MRI for Technologists

MR Image Postprocessing and Artifacts

PROGRAM INFORMATION

MRI for Technologists is a training program designed to meet the needs of radiologic technologists entering or working in the field of magnetic resonance imaging (MRI). These units are designed to augment classroom instruction and on-site training for radiologic technology students and professionals planning to take the review board examinations, as well as to provide a review for those looking to refresh their knowledge base in MR imaging.

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This material will be reviewed for continued accuracy and relevance. Please go to www.icpme.us for up-to-date information regarding current expiration dates.

OVERVIEW

The skill of the technologist is the single most important factor in obtaining good quality diagnostic images. A successful MRI examination is the culmination of many factors under the direct control of the technologist.

MRI for Technologists: MR Image Postprocessing and Artifacts describes the benefits of each type of postprocessing technique, which yield the best effect, and the major types of MR artifacts, their causes, and how to correct or eliminate them.

After completing this educational material, the reader will be able to:

- List postprocessing techniques and describe the benefits of each
- Assess which postprocessing technique is most effective for specific MR scans
- Identify standard MR system evaluation features and their appropriate uses
- Define the major types of MR artifacts and their causes
- Identify which MR artifacts can be corrected or eliminated and which can only be minimized
- Appraise the best method for correcting MR artifact for a specific MR examination

EDUCATIONAL CREDIT

This program has been approved by the American Society of Radiologic Technologists (ASRT) for 2.5 hours ARRT Category A continuing education credit.

HOW TO RECEIVE CREDIT

Estimated time to complete this activity is 2.5 hours. The posttest and evaluation are required to receive credit and must be completed online.

- In order to access the posttest and evaluation, enroll in the online course at www.icpme.us.
- Read the entire activity.
- Log in to your account at www.icpme.us to complete the posttest and evaluation, accessible through the course link in your account.
- A passing grade of at least 75% is required to be eligible to receive credit.
- You may take the test up to three times.
- Upon receipt of a passing grade, you will be able to print a certificate of credit from your online account.

ACKNOWLEDGMENTS

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FDA Drug Safety Communication: FDA warns that gadolinium-based contrast agents (GBCAs) are retained in the body; requires new class warnings

https://www.fda.gov/Drugs/DrugSafety/ucm589213.htm Accessed June 14, 2018.

05-16-2018 Update

In addition to approving the updated prescribing information concerning the gadolinium retention safety issues described in the Drug Safety Communication below, FDA has also approved new patient Medication Guides for all GBCAs.

Health care professionals and patients can access the patient Medication Guides according to the GBCA drug name* on the Medication Guides webpage, or the latest prescribing information by searching in Drugs@FDA.

All MRI centers should provide a Medication Guide the first time an outpatient receives a GBCA injection or when the information is substantially changed. In general, hospital inpatients are not required to receive a Medication Guide unless the patient or caregiver requests it. A health care professional who determines that it is not in a patient's best interest to receive a Medication Guide because of significant concerns about its effects may direct that it not be provided to that patient; however, the Medication Guide should be provided to any patient who requests the information.†

*The brand names of the GBCAs can be found in Table 1 below.
†For more information on distribution of Medication Guides, see the Guidance Document, the Drug Info Rounds Video, or the Code of Federal Regulations at 21 CFR 208.26.

This is an update to the FDA Drug Safety Communication: FDA identifies no harmful effects to date with brain retention of gadolinium-based contrast agents for MRIs; review to continue issued on May 22, 2017.

12-19-2017 Safety Announcement

The U.S. Food and Drug Administration (FDA) is requiring a new class warning and other safety measures for all gadolinium-based contrast agents (GBCAs) for magnetic resonance imaging (MRI) concerning gadolinium remaining in patients' bodies, including the brain, for months to years after receiving these drugs. Gadolinium retention has not been directly linked to adverse health effects in patients with normal kidney function, and we have concluded that the benefit of all approved GBCAs continues to outweigh any potential risks.

However, after additional review and consultation with the Medical Imaging Drugs Advisory Committee, we are requiring several actions to alert health care professionals and patients about gadolinium retention after an MRI using a GBCA, and actions that can help minimize problems. These include requiring a new patient Medication Guide*, providing educational information that every patient will be asked to read before receiving a GBCA. We are also requiring manufacturers of GBCAs to conduct human and animal studies to further assess the safety of these contrast agents.

GBCAs are used with medical imaging devices called MRI scanners to examine the body for problems such as cancer, infections, or bleeding. GBCAs contain gadolinium, a heavy metal. These contrast agents are injected into a vein to improve visualization of internal organs, blood vessels, and tissues during an MRI, which helps health care professionals diagnose medical conditions. After being administered, GBCAs are mostly eliminated from the body through the kidneys. However, trace amounts of gadolinium may stay in the body long-term. Many GBCAs have been on the market for more than a decade.

Health care professionals should consider the retention characteristics of each agent when choosing a GBCA for patients who may be at higher risk for gadolinium retention (see Table 1 listing GBCAs). These patients include those requiring multiple lifetime doses, pregnant women, children, and patients with

inflammatory conditions. Minimize repeated GBCA imaging studies when possible, particularly closely spaced MRI studies. However, do not avoid or defer necessary GBCA MRI scans.

Patients, parents, and caregivers should carefully read the new patient Medication Guide* that will be given to you before receiving a GBCA. The Medication Guide explains the risks associated with GBCAs. Also tell your health care professional about all your medical conditions, including:

- If you are pregnant or think you might be pregnant
- The date of your last MRI with gadolinium and if you have had repeat scans with gadolinium
- If you have kidney problems

There are two types of GBCAs based on their chemical structures: linear and macrocyclic (see Table 1 below). Linear GBCAs result in more retention and retention for a longer time than macrocyclic GBCAs. Gadolinium levels remaining in the body are higher after administration of Omniscan (gadodiamide) or OptiMARK (gadoversetamide) than after Eovist (gadoxetate disodium), Magnevist (gadopentetate dimeglumine), or MultiHance (gadobenate dimeglumine). Gadolinium levels in the body are lowest after administration of Dotarem (gadoterate meglumine), Gadavist (gadobutrol), and ProHance (gadoteridol); the gadolinium levels are also similar across these agents.

*The Medication Guide will be posted once it is approved.

Table 1. FDA-Approved GBCAs*

Brand name	Generic name	Chemical Structure
Dotarem†	gadoterate meglumine	Macrocyclic
Eovist	gadoxetate disodium	Linear
Gadavist†	gadobutrol	Macrocyclic
Magnevist	gadopentetate dimeglumine	Linear
MultiHance	gadobenate dimeglumine	Linear
Omniscan‡	gadodiamide	Linear
OptiMARK‡	gadoversetamide	Linear
ProHance†	gadoteridol	Macrocyclic

*Linear GBCAs result in more gadolinium retention in the body than macrocyclic GBCAs.
†Gadolinium levels remaining in the body are LOWEST and similar after use of these agents.
‡Gadolinium levels remaining in the body are HIGHEST after use of these agents.

To date, the only known adverse health effect related to gadolinium retention is a rare condition called nephrogenic systemic fibrosis (NSF) that occurs in a small subgroup of patients with pre-existing kidney failure. We have also received reports of adverse events involving multiple organ systems in patients with normal kidney function. A causal association between these adverse events and gadolinium retention could not be established.

We are continuing to assess the health effects of gadolinium retention in the body and will update the public when new information becomes available. We are requiring the following specific changes to the labeling of all GBCAs:

- A *Warning and Precaution*
- Changes related to gadolinium retention in the *Adverse Reactions, Pregnancy, Clinical Pharmacology*, and *Patient Instructions* sections

We urge patients and health care professionals to report side effects involving GBCAs or other medicines to the FDA MedWatch program.

MR Image Postprocessing and Artifacts

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IMAGE POSTPROCESSING

Explosive growth in the variety of clinical MR applications has stimulated an associated growth in postprocessing techniques. Postprocessing is essential for extracting vital diagnostic information from an MR data set. All commercial MR systems or their satellite console workstations include postprocessing software that is continually being improved, and it can be a challenge to keep abreast of the latest software packages and upgrades. The most common postprocessing applications include:

- Image subtraction
- Maximum intensity projection (MIP) reconstructions and targeted (subvolume) MIP reconstructions
- Minimum intensity projection (MinIP) reconstructions
- Multiplanar reconstructions (MPR)
- Exoscopic and endoscopic virtual surface rendering
- Brain perfusion mapping
- Brain diffusion
- BOLD
- MR spectroscopy
- Image filtering

This material discusses how these applications enhance and improve the diagnostic capabilities of MRI and improve data presentation.

IMAGE SUBTRACTION

Image subtraction is a postprocessing technique that eliminates unwanted background tissues or signals. The most common method of implementing image subtraction consists of taking the postcontrast images and digitally "subtracting" the precontrast images from them. Image subtraction is very useful in a variety of MR applications.

Image subtraction in gadolinium-enhanced MR imaging can be used with any field strength magnet. It is a quick technique that does not add examination time for the patient. Image subtraction is a way to obtain fat suppression that avoids the potential limitations arising from magnetic field inhomogeneity (imperfections in the main magnetic field) or low field strength.

The subtraction postprocessing technique increases the conspicuity of enhancing lesions, tissues, organs, and vessels, especially against a background that is hyperintense on unenhanced T1-weighted sequences, such as fat, proteinaceous fluid, and subacute blood products. From a technical standpoint, image subtraction requires some of the imaging parameters—such as FOV, slice thickness, and pixel dimensions—to be identical on both pre- and postcontrast images. Patient motion and spatial misregistration between pre- and postcontrast data sets must be kept to a minimum. An IV should be placed before moving the patient into the magnet to minimize the likelihood of patient motion between the pre- and postcontrast image acquisitions.

When using breath-hold acquisitions for chest and abdominal MRI and MRA, patients should hold their breath at the same phase in the respiratory cycle—either full inspiration or expiration—as long as both pre- and postimaging series are the same. This breath-holding technique yields the best reproducibility of organ location during repeated scans.

After MR scanning is completed, the image subtraction software is selected and precontrast and postcontrast data sets are selected. Scanners from some vendors provide image subtraction options as part of their pulse sequence setup menus. In this case, image subtraction is performed automatically after both data sets are acquired. The most widely used subtraction algorithms are performed slice-by-slice. Alternatively, complex subtraction can be performed before image reconstruction (complex subtraction includes effects due to phase differences between the pre- and postcontrast images). The computer creates a new series from the subtracted images that can be viewed on the console, printed, or sent to the picture archiving communication system (PACS). It is imperative that both pre- and postcontrast pulse sequences have identical parameters. The entire subtraction technique can be performed in a relatively short period of time, depending on the type of software available (Figure 1).

Image misregistration is a potential limitation of subtraction techniques. Misregistration occurs when the patient changes position or moves between the pre- and postcontrast image acquisitions or, in the case of breath-held acquisitions, the breathing is held differently from pre- to postcontrast image. Constant vigilance is required to detect evidence of image misregistration because it decreases the diagnostic utility of the subtracted images. Several vendors have developed software that attempts to correct misregistration in either rigid or deformable motion.

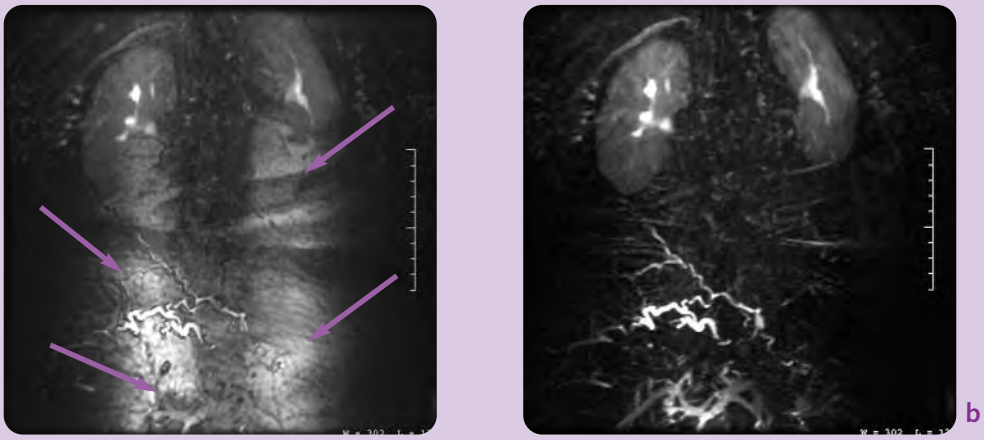


Figure 1a-b. **(a) An unsubtracted image in the coronal plane of the posterior lumbar region following contrast injection to visualize a suspected arteriovenous malformation (AVM). Note the bright signal from fat due to the close proximity of the surface coil, even with the use of fat saturation techniques (arrows). (b) The subtracted image using a precontrast image as a mask. Note the greatly increased visualization of the AVM in the subtracted image.**

APPLICATIONS OF IMAGE SUBTRACTION

Magnetic Resonance Angiography (MRA)

Image quality in gadolinium-enhanced 3D MRA benefits from subtraction postprocessing.

When obtaining an MRA, image subtraction can help to:

- eliminate background signal and improve depiction of small vessels
- eliminate signal folding (aliasing) of nonenhancing background structures onto the angiographic image

Blood vessels that are reconstructed using the MIP algorithm can potentially be obscured by bone marrow, soft tissue, or fat having high signal on T1-weighted sequences. These high signal background structures can limit the conspicuity and evaluation of small vessels in the distal extremities and/or narrowed or diseased vessels. If the FOV is too small, causing aliasing to "fold over" onto vessels of interest, subtracting the image sets can eliminate the aliasing and allow better visualization of the vessels of interest. Subtraction imaging allows optimal visualization of blood vessels and obviates the need for fat suppression techniques.

Magnetic Resonance Venography (MRV)

MRV can be performed by using one of several techniques, including 2D TOF, low-dose gadolinium-enhanced 3D MRV, and multiphase high-dose gadolinium-enhanced 3D MRA with image subtraction. In the latter technique, arterial phase MRA images, which contain only arteries, are subtracted from venous phase images, which contain both arteries and veins. Subtraction postprocessing to display only veins is performed as follows:

Data set with arteries and veins
minus (–)
Data set with arteries only
equals (=)
Subtracted data set with veins only
(the MR venogram)

Head and Spine Imaging

Subtraction techniques can aid in defining the patterns and degree of enhancement of central nervous system (CNS) tumors and complex tumors at the base of the skull. It also assists in depicting and defining residual tumor and distinguishing between tumor and hemorrhage or radiation changes, which can also appear as high signal. In the spine, subtraction imaging can be helpful in defining meningeal disease, extension of tumor into neural foramina and epidural and paraspinal areas, as well as distinguishing between nonenhancing herniated disc and enhancing epidural fibrosis in postoperative cases.

Body Imaging

Subtraction imaging is useful for evaluating solid and complex cystic masses. For example, renal or ovarian masses that contain hemorrhagic or proteinaceous fluid often exhibit increased signal on precontrast T1-weighted images. Such masses may be difficult to assess for internal enhancement which, if present, are frequently a harbinger of malignancy. Conversely, the absence of internal enhancement suggests a benign etiology. Subtraction postprocessing helps determine the presence or absence of enhancement (Figure 2).

Subtraction postprocessing is useful for assessing the extent of tumor necrosis and residual viable tumor in hepatocellular carcinoma or liver metastases after chemoembolization or radiofrequency ablation. Subtraction imaging can serve as an indicator for the success of therapy and can also help to localize residual

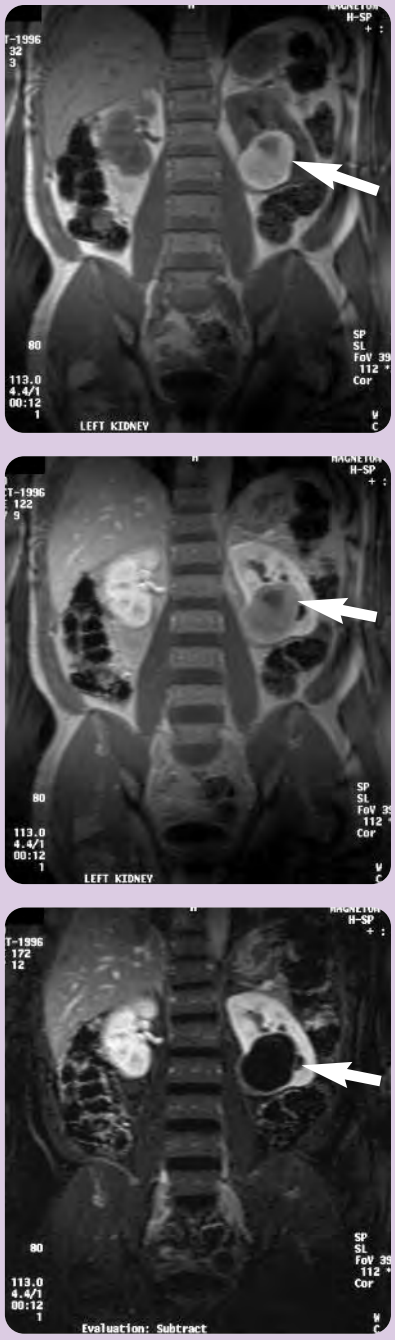
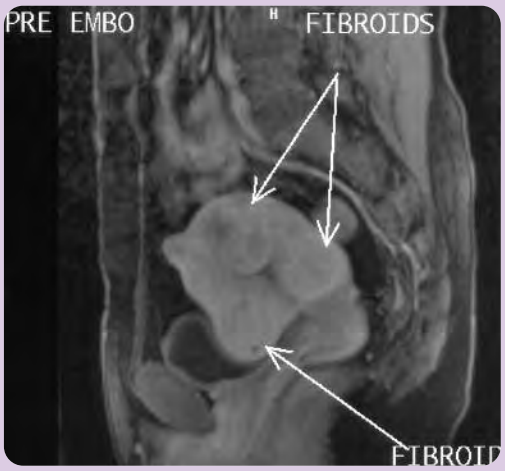


Figure 2a-c. **(a) Coronal T1 FLASH breath-hold noncontrast image through the kidneys. Complex mass in the left kidney. (b) Coronal T1 postgadolinium T1 FLASH breath-hold image. Complex mass in the left kidney. (c) Subtraction (postgadolinium minus pregadolinium) T1 FLASH image. Complex renal mass shows no internal enhancement.**

or recurrent viable tumor in cases of suspected therapeutic failure. Subtraction imaging can also be used to distinguish between viable and nonviable tumor and to help localize areas for biopsy.

