MR Imaging Artifacts and Parallel Imaging Techniques with Calibration Scanning: A New Twist on Old Problems[1]

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Abbreviations: FOV = field of view, RF = radiofrequency, SENSE = sensitivity encoding, 3D = three-dimensional, 2D = two-dimensional

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ONLINE-ONLY SA-CME LEARNING OBJECTIVES

After completing this journal-based SA-CME activity, participants will be able to:

- Discuss how parallel imaging techniques that use calibration images alter the appearance of major artifacts at conventional MR imaging.
- Describe the origin of major artifacts unique to parallel MR imaging.
- Identify methods to correct artifacts at parallel MR imaging.

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TEACHING POINTS

See last page

The application of parallel magnetic resonance (MR) imaging is increasing as clinicians continue to strive for improved spatial and temporal resolution, benefits that arise from the use of fewer phase encodings during imaging. To reconstruct images, extra information is needed to map the spatial sensitivity of each coil element, which may be accomplished by acquiring a calibration image in one common implementation of parallel MR imaging. Although obtaining a quick calibration image is an efficient method for gathering this information, corruption of the image or disharmony with subsequent images may lead to errors in reconstruction. Although conventional MR imaging sequences may be employed with parallel MR imaging, the altered image reconstruction introduces several new artifacts and changes the appearance of conventional artifacts. The altered appearance of traditional artifacts may obscure the source of the problem, and, in some cases, the severity of artifacts associated with parallel MR imaging may be exacerbated, hindering image interpretation. Several artifacts arise in the context of parallel MR imaging, including both traditional artifacts and those associated with parallel MR imaging.

Introduction

In clinics, parallel magnetic resonance (MR) imaging is an important tool that allows for quicker acquisition of imaging data, shorter examination times, increased spatial and temporal resolution, reduced specific absorption rates, and, in some cases, improved image quality. To enable faster image acquisition, fewer phase-encoding steps are performed, leading to undersampling of a k-space for a given field of view (FOV). The time savings is in direct proportion to the number of omitted phase encodings, which are characterized with an acceleration factor that is given as the inverse of the fraction of phase encoding lines acquired; thus, if one-half as many phase encodings are acquired, the image acquisition rate is twice as fast as the conventional rate, for an acceleration factor of two. The other benefits of parallel MR imaging arise from how this increase in speed is used. For example, if examination time is not the highest priority for a particular study (eg, MR angiography), a faster acquisition time may be used to increase the image resolution in the same amount of time as a conventional image. The omitted phase encodings are distributed throughout the k-space, which, at conventional imaging, leads to aliasing artifacts. In parallel MR imaging, the loss of information from an undersampled k-space is replaced with a variation in spatial sensitivity that arises

from a physical arrangement of multiple (eg, one) coil elements, each of which is read out in parallel through separate receiver channels. To use this variation during the image reconstruction process, extra information is needed to map the spatial sensitivity of each coil element (1).

Traditionally, parallel MR imaging techniques were classified into two families that are differentiated by the data-processing domain, in which the data from each coil element are untangled to produce the final image: the raw data domain and the image data domain. Simply described, image-based parallel MR imaging (eg, sensitivity encoding [SENSE]) techniques are first used to reconstruct aliased images from each coil element before the whole image is assembled, whereas k-space–based parallel MR imaging techniques (eg, simultaneous acquisition of spatial harmonics [SMASH]) use partially filled k-spaces from all elements to construct one or more filled k-spaces that may be transformed into a whole image (1,2). Practically, manufacturers have taken advantage of the benefits of both of these approaches, resulting in the hybridization of parallel MR imaging approaches (such as sensitivity profiles from an array of coils for encoding and reconstruction in parallel [SPACE RIP] and combinations of SENSE and generalized autocalibrating partially parallel acquisition [GRAPPA]) that cannot be precisely described with an image- or k-space–based framework (3,4). Similarities, differences, and equivalency in both approaches have previously been discussed in detail (5).

Because the implementations—each of which has its own subtleties—of parallel MR imaging are diverse, a different functional classification of parallel MR imaging techniques on the basis of calibration procedures may provide insight into how to appropriately apply parallel MR imaging. Namely, one class of techniques requires that a separate calibration image be obtained at the beginning of an examination (eg, SENSE), whereas the other class relies on acquiring a small number of extra phase encodings near the center of the k-space for calibration, a method known as autocalibration (eg, GRAPPA) (1,6). Whether it is explicitly (as in the first case) or implicitly (as in the second case) acquired, the calibration information effectively specifies the spatial sensitivity for each coil element necessary for image reconstruction. Performing calibration scanning at the beginning of an examination may decrease acquisition time; for this and other reasons beyond the scope of this article, SENSE-like parallel MR imaging techniques are widely used. In this article, we confine our discussion to SENSE-like parallel MR imaging techniques.

One benefit of classifying parallel MR imaging techniques according to their calibration method is that artifacts may differ in appearance and frequency with each technique. Parallel MR imaging introduces new artifacts that are not seen on conventional images, and it may modify the appearance of traditional artifacts. Some traditional artifacts that typically do not affect an image (eg, zipper artifacts) may have a wider global effect at parallel MR imaging, which increases the practical need for improved image quality (7). Furthermore, altering traditional artifacts may obfuscate the source of the problem. Nevertheless, an understanding of the origins of parallel MR imaging artifacts is critical. Because artifacts may be mistaken for a pathologic condition, it is essential that radiologists be able to identify and ignore artifacts to accurately interpret images. In addition, recognizing artifacts may lead to improved parallel MR imaging protocols because many conventional artifacts may easily be corrected.

Various components are involved in creating a parallel MR image with explicit calibration images (Fig 1). First, the calibration images for each element are acquired by using a quick, dedicated pulse sequence. Each time these images are acquired, a sensitivity matrix *(S)* may be calculated. Frequently, these images are acquired only once, at the beginning of an examination, by using the patient as the target object. The processing stages that occur between the acquisition of calibration images and the calculation of sensitivity maps and the sensitivity matrix vary, depending on the type of scanner used. The sensitivity map for an element may be qualitatively visualized by comparing the image from a single channel with an image from a volume coil, which, by design, has good image uniformity across the FOV. Quantitatively, images from each coil element are divided by an additional image that is acquired with the body coil to derive a coarse sensitivity map. The process of division creates erroneously high sensitivities in regions with air, even though the coil detects no signal in that region. Air is eliminated by multiplying the coarse map by a binary mask that is created by applying an image intensity threshold to the body coil image. Two problems remain with this thresholded sensitivity map: It may contain gaps from bone or sinuses, and it tightly conforms to the boundary of tissues. Both of these problems increase image artifacts in the calibration images if the patient moves slightly. To alleviate problems with small motion, the thresholded map is fitted to a two- (2D) or three-dimensional (3D) polynomial function, which allows spatial sensitivity in image gaps and around the boundary of tissues to be extrapolated. Once sensitivity maps are obtained and recomposed

