MRA. *Invest Radiol*. 2007;42(6):392-98.](image)

![Fig. 138.3 Volume rendering, CE-MRA. Used with permission from Nael K, Fenchel M, Krishnam M, Finn JP, Laub G, Ruehm SG.](image)
An imaging protocol for a given section of the body includes multiple sequences with different contrast weighting (T1, T2, etc.) and orientation (axial, sagittal, coronal, or oblique), each with a given tilt and alignment with respect to specific landmarks. Other factors that may also be optimized include number of slices, FOV, and coverage area. Most practices have established specific imaging protocols for each body part, which are rarely modified depending on the specific clinical question. Standardization of these protocols with different MRI systems and technologists is desired to try and produce uniform imaging sequences for the clinical radiologist. An effective imaging practice strives to obtain high-quality standardized images within an efficient time period.

To help practices reach this goal, the different manufacturing companies have developed tools to help the technologist automate protocol selection and image alignment independently from the body part positioning, patient age, or disease. One such tool is AutoAlign, which works by obtaining a fast low-resolution nondiagnostic 3D VIBE opposed-phase scan of the area of interest. Opposed-phase scans are used as they better delineate boundaries. This acquisition can start as soon as the patient is placed in the magnetic bore, and can even be acquired while the technologist is walking back from the MRI scanner to the operator’s console. The software will automatically detect the correct axial, coronal, and sagittal axis for the body part; display corrected localizer images in all three planes; open the next sequence; set the imaging slices; and prompt the technologist to accept the parameters. The technologist can then either accept or modify them.

Fig. 139.1 shows in the first column original uncorrected images of the knee (opposed phase VIBE), the second column shows the corrected localizer images as would be displayed to the technologist (axial, sagittal, and coronal).

Automatic image alignment may be performed by the identification of landmarks:
for example, the patella, tibial plateau, or femoral condyles in the knee or femoral heads, greater and lesser trochanters, and iliac bone in the hip. The different imaging orientations are then obtained based on predefined field of views and angulations established in reference to these structures. However, in the head, a different approach is taken. A non-rigid registration between the individual patient head and a standardized head atlas is performed; clouds of landmarks that were specified in the head atlas are then transformed back into the patient anatomy. There, the transformed landmarks are used to determine a set of subregion-specific “rigid” AutoAlign references. Fig. 139.2 shows in the first column original uncorrected axial and sagittal images through the brain, the second column shows the corrected localizer information, and the third column shows the final axial T2-weighted and sagittal T1-weighted images obtained.

AutoAlign for the spine is also quite complex as it supports detection of individual vertebral body and disk space alignment. The final slice position results from the application of an additional fitting algorithm, which detects the spine’s geometry. Vertebral body labeling is also provided.

AutoAlign currently includes protocols for the brain, spine, knee, shoulder, hip, and heart with multiple specific applications in individual areas. For example, for the basic brain application, internal auditory canal, orbit, optic nerve, and temporal lobe protocols are also included. In the knee, besides the standard knee protocol, the software setup includes specialized imaging (orientation, angulation) for the meniscus, patellar cartilage, femoral cartilage, anterior cruciate ligament, and posterior cruciate ligament. Potential pitfalls for the use of such software include patient movement once scan parameters have been set and oversight by the technologist who is ultimately responsible for image quality and slice selection.

Fig. 139.2 Automatic image alignment for a screening brain exam.
Due to increasing cost pressure and quality requirements, health care providers are working to become more efficient and effective. As a result, standardization is steadily gaining more importance in the form of standard operating procedures (SOPs). This chapter focuses on the requirements of MR scanning in the context of a larger clinical process and discusses available digital solutions that can help achieve consistent high-quality studies.