Subtraction postprocessing can be quite helpful in the area of uterine artery embolization (UAE), a relatively common procedure that provides a nonsurgical treatment option for symptomatic uterine fibroids. Before performing UAE, MRI with subtraction imaging can accurately detect and define fibroids as vascular (enhancing) or nonvascular. Research has shown that vascular fibroids are more likely to respond to UAE therapy. After UAE, MRI with subtraction imaging helps to define those fibroids that are no longer viable (nonenhancing), as well as fibroids or components of fibroids that still enhance, indicating partial or complete viability. Thus, MRI is useful as an indicator for the success of therapy. Fibroids that continue to enhance after UAE can be localized with subtraction imaging for possible re-embolization or other therapies (Figure 3).

Subtraction postprocessing can be combined with opposed phase imaging to increase the conspicuity of fat content in lesions that contain both fat and water, such as adrenal adenomas and fatty liver. In these cases, intra-



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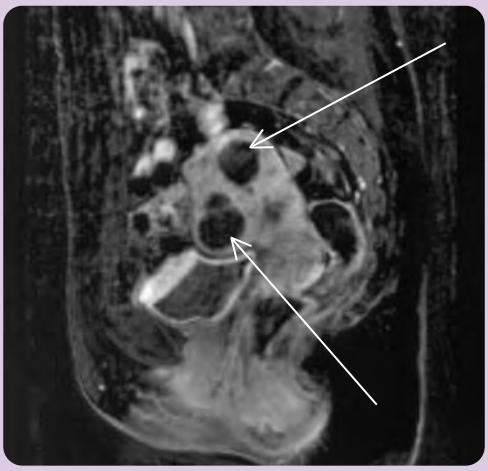

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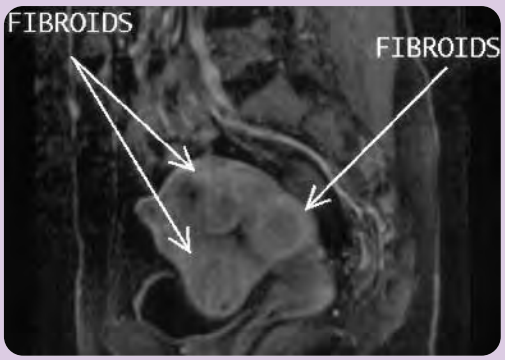


b

Figure 3a-c. **(a) A sagittal T1-weighted image of the pelvis revealing fibroid tumors. (b) Another sagittal T1-weighted image after precontrast images have been subtracted from postcontrast images, prior to embolization therapy. (c) After embolization therapy, the sagittal T1-weighted subtraction image shows areas of necrosis and nonviability of fibroid tumors.**

venous contrast is not required. To perform subtraction postprocessing, out-of-phase T1-weighted gradient echo images are subtracted from in-phase images acquired with identical imaging parameters (except for the echo time). Most scanner manufacturers now offer a dual-echo gradient echo sequence, which enables the in-phase and out-of-phase images to be acquired without the possibility of image misregistration.

Musculoskeletal System

In the evaluation of bone marrow and soft tissues in the musculoskeletal system, subtraction imaging can improve the definition of tumors and inflammatory lesions, fluid collections, and soft tissue or osseous involvement by suppressing high signal fat in bone marrow and in soft tissues. Subtraction imaging can also confirm the presence or absence of contrast enhancement in lesions.

MAXIMUM INTENSITY PIXEL PROJECTION (MIP) RECONSTRUCTIONS AND SUBVOLUME MIP RECONSTRUCTIONS

MIP and subvolume MIP reconstructions are commonly used to postprocess MR angiograms. With newer and more advanced software applications and enhanced magnetic gradients, far more MRA studies (renal, peripheral, aortic, and carotid) are being performed using 3D techniques with intravenous contrast media. Three-dimensional MRA data sets acquired before and after intravenous contrast contain information about blood vessels, as well as enhancing and nonenhancing stationary tissues. To review these data, the technologist or radiologist can scroll through the slices of raw data in sequence. However, vascular structures are better displayed and evaluated by taking advantage of the sharp contrast between the brightly enhanced vessels and the surrounding stationary tissues.

MIP is a postprocessing technique that mathematically filters the strong vascular signals from a 3D volume data set and projects these signals into a plane. The MIP technique is simple and fast.

MIP postprocessing gives MRA images the appearance of a conventional angiogram. MIP and subvolume MIP reconstructions help to illustrate and record vascular morphology. MIP images are produced using a ray-tracing algorithm. This algorithm, which is incorporated into the standard software of almost all MR systems, essentially creates a 2D image projection from 3D image data. To understand this algorithm, consider a straight line or "ray" coursing in a specific direction through the many voxels in a 3D volume. Each voxel has a different signal intensity value. Rays are projected through the 3D imaging volume in the user-defined plane. Of all the voxel signals encountered along each ray, only the highest signal value (maximum pixel intensity) is retained and assigned to the relevant pixel in the 2D projection image (Figures 4-8). Because flowing or contrast-enhanced blood generates high signal, the 2D projection of the brightest voxels effectively produces an image of the vascular system. Vessels are visualized while surrounding tissue is suppressed into the background.

When performing and using MIP reconstructions, it is helpful to perform MIP repeatedly and to slightly change the direction of the projection each time. This technique yields projections that demonstrate the vessels at slightly different angles. The projections can be displayed in cine mode, and the vessels will appear to rotate in space. If one vessel overlaps or covers another vessel in one pro-

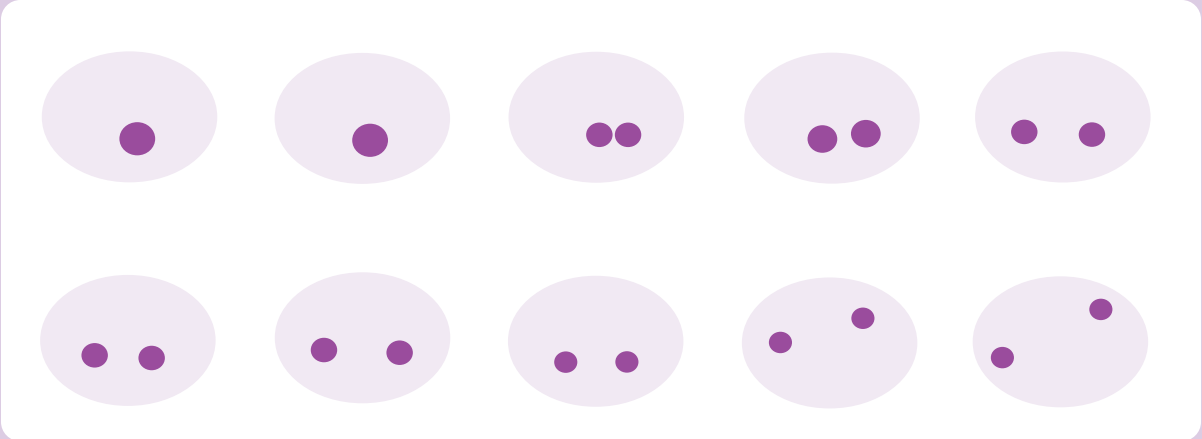


Figure 4. Individual axial images showing bright vascular flow.

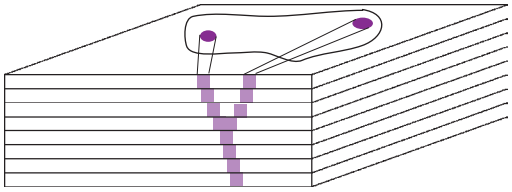


Figure 5. When the axial images are "stacked," the bright vessels are displayed in any direction.

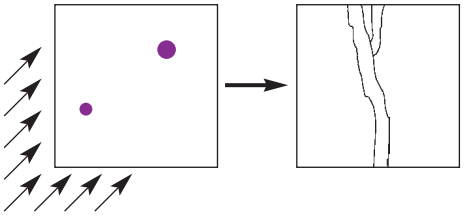


Figure 6. Here the ray is projected from a 45-degree angle. Note the orientation of the resulting MIP.

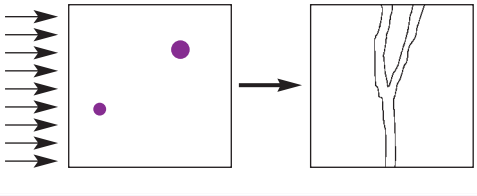


Figure 7. The MIP process "looks through" the data to project the brightest pixels along the direction of the ray. The user selects the best direction(s); in this figure, the ray is coming directly from the left side.

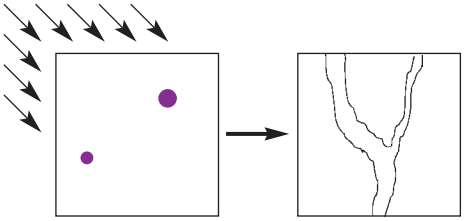


Figure 8. In this figure the ray is projected from the opposite 45-degree angle. Note the orientation of the resulting MIP now.

jection, the projection angle can be changed until the vessel of interest is easily visualized.

The MIP algorithm does have some limitations. These limitations include the loss of information on a MIP reconstruction due to selection of the brightest voxel, such as failure to visualize a small thrombus in a vein, obscured vessels if bright objects are included in the MIP volume, the inability to determine the proximity of a vessel to the viewer on a single projection, and corruption of MIP reconstructions by venous overlap. Each of these limitations can be reduced or even eliminated through careful and complete postprocessing by performing the targeted MIP.

TARGETED (SUBVOLUME) MIP

A targeted (subvolume) MIP will improve image quality and significantly reduce reconstruction time. When using this technique, only a limited portion of the acquired data is used; that portion of data is selected to include vessels of interest. The advantage of targeted MIP is that the projection contains fewer noisy background pixels, such as high signal fat, and can remove overlapping vessels. The only signals considered in the targeted MIP algorithm are those the user chooses to include within the ROI. This technique can be used to isolate specific vessels of interest from others in the projection path, keeping the image uncomplicated. Targeted MIP reconstructions are especially helpful for depicting tortuous, overlapping vessels or vascular pathology at vessel origins and branch points.

"Trimming" the area in a targeted MIP can yield sharper delineation of vessels by eliminating bright volumes that could superimpose areas of interest. The areas of interest are typically drawn by hand so they can be tailored to the patient's unique anatomy for optimal visualization. This is illustrated in Figure 9. Note how the brighter tissue, such as fat, interferes with visualizing the vessels behind it. Removing this unwanted bright signal clearly improves the quality of the vascular structure visualization.

Whether the technologist utilizes a MIP or subvolume MIP, the source images must always be analyzed and presented.

MINIMUM INTENSITY PIXEL PROJECTION (MINIP) RECONSTRUCTIONS

MinIP reconstructions are analogous to MIP reconstructions, except that the algorithm retains the voxel of lowest signal intensity along each ray and assigns it to the relevant pixel in the 2D projection image. MinIP reconstructions are seldom used in daily practice. However, their value becomes evident in special situations that require 2D projectional displays of low signal intensity structures, such as the air-filled lumen of the tracheobronchial tree or metallic vascular or biliary stents. These low signal intensity structures would be hidden on routine MIP reconstructions by high signal intensity voxels in the 3D volume. Other than these specialized studies, the MinIP technique is rarely employed in MR imaging.

MULTIPLANAR RECONSTRUCTION (MPR)

MPR is another postprocessing technique that is part of the standard software on most MR systems. The MPR function interrogates a thin slice from a 3D or multislice 2D image data set and reconstructs a new image. The data taken from the original images generate the

MPR series. Conceptually, the MPR reconstruction is similar to a subvolume MIP reconstruction with the slab thickness reduced to one voxel. MPR reconstructions are especially useful and are most often employed for evaluating source data from MRA and magnetic resonance cholangiopancreatography (MRCP) studies. This postprocessing technique provides supplemental 2D views of single voxel thickness of vascular or biliary structures in a projection different from those of the original MR acquisition. The technologist or radiologist can load source data from a single acquisition on the MR console or a satellite workstation. The user can then scroll through the data in one projection and simultaneously view it in other standard or oblique projections. MPR reconstructions can be performed in single or double oblique projections. Curved reformations can also be performed. The technique is quick and easy to execute and can be performed interactively. Blood vessels or the biliary tree can be interrogated in

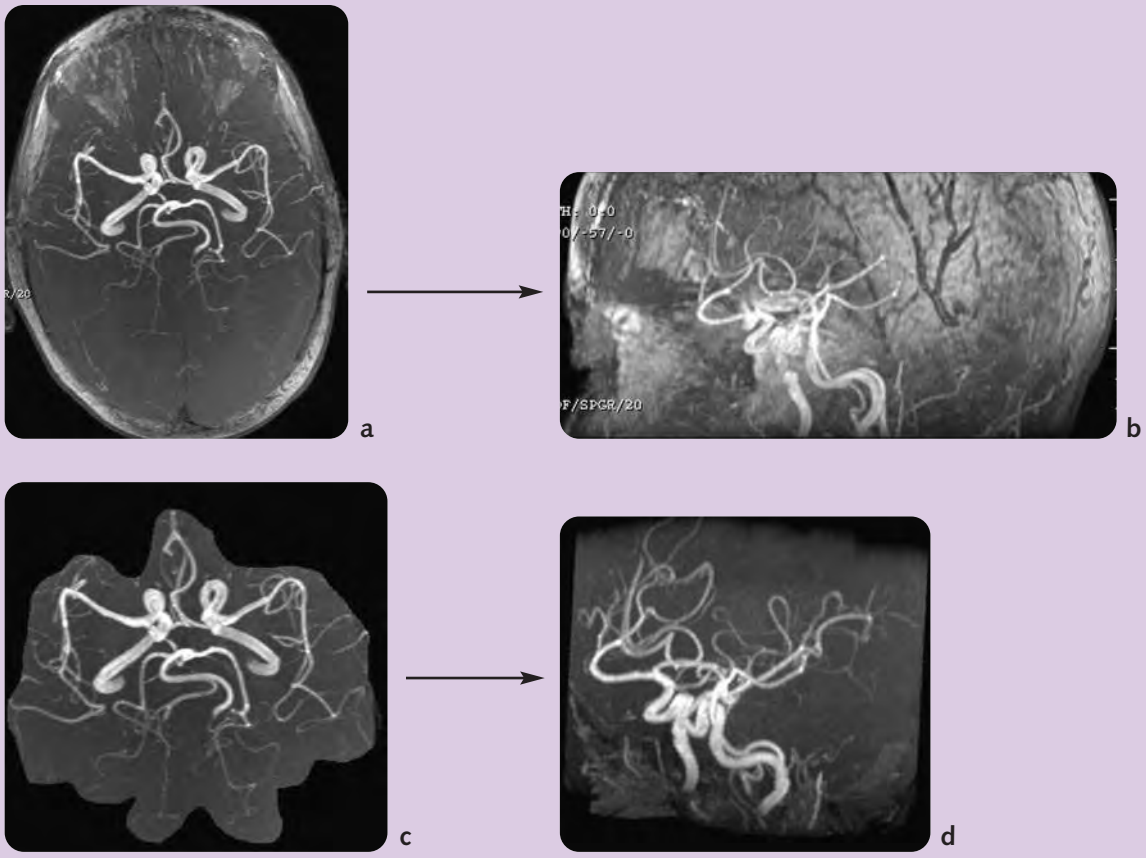


Figure 9a-d. (a) Axial MIP of a 3D TOF image, untrimmed. Since this MIP is performed in the same direction as the acquired slices (in the axial plane) it is commonly referred to as the "collapsed" MIP. (b) The same MIP image rotated to an anterior-sagittal view. Note how the bright signal obscures the vasculature and the smaller cerebral arteries are not seen at all. (c) This image shows the collapsed MIP with the bright signal trimmed away. (d) This image shows the same rotation as in (b). Note the increased visibility of the vasculature, including the clarity of the smaller cerebral arteries.

any desired projection to optimally assess for narrowing, dilatation, or other pathologies.