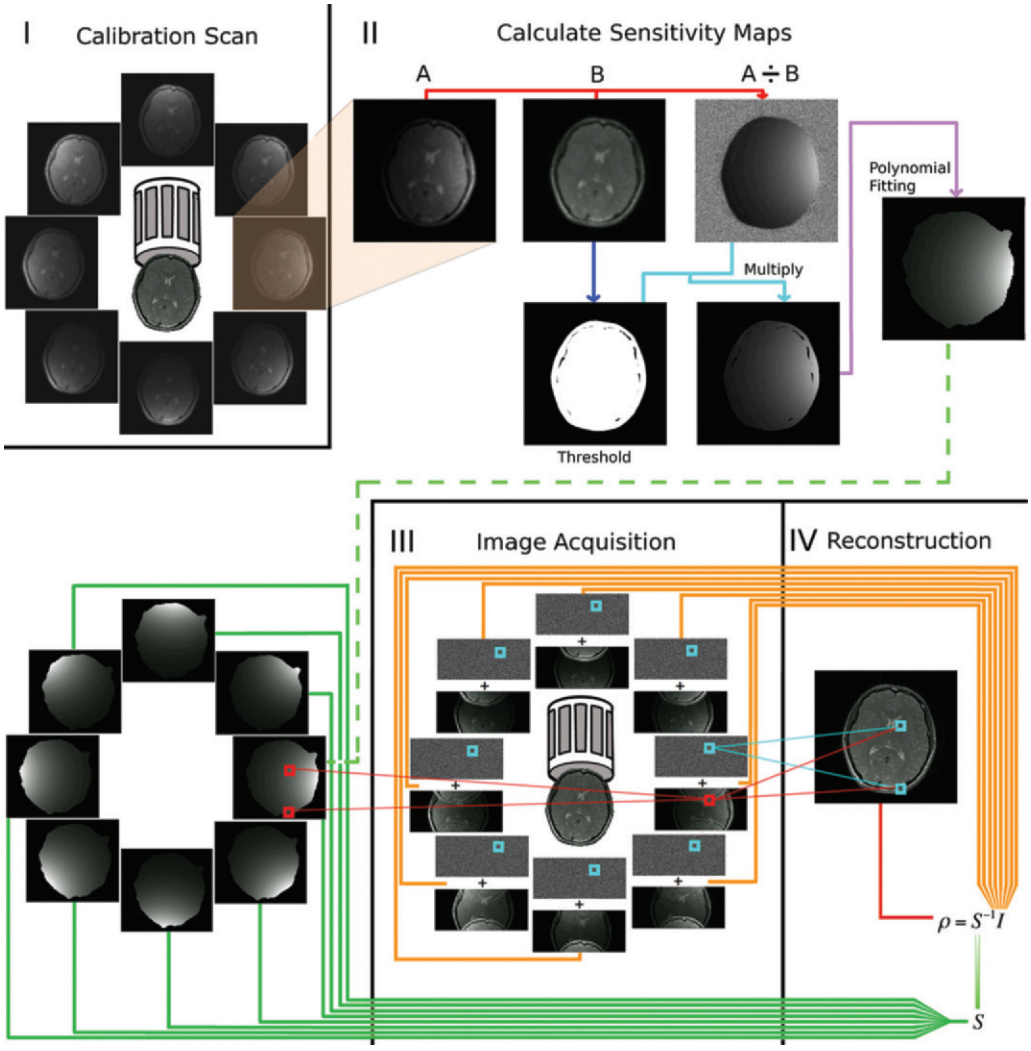


Figure 1. Calibration image–based parallel MR imaging technique. Chart shows the various steps and data components required to create a parallel MR image. In panel I, a low-resolution calibration image of the brain is quickly acquired for each coil element. The body coil may also be used to acquire a homogeneous image to help calculate a sensitivity map. In panel II, the calibration image from each coil element undergoes processing to create a sensitivity map. Each vendor uses different processing procedures; this procedure is based on that outlined by Pruessmann et al (1). Images from each coil element are divided by a global image that is acquired with the body coil to create a rough sensitivity map for each channel. Signal intensity is used to threshold the body-coil image and create a binary mask. Multiplication of the binary mask and a rough sensitivity map eliminate much of the noise from air. Finally, the thresholded map is fitted with a 2D polynomial, which allows extrapolation of the sensitivity map within holes (from thresholding) and to a small distance outside the relevant tissue for motion tolerance. Panel III shows image acquisition, in which parallel MR images are acquired and reconstructed for each channel. Undersampling along the phase-encoding direction results in aliasing, in which the intensity in one pixel is the sum of the intensities (red boxes) at two locations in unaliased images. Noise from the system or from the patient is essentially uniform across the aliased image; however, it may be correlated between channels. Panel IV shows reconstruction, which converts the aliased images from each channel into one unaliased image *(ρ)*. Sensitivity maps for all channels are organized into a sensitivity matrix *(S)*, which is inverted and multiplied by a matrix that contains the image data *(I)*. The inversion process inherently corrects image uniformity according to the sensitivity maps. Although the signal in each pixel is unaliased after reconstruction, the noise in the aliased images is distributed from all channels into multiple locations across the image (blue boxes). As a result, the distribution of noise is not uniform across the image.

into the sensitivity matrix, parallel MR imaging–enabled sequences are performed and undersampled images acquired, with a separate image for each coil element. These undersampled images show aliasing artifacts across the reduced FOV, with noise uniformly distributed. Finally, given the matrix inversion of *S*, individual coil element images are combined to form a final, unaliased image, with nonuniform noise amplification in certain regions.

Corruption of data, at any stage in the process, may result in artifacts. For example, corruption of calibration images leads to a faulty sensitivity matrix and, often, duplicate ghosts of the image, with variable signal intensity across the FOV. Considering the multiple operations that are required to create a sensitivity matrix, mistakes in processing the calibration images may occur at various steps. (For example, thresholding may not mask air if substantial noise or a conventional artifact is present in the calibration image for one coil.) On the other hand, an increase in the intrinsic noise in the system for one coil element may increase the noise magnification at focal regions in the image, as well as ghosting. A lack of consistency between the calibration and subsequent images, which occurs when the patient moves between imaging series, may cause tissues to migrate to regions that were thought to contain only air. In this case, during the reconstruction process, it is assumed that the migrant tissue was located somewhere other than a space that was thought to contain only air (where the sensitivity should be zero), which results in a chopped, disjointed appearance on images.

In this study, we discuss the appearances of several conventional artifacts that were altered during parallel MR imaging and new artifacts that are particular to parallel MR imaging, examples of which were identified either in clinical images or in images of a volunteer (7–12). The appearances of these artifacts were theoretically demonstrated with computer simulation and validated at phantom imaging under appropriate conditions.

Methods

Clinical images were acquired with an HDxT 3T (General Electric Medical Systems, Milwaukee, Wis) or Ingenia 3T (Philips Medical Systems, Best, The Netherlands) platform with a variety of different coils and 2D and 3D pulse sequences. Parallel MR imaging techniques—array spatial sensitivity encoding technique (ASSET; GE) and SENSE (Philips)—were enabled, with acceleration along only one phase-encoding direction. All phantom images were acquired on the GE HDxT 3T system with an eight-channel head coil and an acceleration factor of two. Nominally, a GE head-sphere phantom filled with doped water was used for all artifacts except chemical shift, for which a pork loin was used as a phantom.

MR imaging simulations and parallel MR imaging reconstruction algorithms were programmed with Matlab (Mathworks, Natwick, Mass). Sensitivity maps for each coil element were created by using the principle of reciprocity and the Biot-Savart law, with the assumption that coils had eight circular elements arranged in a ring (13,14). For a FOV of 24 cm, each coil element had a radius of 7.5 cm, and all coils were assembled in a ring with a 21-cm diameter. In regions of air, all sensitivity maps were masked conservatively, far from the boundary of a simulated phantom object, a 14-cm sphere with two regions: an inner region that represented water and an outer region that represented fat. The phantom comprised a magnitude image and a phase image (matrix, 128 × 128) that were multiplied by sensitivity maps and Fourier transformed to create eight separate k-space representations. Artifacts were introduced by either first altering the magnitude and phase images or by directly manipulating the k-space, depending on the type of artifact. Afterward, every other line of the k-space was removed to simulate parallel imaging with an acceleration factor of two. Images were reconstructed with a SENSE-type algorithm (1).

Conventional Artifacts Modified by Parallel MR imaging

Zipper

Zipper artifacts are common in conventional MR imaging and originate from contamination of the nuclear MR imaging signal by spurious radiofrequency (RF) noise, a result of either a compromised Faraday cage (eg, a breach in shielding material that surrounds the scanner, or an open door to the scanning room, causing RF to leak into the room from an outside source) or faulty equipment within the scanning room. A conventional zipper artifact appears as one line or a series of alternating black and white lines, giving the artifact its name. Because a zipper artifact results from a set of contaminant RFs, it fills one or more lines in the image at a given location along the frequency-encoding axis.

At parallel MR imaging, zipper artifacts appear along the frequency-encoding axis (Figs 2, 3). RF noise also leads to reconstruction errors in the form of subtle ghosting, which may overlap with areas beyond the zipper artifact. If it is not severe, a zipper artifact may not impede interpretation of conventional MR images, which may cause the

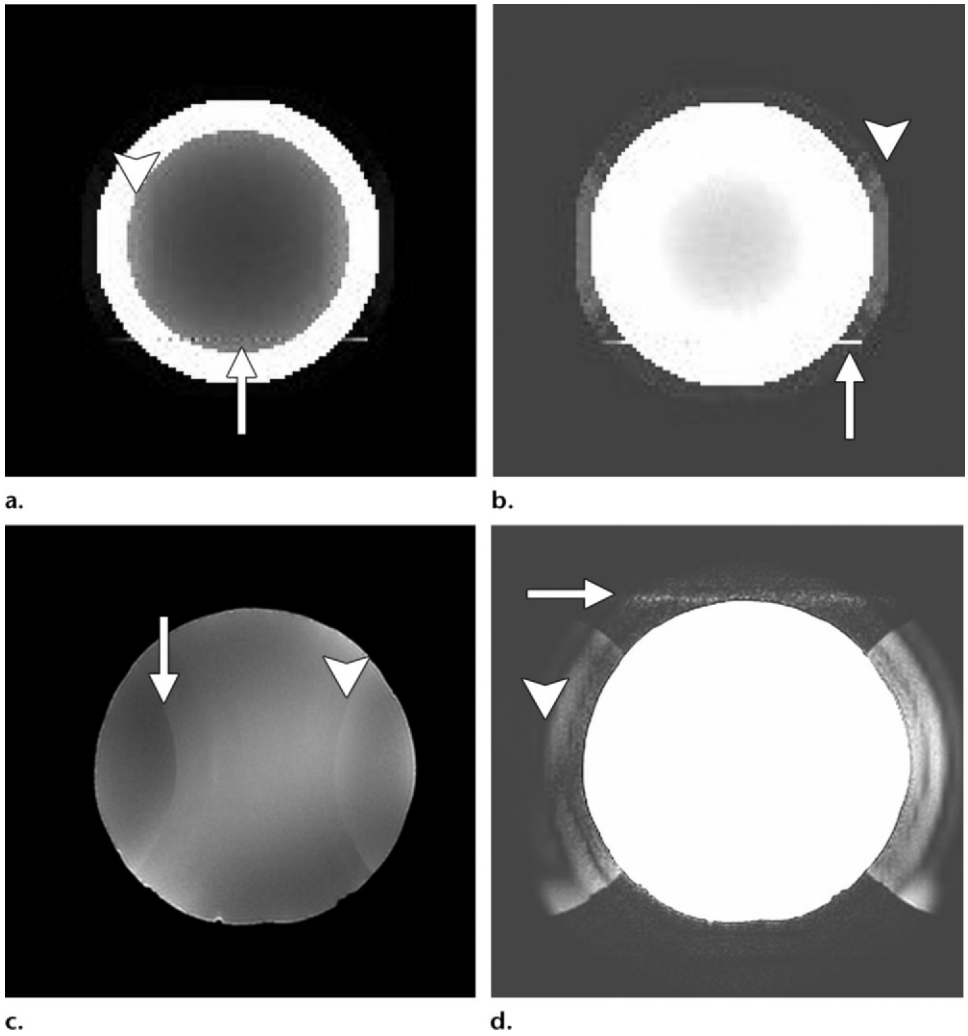


Figure 2. Zipper artifact. Simulated **(a, b)** and phantom **(c, d)** MR images show zipper artifacts (arrow). In parts **b** and **d**, contrast and brightness were adjusted to better depict ghosts (arrowhead) along the phase-encoding axis (right to left). In the phantom images, a faulty infusion pump was used to generate zippers. In both cases, no zipper was present on the calibration image.