The workflow process begins with the referring physician placing an order which is then protocoled by the radiologist. The patient is scheduled usually by an administrative person. Once the patient arrives to the appointment, all this information must be available to the technologist to perform the correct examination. Paper coding runs the risk of errors such as double coding, misreads, or misplacement of the orders. Such errors can lead to delays in scanning or, in the worst case, the wrong examination being performed, contributing to longer waiting times and a decrease in quality. Digital information exchange standards such as DICOM (Digital Imaging and Communication in Medicine) and HL7 (Health Level 7), which are globally accepted, can ensure a consistent digital communication flow between different systems in the clinical workflow eliminating these risks.

Patient information can also be transferred to the scanner console, and specifically the technologist, to reduce the preparation time in the exam room. Guidance at the scanner can range for example from patient positioning information to displaying of grid coordinates for a breast biopsy. Such visual guidance can play a significant role in ensuring quality standards while at the same time accelerating the patient preparation process.

Separation of tasks in technologist operation of the MR system is important for improve performance and task tracking. One example is separation of scanning and viewing processes at the operator’s console. Software can be designed to group patient registration, scanning, and protocol management for example on one screen with image construction, post-processing, and interface with PACs and other DICOM processes on a separate screen.

Integrated coil systems, which allow coils to be combined easily and examinations to be performed without repositioning the patient further, drive efficiency gains in patient preparation. Lightweight and easy-to-handle coils ensure that technologists can perform the setup quickly.

Institutions spend a lot of time defining and documenting their SOP. For scanning, the SOP often contains standard steps to perform an examination. However, patients vary and workflow is not always straightforward. Therefore, the SOP also needs to define decision points (e.g., does the scan require a contrast agent?) as well as alternative protocols for special cases (e.g., if patient cannot hold their breath or lie still). This documentation is often supplemented by image examples or text information to ensure that the examination is performed according to the institutional standards of care.

Integrating this information into the scanner-user interface is another example of how digitization of information can significantly improve processes. A simple example of SOP and workflow integration is the use of automated image alignment (see Chapter 139) to select all imaging planes and then proceed to a simplified decision tree allowing
the technologist to select from different scan options such as standard, resolution focus, speed focus, or limited patient capability.

Features such as automatic alignment of slices or automatic selection of slice plane and coverage increases consistency, not only in follow-up examinations, but also across patient populations. Automated bolus detection combined with automated voice commands increase bolus timing accuracy, which is essential for contrast-enhanced MRA and dynamic enhanced examinations of the liver. Automation not only improves the scanning process itself, but also improves image interpretation by supporting consistent image quality and output. After the exam, information about the performed procedure can be transmitted back for billing using the DICOM service “modality performed” procedure step.

Recent software and hardware advances continue to drive workflow optimization. In terms of acquisition software, simultaneous multi-slice and compressed sensing are important advances that, although still early in their development, can already be utilized in routine clinic operations to reduce scan times and improve patient throughput. The latter technology enables high quality free-breathing exams, both for the liver and for the heart, which will be further optimized and exploited with time. In terms of hardware advances, integrated shimming elements within complex high density multi-element coils can be used to provide improved $B_0$ homogeneity on an individualized patient basis. Slice specific shimming is now also available, for improved image quality in many exams, for example to reduce susceptibility effects in whole body diffusion weighted imaging.

At the scanner, during the exam, the trend is clearly moving toward automated processing instead of post-processing. The technologist can predefine the processing that should be performed and the system begins this work immediately after data acquisition without user interaction. Examples include automatic inline MIP image generation from MRA studies, multiplanar reconstruction (MPR) of 3D datasets and image composing after multi-station exams (see Chapter 125). Client server solutions allow fast access to more advanced data and processing applications from the scanner workplace, and provide users with the ability to begin processing and interpreting images directly in the scan room, or from any workstation that has installed the client software.

As health care continues to become more digital, the gains in efficiency and quality that the “digitization” of information presents will also continue to grow. Those organizations that take advantage of these technologies will be providing the optimal service to their referrers and the best diagnoses for their patients.
Listed in this appendix are only the acronyms that differ from vendor to vendor and are specifically discussed in this text.

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