Vascular and MRCP studies are just two applications for MPR processing. The MPR technique can be employed for any application where data scanned in one direction (axial, for example) needs to be visualized in another, where the other direction does not exist. For example, axial and coronal T1-weighted images of the brain are acquired, as well as axial T2s. If the patient is unable to complete the exam and no coronal T2s were acquired, coronal MPRs can be processed from the axial T2s for visualization. However, it must be noted that the quality of any MPR is directly proportional to how isotropic the voxels are. A true isotropic voxel is a cube, in that its dimensions are all equal for height, width, and length. Such a voxel can be rotated in any direction while maintaining its voxel size and spatial resolution (Figures 10 and 11). Note the axial image in Figure 10a and the sagittal reconstruction of the same image in Figure 10b; the voxel size is the same except for the length dimension, since the slice thickness is greater in one than the other. Note the difference in quality of the coronal MPR from each sagittal image, particularly the greater detail and quality from the axial data set where the voxel was closest to isotropic.

When performing MPR reconstructions, the technologist selects a reference image (active or base image) and uses it as the basis for each type of cut calculation. Two other segments display orthogonal cuts through the base image. Once the 3D pointer is shifted on the base image, cuts are recalculated in real time and displaced in the other two segments. Positioning can be performed obliquely or double obliquely, and a series of reconstructed images can be obtained in different obliquities. Slices can be thin or thick. These cuts can be reconstructed and saved as

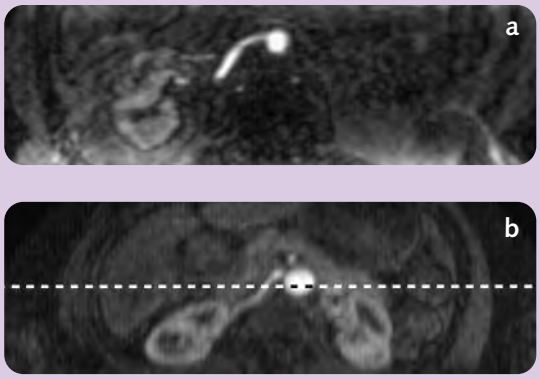


Figure 10a-b. MPR through the right renal artery.

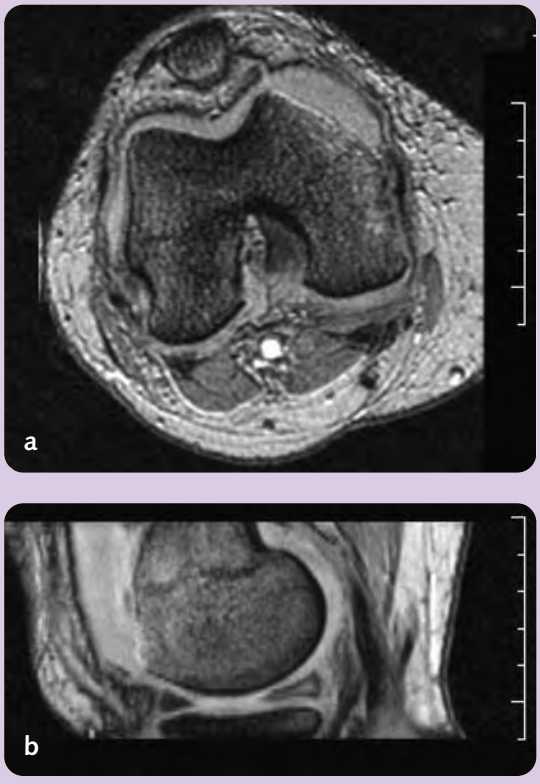


Figure 11a-b. (a) Axial 3D gradient echo of the knee with a 1 mm slice thickness. (b) This image shows the sagittal reconstruction of the axial 3D image in (a). Note the sagittal reconstruction maintains its high resolution because the voxels are nearly isotropic. Compare these images to Figure 12.

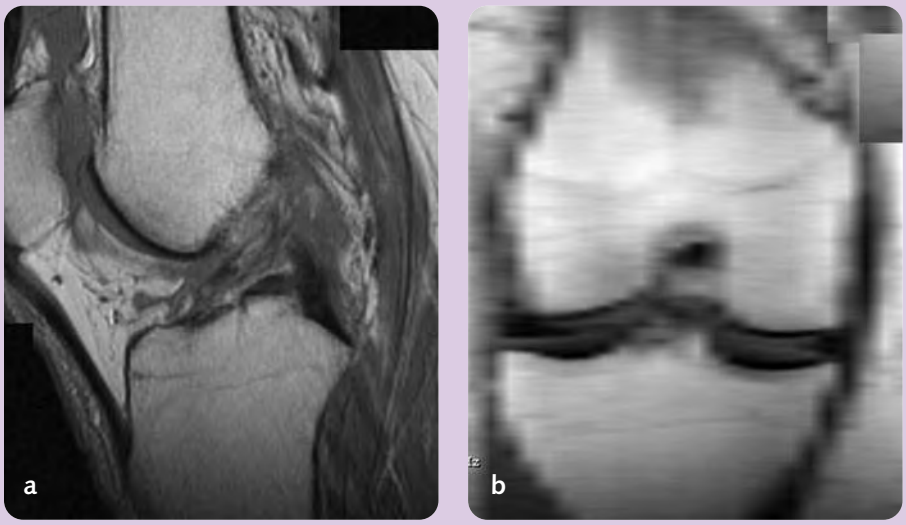


Figure 12a-b. (a) A 2D sagittal image of the knee with a 3 mm slice thickness. Note the high in-plane image quality. (b) This image shows the coronal reconstruction of the image in (a). Note the lack of sharpness and resolution resulting from the voxels not being isotropic.

a separate image series, which can then be viewed as a yo-yo or cine loop.

MPR is an excellent tool for evaluating tortuous, kinked, or branching vessels, and it can also help assess pathology in overlapping vessels. MPR is useful for accurately evaluating stenoses and aneurysms, providing accurate measurements of aneurysms, determining the anatomic relationship of branch vessels to the aneurysm, and defining the aneurysm neck. Other advantages of MPR include evaluation of small vessels, depiction of atheromatous plaque and mural thrombus, delineation of complex vascular anatomy in multiple planes, and definition of luminal detail.

EXOSCOPIC AND ENDOSCOPIC VIRTUAL SURFACE RENDERING TECHNIQUES

Exoscopic Virtual Surface Rendering

Volume or surface rendering uses a ray-tracing algorithm that selects visible voxels by tracing rays from an instantaneous viewing position. The walls are identified by a thresholding technique, defining the surface of the object under evaluation. Once the surface of the object is demarcated, all other data are discarded. Surface contours are modeled as a collection of polygons. Display is with surface shading. The image obtained is simplified and can potentially misrepresent a structure, especially if the surface of the object is not easy to determine or define. When converting data from volume to a surface, significant amounts of the data are lost in exchange for faster, easier computation. Exoscopic virtual surface rendering provides very little information about anatomy or pathology of the internal luminal wall of a blood vessel. Surface rendering postprocessing is included in most MR software packages or is available separately as 3D medical imaging software (Figure 13).

Volume Rendering

Volume rendering is a postprocessing technique that provides the entire volume of data, not just the surfaces. This technique provides more information than the shaded surface model. Volume rendering adds the contributions of each voxel along a line from the

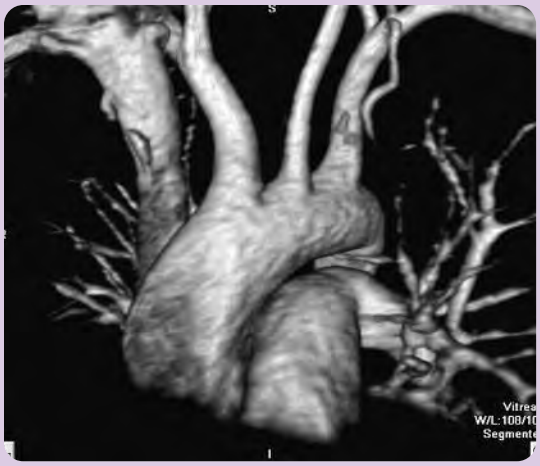


Figure 13. This is a volumetric surface rendering of contrast-enhanced 3D TOF of the aortic arch and great vessel.

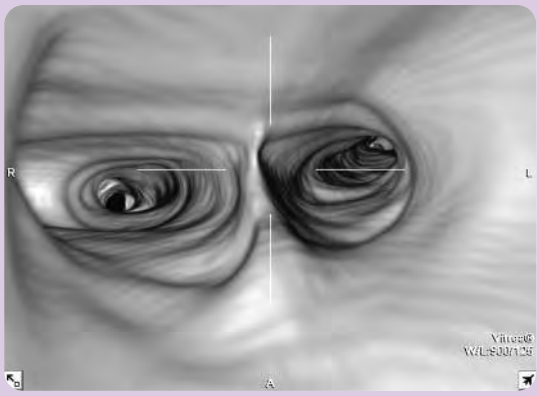


Figure 14. This image shows an endoscopic fly-through of the internal carotid artery facing the bifurcation of the internal and external carotid arteries. The original data set is a contrast-enhanced 2D TOF image.

viewer's eye through the data set. The process is repeated to determine each pixel value in a displayed image. Volume rendering significantly enhances the structure of a data set, especially when compared to surface rendering, and reveals structures that would be hidden with surface rendering techniques. This technique is useful for diagnosis and surgical and postoperative planning. Because each voxel is used in its entirety, and because the data sets can be very large due to processing many slices with high resolution, the amount of processing required can take minutes to reconstruct using volume rendering versus seconds for a MIP. Computers equipped with the latest high-end processors can greatly reduce the processing time to near-real time.

Endoscopic Virtual Surface Rendering

Virtual endoscopy allows the operator to interactively navigate within a vascular lumen with 3D visualization of the internal luminal surface. The process of reviewing a 3D MRA data set using virtual surface rendered endoscopy is called "fly-through." It remains to be determined whether virtual endoscopy provides unique diagnostic benefits. Figure 14 shows an endoscopic view of the common carotids from 2D TOF data.

ADVANCED IMAGE ANALYSIS AND QUANTIFICATION

Many advanced MR studies require special software analysis tools for accurate and complete diagnostic interpretation. This software is usually available from MR scanner manufacturers or third party vendors. Below are examples of MR studies that require advanced postprocessing and the specific functions of the analysis software.

BRAIN PERFUSION

MR brain perfusion mapping evaluates regional blood flow and volume to the brain and has become an important diagnostic tool in patients with suspected stroke. In this test, MR images are acquired rapidly and repeat-

edly during intravenous administration of a bolus of gadolinium-based contrast agent. The magnetic susceptibility of the intravascular gadolinium causes signal loss in perfused areas of the brain on T2*-weighted MR images. In healthy brain tissue, contrast tends to be absorbed rapidly (called "wash-in") and then be excreted ("wash-out") rapidly. Brain tissue that lacks good vascularity, either through compromised blood flow or tissue death, lacks this wash-in/wash-out effect. To quantify brain perfusion, the relevant MR data sets are loaded into specialized analysis software, and the user selects one or more ROI. The software calculates several perfusion parameters, including mean transit time (MTT), enhancement up-slope, relative cerebral blood volume (rCBV), peak enhancement, and area under the curve (AUC). The results can be displayed graphically or as a pseudocolor overlay on anatomic brain images.

Applications for brain perfusion and its required postprocessing typically include assessment of acute stroke where the differential diagnosis is brain tissue death versus at risk of infarction, and tumor versus necrotic tissue questions.

BRAIN DIFFUSION

MR diffusion imaging evaluates the diffusion of intercellular water in the brain. Diffusion imaging is useful for detecting the cytotoxic edema associated with stroke and, like perfusion mapping, is a mainstay for evaluating stroke patients to determine which are candidates for thrombolytic therapy. The MR diffusion sequence is typically run four times, once without the diffusion gradient, then one time each using three different orthogonal diffusion gradients. Therefore, each slice location is scanned four times: one time as a T2*-weighted image and one time in each of the X, Y, and Z diffusion directions. Typically the three sets of diffusion images for each slice are automatically combined during the reconstruction process and displayed as a single diffusion-weighted image.

The degree to which the pulse sequence is sensitized to molecular water motion is determined by the strength of the diffusion gradient, which is known as the "b-value." The stronger the gradient (b-value of 2000 versus 1000 for example), the greater the sensitivity to water motion, or lack thereof. However, increasing the b-value also increases the TE of the sequence, leading to greater loss of SNR. Therefore, a good balance must be found between b-value and SNR. Typical b-values for diffusion weighted imaging at 1.5 T range from 1000 to 2000, although most scanners are capable of far higher values.

It is often advantageous to calculate a map of apparent diffusion coefficients (ADC). The ADC map circumvents a pitfall of diffusion imaging known as "T2 shine-through." Areas of pathology on diffusion-weighted images are bright. However, normal tissue having long T2 times can sometimes remain bright and "shines through" on the diffusion-weighted images, mimicking or exaggerating the pathology. ADC maps provide a grayscale representation of relative diffusion coefficients, and the maps are clinically useful for identifying areas of abnormal high or low diffusion in the brain. The ADC map renders the pathology darker while leaving the T2 signal bright, thus differentiating the two. Exponential ADC maps (eADC) can reverse the ADC process one more time, returning the pathology to its more common bright contrast and making the T2 signal dark, thus eliminating any T2 shine-through.

A related MR technique called diffusion tensor imaging (DTI) detects diffusion

anisotropy (directional differences in the diffusion of water). In DTI images, the diffusion gradients are not limited to the X, Y, and Z directions. Rather, they can be applied in multiple directions; the number of directions typically used is 25. The use of multiple directions allows DTI to demonstrate clinical efficacy in depicting and defining normal and abnormal nerve fiber tracts in the brain, an application known as tractography. Most MR scanner manufacturers offer advanced software to postprocess the DTI data and create fiber tract maps of the white matter fibers in the brain.

Functional MRI

Functional MRI, or *f*MRI, is a class of imaging techniques that is characterized by the unique purpose of the examination. *f*MRI is used to evaluate the *function* of an organ, but not necessarily its visual structure. The term *f*MRI has slightly different meanings depending upon the context in which it is used. It is widely, though incorrectly, thought to refer only to a type of imaging known as BOLD imaging, discussed later.

The premise on which *f*MRI is based is that certain disease processes may significantly alter the function of an organ without necessarily altering its visual structure. Additionally, a structural change to an organ can often be caused by numerous different disease processes yet have the same end appearance, making diagnosis difficult, if not impossible, on the basis of appearance alone. For example, an area of metastasis in the brain can appear very similar to a multiple sclerosis lesion and/or can also resemble small ischemic areas. The role of *f*MRI imaging in this case would be to aid in the evaluation of what specific disease process is occurring. Examples of *f*MRI include BOLD imaging, MR spectroscopy, T1, and T2* perfusion imaging.

MR Spectroscopy

MR spectroscopy (MRS) is the oldest form of *f*MRI; it is also the oldest form of MR imag-

Spectroscopy Disease Signatures

Disease	NAA	Creatine	Choline	Lactate	Lipid
Radiation necrosis	decreased	decreased	decreased	increased	increased
Infarct	decreased	—	—	increased	increased
Multiple Sclerosis	decreased	decreased	increased	—	—
Epidermoid	—	—	—	increased	—
Solid tumor	decreased	decreased	increased	—	—
Necrotic tumor	decreased	—	increased	increased	increased
Abscess	decreased	decreased	decreased	increased	increased

Table 1. Spectroscopy disease signatures, indicating typical responses of common metabolites to certain disease states. The table refers to the height of the spectroscopy signal as compared to the normal spectroscopic display.

ing. The chemical constituents of tissues and organs can be evaluated with MRS. Clinical applications of MRS primarily examine hydrogen protons, although other nuclei, such as phosphorus-31, carbon-13, fluorine-19, and sodium-23, can also be studied. The real clinical value of MRS derives from extensive research showing that tumors and other pathologies often exhibit a characteristic metabolic fingerprint. Table 1 provides the typical MRS signatures for common pathologies. MRS sequences are roughly divided into single voxel and multivoxel techniques. Postprocessing software is required to convert the raw MRS data from each voxel into a spectrum (Figures 15-18). Most MRS software packages also permit spectral editing and measurement of spectral peaks, the latter

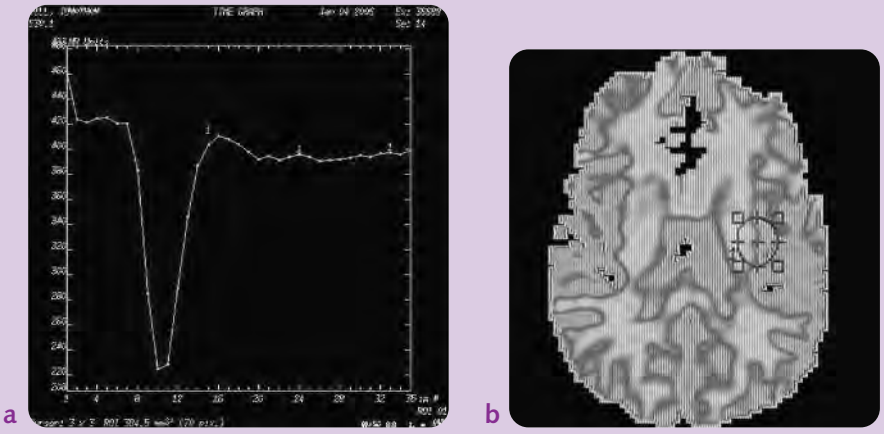


Figure 15.
(a) This figure shows a graphic depiction of the wash-in/wash-out effect of gadolinium on normal brain tissue. (b) A map of the brain demonstrating wash-in/wash-out over the entire slice of brain.