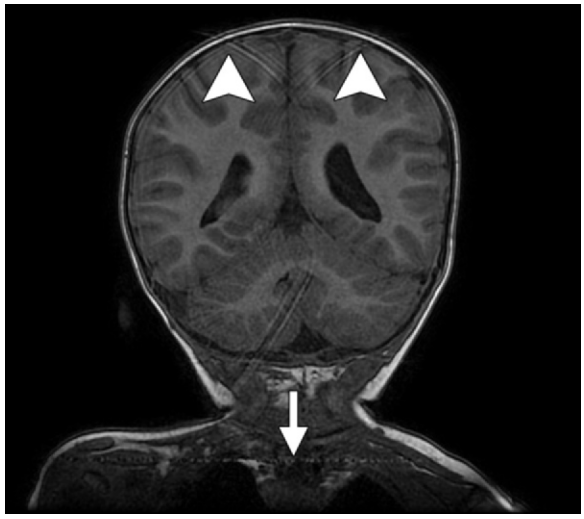
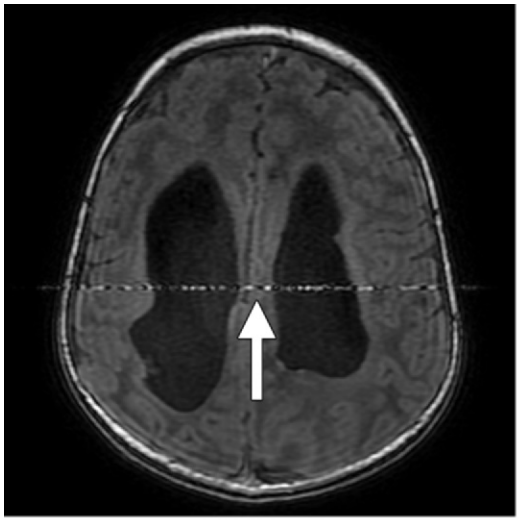


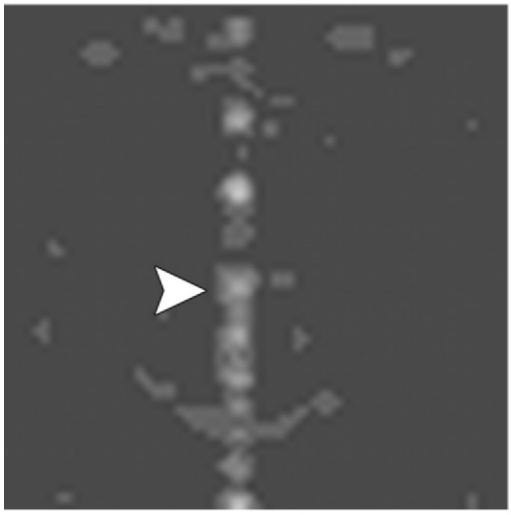
Figure 3. Zipper artifact. Coronal accelerated 3D fast-spoiled gradient-recalled echo (FSPGR) MR image shows an ASSET reconstruction error (arrowheads) arising from a zipper (arrow). A faulty injector pump was the origin of this noise.

origin of the artifact to be ignored. However, the ghosting associated with zipper artifacts at parallel MR imaging may, in certain circumstances, exacerbate problems with image interpretation.

In parallel MR imaging, RF contamination may originate from two different sources during image acquisition. Depending on the source, RF noise and the subsequent zipper artifact may disappear and reappear at different times during the examination. First, a zipper may be visible in the image itself. Reconstruction programs recognize zippers as anatomy that should be present in the sensitivity map. Second, in addition to being seen on images, a zipper may appear during calibration scanning, which leads to a faulty sensitivity map (Fig 4). Both cases may lead to problems in reconstruction. In the second case, differences between calibration and imaging protocols, such as bandwidth and pulse sequence timing, change the position and appearance of zipper artifacts.



a.



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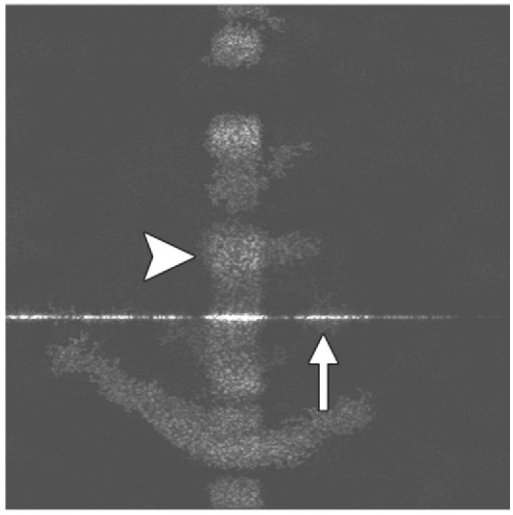


c.

Figure 4. A zipper artifact may directly affect the calibration image. **(a)** Axial T1-weighted parallel MR calibration image shows a zipper artifact (arrow) with frequency encoding along the anterior-posterior axis. **(b)** Top section of a global sensitivity map created by summing the calibration images from each coil element shows that the orientation of the zipper artifact (arrowhead) is flipped and that the artifact is not masked by the algorithm used to create the sensitivity map. **(c)** Top section of the T1-weighted parallel MR image seen in **a** shows the artifact (arrow) and an absence of masking (arrowhead) near the artifact on the calibration image.

Mitigating zipper artifacts at parallel MR imaging is difficult; changing scanning parameters such as the FOV and acceleration factor does not suppress them. The ideal solution is to discover the source of the RF leak, such as an open door or faulty equipment. If RF noise periodically disappears, repeating calibration scanning or parallel MR imaging may result in a better image, but this may not help in all circumstances. The frequency- and phase-encoding directions may also be flipped to alter the direction of ghosting; however, aliasing may be introduced, which creates additional artifacts (see the section on small FOV artifacts) (10). If ghosting artifacts are severe and the spatial extent of the zipper artifact is constrained at conventional imaging, turning off parallel imaging may be the best option.

Flipping the phase- and frequency-encoding directions may or may not be an option, depending on the coil being used. Practically, parallel MR imaging acceleration requires two or more coil elements that span the phase-encoding axis. If the coil-element layout does not include two or more elements along the coil in the desired phase-encoding direction, the scanner hardware may prohibit switching of phase- and frequency-encoding directions. An example of this is an eight-channel head coil with its elements encircling the head. Looking at a projection of the ring in different directions, four elements span from left to right and from anterior to posterior, but the projection of the ring along the superior-inferior axis is only one element deep; thus, with this coil, phase encoding along the superior-inferior axis does not support undersampling. Although, at first glance, flipping the phase- and frequency-encoding directions may seem to be a poor choice (eg, the neck and chest outside the FOV could alias into the head and create myriad parallel MR imaging artifacts), with 3D imaging, two phase-encoding directions, including the

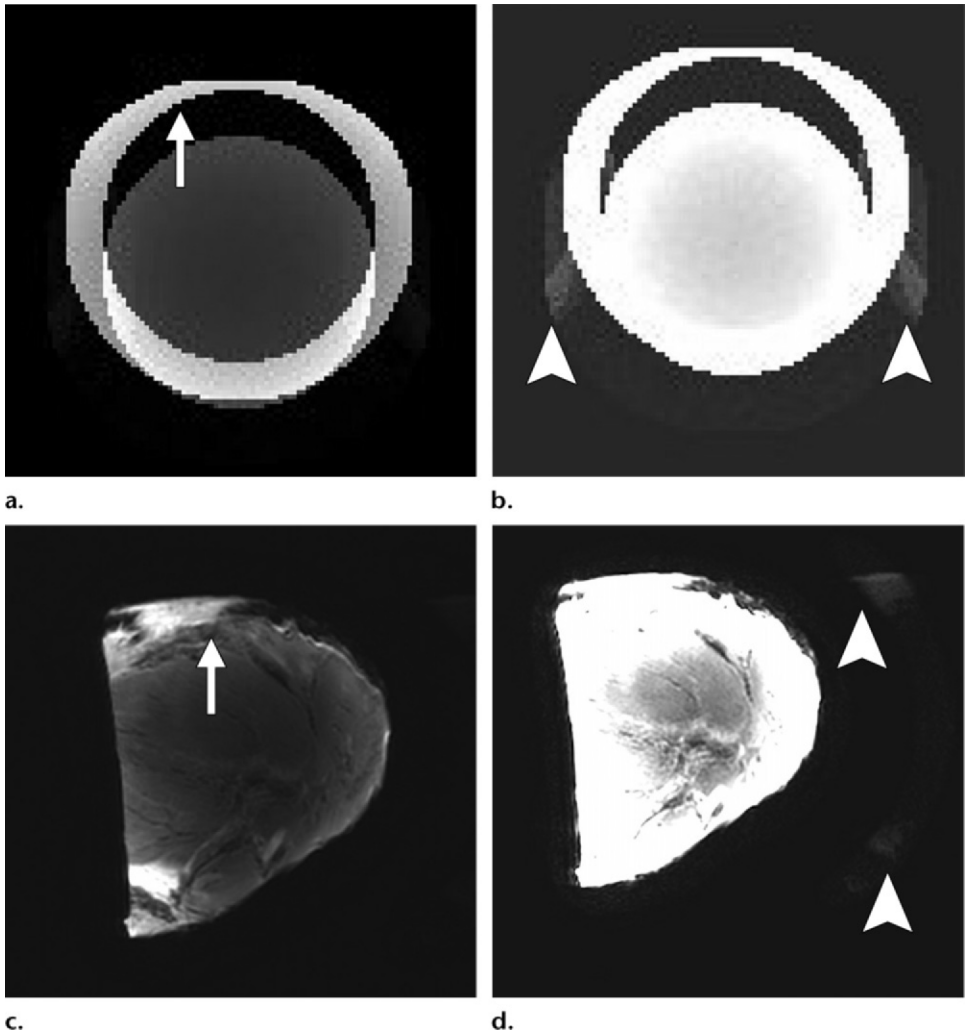
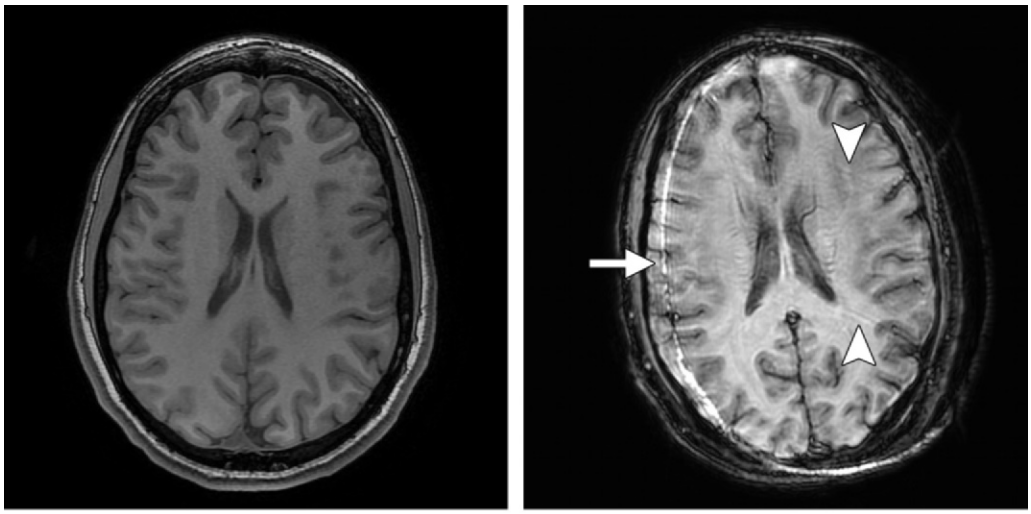


