5 Techniques for Performing Abbreviated Breast Magnetic Resonance Imaging

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5.1 Introduction

The goal of abbreviated breast magnetic resonance imaging (MRI) is to maximize sensitivity to breast cancer in a short (<5-minute) imaging session that screens women for breast cancer.\(^1,2\) The brief imaging duration must include acquisition of a localizing series, a single T1-weighted precontrast series, contrast injection, and acquisition of a single T1-weighted postcontrast series. To perform the entire imaging session in less than 5 minutes, each pre- or postcontrast series should image all breast tissue in 2 minutes or less (▶ Fig. 5.1). Since interpretation of abbreviated MRI is done primarily from the subtracted series (precontrast subtracted from postcontrast series, voxel by voxel) and maximum intensity projection (MIP) reconstructions of the subtracted series, pre- and postcontrast series should be identical and acquired without motion during, or misregistration between, the two series. Modern breast MRI equipment and fast gradient-echo techniques make this feasible while maintaining the essential features that make breast MRI highly sensitive to breast cancer: high spatial resolution, thin slices, and good signal-to-noise ratios (SNR) over both breasts.

Abbreviated breast MRI was initially proposed and validated by Kuhl et al, using a two-dimensional (2D, or planar) non-fat saturation pre- and postcontrast multislice gradient-echo series pair acquired in a total imaging time of 3 minutes, with breast immobilization.\(^3\) Their results showed that the high sensitivity, specificity, and negative predictive value of a full diagnostic breast MRI protocol could be maintained with an abbreviated MRI protocol that significantly shortens both image acquisition and interpretation times. Subsequent studies have demonstrated that similar high sensitivity to breast cancer can be maintained using three-dimensional (3D, or volume) fat-saturated approaches to abbreviated breast MRI.\(^3,4,5\)

![Fig. 5.1 Schematic of abbreviated breast magnetic resonance imaging (MRI) protocol and example images.](image-url)
This chapter will discuss the specific equipment and imaging techniques needed to perform abbreviated breast MRI and the technical aspects of breast MRI that can maximize sensitivity in the process. Examples of ideal and less-than-ideal image acquisitions will be presented to illustrate techniques and pitfalls of abbreviated breast MRI. Finally, contrast agents suitable for abbreviated breast MRI and some practical considerations in the delivery of contrast agents will be discussed.

5.2 Essential Equipment Requirements for Abbreviated Breast MRI

Essential equipment requirements for abbreviated breast MRI are identical to those for high-quality diagnostic breast MRI. They include the following:

- A high magnetic field strength (1.5 T or higher), high homogeneity MRI system.
- A dedicated bilateral breast receive or transmit–receive coil with prone patient positioning.
- Strong magnetic field gradients with short gradient rise times.
- A system capable of obtaining good fat suppression over both breasts.

While these equipment requirements have been described in detail elsewhere for breast MRI, each is described briefly below.

5.2.1 High Magnetic Field Strength, High Homogeneity MRI System

While MRI systems approved by the Food and Drug Administration (FDA) for clinical use have magnetic field strengths ranging from 0.064 up to 3.0 T, systems used for abbreviated MRI should have magnetic field strengths of 1.5 T or higher. There are several reasons for requiring a high-field system. Above about 0.3 T, image SNR per voxel increase approximately linearly with magnetic field strength. Consequently, higher static magnetic field strength ($B_0$) provides higher SNR per voxel. High SNR per voxel is needed to permit rapid imaging of both breasts with high in-plane spatial resolution and thin slices without losing more subtly enhancing non–mass-like lesions, such as some ductal carcinomas in situ (DCIS).

A second reason for performing breast MRI at high magnetic field strength is to ensure higher static magnetic field homogeneity over the entire imaged volume. High magnetic field homogeneity for breast imaging requires that the static magnetic field strength should remain uniform within about 1 part per million (ppm) across both breasts. At 1.5 T, the resonant frequency of the static magnetic field is about 64 MHz, so uniformity at 1 ppm corresponds to a 64-Hz frequency shift across the entire imaged volume. It is difficult for scanners of magnetic field strengths lower than 1.5 T to maintain a uniform magnetic field at the level of 1 ppm across the volume of both breasts (30–35 cm left to right and posterior to anterior). The main reason for maintaining this high level of magnetic field uniformity is to obtain good fat suppression based on the chemical shift between water and fat over the entire imaged volume, as will be discussed below.