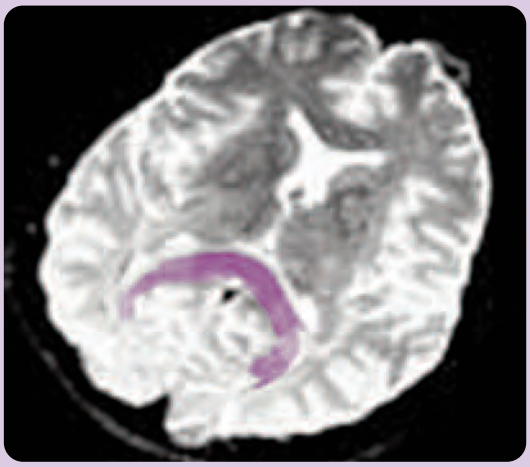


Figure 16. This image demonstrates the colorized white matter tracks for a diffusion tensor image, or tractography. The information provided in this application is useful in surgical planning and tumor staging.

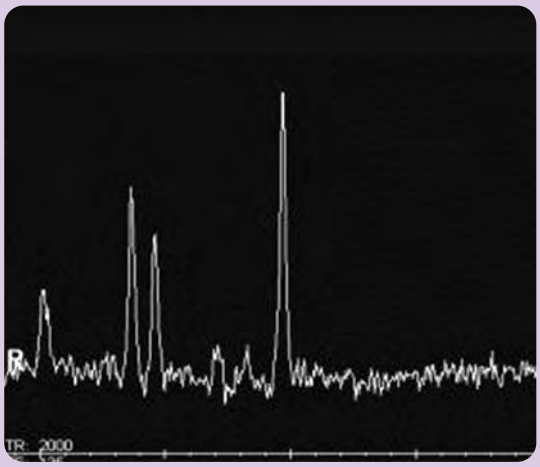


Figure 17. MRS can be performed as a single large voxel of an ROI, producing a single spectrum.

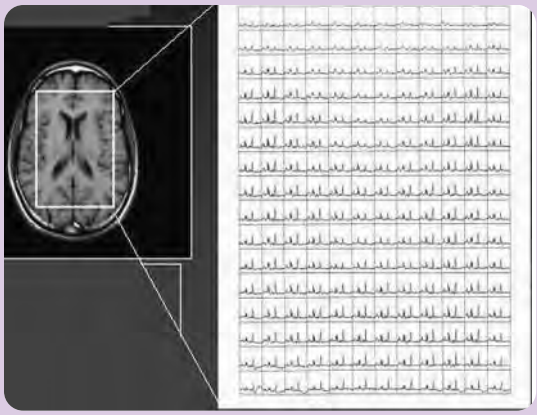


Figure 18. MRS can be performed over the entire slice plane and then analyzed in a grid of spectra. This is also referred to as multivoxel MRS.

providing an estimate of the relative concentrations of the various metabolites. Metabolic mapping provides a convenient, color-coded, visual display of a single metabolite's concentration overlaid on an anatomic MR image (Figure 19). Most MR scanner manufacturers offer a variety of MRS software packages.

BOLD

BOLD imaging is just one method of *f*MRI, although it is often incorrectly labeled as simply *f*MRI. BOLD imaging is a method for visually capturing areas of the brain in action. In other words, BOLD imaging allows the technologist and radiologist to literally watch the brain function during a mental task. BOLD imaging uses the concept of magnetic susceptibility and requires careful application of a mental stimulus to capture minute increases and decreases in susceptibility. When performed appropriately, BOLD imaging techniques can provide a tremendous amount and quality of information about the functional integrity of the affected area. Such

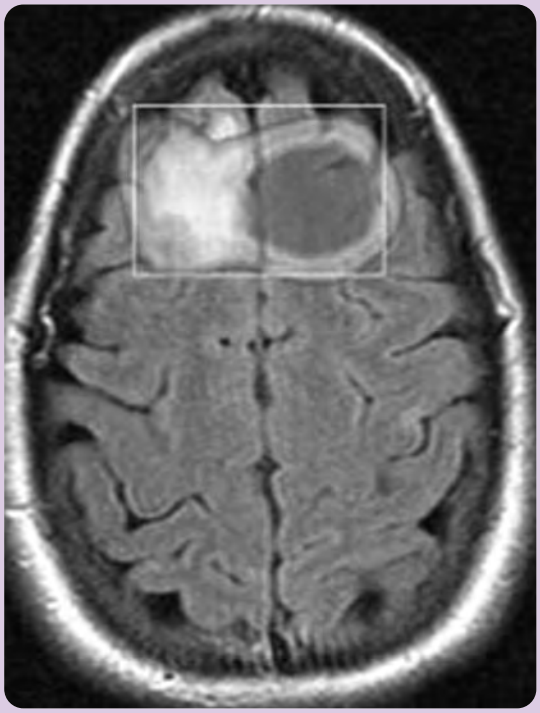


Figure 19. Recently, the ability to visually map the MR spectrum has led to the use of metabolite mapping where the spectrum can be visualized as a color map indicating higher levels of the metabolite in question. This is a map of N-acetyl aspartate (NAA) following MRS examination. Note the lack of NAA on the left side, which is an important marker in certain pathologies.

information enables treating physicians to carefully map out an appropriate course of action, such as in surgical planning.

Performing BOLD imaging requires the use of some stimulus or task (finger tapping is a simple example) that is turned on and off at regular intervals, so that the area of the brain controlling or processing that task shows blood flow to that area increasing and decreasing while scanning is taking place. In the brain, as with many parts of the body, when an area is "working," blood flow will

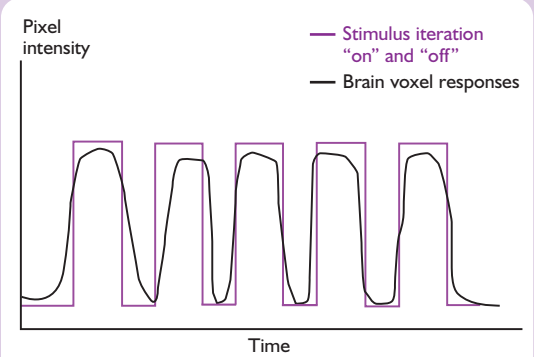


Figure 20. An illustration of the task iteration of "on" and "off" with a graph of the voxel value of an ROI. The graph demonstrates how that area of the brain experiences increases and decreases in the blood flow along the same temporal pattern as the on-off task.

increase to that area. When the brain stops working, the blood flow returns to normal and equilibrium is restored. The contrast mechanism relies on the principle of magnetic susceptibility. Increased levels of oxyhemoglobin to any area of the brain causes a change in the local magnetic susceptibility, due to the slight paramagnetic properties versus deoxyhemoglobin. The change in susceptibility causes the signal values to increase or decrease on a minute level. While the human eye cannot discern these small changes, a computer is able to, provided it has enough information for statistical analysis. Typical BOLD sequences can yield from 500 to more than 7000 images (Figure 20).

Breast

MRI of the breast can detect cancers that are not palpable on physical examination or visible on mammography, making breast imaging a rapidly growing area of MRI. Emerging clinical applications of breast MRI include breast cancer screening for high-risk patients and local cancer staging both pre- and postlumpectomy. Malignant lesions seen on breast MRI show signature features of contrast enhancement and irregular or spiculated margins. A typical breast MRI study consists of precontrast T1-weighted imaging followed by imaging at multiple time points after administration of intravenous gadolinium-based contrast. The time-course of lesion enhancement—also called enhancement kinetics—provides diagnostic information about the likelihood of malignancy. Software packages are available from MR scanner manufacturers and third party vendors for postprocessing breast MRI data, including calculation of various kinetic parameters, eg, signal enhancement ratio, up-slope, and wash-out, and display of color-coded breast perfusion maps.

Once the imaging sequence is complete, all the images (as many as 700) are loaded into the postprocessing software program. The program is capable of quickly analyzing each pixel in each image for its contrast change over time. When the pixel intensity matches the preselected algorithms up-slope, wash-out, etc., the pixels are colorized so that they are easily seen against all the others. The postprocessing operation is typically fast and requires little technologist interaction, masking the sophistication of the software programs and their analytic capabilities.

Cardiac

Advances in MR technology have led to many new quantitative cardiac MR applications for determining ventricular volume, wall thickening, function and mass, first pass myocardial perfusion, myocardial viability, and flow-based measurements of valvular function and cardiac shunts. Considerable image postprocessing is necessary to calculate the various

parameters. Fortunately, these tasks are facilitated by cardiac software packages included with many MR systems or available from third party vendors. Cardiac MR postprocessing software is usually semiautomated, requiring the user to draw or confirm multiple ventricular or vascular contours. The standard outputs of analysis software are graphical and tabular. "Bull's eye" plots offer a convenient topographical display of left ventricular wall thickening, perfusion, and viability.

IMAGE FILTERING

The filter function of the MR system permits the technologist to proactively or retroactively manipulate certain features of the MR images. The filter function works to smooth contiguous areas and enhance contours. This function also enhances SNR without loss of information about structural anatomy. The filter is used as an adaptive filter. Its function is based on local distribution of the signal intensity and the plot of local tissue structures.

Various low and high pass filters are used to assess whether anatomical structures are coincidental or connected with other structures. Filter sharpeners can be selected. Filter settings include smooth, medium, and sharp, and combinations of all three, such as "heavy smoothing with little sharpening" and "medium sharpening with low smoothing." Various MRI systems manufacturers make recommendations for individual filter values. Some settings will result in increased structure enhancement, emphasizing edges and lines, which is useful for MRA images (Figure 21).

Other settings may yield good noise suppression and low signal amplification. The appearance of filtered images may be very subjective, depending upon the interpreter.

Whether a high image filter is used (or any filter at all) is completely dependent on the individual tastes of the person interpreting the images. It must also be noted that most manufacturers apply some degree of image filtration in the reconstruction process and allow for further filtration as a postprocessing feature. In these cases the images are "double" filtered. While heavy filtering rarely alters the diagnostic quality of the image, it can give the image an "airbrushed" appearance. This appearance, while completely acceptable to some radiologists, is rejected by others. Image quality is often subjective.

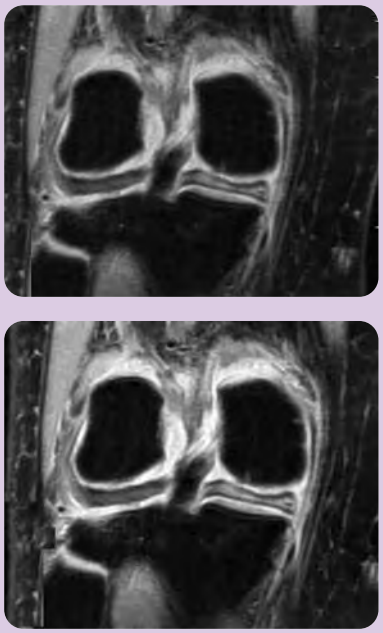


Figure 21a-b. (a) This image shows a coronal fat-suppressed proton density image of the knee without any postprocessing filtration. (b) This is the same image, filtered after initial reconstruction to sharpen edges and suppress noise. Note the increased edge definition, particularly in the area of the medial meniscus.

OTHER IMAGE ANALYSIS AND MANIPULATION TOOLS

Every MR system comes with numerous standard image manipulation and display features. While some of these are used rarely, such as calculating pixel value, others, like image magnification, are used quite extensively. Other features standard on MRI systems include:

- **Manipulation of grayscale values (window-level) of selected images for viewing**
  By far the most often used image manipulation feature of any MR scanner. Typically referred to as "window/leveling," this tool allows the technologist or the radiologist to change the brightness as well as the contrast of the overall image. Note that this tool does not allow for the altering of individual structures in the image, only the image in its entirety.

- **Image magnification**
  Used in filming as well as viewing images electronically that have a rather large FOV, where structures of interest are small in the unmagnified image. An example is imaging of the pituitary gland.

- **Image scrolling**
  Allows the technologist or radiologist to view a stack of images one at a time in progression.

- **Measure distance and angle**
  Used for measuring the dimensions of structure of interest, often to gauge changes in the structure's size during treatment regimens such as radiation and/or chemotherapy.

- **Image annotation**
  Used for labeling images to provide needed information about the image to the radiologist. Examples include annotating a left or right extremity and annotating the targeted MIP of a left or right middle cerebral artery.

- **Statistically evaluate the distribution of pixel grayscale values using ROI calculations of mean value, standard deviation, and minimum/maximum values and display the ROI frequency distribution of pixel values as a histogram, as well as in graphical or tabular form**
  Though not used often, this tool can be used to determine the precise timing of the peak amount of contrast delivered to an area of interest. An application for this is in breast cancer imaging where the contrast uptake is measured for time required to reach a suspected lesion as well as how long it takes the absorbed contrast to wash out.

SUMMARY

In conclusion, there is an array of postprocessing techniques available to the radiologist and MR technologist. Each has specific utility in evaluating MR data and enhancing the diagnostic yield and accuracy of MRI. Postprocessing techniques are generally fast and user-friendly and do not add to patient examination time. One of MRI's most useful applications is the MR angiogram. This type of study is faster, safer, and less painful for the patient compared to conventional angiography. Yet, it is the postprocessing of the acquired images that makes MRA practical and useful. Without the ability to subtract data and create targeted MIPs, the raw MRA examination data is of limited value in diagnosing vascular abnormalities.

Along with the postprocessing performed for MRA imaging, the more powerful computers in use today, along with sophisticated imaging analysis software packages, have caused an explosion of new applications, such as cardiac and breast cancer analysis. Both require voxel manipulation and analysis to provide not only qualitative information, such as image reformations, but also quantitative information, such as contrast uptake curves and analysis.

There is no question that MR postprocessing tools have been the enablers for numerous important applications; there is little doubt that future applications will require new and more complex postprocessing tools.

MRI IMAGE ARTIFACTS

Even under the best of imaging circumstances, image artifacts will often arise in the course of a clinical examination. The analysis, quantification, and interpretation of a study is only as good as the quality of the source images. Thus, any type of artifact is undesirable because it reduces the quality of the source images.

A wide range of factors can be responsible for causing artifacts, but most factors fall into one of three main categories: faults or limitations in the hardware, less than optimal scan parameter selection, and patient motion.

In some cases artifacts can be eliminated completely by updating imaging parameters or repairing the faulty hardware. In other cases some level of artifact must be tolerated, but careful selection of parameters can help to minimize their effects on image quality. Artifacts related to patient behavior may be improved with careful instructions to the patient and reassurance to those who are anxious about the examination.

The sections below provide an overview of the types of artifacts that can arise in the course of a typical clinical scan and strategies for eliminating or minimizing many of these artifacts.

MOTION ARTIFACTS

Motion artifacts are caused by patient movement during the acquisition of an image. Patient motion may be voluntary, such as a patient trying to get into a more comfortable position, or it can be involuntary, such as movement caused by the patient's breathing. Types of motion that can cause artifacts include:

- gross patient motion, such as involuntary jerk motion and swallowing
- respiratory motion
- cardiac motion
- peristaltic motion

Artifacts related to fluid flow can also be considered a form of motion artifact but will be discussed in a separate section.

When motion occurs between the acquisitions of different phase-encoding steps, the shift in position means that the final image will be made up of data from the object being scanned in two or more different positions, and the reconstructed image will contain data from a mix of positions. Since the data are acquired in k-space, the result may not always be a simple blurring of the image, as it would be for a long-exposure film. Periodic motion, in particular, yields a ghosting effect, which appears like a faded but misplaced duplicate image in the background along the phase-encoding direction.

Gross Patient Motion

Patient motion is one of the most unpredictable types of motion, and the artifacts arising from patient motion can be difficult to correct. Despite clear instructions and

reassurances about the safety of the examination, patients may still move or shift during the course of an examination. The cause may be physical pain or discomfort from positioning in the scanner or unrelated conditions. Some patients may have disorders that make it difficult to lie still for the duration of a study. Anxiety may also prevent patients from remaining motionless during the examination.

Effects of Gross Patient Motion

Blurring. Any shifting of position or movement during the examination will probably cause blurring on the images of the anatomy that has moved. The magnitude of the artifact will largely depend on the degree of motion. At a minimum, blurring causes an apparent loss of resolution and inconclusive image quality.

Misregistration. When the technologist or radiologist reviews stacks of 2D images, the anatomy should appear in alignment from image to image. Gross patient motion may cause the anatomy to be misaligned or misregistered. Depending on the timing and duration of the motion, misregistration or ghosting (in the case of a periodic motion) of the images may result. Misregistration is especially likely in protocols that acquire time sequences of images or multiple slices over a series of acquisitions, such as in breath-hold acquisitions.

Remedies for Gross Patient Motion

Careful attention to patient comfort during positioning and placing positioning pads around the area of interest for stability can help. Clear instructions and reassurance can also help alleviate the patient's anxiety.

Respiratory Motion

Respiratory motion arises from the motion of the organs of the chest and abdomen during normal breathing. See Figure 22 for an example of respiratory motion artifacts.

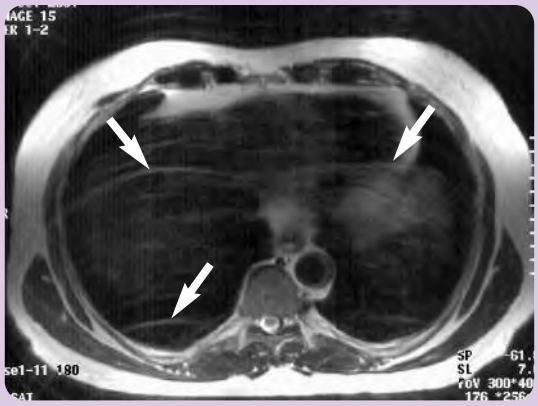


Figure 22. Respiratory artifact shown on an axial T2-weighted imaged.