Figure 5. Chemical shift artifact. Simulation **(a, b)** and phantom (pork loin) **(c, d)** MR images show chemical shift artifact, with fat shifted (arrow in **a** and **c**) along the frequency-encoding direction (y-axis), resulting in ghosts (arrowheads in **b** and **d**). In **b** and **d**, contrast and brightness were adjusted to better depict ghosting along the phase-encoding axis (right to left).

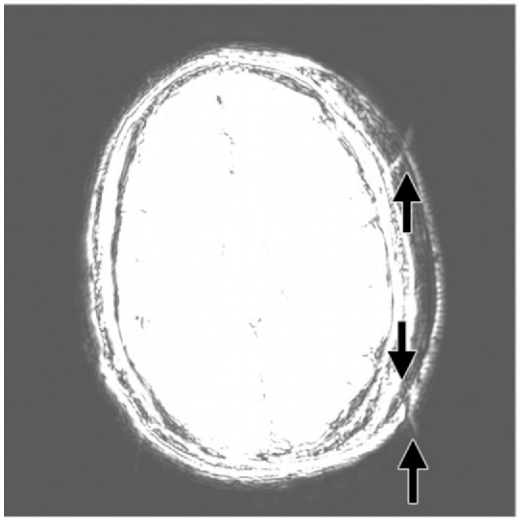
through-plane axis, may be used. In conventional imaging, axially prescribed 3D nonparallel MR imaging is possible without substantial artifacts if it is correctly prescribed. With this eight-channel coil, an axially prescribed 3D series cannot accelerate in the superior-inferior direction. For acceleration in two dimensions, a sagittal or coronal 3D prescription, in which the through-plane axis may be phase-encoded and accelerated, must be used, leaving one other axis that may be accelerated within the plane.

Chemical Shift

Chemical shift is present, to some degree, in all images and results from a difference in resonance frequency between water and fat. At conventional imaging, fat tissue shifts along the frequency-encoding axis, where the magnitude of the shift is controlled by the image series receiver bandwidth, which may be lowered to increase the signal-to-noise ratio in exchange for increased chemical shift. Because the bandwidth is the inverse of the duration of the readout period, it may be constrained by the type and parameters of a sequence. For example, substantial chemical shift may arise in echoplanar imaging. Although the bandwidth along the frequency-encoding axis is high, echoplanar imaging sequences have an effective bandwidth in the phase-encoding direction because of continuous encoding between lines in the k-space. The phase-encoded bandwidth may be fairly low (approximately 10–20 Hz/pixel), which may substantially shift fat in that direction. Ordinarily, because some form of fat suppression is applied during echoplanar imaging, negligible fat is observed. If fat suppression is poor or incomplete, chemical shift may be a problem.



a. b.



c.

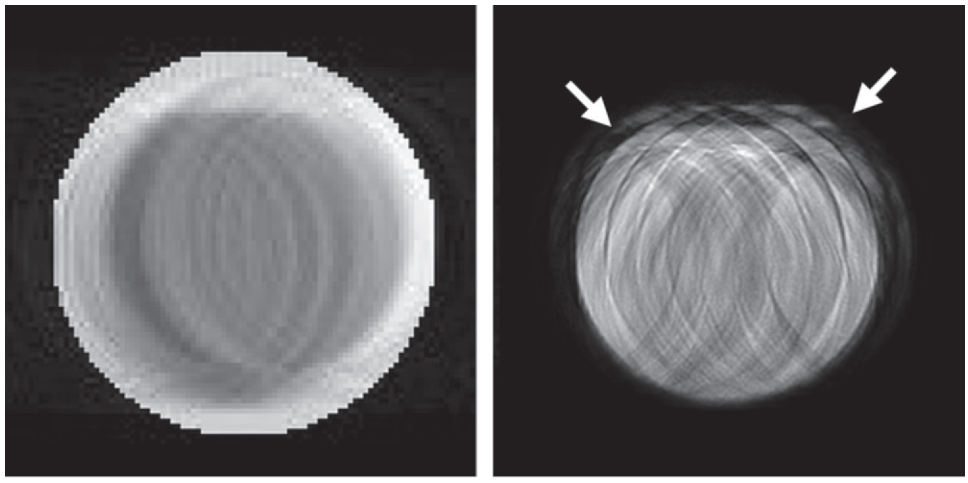
Figure 6. Chemical shift artifact. **(a)** Axial T1-weighted 3D FSPGR parallel MR image shows no artifacts. **(b)** Axial T1-weighted 3D FSPGR parallel MR image obtained with a small bandwidth shows substantial chemical shift (arrow) and ghosting (arrowheads) artifacts. **(c)** Axial T1-weighted 3D FSPGR parallel MR image with adjustment of brightness and contrast shows ghosting (arrows) along the phase-encoding direction.

In parallel MR imaging, substantial chemical shift may lead to ghosting (Figs 5, 6). Typically, calibration images are acquired with a high, preset bandwidth that often may not be adjusted by the user. If, during an image series, chemical shift is large enough to shift fat into areas that normally contain air and are masked in a calibration image, reconstruction errors may occur because the calibration images do not accurately match anatomy. Although chemical shift artifacts demonstrate fat shifts along the frequency-encoding axis at MR imaging with most conventional (but not echoplanar) sequences, at parallel MR imaging, such artifacts lead to ghosting in the phase-encoding direction. On echoplanar images, chemical shift is dramatic and occurs in the phase-encoding direction at conventional MR imaging; parallel MR imaging may create more profound artifacts along the same axis.

Fat shifting that is substantial enough to affect parallel MR image quality is rare but possible under certain circumstances. Usually, alleviating the artifact is simple; use of a higher bandwidth is recommended. Alternatively, to ensure that fat does not shift outside the areas of tissue determined at calibration scanning, flipping the phase- and frequency-encoding directions may solve the problem, a solution that caused ghosting artifacts in a phantom to disappear (Fig 5). The use of better fat suppression methods, such as spectral-spatial pulses, in echoplanar imaging may also help eliminate chemical shift (15).

Ghosting

At conventional MR imaging, motion during scanning, whether physical (ie, that of a body part) or pulsatile (ie, blood flow), results in ghosting in the phase-encoding direction. Sometimes, the motion is slight enough or the flow is spatially constrained enough to allow images to be interpreted. Motion has a similar appearance at parallel MR imaging (Figs 7, 8). However, no ghosting occurs in areas that appear as air on calibration images. In parallel MR imaging, findings on calibration images do not indicate how information outside tissue boundaries should be reconstructed because there is no signal to excite in these areas. Consequently, motion artifacts at parallel MR imaging may be more severe, with the entire artifact within the tissue boundaries, further decreasing conspicuity of features.



a. b.

Figure 7. Movement during scanning. Simulation **(a)** and phantom **(b)** MR images show ghosting (arrows in **b**), which is restricted to regions within or near the imaged object and is a result of motion during scanning.

Similar to conventional MR imaging, motion artifacts at parallel MR imaging may be reduced by asking patients not to move. If pulsatile motion is present, switching the phase- and frequency-encoding directions may flip the direction of vessel ghosting. However, as was previously mentioned, aliasing may be introduced and lead to additional artifacts. (See the section on small FOV.) If substantial movement occurred, a new calibration image may be needed because it may not overlap with the patient's current position. Finally, because parallel MR imaging increases the severity of motion artifacts, use of a nonparallel MR imaging technique may be best.

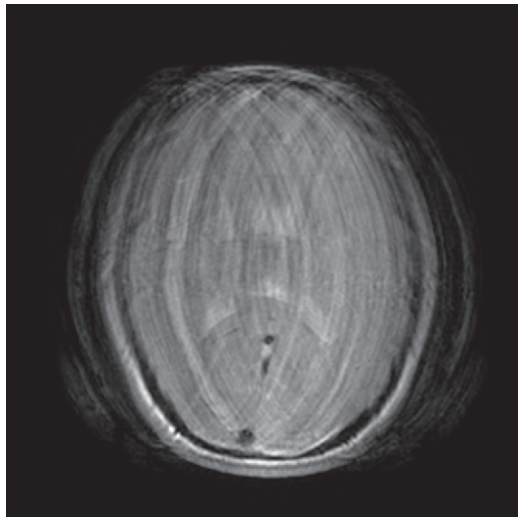


Figure 8. Motion artifact. Axial T2-weighted 2D parallel MR image shows a motion artifact.