5.2.2 Dedicated Bilateral Breast Coil with Prone Positioning

Dedicated bilateral receive or transmit–receive breast coils have been shown to yield SNR 5 to 10 times that of the body coil (Fig. 5.2). When dedicated breast coils are receive-only coils, the radiofrequency (RF) transmit coils are those built...
into the scanner bore, the same transmit coils used for whole-body imaging or with other receive-only coils, such as spine coils, abdomen coils, and dedicated coils used for most other anatomical regions. When the breast coil is a transmit–receive coil, the breast coil is designed both to send RF signal into the breasts (to excite measurable signals from hydrogen nuclei) and, a short time later, to measure the signals emitted from breast tissues.

Current breast coils, whether transmit–receive or receive-only coils, have multiple receive channel elements. Bilateral breast coils have between two channels (one channel for each breast) and 18 channels. Multichannel coils record received signals from each channel simultaneously using a different amplifier and analog-to-digital converter for each channel. Like other coils, multichannel breast coils must be matched to the data-handling capabilities of the MR scanner hardware and software to simultaneously record and process multiple channels of data.

When a woman has both breasts, bilateral breast imaging is recommended for a number of reasons. First, comparison of both breasts helps identify focal and asymmetric enhancement patterns that may signal breast cancer (Fig. 5.3). This helps prevent false-positive interpretations due to physiologic enhancement, which tends to be bilaterally symmetric, especially in premenopausal women and in postmenopausal women on hormone replacement therapy. Second, when a breast cancer occurs, there is about a 3 to 5% chance that breast MRI will detect a mammographically occult breast cancer in the contralateral breast, so it is important to examine both breasts. Third, bilateral breast imaging eliminates some of the artifacts that can occur in unilateral imaging, such as wrap artifacts from the contralateral breast when the field of view (FOV) is set for only a single breast.

Prone positioning in a dedicated bilateral breast coil helps reduce breast motion and respiratory and cardiac pulsation artifacts, without distorting the breasts. The screened woman is supported in the bilateral breast coil at the sternum, lateral chest, and above and below the coil, with breasts pendent within the coils. Providing mild compression to the breasts without deforming them can further minimize breast motion during imaging and prevent misregistration between pre- and postcontrast scans. Any motion between pre- and postcontrast scans or during scanning can cause blurring of images, motion artifacts, and misregistration artifacts in both subtracted images and the MIP images reconstructed from the volume of subtracted images. Mild compression parallel to the...
acquisition plane also can help limit the number of slices needed for the screening study. By positioning the patient comfortably and by properly instructing the patient prior to the pre- and postcontrast T1-weighted series pair, rather than after the precontrast scan and just prior to contrast injection, the MR technologist plays an important role in minimizing motion and misregistration. The 3- to 5-minute imaging time of abbreviated breast MRI also should limit patient motion during scanning. Technologists performing either conventional breast MRI or abbreviated breast MRI should be aware of techniques to optimize patient positioning.19

5.2.3 Strong Magnetic Field Gradients with Short Gradient Rise Times

Magnetic field gradients are required in each perpendicular direction (x, y, and z) to resolve the volume of tissue within the bore of the scanner into individual voxels. Magnetic gradients are turned on and off quickly many times during a pulse sequence series to select a slice or volume of tissue and to resolve signals from a slice into pixels (in 2D acquisitions) or to resolve signals from a volume into voxels (in 3D acquisitions).5 Two parameters characterize the performance of magnetic field gradients: maximum gradient strength, which determines the maximum change in magnetic field over a given distance along the x, y, or z axis, and gradient rise times, which describe the time interval needed to turn a magnetic field gradient from zero to maximum strength. Magnetic gradient strength helps determine how small voxels can be made, while gradient rise times determine how quickly the pulse sequence that resolves signals into pixels or voxels can proceed. The shorter the gradient rise times, the shorter repetition time (TR) and echo time (TE) can be made. This is especially important in 3D gradient-echo imaging for abbreviated breast MRI, because short TRs and TEAs are required to produce strongly T1-weighted images in the limited time available for a full 3D series to be obtained (1–2 minutes). Modern MR scanners have maximum gradient strengths of 40 to 50 mT/m and gradient rise times as short as 200 µs, yielding TR values as short as 4 ms and TE values as short as 1 ms. To obtain adequate spatial resolution over both breasts in 1 to 2 minutes per 3D gradient-echo series, TR needs to be less than 6 ms and TE needs to be less than 3 ms.