Effects of Respiratory Motion

Respiration is a periodic motion, and the artifact will usually appear as a ghosting artifact. However, changes in breathing rate or depth can make the artifact more complex.

Remedies for Respiratory Motion

Abdominal Binder. An abdominal binder restricts the motion of respiration, thus reducing artifact. Scanning the patient in a prone position can also minimize respiratory motion, but it may be uncomfortable for many patients, and the prone position is not suitable for patients with pre-existing breathing difficulties. Another, more practical, method for achieving this goal is to bind the area that is being scanned to force breathing to occur in another body area. For example, if the abdomen or pelvis is being imaged, place a wide waist wrap (always provided by the manufacturer) around the waist to make the patient breathe in the chest area. The technologist must be careful not to bind the

waist too tightly because it may be uncomfortable for patients and increase their level of anxiety.

Signal averaging. The signal averaging technique works by acquiring the phase-encode lines multiple times so that the motion is averaged out over several acquisitions, since the respiratory position is likely to be different during different averages. While this method works very well, it also dramatically increases scan time, often prohibitively.

Respiratory triggering. During respiratory triggering, the patient's respiration is monitored with navigators or a respiratory bellows, and data are only acquired during a part of the respiratory cycle. The acquisition point is usually at the end of expiration, when the motion is minimal and there may be a quiet period of no motion before the next breath. Since no data are collected during the remainder of the respiratory cycle, the scan time can be increased by as much as a factor of two. More efficient FSE imaging has reduced the scan time increase significantly. Another important factor is that when respiratory triggering is used, the control of the TR is determined by the patient's respiratory rate, resulting in long TRs and making standard T1 imaging with respiratory triggering nearly impossible. For this reason respiratory triggering is typically limited to proton density and T2-weighted imaging.

Respiratory pseudotriggering. By selecting the TR to match the rate of respiration, data acquisition can be timed to the same phase of respiration for each phase-encode. This timed acquisition assumes that the respiration rate is consistent throughout the scan, which often is not the case. Because of the difficulty in assuring a consistent respiratory rate, pseudo-triggering is not often employed.

Respiratory ordered phase-encoding. In this technique, data are collected continuously throughout the respiratory cycle. For each phase-encode line acquired, the respiratory position at that time is also recorded. Upon completion of the scan, the acquired data are shuffled to correspond to an acquisition with a long, slowly varying respiratory pattern. The resulting images have reduced ghosting, and the scan time does not need to be significantly lengthened.

Gradient moment nulling. Though more commonly used for flow compensation, gradient moment nulling can also be applied to reduce some of the effects of respiratory motion. Extra gradient pulses are applied to cause the phase changes due to motion to refocus at the echo time. This helps restore the signal loss from intravoxel dephasing and reduce the constructive interference that causes the motion ghosts. As with flow compensation, gradient moment nulling is applied in the readout and slice select directions.

Gradient reorientation. Since the motion artifact yields ghosting in the phase-encoding direction, a swap of frequency and phase-encoding directions can throw the artifact into a different—and perhaps less detrimental—area of the image. The idea is not to eliminate the ghosting but to have the ghosts placed where they will not harm the diagnostic portion of the image. Motion blurring will not be improved with this technique, but its direction will be changed. The gradient reorientation technique is often applied when scanning the thoracic spine in the sagittal plane where respiratory motion (and also cardiac motion) can ghost across the image in the phase direction, obscuring the spine. When the phase direction is swapped to the frequency direction, the motion artifacts run top to bottom (superior to inferior) and anterior to the spine. Also, care must be taken to assure that suffi-

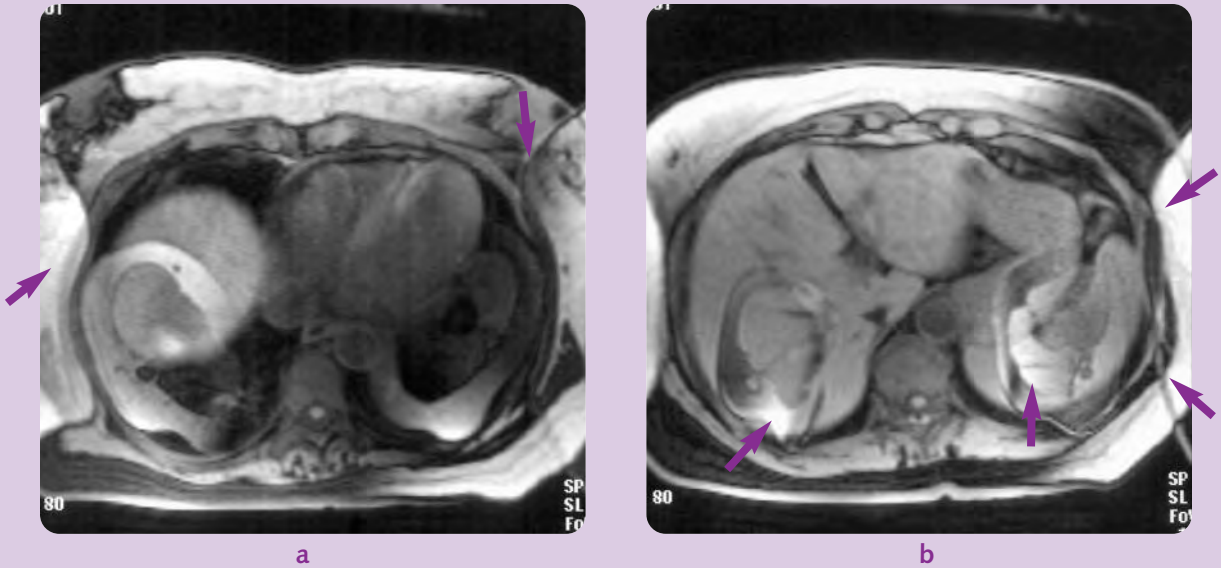
cient sampling is maintained in the new phase-encode direction to prevent wrap artifact (Figure 23).

Spatial presaturation bands. The intensity of ghosting artifact will be proportional to the intensity of the tissue that is moving. Often, the high signal of abdominal fat is a significant contributor to artifact. The saturation pulses eliminate much of the signal from the moving tissue and, therefore, eliminate much of the ghosting signal as well. Saturation pulses should not be used on the anatomy of interest for the examination. The addition of saturation pulses may impact some of the other imaging parameters, such as TR or the number of slices acquired. An example of spatial presaturation banding is to place thin saturation bands over the anterior abdominal wall, thus saturating signal that might otherwise ghost across the image.

Fat suppression. Fat suppression techniques use much the same idea as presaturation, but they work by chemically selective elimination of signal (from fat) rather than by spatial saturation. Overall SNR may be reduced, and nonfat tissues will still give rise to artifact if in motion.

Bellows. A respiratory bellows is similar to a rubber belt placed around the patient's chest or abdomen to monitor respiration via the expansion and contraction of the chest cavity. The bellows serves as the monitoring device for performing respiratory triggering or respiratory ordered phase-encoding and is also used to verify that patients do, in fact, hold their breath during breath-hold examinations.

Navigators. An alternative method for monitoring respiratory motion is the use of navigator echoes. A navigator is basically a one-dimensional image that includes the



a b

Figure 23a-b. Axial T1-weighted breath-hold images with gradient reorientation. The phase-encode direction is now right to left. Wraparound artifact is seen (arrows) because the FOV was not appropriately increased.

diaphragm. Between each phase-encode acquisition, a brief excitation and one-dimensional data readout is taken and rapidly reconstructed. The sharp contrast between the diaphragm and lungs can be reliably detected and used to measure the motion of the diaphragm during breathing. This positional information can then be used to estimate the respiratory phase (inspiration or expiration) as with the bellows. Alternatively, the motion information can be used to track the anatomy by shifting the slice location by the same amount to track the organs during the respiratory cycle.

Ultrafast imaging. Rapid and ultrafast imaging techniques can eliminate most respiratory motion artifacts by collecting images much more rapidly than the rate of respiratory motion. The short duration of these scans mean they have limited resolution and/or SNR, so image quality may not be sufficient for an examination.

Breath-hold imaging. The above techniques have varying degrees of success in suppressing respiratory motion, but none can completely eliminate artifact; the best method for eliminating breathing motion is to eliminate breathing in the first place. In breath-hold imaging, the patient is given breathing instructions prior to the beginning of the acquisition and suspends breathing for 10 to 30 seconds while data are collected. This method can completely eliminate the motion artifact. The drawbacks are that only a limited amount of data can be collected in a single breath-hold (limiting the resolution that can be achieved), and many breath-holds may be necessary to complete the coverage required. Multiple breath-holds can be fatiguing for the patient. The breath-hold position may also be inconsistent, yielding misregistration of data among the acquired slices. Parallel imaging, which reduces the number of phase-encodes required, can help to reduce the duration of breath-holding required. Other advanced rapid imaging methods, such as fast 2D and 3D gradient echo pulse sequences, now enable whole-liver imaging to be performed in a single breath-hold.

Cardiac Motion

Cardiac motion comes not only from the complex motion of the heart as it beats in the chest but also, and more importantly, from the extreme motion of the blood as it flows through it. Substantial motion is caused not only by the amount of blood moving through the heart but also from the numerous directions in which it is moving.

Effects

Like respiration, the motion of the heart is periodic and has a regular rhythm, but unlike respiratory motion, the patient has no voluntary control over the heart's rhythm. The artifacts caused by cardiac motion are qualitatively similar to respiratory motion artifacts. Ghosting artifacts are seen along the phase-encoding direction, and general blurring is also displayed (Figure 24). Cardiac pulsation can also be transmitted and cause

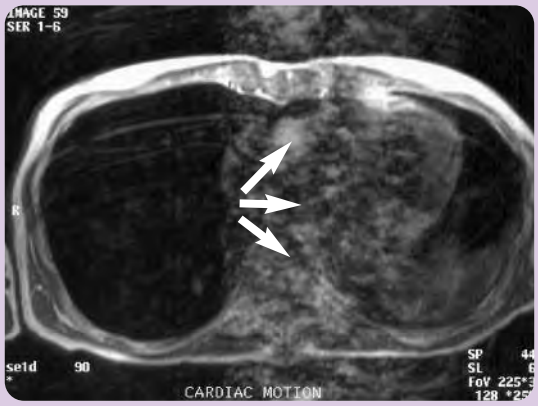


Figure 24. Cardiac motion (arrows).

signal changes in other anatomy, such as the liver or thoracic spine. Cardiac motion suppression is accomplished by synchronizing the image acquisition process to the heart cycle through cardiac gating or electrocardiogram (EKG) gating.

EKG Gating

The most common form of cardiac gating uses EKG leads on the patient's chest and/or back to generate an EKG tracing that is sent to the scanner. The signal from the EKG is used to trigger each repetition of the pulse sequence. The R-wave is the most visible feature of the EKG tracing and the most reliable for triggering.

When cardiac gating is applied, the repetition time of the scan will become the interval between successive heartbeats (the R-R interval). The number of slices and other parameters of the examination will be dependent on the heart rate. It is important to realize that the heart rate may increase during the scan, so the R-R interval (and the effective TR time) may get shorter. Some margin of safety should be added to ensure that all of the data can be acquired within a single heart cycle in case the R-R interval becomes significantly shorter.

If properly triggered, each acquisition will be taken at the same point in the heart cycle. This means that the heart should be in the same position as each phase-encode is taken, effectively "freezing" the motion of the heart.

Some of the parameters involved in cardiac gating include:

Trigger window. The trigger window is the period between the last activity of the scanner triggered by the last heartbeat and the expected arrival of the next cardiac trigger. The trigger window provides a margin of safety for the heart rate to increase without missing heartbeats. For example, if the heart rate is 60 beats per minute, the R-R interval will be 1000 ms, and the trigger window might be set at 150 ms; 10–20% of the R-R interval is typical. This means that data may be acquired for 850 ms, at which point the scanner will stop and wait for the next cardiac trigger before continuing. If the heart rate increases to 65 beats per minute (an R-R interval of about 920 ms), the scanner will still start looking for the next trigger after 850 ms and will still receive the next heartbeat. However, if the heart rate were to increase to 75 beats per minute (R-R interval of 800 ms), the gating system will miss the next trigger and wait for the following heartbeat. The selection of the trigger window is a compromise between efficient use of scan time and reliable triggering.

Trigger delay. The trigger delay is the time delay from the detection of the R-wave trigger to the beginning of actual data collection. Usually, this will be a few milliseconds; however, it may be set longer to image the heart in a different phase of contraction, such as diastole (Figure 25).

Multislice acquisition order. Depending on the application, cardiac gated images may be acquired of a single slice at many time points (phases) in the cardiac cycle. If time-resolved imaging is not needed, the imaging time may instead be used to acquire multiple slices during the heart cycle. Each of the slices will be acquired at a different delay time from the R-wave to freeze the motion in each individual slice. This makes for an efficient use of the available imaging time at the expense of some misregistration between the slices as they are acquired at different stages of contraction.

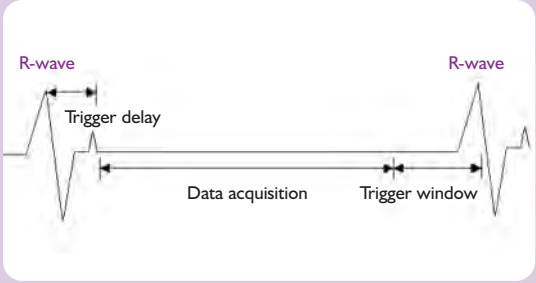


Figure 25. EKG gating. In cardiac gating, the patient's cardiac cycle is monitored. Once reliable monitoring by the MR system is established, the timing of the pulse sequence can be synchronized to the patient's cardiac cycle, greatly reducing artifacts from cardiac motion. In a typical gating sequence, the R-wave is detected and data acquistion begins after a time delay of a few milliseconds and continues for a predetermined amount of time. Once data acquisition is complete, the trigger window opens and the system begins sensing for the next R-wave and the cycle begins again.

Peripheral gating. Peripheral gating is another way to synchronize image acquisition with the heart cycle. In this case, the signal used for gating comes from a photosensor placed on the finger or toe. The sensor detects the pulsation of the blood flowing in the capillaries in the extremity, which changes the reflection of light back to the sensor. The resulting waveform is similar to an EKG waveform and is also suitable for generating cardiac triggers. It is important to recognize that the timing of the peak of this pulse is significantly delayed from the EKG R-wave by as much as 250 ms. This delay must be accounted for in the trigger delay if a specific phase of the cardiac cycle is to be acquired with peripheral gating. For this reason, peripheral gating is not recommended for cardiac imaging but as a method for reducing cardiac motion artifacts in other scans. Peripheral gating should be used for cardiac imaging only when direct cardiac gating methods fail.

Retrospective gating. Much like respiratory gating described above, respiratory cardiac gating is also possible by continuously acquiring image data and simultaneously recording EKG waveforms. As above, the data acquired are re-sorted into group acquisitions acquired at similar delays from the R-wave to create the appearance of a gated and motion-suppressed image.

Flow and Related Artifacts

At the most basic level, artifacts caused by flow of blood or other body fluids are just a specific case of general motion artifacts. However, they are generally given separate consideration. Flow differs in that the moving material follows a well-defined path (usually through a blood vessel), and the motion is consistently in one direction along that path, as compared to the periodic motion of the tissue in respiratory and cardiac motion or the short distance of motion or rotation in gross patient motion.

Vascular Pulsation

Vascular pulsation artifacts occur in larger vessels where the flow patterns are periodic rather than relatively constant and continuous. The aorta is the primary vessel that manifests a vascular pulsation artifact. Vascular pulsation can cause poor display of the vessel of interest and can generate artifacts that look like aortic dissection, or ghosts, that may appear to be pathologies in other areas of the image. Ghosting artifacts will always appear in the phase-encoding direction and will be more severe on short TE and gradient echo sequences. Typical areas where this type of ghosting is seen are on axial imaging where the cross-sectional slices are perpendicular to the flow of major vessels, such as the carotids

in axial imaging of the cervical spine or neck (Figures 26-27). Axial abdominal or lumbar spine anatomy can be obscured by the aorta or vena cava.

Cerebrospinal fluid (CSF) Pulsation

CSF pulsation artifact arises from cardiac pulsations transmitted to the subarachnoid space. The artifact mainly appears on images of the cervical spine. Ghosting, blurring, and areas of decreased intensity of CSF can be seen, especially on T2-weighted images (Figure 28). The techniques for suppressing pulsatile flow artifacts are similar to some of the general motion-suppression techniques.

Saturation. In using saturation pulses to suppress flow-related artifacts, the goal is to eliminate signal from blood or fluid that may flow into the slice during the acquisition. To ensure complete suppression, saturation pulses need to be applied on both sides of the imaged slice to eliminate signal from both arterial and venous inflowing blood. In most

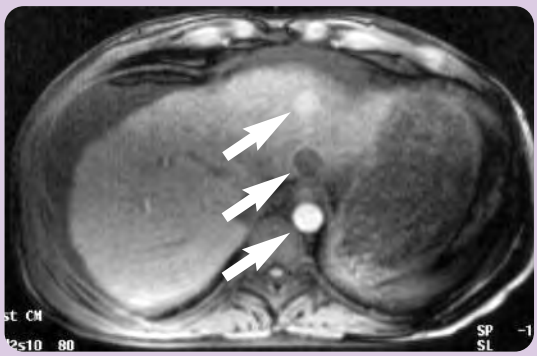


Figure 26. Axial T1 breath-hold image. Aortic vascular pulsation artifact (arrows).