Susceptibility

The uniformity of the resonance frequency of neighboring spins depends on the local field homogeneity, which may be perturbed by variations in local magnetic susceptibility, which affects the ability to map spins to a particular place by using frequency or phase encoding. In particular, dental hardware, bone screws, and metallic implants may cause substantial susceptibility artifacts. Air in the sinuses and colon may cause artifacts to a lesser degree. Substantial inhomogeneity in the local magnetic fields produces signal dropout and distortion on conventional MR images. The magnitude of susceptibility artifacts depends on the type of pulse sequence (eg, spin echo is less prone to such artifacts) and on sequence parameters (eg, echo time and bandwidth). In some cases, susceptibility artifacts may be ignored because they do not obstruct pertinent regions.

At parallel MR imaging, field inhomogeneity may lead to errors in image reconstruction. Nominally, calibration images should correspond to the patient's position throughout the examination. Profound susceptibility artifacts usually cause calibration images to be corrupted differently than other images obtained during the examination because differences in bandwidth and echo time lead to errors in reconstruction, a result of calibration images that do not correspond to anatomy. At parallel MR imaging, susceptibility artifacts may mirror other locations and impede diagnosis (16). Susceptibility artifacts appear as ghosting along the phase-encoding axis and duplicates of the artifactual region in other areas of the image (Figs 9–11). The position of the artifacts depends on the phase-encoding axis and the amount of parallel MR imaging acceleration used. Because these duplicates shift to other regions, susceptibility artifacts that occur at par-

Teaching Point

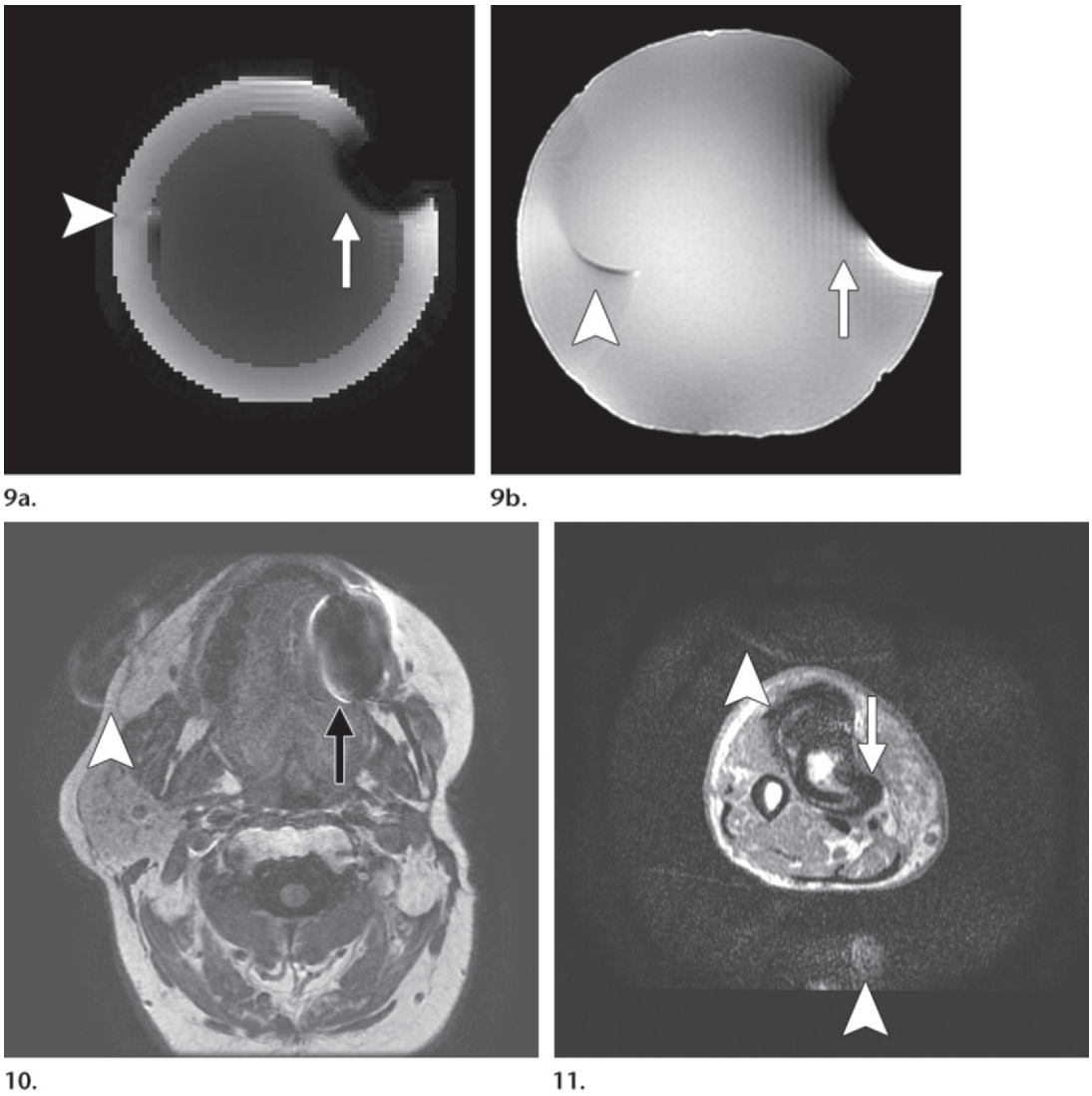


Figure 9–11. Susceptibility artifacts. **(9)** Axial simulation **(a)** and phantom **(b)** parallel MR images show a ghost (arrowhead) mirroring a susceptibility artifact (arrow). **(10)** Axial T1-weighted 2D gradient-echo parallel MR image shows a susceptibility artifact (arrow) and ghosting (arrowhead), results of parallel imaging, arising from dental hardware. **(11)** Axial T1-weighted 2D gradient-echo parallel MR image shows susceptibility (arrow) and ghosting (arrowhead) artifacts resulting from parallel MR imaging in a subject with a metallic leg implant. Image contrast and brightness were adjusted to enhance the regions outside the leg.

allel MR imaging may obstruct regions far from the areas of susceptibility variation.

In most cases with a metallic implant or profound susceptibility artifacts, parallel MR imaging should be avoided. The use of autocalibrating parallel MR imaging methods, such as GRAPPA, reduces such artifacts, although results may be unpredictable because they depend on the magnitude of regional susceptibility variation. Sequences that are more sensitive to susceptibility artifacts, such as gradient-echo sequences, exhibit more susceptibility artifacts and, thus, more parallel MR imaging susceptibility, and they should not be used when susceptibility artifacts are expected. (The same is true for conventional MR imaging.)

The concept that parallel MR imaging hinders the reconstruction of images with susceptibility artifacts is primarily predicated on the corruption of calibration images. However, circumstances exist in which an imaging study with an uncorrupted calibration image may show mild susceptibility artifact. In the presence of a mild susceptibility artifact, parallel MR imaging may be warranted as a method to alleviate artifacts from regions such as the auditory canal and near the nasal sinuses. To determine whether the susceptibility artifact is profound or minor, examine a spin-echo image that was conventionally acquired. If no susceptibility is seen, the calibration image probably is not corrupted, and parallel