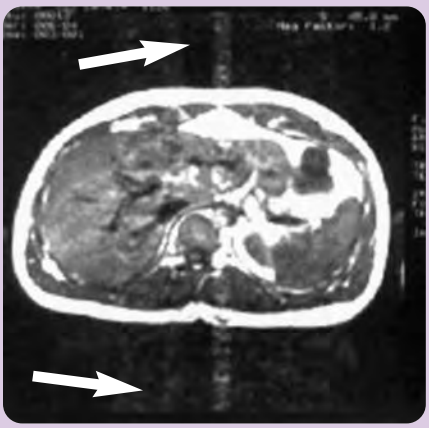


Figure 27. Artifact caused by pulsatile flow from the abdominal aorta (arrows). Presaturation pulses applied in the superior-inferior direction would greatly reduce this artifact from both the aorta and the inferior vena cava.

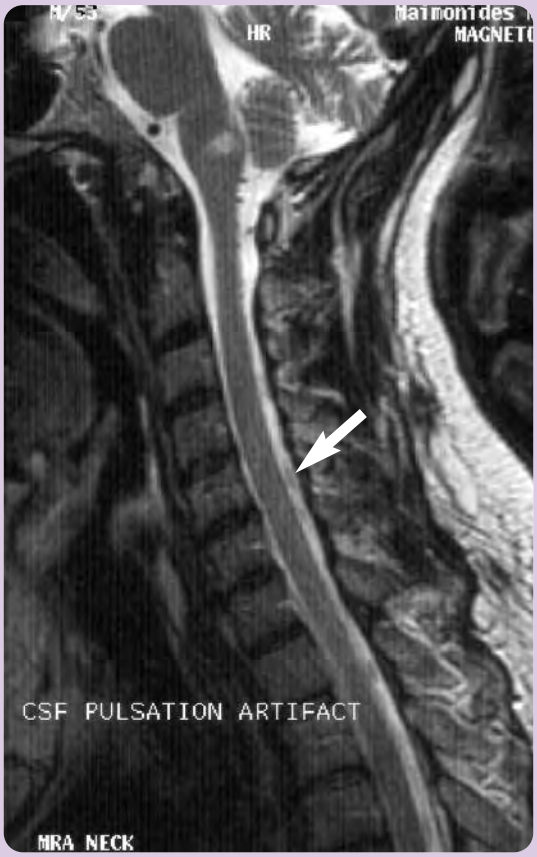


Figure 28. Sagittal T2-weighted image through the cervical spine. CSF pulsation artifact (arrow).

cases these pulses are placed "upstream" from the imaging volume. For example, to eliminate vascular flow from axial images, place a superior saturation pulse to saturate aortic flow. To eliminate flow from the vena cava, place an inferior saturation pulse.

Gradient moment nulling. This technique is the same as described in the section on general motion artifacts. In gradient moment nulling, the frequency and slice-select gradients are pulsed in positive and then negative directions while blood or CSF flows. The effects of the gradient pulses are to create a phase shift in flowing tissues, such as blood or CSF, and then to reverse that phase shift so the net phase shift at the end of the scan is zero. A net phase shift of zero equates to no motion artifact in the image. Gradient moment nulling works best with slow to intermediate velocity flow. The effectiveness on fast flowing blood is far less substantial. The extra gradient time used may require a longer TE in these cases.

Magnetic Susceptibility Artifact

Magnetic susceptibility refers to the ability of a material to be magnetized. Magnetic field gradients can be generated by the interfaces between air and tissues, as well as by the presence of metal and can significantly alter the magnetic field in the area around such a region. The change in magnetic field further causes the resonant frequency to be altered, leading to dephasing of spins. The net effect in regions of altered magnetic susceptibility is to cause dephasing in the region, yielding significant signal loss and dropouts. The field gradients that may be set up can also result in geometric distortions in the images.

Several factors influence the appearance and severity of susceptibility artifacts:

Field Strength

Susceptibility effects are a function of field strength. As magnetic field strength increases, susceptibility effects and their artifacts increase as well.

The increased susceptibility effect at higher field strengths is actually an advantage in some applications. *f*MRI relies on this effect. In BOLD imaging, susceptibility effects are used to determine what areas of the brain are activated when stimulated with some task. When areas of the brain are stimulated, minute changes in blood flow occur in those areas. Areas of the brain under stimulation have levels of oxyhemoglobin and deoxyhemoglobin that change as the stimulus is turned "on" and "off." This change, while very minute, causes corresponding small but detectable changes in the susceptibility of the brain tissue. The increased susceptibility effect makes BOLD imaging more sensitive to activation for such experiments.

Echo Time

Since susceptibility artifacts cause dephasing effects, longer echo times create more severe artifacts. Minimizing echo time will reduce the severity of the artifact but may not eliminate it completely.

GRE/Echo-planar vs. SE/FSE/TSE

Spin-echo–based sequences are less affected by susceptibility artifacts because the dephasing effects of the field changes are rephased in such sequences. By contrast, gradient echo (GRE) and echo-planar (EPI) sequences may experience large areas of signal loss and dropout from the field changes and dephasing effects. In severe instances, these sequences may be completely unusable.

Once again, the susceptibility effect can be useful under certain circumstances. The hemoglobin in blood contains enough iron to cause local susceptibility changes that can yield signal changes on gradient echo images, useful for detecting hemorrhage. Iron deposition in the liver can also be assessed in this way.

Metal

A major cause of susceptibility is metal from orthopedic implants and other sources. All metal should be removed from the patient where possible, including jewelry and certain kinds of makeup or hair products. When imaging near metal implants (after verifying their safety for scanning), FSE sequences with minimal echo spacing may be most suitable to acquire reasonable images around the metallic device (Figure 29).

Chemical Shift Artifacts

Protons that are part of fat molecules actually spin at a slightly slower frequency than protons that are part of water molecules. At 1.5 T, this difference amounts to 220 Hz, but it is less at lower field strengths. The localization of tissues along the frequency encoding direction uses the spin (precession) frequency to determine spatial position.

Chemical Shift Misregistration

The protons of fatty tissues precess at slower rates than protons of water-based tissues. The slower rate of precession of fat protons contained in the same voxel as water protons means that the signal from fat protons will be displayed in the image incorrectly and shifted relative to its correct position. The shift in fat signal creates either a bright or dark band at interfaces between fat and water. The fat signal shift can also cause structures to look thicker than normal, or the shifted fat signal may overlap important structures or abnor-

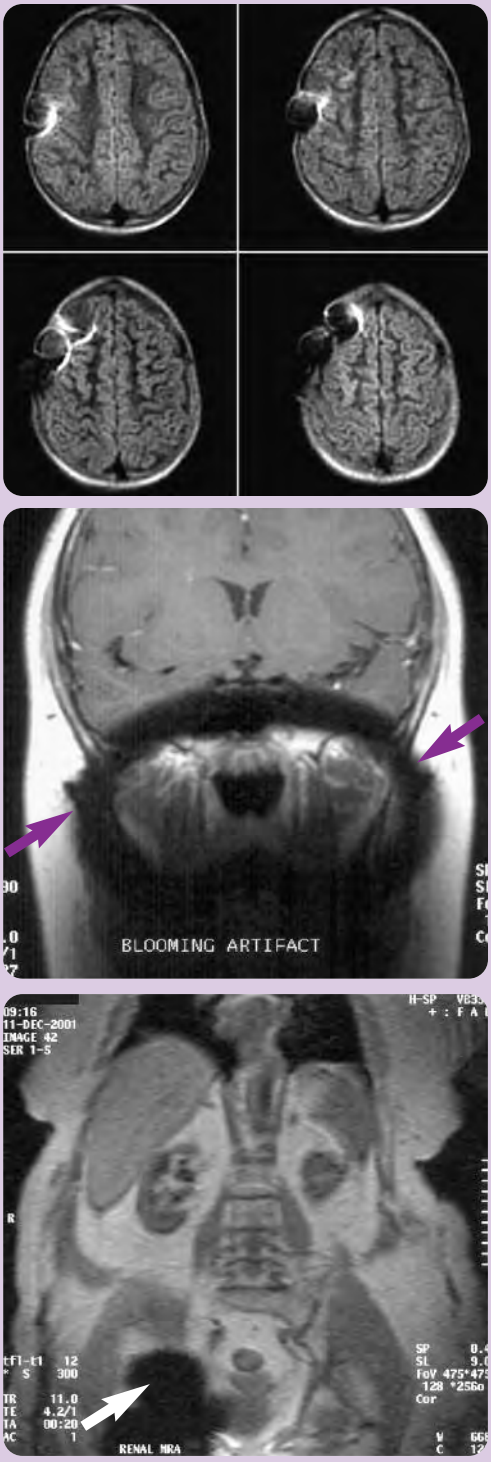


Figure 29a-c. (a) Susceptibility artifacts caused by metal. (b) Coronal MRI of the brain; artifact secondary to patient braces. (c) Coronal image shows artifact due to a right hip prosthesis.

malities in the signal, resulting in a nondiagnostic image (Figure 30).

The distance of the shift depends on receiver bandwidth, with lower bandwidths yielding greater shifts. Increasing the bandwidth can help reduce the magnitude of the effect, as can higher resolution (smaller vowel size). If the fat signal is not needed for diagnosis, fat suppression techniques can eliminate the signal and its obscuring effects. The amount for voxel shift can be calculated by the following:

- The chemical shift between fat and water at 1.5 T is 220 Hz.
    1. Calculate the number of Hz/voxel by dividing the number of frequency encoding steps by the receive bandwidth (RBW).
        - if a +/-16,000 Hz RBW is used, there are 32,000 Hz across the image.
    2. Calculate the chemical shift by dividing the total number of Hz by the number of frequency encoding steps (32,000 ÷ 256).
        - if there are 256 frequency encoding steps in the image, each voxel contains 125 Hz/voxel.
    3. Since the normal chemical shift of fat and water is 220 Hz at 1.5 T, calculate the actual shift per voxel by dividing 220 Hz by number of Hz/voxel (220 ÷ 125).
        - the chemical shift of the fat and water signals will be shifted in the image 1.76 voxels.
- If the RBW is changed from +/-16,000 Hz to +/- 32,000 Hz, calculation changes to 64,000 Hz across the image; therefore 250 Hz/voxel (and also the chemical shift) will be less than 1 voxel.

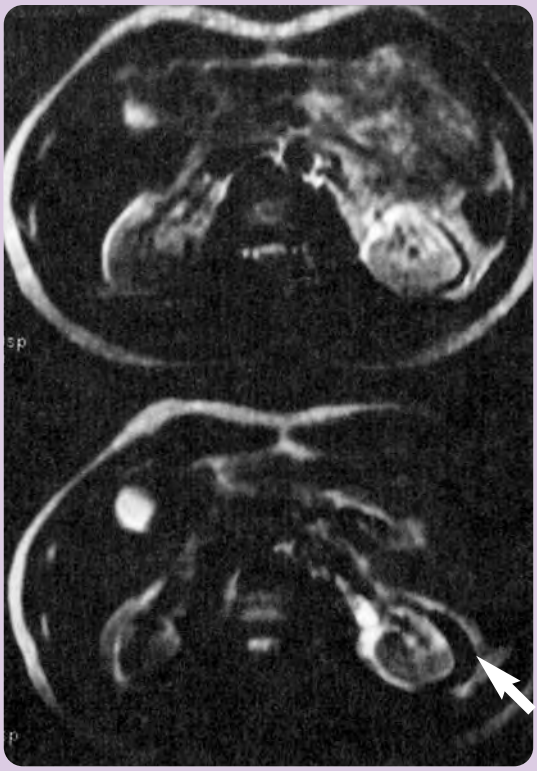


Figure 30. Chemical shift misregistration artifact (arrow).

Chemical Shift Phase Cancellation Artifact

In boundary areas between fat and water, there are inevitably voxels containing a mix of the two tissues. Because these protons spin at different rates, the relative phases of the two signal components will vary. For example, at 1.5 T with an echo time of about 4.2 ms, the components will be in-phase and will add constructively, yielding a bright signal. At an echo time of 2.1 ms, the two tissues are 180 out-of-phase and cancel each other out. The effect will be to create a dark band around fat/water interfaces where these two components are joined in the same voxel. The dark band can obscure important details at these types of interfaces (Figure 31).

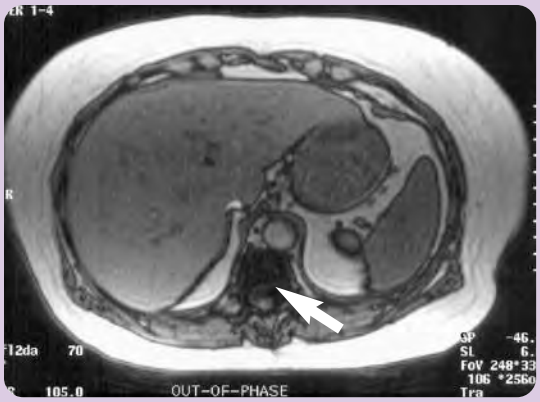


Figure 31. Axial T1 out-of-phase image through the liver. Chemical shift phase cancellation in the bone marrow (arrow).

Smaller voxel sizes, spin-echo–based sequences, and adjustment of echo time to an in-phase value are all methods of eliminating this artifact. There are applications where the out-of-phase images are useful, so this type of acquisition is not always an artifact.

Fat Saturation Artifacts

Some fat saturation techniques rely on the difference in resonant frequency between fat and water to selectively saturate fat or excite water. In areas of inhomogeneous field, the freqency shifts may be altered significantly enough to interfere with the fat suppression. In these cases, alternative fat suppression techniques, such as inversion recovery that relies on T1 rather than chemical shift, can improve the performance of fat suppression.

Artifacts Related to Slice/FOV/Readout

Incorrectly prescribed parameters describing the FOV of the acquisition can cause significant artifacts from aliasing or "wrap." These artifacts can be readily corrected with more suitable geometric parameters for the acquisition.

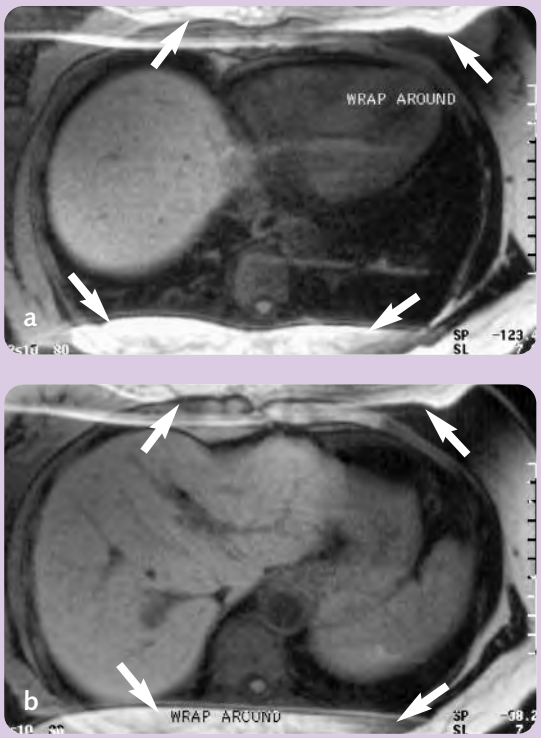


Figure 32a-b. Axial T1 breath-hold images showing aliasing (wraparound) artifact.

Aliasing Artifacts

Aliasing artifacts, also known as wraparound artifact or just wrap, are caused by a FOV too small to contain the object being imaged. The signal from outside the FOV is not uniquely encoded by the phase or frequency encoding gradients, so it is mismapped to the opposite side of the image (Figure 32). Aliasing artifact can occur along either phase or frequency directions or even the second phase-encoding direction in a 3D scan.

PHASE WRAP

Phase wrap refers to aliasing in the phase-encoding direction, the most commonly seen form of this artifact (Figure 33). As mentioned above, it is caused by insufficient FOV in the phase-encoding direction. The "no phase

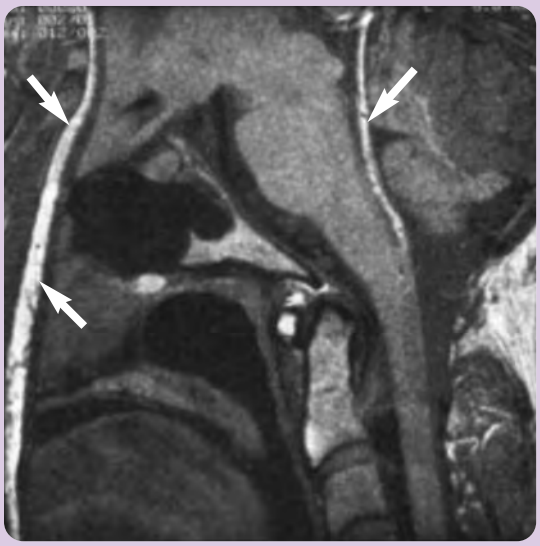


Figure 33. MRI image of the brain shows phase wraparound or "aliasing" artifact (arrows). Note in the small FOV sagittal head how the posterior part of the head "wraps" to the front and the anterior part wraps to the back. This is because the FOV used is smaller than the area being excited.

wrap" option is used to eliminate this artifact. The no phase wrap option simply increases the FOV or oversamples the phase-encoding direction to correctly encode the area outside the area of interest. To prevent the associated doubling of scan time, the number of signal averages (NEX or NSA) is reduced by half. To prevent a reduction in SNR, the FOV in the phase direction is doubled. There is no loss in SNR and no increase in scan time. During the reconstruction phase of the image, the extra phase-encoding steps are eliminated, and the image is constructed and displayed at its prescribed FOV with no aliasing. While this may seem like a "no penalty" method of elimination, the technologist must remember that one of the most effective ways to reduce motion artifacts is to increase the number of signal averages. Using the no phase wrap tool reduces signal averaging by half, thereby increasing the risk of motion artifacts.