Teaching Point

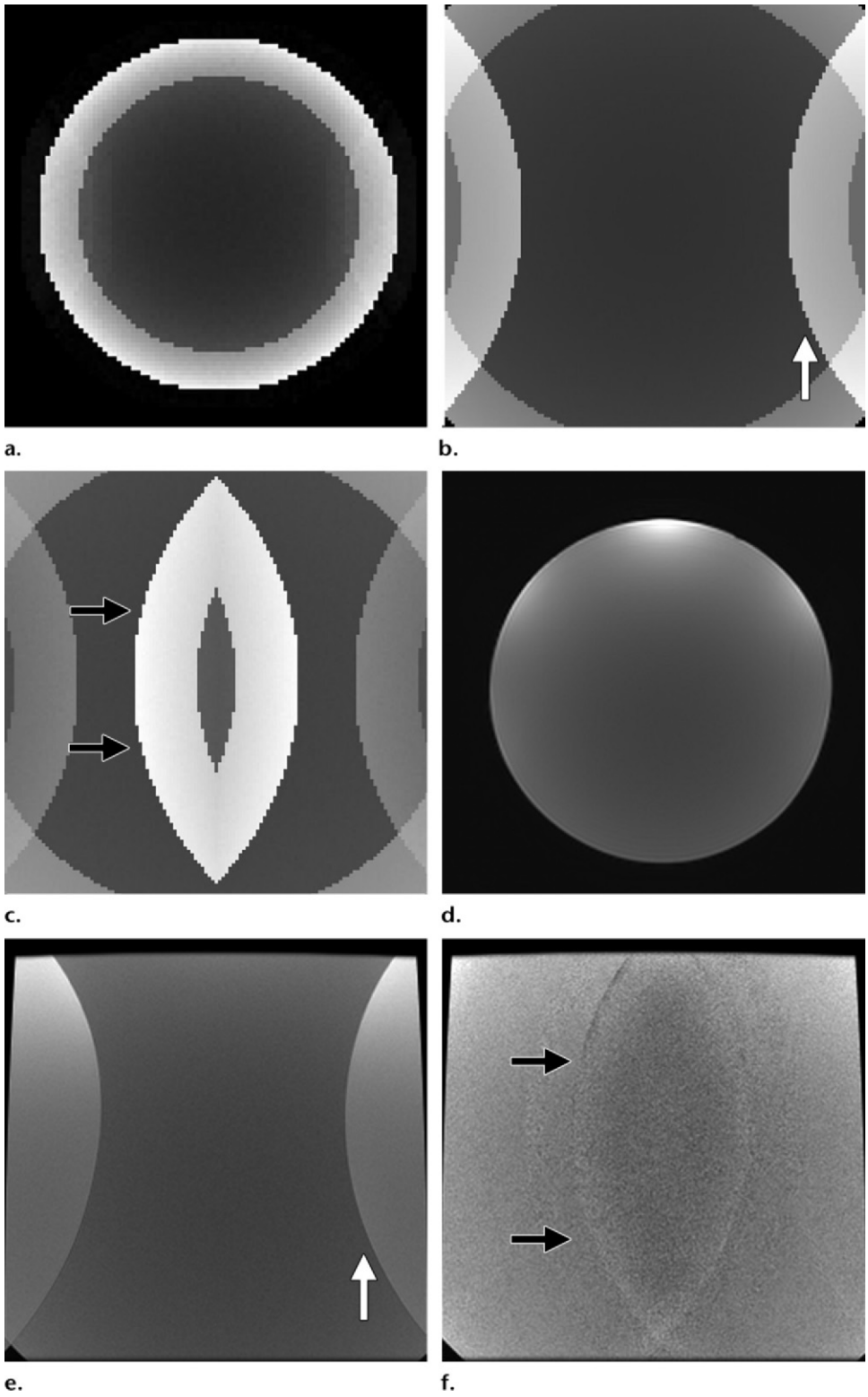


Figure 12. Aliasing artifact. Simulation **(a–c)** and phantom **(d–f)** MR images obtained with a normal FOV **(a, d)**, a small FOV **(b, e)**, and a parallel MR imaging technique with a small FOV **(c, f)** show an aliasing artifact wrapping (white arrow) into the middle of the FOV (black arrows). The phase-encoding direction is from right to left.

MR imaging likely will improve image quality for some sequences (eg, gradient echo). An example of the use of parallel MR imaging to alleviate artifacts involves echoplanar imaging, which is strongly affected by susceptibility variation because its long readout window amplifies distortion. Parallel MR imaging allows for a reduced readout window, which is useful for mitigating

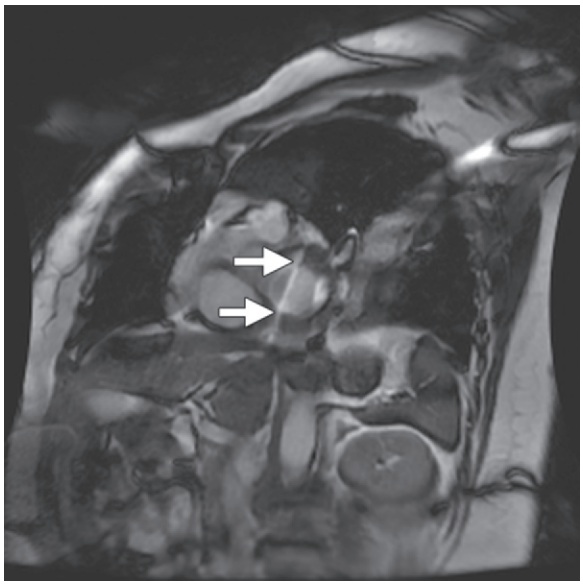


Figure 13. Small FOV artifact. Cardiac short-axis gradient-recalled echo cine parallel MR image shows wrapping of a body cavity aliasing artifact into the center of the FOV (arrows).

image distortion, and, subsequently, reduced echo time. Parallel MR images have decreased signal-to-noise ratio (SNR) compared with their conventional counterparts; however, in some cases, a shortened echo time may recover some or all of the decreased SNR, a result of decreased T2 weighting (eg, diffusion imaging). Another method to counteract a loss of SNR is to acquire multiple images for averaging. Because echoplanar imaging sequences often take little time to execute, slowing the acquisition time may be a viable alternative for increasing the SNR.

Small Field of View

When using a small FOV at conventional MR imaging, tissues outside the FOV and along the phase-encoding direction alias into the FOV. One way to alleviate this aliasing is to sample outside the FOV along the phase-encoding axis and throw away the imaging data outside the FOV, a method referred to as "no-phase wrap," "fold-over suppression," or "phase oversampling," which increases scanning time. Conventional aliasing appears to "wrap" features to the other side of the image if oversampling is inadequate.

Aliasing may drastically affect the appearance of images when paired with parallel MR imaging techniques that require a calibration image (Fig 12). Aliased anatomy that wraps to the boundaries of conventional MR images leads to mismatched anatomy and increased noise in the center of parallel MR images (8,10). Under certain circumstances, conventional aliasing may be ignored if the overlap is minor, such as aliasing of a body cavity at the borders of a cardiac image. The central location of aliasing on parallel MR images usually makes image interpretation difficult. This effect results from the tendency of parallel MR imaging to unwrap images that become aliased as a result of undersampling. If any aliasing is present before parallel MR imaging undersampling is performed, the initial aliasing may not be separated from ordinary anatomy in the FOV during image reconstruction, resulting in obstructive placement of aliased anatomy (Fig 13).

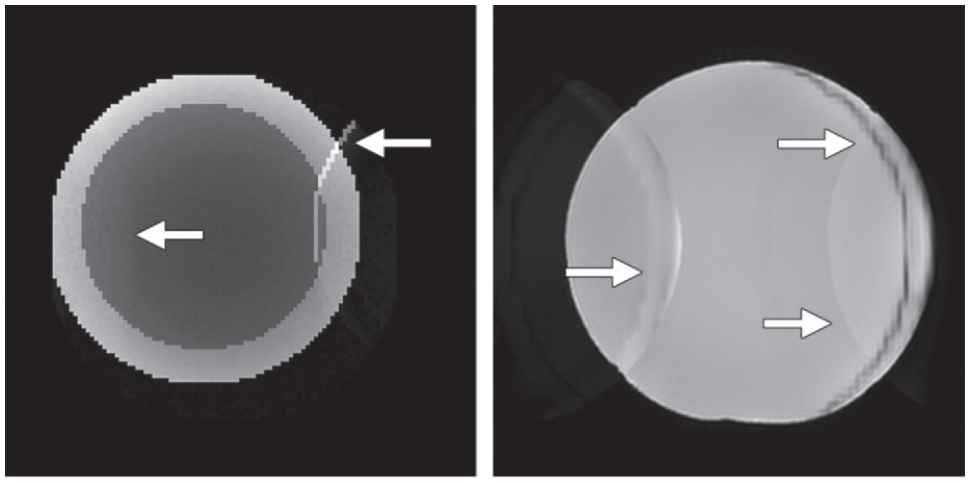
Teaching Point

The solution for aliasing with a small FOV at parallel MR imaging is similar to that at conventional MR imaging: Avoid the use of a small FOV. Aligning the phase-encoding direction with the small anatomic axis may also eliminate aliasing. If use of a small FOV cannot be avoided (eg, because of a desired protocol), parallel MR imaging should be turned off. A final alternative is to use an autocalibration parallel MR imaging method, such as GRAPPA, because it is not affected by aliasing to the same degree (5).

Artifacts That Are Unique to Parallel MR Imaging

Calibration and Diagnostic Image Mismatch

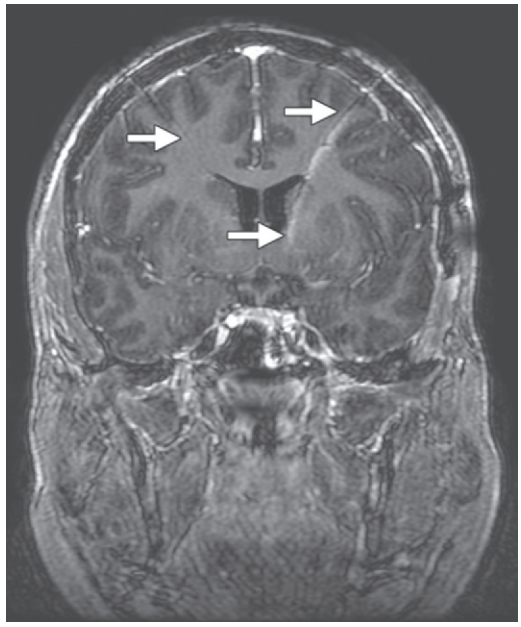
Mismatches between the anatomy on calibration images and that on diagnostic images are the cause of some of the artifacts that were previously discussed, such as those due to chemical shift and patient movement during calibration scanning. Patient movement during conventional MR imaging always decreases image quality. Accordingly, patient movement during parallel MR imaging leads to poor images. Movement between different image series obtained with parallel MR imaging and calibration scanning also leads to poor images. At image reconstruction, tissue is expected to be in the same location as on the calibration image. If it is not, errors in reconstruction will occur.



a. b.

Figure 14. Simulation **(a)** and phantom **(b)** MR images show artifacts (arrows) from motion that occurred after the calibration image was obtained.

Figure 15. Coronal T1-weighted 3D FSPGR parallel MR image shows a motion artifact (arrows) resulting from head movement after calibration.



Motion that occurs after calibration scanning results in wrapping of peripheral tissues as a ghost onto more central regions of images (Figs 14, 15). This appearance is similar to that of aliasing, with two differences: First, aliasing requires a small FOV, whereas this ghosting effect occurs even when the anatomy is substantially smaller than the FOV (Fig 16). Second, motion that occurs after calibration scanning usually results in dark and light lines that mirror one or more of the anatomy boundaries (Figs 14, 15).

One example of a mismatch between calibration and diagnostic images that is of particular note is the interaction between parallel MR imaging and breathing. Breath holding may eliminate motion artifacts during image acquisition, but it may cause other artifacts. If the position of the chest wall during a breath hold at calibration imaging is different from that at image acquisition, parallel MR imaging artifacts will appear in the middle of the FOV (Fig 17). In general, artifacts are worse if the chest inflates to a smaller degree at calibration scanning than at parallel MR imaging. In particular, artifacts are more numerous if breath holds at both inhalation and exhalation are included.

Teaching Point

Avoiding a mismatch artifact resulting from breath holding is straightforward; reacquiring calibration images usually clears it up, provided that patients are not moving in perpetuo, and it usually takes only a few seconds (Fig 15). For patients who move frequently (eg, children and adults with dementia), a calibration image may be obtained with every sequence. Breath holding must be consistent each time it is performed in a single patient. Rehearsing breath holding before imaging may improve performance in those who are capable. Requiring a few deep breaths before a breath hold may also help patients hold their breath longer and with more uniformity. Finally, if patients are unable to stay still for any reason (eg, they are not cognizant or illness prevents them from holding still), parallel MR imaging with calibration scanning is not a good idea. Autocalibrated parallel MR imaging, in which calibration information is acquired during imaging, may eliminate these artifacts as long as patients can hold still during scanning.

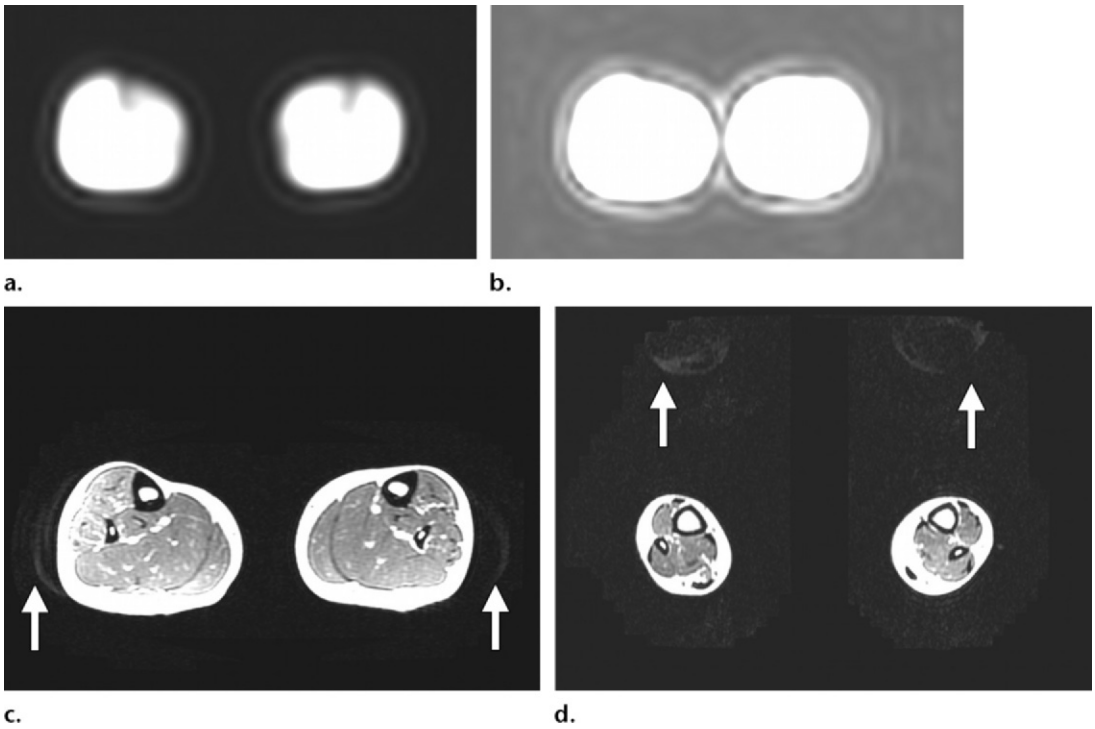


Figure 16. Correction of motion artifacts. **(a)** Original calibration image shows the legs. **(b)** Repeat calibration image obtained after motion was detected clearly shows the legs in a different position and substantial noise, which will insufficiently mask air from the sensitivity map. **(c)** Axial T1-weighted gradient-echo parallel MR image shows the legs near their original position and artifacts (arrows) resulting from disharmony with the second calibration image. **(d)** Axial T1-weighted gradient-recalled echo parallel MR image obtained with the phase and frequency directions flipped, which often resolves motion artifacts, shows reconstruction errors (arrows), a result of noise in the calibration image, which leads to faulty masking on sensitivity maps.

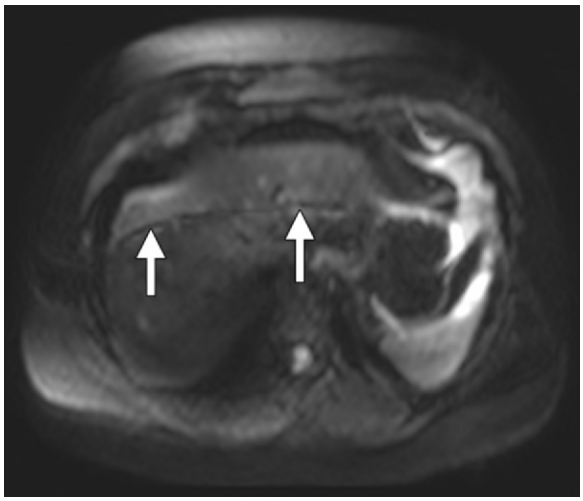


Figure 17. Axial diffusion-weighted parallel MR image of the abdomen obtained with breath holding shows artifacts (arrows), a result of breath holding distending the abdomen to different degrees at calibration and diagnostic imaging.

"Bright Structure" Reconstruction Errors

Calibration images have intrinsically low resolution, and they are less accurate near small structures with abrupt changes in signal intensity (eg, the cochlea and ear lobes). To understand this characteristic, two issues are important to consider: First, in a low-resolution image, the signal from a thin structure is volume-averaged with the signal from surrounding air, which decreases the apparent sensitivity of the coil in that region. Second, many algorithms used to create the sensitivity matrix from calibration images use a threshold, which may mask such structures outside the body habitus if volume averaging is severe (Fig 1). In either case, sensitivity in the region of these structures relies on extrapolation, which leads to ghosting, or at worst, is set to zero, which causes the structures to be repositioned in the wrong place.

These bright, small structures are "mirrored" as a subtle ghost that is displaced along the phase-encoding direction into a more central region of the image (Figs 18, 19). These artifacts may not be completely eliminated, so one should expect to see them on occasion and know that they will arise in certain places. For example, in many image series, structures in the internal auditory canal may be seen as ghosts.

Figure 18. Reconstruction errors. Simulation **(a)** and phantom (obtained with a vitamin tablet taped to a thick pad on the phantom) **(b)** MR images show reconstruction errors caused by bright, peripheral structures (arrow), which form a duplicate artifact along the phase-encoding direction (arrowhead), where the number of artifacts equals the acceleration factor.

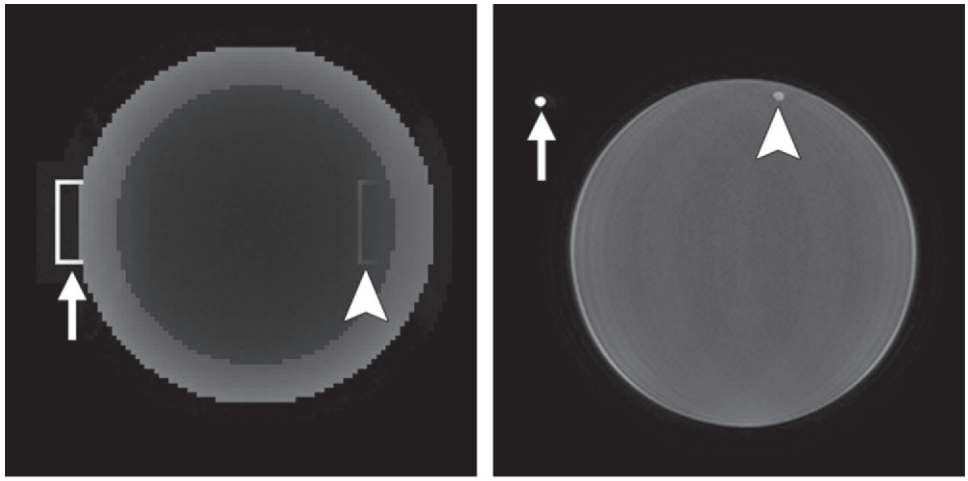
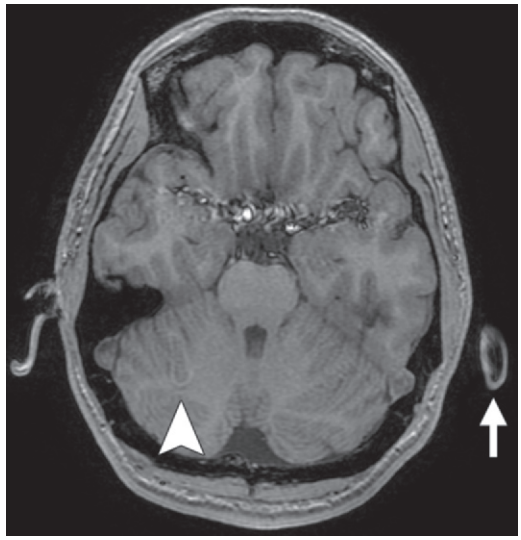


Figure 19. Reconstruction error. Axial T1-weighted 3D FSPGR parallel MR image shows the right ear, which appears bright and thin (arrow) and is mirrored as an artifact (arrowhead) into the cerebellum. No artifact from the other ear is seen.



If these small structures mask the underlying anatomy, they may be relocated by swapping the phase- and frequency-encoding directions or by using a larger FOV.

Coil Element Failure

Failure of elements in a coil results in signal dropout at conventional MR imaging. At parallel MR imaging, variations in coil element output (eg, a failing element) lead to poor mapping of spatial sensitivity at calibration scanning. Faulty coil elements may cause apparent mismatches between calibration and diagnostic images, resulting in failed reconstruction and ghosting in the vicinity of the faulty element (Figs 20, 21).