FREQUENCY WRAP

Frequency wrap can also occur because the FOV is too small in the frequency encoding direction. In this case, the sampling frequency can be increased. On many systems, the sampling rate in the frequency direction is automatically set high enough to assure that only signal from the prescribed FOV is retained. The signal from anatomy outside the FOV is filtered out in the RF signal chain.

Cross-talk / Slice Overlap

The RF pulse used to excite a slice is of limited duration. As a result, the profile of the slice excitation may slightly excite some small amount of tissue outside the prescribed slice thickness, or it will fail to excite some of the tissue within the slice. This is particularly true when the profile of the slice excitation pulse is not "square." Figure 34 illustrates the difference between square and nonsquare RF excitation pulses. Certain pulse sequences have RF excitation pulses that are more square and therefore less "lobular." For example, FSE sequences typically use RF excitation pulses that are far more square than standard spin-echo sequences. Moreover, older MR systems may use RF excitation pulses that are less square than later MR systems. It is important for the technologist to know which pulse sequences are more suspectible to cross-talk and which are not. The tissue on the boundary between slices may actually be excited in more than one slice excitation and become saturated, reducing the signal within the slice. Cross-talk is visualized by a characteristic light-dark-light-dark appearance to the images. One available option is to include a gap of 10% or more between the slices to skip over the saturated tissues. In a multislice acquisition, the slices can be ordered such

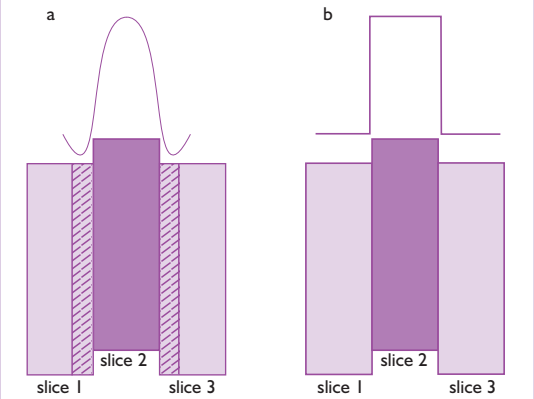


Figure 34a-b. (a) An RF pulse with non-square lobes that can inadvertently excite adjacent slices and cause the slices to be partially saturated with RF. As a result, slices 1 and 3 will be darker than slice 2. (b) An RF pulse with very square lobes, ensuring that each slice "sees" the same amount of RF, and consequently the same brightness and contrast.

that adjacent slices are acquired far apart in time to permit the out-of-slice tissues to recover their magnetization. In the extreme case, the acquisition can be separated into two blocks, with the first block collecting odd numbered slices and the second block collecting even numbered slices. With continued improvements in digital RF transmission, modern MR scanners often use RF pulses that are sharp enough to virtually eliminate cross-talk. The technologist should take care to know what pulse sequences use the sharpest RF pulse profiles. For example, FSE pulse profiles are sharp enough that a slice gap is not necessary.

Partial Volume Averaging

The limited slice resolution that can be achieved in a 2D scan means that often a mix of tissue types is represented within a single voxel. Accordingly, the mix of tissue types may cause the intensity of a region to comprise a combination of the tissues present, and therefore the region will not resemble any single tissue type. This effect can even resemble pathology; it can also prevent visualization of small or low contrast abnormalities and make the image look blurry.

Partial volume effects increase with increasing slice thickness, so minimizing the thickness or performing a 3D examination with higher slice direction resolution can alleviate the problem. Thinner slices in 2D will reduce SNR, however, and 3D acquisitions may take more imaging time. Acquisition of images in different (orthogonal) planes through the affected area can also help determine whether the intensity anomaly is due to partial volume averaging or true pathology.

Truncation (Gibbs or Ringing) Artifact

Truncation artifact appears near sudden transitions between high and low intensity (strong edges) and creates a low intensity band within the high intensity region (Figure 35). The problem arises from insufficient sam-

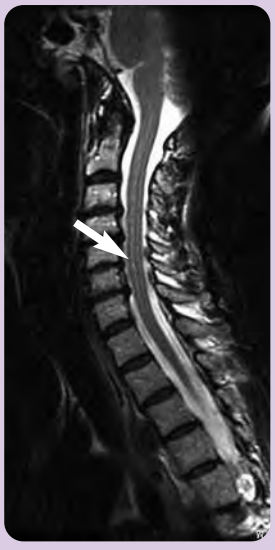


Figure 35. This image displays the classic truncation artifact (also known as Gibbs artifact). Note the lines of high and low intensity in the cord. Increasing the matrix number will reduce or prevent this artifact.

pling, usually in the phase-encode direction. The number of phase-encode samples is too small to accurately represent the rapid intensity differences at the interface. The artifact can be eliminated by increasing sampling in the phase-encoding direction, at the expense of a longer scan time.

Sequence-specific Artifacts

Banding Artifact (SSFP/TrueFISP/Fiesta/bFFE)

Steady state free precession (SSFP) imaging (also known by any of the names above) has become popular for its high contrast and rapid image acquisitions, especially for use in cardiac imaging. Its limited use in years past was due to its high sensitivity to off-resonance and field inhomogeneities. Off-resonance behavior in SSFP results in dark bands in the image where the resonant frequency differs by 1/TR between acquisitions. These dark bands are particularly marked in regions of high susceptibility, where there are significant changes in resonant frequency. Minimizing banding artifacts requires accurate center frequency selection and shimming. The TR should also be set at the minimum to make the banding frequencies large.

EPI Phase Misalignment

In echo planar imaging, many (or even all) phase-encode lines are acquired after a single excitation. To accomplish acquisition as rapidly as possible, the phase-encodes are acquired in a "zig-zag" fashion, alternating directions through k-space. Because the alternating lines, that is, the even and odd lines, are acquired in opposite directions, there may be slight timing differences in the even and odd lines with respect to when they cross the k-space origin. The slight timing differences cause phase differences in the even and odd images. Effectively, two slightly different images are formed, each with one half of the FOV. When the timing is perfect, the aliased components will cancel out, leaving a single full FOV image with no artifact. If there are timing and phase mismatches, however, the aliased components will not completely cancel and ghost artifacts will remain, offset by half the FOV in the phase-encoding direction (Figure 36). The phase misalignment may be a result of either the timing of the gradient pulses or eddy currents that cause the actual gradient waveform to be distorted from the desired waveform. In either case, accurate system calibration may be needed to compensate for the artifact-producing effect.

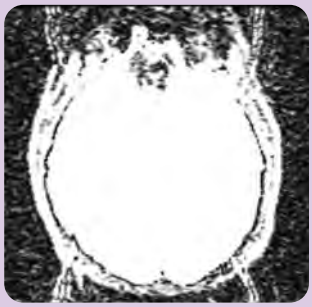


Figure 36. This brain image displays the characteristic misalignment artifact possible in EPI. The image contrast and brightness is adjusted in order to fully display the artifact.

Spiral Imaging Artifacts

Spiral imaging is a rapid acquisition technique that can be used as an alternative to EPI. In spiral imaging, data is not acquired along straight phase-encoded lines but obtained in a spiral fashion starting at center of k-space and moving towards the edges. This curved path through k-space results in different artifact characteristics. Spiral acquisitions are more resistant to flow and motion artifacts, and when these artifacts do occur, the result is blurring instead of ghosting. One drawback of spiral imaging is that there is no distinct frequency and phase-encoding direction. There-

fore, the FOV cannot be limited along a frequency encoding direction by a change in sampling bandwidth. Spiral acquisitions are therefore more prone to artifacts resulting from insufficient FOV. These artifacts have a characteristic "wisp" appearance forming a circular pattern around the outside of the image. Correction for this artifact is done by increasing the overall FOV at the potential expense of a longer scan time.

Artifacts Related to Hardware and RF

The artifacts noted below are caused by problems with the RF subsystem of the scanner and other hardware failures. Most of these can be corrected only by repair of the system component causing the problem.

Image Warping

Image warping refers to geometric distortions of the image that cause the image to appear "bent" or "warped" and distorted. Geometric distortions cause problems with image interpretation and quantitative measurements.

GRADIENT NONLINEARITY

Imperfections in the gradient system performance can be one cause of a gradient nonlinearity artifact. Gradients are used for spatial localization of the tissues being imaged, and the reconstruction algorithms assume that the gradient is uniform over the entire image volume. If the gradient is not uniform (which particularly occurs around the edges of the image), then voxels will be misregistered and distorted, depending on the degree of gradient nonuniformity. These artifacts can sometimes be corrected if the characteristics of the gradients are measured and a suitable correction process is available.

MAGNETIC FIELD INHOMOGENEITY

Image warping effects can also be seen when there are significant degrees of inhomogeneity in the main magnetic field. One way to see the effect of inhomogeneities on image warping is to note that variations in the magnetic field can be approximated by a sum of a constant offset (a shift in the center frequency), a component that varies with position (which acts like a gradient), and additional components that act to distort this "gradient." Therefore, the presence of these nonlinear variations in the magnetic field can have the same effect as gradient nonuniformities, leading to geometric distortions and warping.

Zipper Artifact

Zipper artifacts are caused by the RF system receiving signals from outside of the scan room (Figure 37). The RF acquisition system records all signals received in the frequency range of the acquisition, regardless of their

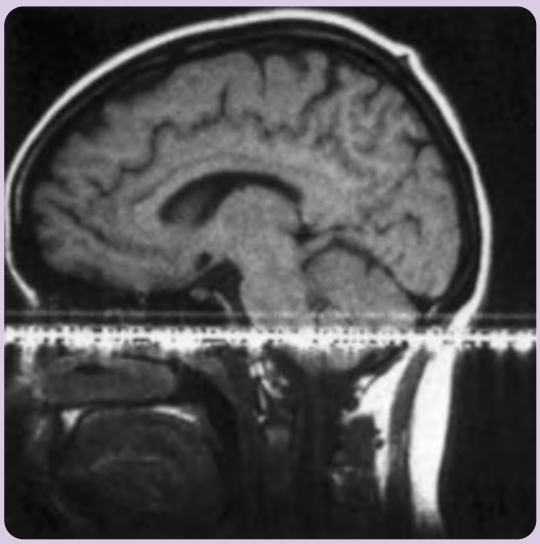


Figure 37. Zipper artifact is caused by an RF leak.

origin. If RF signals are generated by other devices—or even received from broadcast sources—and the scan room is not properly shielded from these signals, they will generate zipper artifacts in the image. The simplest explanation may be that the scan room door remained open during the examination, defeating the RF shielding in the door. If this is not the problem, there may be failures in the overall room RF shielding that need to be identified.

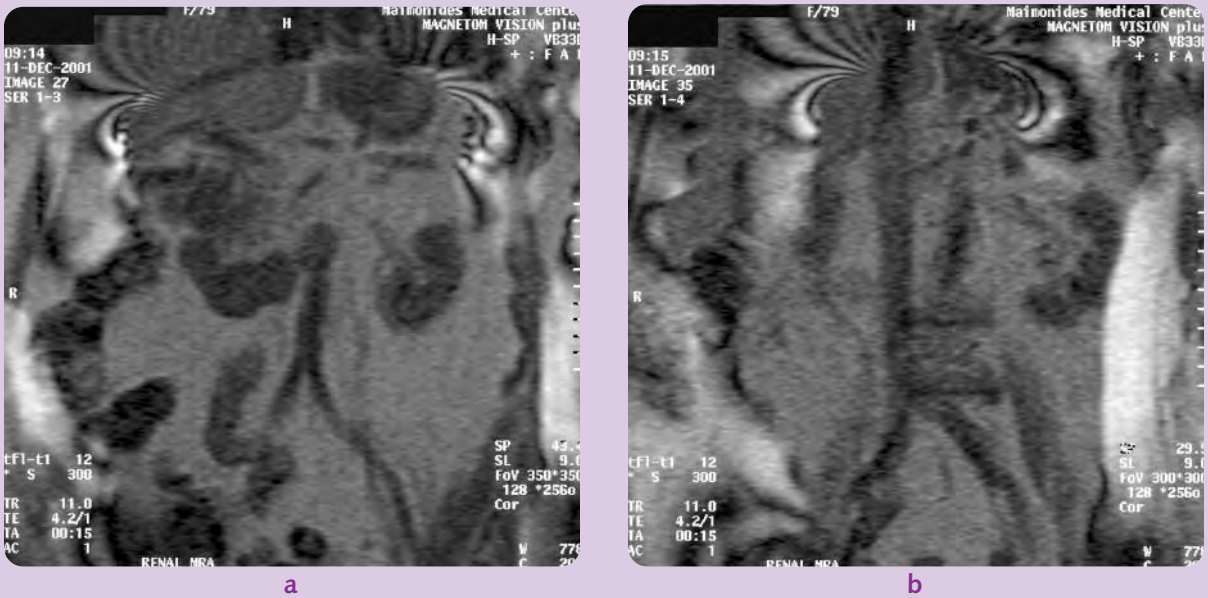
EXTERNAL RF INTERFERENCE

The zipper artifact is related to external RF interference problems. RF-related artifacts may take other forms as well, such as solid lines running through the image. These types of artifacts are generally displayed in the frequency encoding direction. Apart from checking the open door problem, an engineering diagnosis and treatment are usually necessary.

Zebra Stripes Artifact

There are actually two artifacts occasionally referred to by the term "zebra stripes" artifact. The first can result when a portion of the k-space acquisition is improperly set to zero. This might be due to either a computer hardware failure or an incomplete acquisition of some kind. The abrupt change in signal values creates stripes in the image.

The other type of artifact creates stripes that are not necessarily parallel to each other. The artifact may occur when the patient touches the coil or as a result of cancellations due to phase wrap. It is most notable on gradient echo scans where there are susceptibility changes at air tissue interfaces at the margins of the image. Adjusting the coil position or setting no phase wrap may reduce this version of the zebra stripes artifact (Figure 38).



a b

Figure 38a-b. Coronal gradient echo scout images demonstrating zebra stripe artifact.

Spike Noise Artifact and Corduroy Artifact

Spike noise occurs when one or more points in k-space contain erroneous data and have much higher amplitudes than the rest of k-space. The resulting image has a marked striped appearance much like corduroy fabric with the intensity and spacing of the stripes depending on the precise location in k-space where the noise occurs. In theory it may be possible to process the image to remove most of the artifact, but in practice a repeat of the scan is the most effective course of action.

Coil-related artifacts

Artifacts Related to Parallel Imaging

Image artifacts related to parallel imaging result mostly from an inaccurate characterization of the coil sensitivities. Parallel acquisitions achieve their acceleration in imaging time by sampling fewer than the required number of phase-encode lines and using the coil sensitivity information to fill in the missing information that the phase encoding lines would typically provide. If the coil sensitivity information is not accurate, the parallel imaging reconstruction will not be able to resolve the aliased (undersampled) images acquired by each individual coil. Also, noise from the image may be strongly amplified by the incorrect calibration. The result is ghosting or the appearance of wrap, along with poor SNR. Correction of these problems involves orienting the coil to properly align with the phase-encoding direction or reducing the acceleration factor and acquiring more data (Figure 39).

Shading Artifact

Birdcage and saddle RF coil designs are supposed to have a nearly uniform RF response. Malfunctions in the coil can result in an inhomogeneous transmission or reception of the

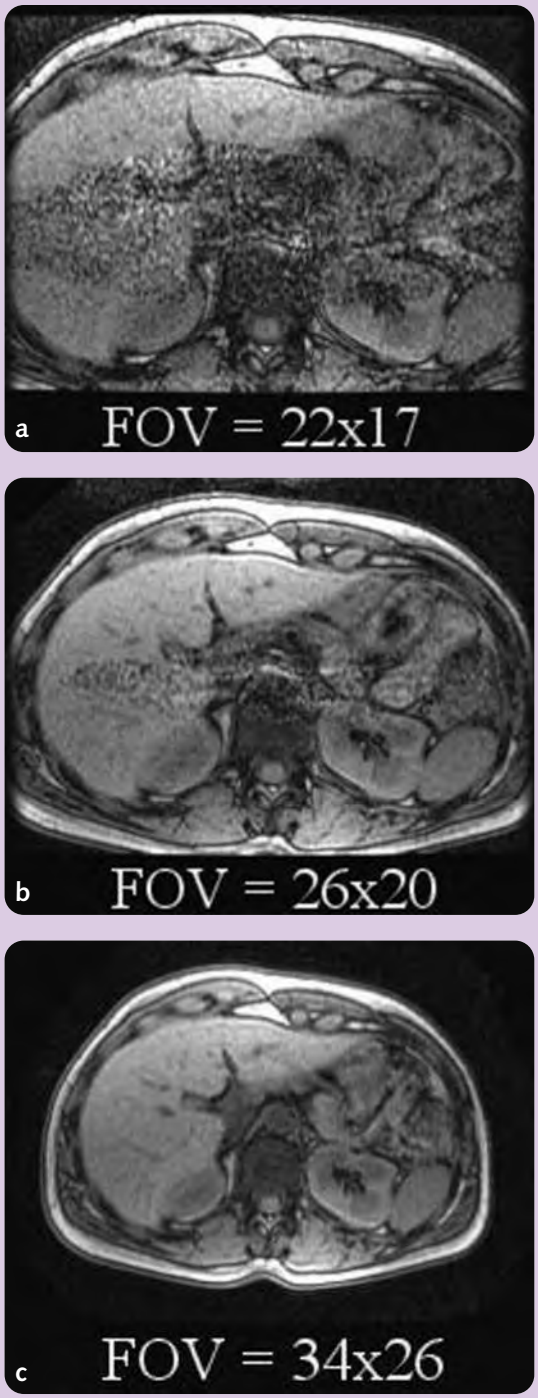


Figure 39a-c. (a-b) The FOV in these images of the liver is too small (22x17 and 26x20). Consequently a noise band can be seen, and the result is a parallel imaging wraparound effect. (c) Increasing the FOV reduces, and then eliminates (34x26), the artifact.