Teaching Point

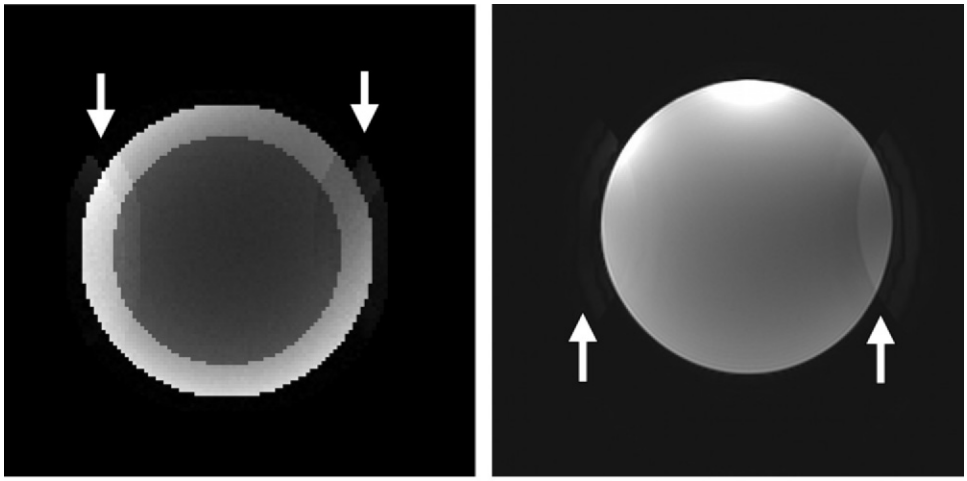
Coil elements may fail and still produce usable images (Fig 21). By virtue of the sensitivity matrix, which contains all variations of spatial coil responses—whether from the spatial position of the coil or attenuating sensitivity with distance or dielectric effect—correction of natural signal intensity is achieved during reconstruction of parallel MR images. Following this principle, image reconstruction may be worse if a coil element fails intermittently rather than completely. If a dead coil element is present, its "empty" calibration image shows an absence of signal from that element; thus, it is excluded.

The solution for failing-element artifacts is twofold: First, because coils sometimes slowly degrade, regular preventative maintenance and quality assurance measures should be performed on all coils to identify such problems. Second, while faulty coils undergo repair or replacement, use of substitute coils will ensure good image quality.

Conclusions

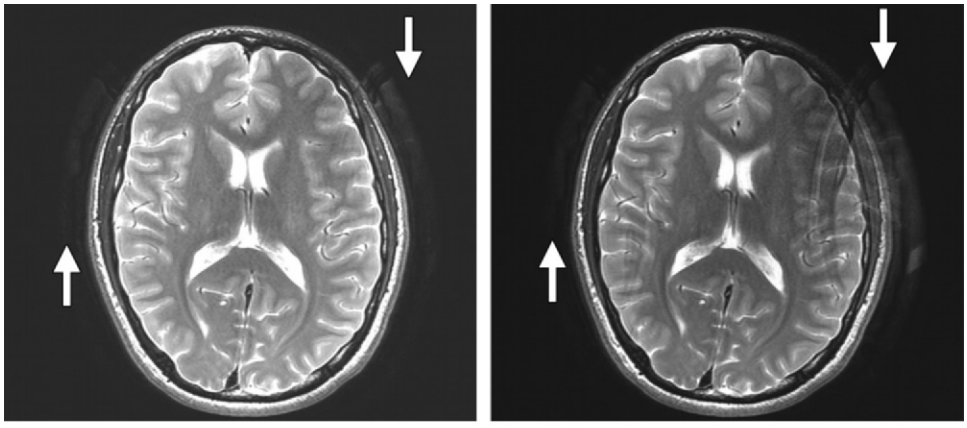
The method of image acquisition and reconstruction creates new artifacts specific to parallel MR imaging and modifies the appearance of artifacts at conventional MR imaging. Many of these artifacts may be avoided by modifying protocols

Figure 20. Faulty coil element. Simulation **(a)** and phantom **(b)** MR images obtained with one faulty coil element in the upper right quadrant show ghosting (arrows) resulting from absent signal from one coil element during image acquisition. Calibration images were normal.



a. b.

Figure 21. Coil element dropout in an eight-channel coil. Calibration images were normal. Individual coil elements were disconnected from the receiver before image acquisition. **(a)** Axial T2-weighted parallel MR image of the brain shows dropout from one coil element in the upper right quadrant and ghosting in the phase-encoding direction (arrows). **(b)** Axial T2-weighted parallel MR image shows dropout from two elements at right and ghosting in the phase-encoding direction (arrows). **(c)** Repeat calibration image with dropout from one coil element shows substantially improved image quality. The other coils were sufficiently sensitive to image the missing quadrant. However, magnification of the image shows more noise in the region of the missing element (right-hand pane). For larger FOVs and similar numbers of coil elements, increased noise worsens.



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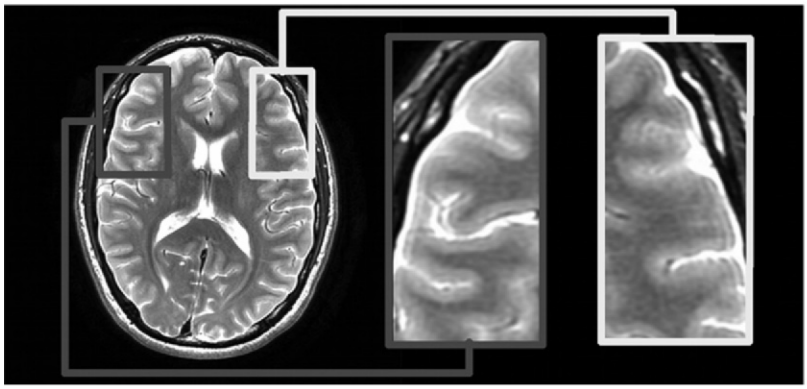

c.

while continuing to perform parallel MR imaging, whereas others require an understanding of the limitations of parallel MR imaging and when it should not be employed.

References

1. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. Magn Reson Med 1999;42(5):952–962.
2. Sodickson DK, Manning WJ. Simultaneous acquisition of spatial harmonics (SMASH): fast imaging with radiofrequency coil arrays. Magn Reson Med 1997;38(4):591–603.
3. Kyriakos WE, Panych LP, Kacher DF, et al. Sensitivity profiles from an array of coils for encoding and reconstruction in parallel (SPACE RIP). Magn Reson Med 2000;44(2):301–308.
4. Blaimer M, Breuer FA, Seiberlich N, et al. Accelerated volumetric MRI with a SENSE/GRAPPA combination. J Magn Reson Imaging 2006;24(2):444–450.
5. Lustig M, Pauly JM. SPIRiT: iterative self-consistent parallel imaging reconstruction from arbitrary k-space. Magn Reson Med 2010;64(2):457–471.
6. Griswold MA, Jakob PM, Heidemann RM, et al. Generalized autocalibrating partially parallel acquisitions (GRAPPA). Magn Reson Med 2002;47(6):1202–1210.
7. Heiland S. From A as in aliasing to Z as in zipper: artifacts in MRI. Clin Neuroradiol 2008;18(1):25–36.
8. Griswold MA, Kannengiesser S, Heidemann RM, Wang J, Jakob PM. Field-of-view limitations in parallel imaging. Magn Reson Med 2004;52(5):1118–1126.
9. Noël P, Bammer R, Reinhold C, Haider MA. Parallel imaging artifacts in body magnetic resonance imaging. Can Assoc Radiol J 2009;60(2):91–98.
10. Goldfarb JW. The SENSE ghost: field-of-view restrictions for SENSE imaging. J Magn Reson Imaging 2004;20(6):1046–1051.
11. Zhuo J, Gullapalli RP. AAPM/RSNA physics tutorial for residents: MR artifacts, safety, and quality control. RadioGraphics 2006;26(1):275–297.
12. Glockner JF, Hu HH, Stanley DW, Angelos LA, King K. Parallel MR imaging: a user's guide. RadioGraphics 2005;25(5):1279–1297.
13. Ibrahim TS. Analytical approach to the MR signal. Magn Reson Med 2005;54(3):677–682.
14. Jin J, Liu F, Weber E, Li Y, Crozier S. An electromagnetic reverse method of coil sensitivity mapping for parallel MRI: theoretical framework. J Magn Reson 2010;207(1):59–68.
15. Block W, Pauly J, Kerr A, Nishimura D. Consistent fat suppression with compensated spectral-spatial pulses. Magn Reson Med 1997;38(2):198–206.
16. Chen W, Beatty P, Koch K, Brau A. Parallel MRI near metallic implants [abstr]. In: Proceedings of the Seventeenth Meeting of the International Society for Magnetic Resonance in Medicine. Berkeley, Calif: International Society for Magnetic Resonance in Medicine, 2009.

MR Imaging Artifacts and Parallel Imaging Techniques with Calibration Scanning: A New Twist on Old Problems

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Page 540
Profound susceptibility artifacts usually cause calibration images to be corrupted differently than other images obtained during the examination because differences in bandwidth and echo time lead to errors in reconstruction, a result of calibration images that do not correspond to anatomy.

Page 541
In the presence of a mild susceptibility artifact, parallel MR imaging may be warranted as a method to alleviate artifacts from regions such as the auditory canal and near the nasal sinuses.

Page 543
Aliased anatomy that wraps to the boundaries of conventional MR images leads to mismatched anatomy and increased noise in the center of parallel MR images (8,10).

Page 544
If the position of the chest wall during a breath hold at calibration imaging is different from that at image acquisition, parallel MR imaging artifacts will appear in the middle of the FOV (Fig 17).

Page 546
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