MR signal. The result is that for an excitation pulse, the flip angle actually generated by the coil may vary over the volume of the coil, resulting in corresponding variations in the signal intensity causing areas of the image to be darker or brighter than what would be correctly displayed in the image, particularly in areas of anatomy close to the RF coil itself.

SUMMARY

MR imaging artifacts are inevitable, and their causes are widely varied. The MR technologist has the responsibility of being aware of a wide range of common and uncommon image artifacts that can arise in the course of routine scanning. A thorough understanding of artifacts, their causes, and the means that can be used to correct them are important to assure that high quality data of diagnostic quality will be consistently obtained. Artifacts that arise from physiologic causes (breathing motion, flow motion) can always be prevented, eliminated, or reduced by using tools provided on every MR system. The technologist must understand when and how to use these tools to best effect. Artifacts arising from RF leaks, mistuned surface coils, or poor homogeneity, ie, MR system–related artifacts, are just as challenging for the technologist as those with physiologic causes. However, system-related artifacts are unpredictable and have no technologist-controlled solutions built into the system. The technologist must be able to identify the difference between the two types of artifacts in order to quickly determine the best method of producing high quality diagnostic images in spite of artifacts.

Figures

1, 9, 11, 12, 15, 21	Courtesy of Fairfax Radiological Consultants
2, 3, 10	Courtesy of Mark Flyer, MD, Maimonides Medical Center
4-8, 20	Tom Schrack, BS, ARMRIT
13, 14	Courtesy of Vital Images, Inc.
16-19	Courtesy of GE Healthcare
25, 34	Tom Schrack, BS, ARMRIT
22, 23, 24, 26, 28, 29b-c, 31, 32, 38	Courtesy of Mark Flyer, MD, Maimonides Medical Center
29a, 33, 37, 39	Courtesy of GE Healthcare
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27, 35, 36	Courtesy of Fairfax Radiological Consultants

Tables

1	Tom Schrack, BS, ARMRIT

Additional Reading

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ABBREVIATIONS OF TERMS

ADC	apparent diffusion coefficients
AUC	area under the curve
B_1	the RF transmission field
B_0	the main magnetic field
bFFE	balanced fast field echo
BOLD	blood oxygen level dependent
cm	centimeter
CNS	central nervous system
CSF	cerebrospinal fluid
CT	computed tomography
CTL	cervical-thoracic-lumbar
dB/dt	time-varying B fields (gradient-altered B_0)
dB	change in the main magnetic field (B_0) –OR– decibel
dBA	decibel attenuation
DSC	dynamic susceptibility contrast
DTI	diffusion tensor imaging
eADC	exponential ADC map
EKG	electrocardiogram
EPI	echo-planar imaging
FLAIR	fluid attenuated inversion recovery
***f*MRI**	functional magnetic resonance imaging
FNH	focal nodular hyperplasia
FS	fat suppressed
FSE	fast spin-echo
FOV	field of view
G	Gauss
g-factor	geometry factor
GRE	gradient echo
Hz	hertz
IR	inversion recovery
IV	intravenous
kHz	kiloHertz (1,000 hertz)
KVO	keep vein open
mHz	megaHertz (1,000,000 hertz)
min	minute(s)
MinIP	minimum intensity projection
MIP	maximum intensity projection
mm	millimeter
mmol/kg	millimole per kilogram
mOsm/kg	milliosmoles per kilogram
mOsm/L	milliosmoles per Liter
MPR	multiplanar reconstruction
MRI	magnetic resonance imaging
MRA	magnetic resonance angiography
MRCP	magnetic resonance cholangiopancreatagraphy
MRS	magnetic resonance spectroscopy
MRV	magnetic resonance venography
mT	milliTesla
mT/m	milliTesla per meter
mT/m/s	milliTesla per meter per second
MTT	mean transit time
NAA	N-acetyl aspartate
NEX	number of excitations (also NSA)
nm	nanometer
NSA	number of signal averages (also NEX)
OCP	o-cresolpthalein complexone technique
Osm/kg	osmoles per kilogram
PACS	picture archiving communication system
PNS	peripheral nerve stimulation
r_1	T_1 recovery
r_2	T_2 decay
RBW	receive bandwidth
rCBV	relative cerebral blood volume

RES	reticuloendothelial system
RF	radiofrequency
ROI	region of interest
SAR	specific absorption rate
SE	spin-echo
SNR	signal-to-noise ratio
SPGR	spoiled gradient echo
SPIO	superparamagnetic iron oxide
SSFP	steady state free precession
STIR	short tau inversion recovery
T	Tesla
T1	time for 63% of a tissue's longitudinal magnetization to recovery
T2	time for 63% of a tissue's transverse magnetization to decay
TE	echo time
T/m/s	tesla per meter per second
TOF	time of flight
TR	time to recovery –OR– repetition time
TSE	turbo spin-echo (Siemens Medical Systems term for fast spin-echo)
UAE	uterine artery embolization
USPIO	ultrasmall superparamagnetic iron oxide
W/kg	watts/kilogram

GLOSSARY

absolute zero
the theoretical point at which all molecular motion stops, measured as 0^0 Kelvin

aliasing
a common artifact caused when the FOV selected is smaller than the area of tissue excited; also known as "wrap-around"

algorithm[1]
a step-by-step method of solving a problem or making decisions, as in making a diagnosis. An established mechanical procedure for solving certain mathematical problems

artifact[1]
in radiology, a substance or structure not naturally present in living tissue, but of which an authentic image appears in a radiograph

BOLD (blood oxygen level dependent) imaging
an imaging technique based on the EPI pulse sequence in which the patient is given mental tasks to perform while undergoing MR examination. Increases in blood flow to affected areas of the brain are postprocessed into functional maps indicating brain activity.

bore
the tubular portion of the magnet in which the patient is placed

contraindication
an absolute reason or cause not to proceed with a diagnostic examination or procedure. In MRI it is typically a patient condition, such as an embedded cardiac non-MR compatible pacemaker, that prohibits the patient from undergoing MR examination.

contrast
differences in signal intensity between two adjacent areas on an MR image

cryogen
super-cooled liquid gas used to cool a given material to near absolute zero, thus becoming superconductive; in MR, liquid helium

cryostat
the vessel in which the magnet is immersed in liquid helium

diffusion imaging
an imaging technique usually based on the EPI pulse sequence; it is used to indicate the amount of water absorption across a brain cell membrane

duty cycle
the amount of time (stated in percentage) that an MR system gradients can be "active" before they must be turned off in order to cool; most MR manufacturers report a duty cycle of 100%, indicating the gradients can run at full power continuously

echo planar imaging (EPI)
a very fast pulse sequence characterized by rapid oscillation of the frequency encoding gradient to create an echo train; EPI fills k-space quickly

echo spacing
the time from the peak of one echo in an echo train to the next

echo train
the series of echoes created by a FSE or EPI pulse sequence

electromagnetic spectrum
continuous series of different types of electromagnetic radiation, ordered according to wave-length of frequency

fast spin-echo (FSE)
a rapid pulse sequence characterized by a series of 180^0 RF echo pulses used to create numerous echoes within a single TR, thus filling k-space more quickly

fat suppression
any of the methods used to reduce signal from fat on an MR image

ferromagnetic
materials that react to a magnetic force; all iron and some stainless steel are ferromagnetic

field of view (FOV)
an area of tissue or anatomy to be imaged in an MRI scan

fluid attenuated inversion recovery (FLAIR)
inversion recovery-based pulse sequence that utilizes a relative long TI in order to suppress signal from a long T2 tissue while maintaining heavy T2-weighting throughout the rest of the image

flip angle
the rotation of the amount of RF energy used to excite some portion of protons in the longitudinal plane into the transverse plane

frequency
cycles per unit time; usually measured as cycles per second, or hertz (Hz)

frequency encoding
generation of frequency differences along a particular direction of a tissue slice for use in spatial localization of MR signal

fringe field
the extended magnetic field generated by an MRI magnet system that is outside the magnet bore, or scan area. The fringe field decreases in strength as the distance from the magnet increases

functional MRI (*f*MRI)
any of the imaging techniques that demonstrate function of anatomical structure; examples include spectroscopy, dynamic liver imaging, cardiac imaging, and BOLD imaging

Gauss[1]
[Karl Friedrich *Gauss,* Geman mathematician and physicist, 1777-1855] the cgs unit of magnetic flux density, equal to 10^{-4} tesla. Symbol, G

gradient (magnetic field)
a magnetic field that changes in strength along a given direction

gradient amplitude
the degree to which a gradient field can vary from zero to the peak maximum point; typically measured in milliTesla/meter (mT/m)

gradient coil
one of six coils (3 pair) placed in the three orthogonal planes (denoted X,Y, and Z) that generate small magnetic fields along the plane of the main magnetic field; it is used for slice selection, phase and frequency encoding

gradient echo
pulse sequence characterized usually by a $<90^0$ RF excitation pulse and an echo generated by gradient reversal instead of a 180^0 RF echo pulse

gradient moment nulling
a method to reduce flow artifact in which one, or all three, pair of gradients are pulsed to dephase and rephase spins that flow along the axis of that gradient

homogeneity
the uniformity of any field; in MRI it is the uniformity of the B_0 field

inhomogeneity
absence of homogeneity or uniformity; inhomogeneity in a magnetic field may occur as one area of the field deviates from the average magnetic field strength

inversion recovery
pulse sequence characterized by an initial 180^0 RF inversion pulse

k-space
the domain in which the information from each phase-encoding step is placed during a pulse sequence. Each "filled in" line of k-space corresponds to each phase-encoding step; once the required amount of k-space is filled, image reconstruction can begin

Larmor frequency
the frequency at which magnetic resonance is produced in a sample of hydrogen nuclei, or other types of nuclei used in MRI

ligand[1]
a molecule that donates or accepts a pair of electrons to form a coordinate covalent bond with the central metal atom of a coordination complex

magnetic moment
the net magnetic properties of an object or particle (such as a magnetic dipole)

magnetic shielding
metal surrounding an MR magnet used to contain the main magnetic field fringe field within acceptable limits; magnetic shielding can be passive (steel lined walls) or active (built into the system)

magnetic susceptibility
the degree to which a tissue can become magnetized

maximum intensity projection (MIP)
a ray tracing algorithm where a ray goes through a designated imaging block or volume. Signal intensity is designated based on nearness to the observer

motion artifact
an artifact or signal not naturally present in living tissue, but which appears on MRI film due to movement of muscle or fluid or motion of any body part

multiplanar reconstruction (MPR)
two-dimensional views of a single voxel thickness of vascular structures reconstructed from three-dimensional or multi-slice images

magnetic resonance angiography (MRA)[1]
a form of magnetic resonance imaging used to study blood vessels and blood flow, particularly for detection of abnormalities in the arteries and veins throughout the body

magnetic resonance imaging (MRI)
a method of visualizing soft tissues of the body by applying an external magnetic field that makes it possible to distinguish between hydrogen atoms in different environments

magnetic resonance spectroscopy (MRS)
MR examination in which the data collected are not reconstructed into an image, but into a spectrum of signals based on metabolite presence within the tissue being examined; a type of *f*MRI

nanometer (nm)[1]
a unit of linear measure equal to one-billionth of a meter, 10^{9} meter

number of excitations (NEX)
the number of cycles of completed k-space filling; also known as number of signal averages (NSA)

number of signal averages (NSA)
see number of excitations

paramagnetic
a substance with magnetic properties that may significantly reduce T1 and T2 relaxation times in MRI

perfusion imaging
an imaging technique based on the EPI pulse sequence; gadolinium contrast is used to indicate the amount of blood perfusion across a brain cell membrane

peripheral nerve stimulation (PNS)
activation of a peripheral nerve fiber(s) caused by rapidly switching gradient fields; PNS is not a patient safety concern but a potential patient comfort concern

phase
particular stage or point of advancement in a cycle

phased array coil
a type of surface coil composed of several coils and receivers that are linked together. The signals from each of the coils and receivers are subsequently united to form an image with good SNR.

phase encoding
generation of phase differences along a particular direction of a tissue slice for use in spatial localization of MR signal

pixel
picture element; the smallest discrete part of a digital image display

pulse sequence
a series of events for exciting protons and detecting signals during the MR examination; every pulse sequence includes slice excitation, echo generation, and phase and frequency encoding

quench
sudden and massive expansion of liquid helium into gaseous helium due to an increase of heat from a loss of superconductivity of the magnet

R-R interval
the period of time between each R-wave; one cardiac cycle

R-wave[1]
the initial upward deflection of the QRS complex, following the Q-wave in the normal electrocardiogram

radiofrequency (RF)[1]
the range of frequencies of electromagnetic radiation between 10 kilohertz and 100 gigahertz that is used for radio communication

ramp time
the minimum time required for a gradient field to go from its peak maximum point to its peak minimum point; measured in microseconds

ramp up/ ramp down
the controlled process of bringing the magnet to maximum field or reducing the magnet from its maximum field

receive bandwidth (RBW)
the range of frequencies collected during the frequency encoding portion of the pulse sequence

region of interest (ROI)
a specific defined area, ie, fluid or a portion of an organ or tumor, where a relative signal intensity measurement can be obtained

RF shielding
metal used to prevent stray RF frequencies from entering the magnet room during an MR exam, typically made of copper

rise time
the minimum time required for a gradient field to go from zero to its required maximum; measured in microseconds

resonance
state of a system through which energy may be transferred to another system with the same preferred or resonant frequency; characterized by absorption and dissipation of energy through resonant oscillation

saturation
lack of signal due to overexcitation of protons

signal-to-noise ratio (SNR)
amount of true signal relative to the amount of random background signal (noise) on an image

slew rate
describes overall gradient performance as a function of gradient amplitude and gradient rise time. Slew rate is derived by dividing the amplitude by the rise time and typically is described in units of T/m/sec.

specific absorption rate (SAR)
the FDA-regulated amount of RF heat energy that a patient can absorb during an MRI exam, measured in Watts/kilogram

spin-echo
basic pulse sequence of MR imaging using a 90^0 RF excitation pulse and a 180^0 RF echo pulse

static magnetic field
the large main magnet field generated by the magnet to place the protons into the longitudinal plane prior to the MRI pulse sequence; also known as the B_0 field

short tau inversion recovery (STIR)
inversion recovery-based pulse sequence that utilizes a relatively short TI time in order to obtain heavy fat suppression; the TI is selected based on T1 recovery time of fat

superconductivity
the state where all molecular motion becomes so restricted, due to a lack of heat, that electricity can flow through a conductor without resistance. Materials such as a magnet coil become superconductive after being immersed in liquid helium and reaching a temperature of 4^0 Kelvin

surface coil
specialized antenna for transmitting and/or receiving RF energy to or from the patient during a pulse sequence

T1
the amount of time for 63% of a tissue's protons to recover to longitudinal magnetization

T1-weighting
generation of MR images under conditions that highlight differences in T1 between tissues

T2
the amount of time for 63% of a tissue's protons to decay in the transverse plane

T2-weighting
generation of MR images under conditions that highlight differences in T2 between tissues

Tesla[1]
the SI unit of magnetic flux density, calculated as webers per square meter. It replaces the gauss.

time to echo (TE)
the amount of time selected by the technologist to allow for T2 decay of excited protons; also the amount of time from the beginning of initial slice excitation pulse and the generated echo signal

time-of-flight imaging (TOF)
gradient echo-based pulse sequence that uses flow-related enhancement to greatly increase the contrast between flowing blood and stationary tissue; performed in either a 2D or 3D acquisition, the images are postprocessed into a maximum intensity pixel projection (MIP) to display like an MR angiogram

time to inversion (TI)
the time allowed from the initial 180 degree excitation pulse until the 90 degree pulse in an inversion recovery pulse sequence; TI determines the amount of T1 recovery time for a given tissue

torque[1]
a rotatory force causing a part of a structure to twist about an axis

time to recovery (TR)
the amount of time selected by the technologist to allow for T1 recovery of the excited protons; also the time from the beginning of one pulse sequence to the beginning of the next

trigger window
the time delay before each R-wave

vector
mathematical quantity representing both magnitude and direction, symbolized by an arrow

voxel
a three-dimensional volume of tissue corresponding to a pixel on an MR image, a "volume element"

Items marked[1]**—**from Dorland's Illustrated Medical Dictionary, 29th Edition, 2000, Philadelphia, PA; W.B. Saunders Co. Used with permission.