5.2.4 Good Fat Suppression over Both Breasts

The question of whether fat suppression is advantageous for abbreviated breast MRI is still open. The abbreviated breast MRI proof-of-concept study conducted by Kuhl et al used bilateral 2D gradient-echo imaging without fat suppression.1 Through discussions with Dr. Kuhl, I have learned that their 2D non–fat-saturated approach is sometimes helpful in confirming the presence of suspicious lesions by seeing lesion margins, and in particular spiculations, in precontrast series as lower signal areas in a background of bright fat. Most dynamic breast MRI exams performed in the United States, on the other hand, employ 3D gradient-echo imaging with fat suppression.3,4,5,6 In this approach, lesion margins and spiculations are typically best seen in postcontrast, subtracted, or MIPs of subtracted series. This chapter’s perspective, based on extensive experience with dynamic breast MRI, is that if abbreviated MRI can be shown to be effective using multiplanar 2D gradient-echo imaging acquisitions without fat

Fig. 5.4 Wrap (or aliasing) artifact of the contralateral breast in a unilateral transaxial acquisition, with phase encoding left to right. Note that the wrap artifact is most apparent outside the imaged breast (arrows), but also wraps across and adds structured noise within the imaged breast.
suppression, as it has by Kuhl et al,\(^1\) then it should be at least as effective using appropriate 3D gradient-echo acquisitions with fat suppression. The preliminary abbreviated MRI studies using a 3D fat-suppressed approach seem to support this view, at least as far as sensitivity to breast cancer is concerned.\(^3,4,5\) The effect on specificity of performing abbreviated MRI using 3D, fat-suppressed acquisition techniques on a high-risk or intermediate-to-high-risk cohort remains to be seen.

The inclusion of fat suppression in the pre- and postcontrast T1-weighted pair in abbreviated breast MRI improves the conspicuity of enhancing lesions in postcontrast images and reduces confounding background noise and artifacts that can mimic or mask enhancing lesions in subtracted images and MIP images reconstructed from the set of subtracted images (Fig. 5.5). When pre- and postcontrast images are acquired without fat suppression, misregistration of fat between the two series can result in residual fat signals in subtracted and MIP images that can simulate enhancing lesions, especially non-mass-like lesions, leading to false-positive findings. In addition, residual fat signal in subtracted images adds structured noise that can mask detection of subtle non-mass-like lesions characteristic of DCIS, leading, in some cases, to false-negative interpretations.

Unfortunately, there are no clinical studies that compare similar contrast-enhanced breast MRI techniques with and without fat suppression, in either dynamic breast MRI or abbreviated breast MRI. This is due, at least in part, to the necessity in such studies to image each study volunteer twice, once with and once without fat suppression, each time injecting contrast agent.

The next section describes details of the gradient-echo pulse sequence needed to maximize enhancing lesion conspicuity in abbreviated breast MRI.

### 5.3 Essential Abbreviated Breast MRI Screening Exam

The essential elements of an abbreviated breast MRI exam are as follows:
- Identical pre- and postcontrast T1-weighted gradient-echo acquisitions, each of 1- to 2-minute duration.
- Inclusion of all breast tissue.
- In-plane spatial resolution (pixel size) of 1 mm or less.
- Acquired slice thickness of 3 mm or less.
- Adequately high SNR to detect small (1–2 mm diameter) vessels in MIP reconstructed images.

Each of these essential elements is discussed below.

#### 5.3.1 Identical Pre- and Postcontrast T1-Weighted Gradient-Echo Acquisitions, Each of 1- to 2-Minute Duration

T1-weighting is used in contrast-enhanced breast MRI, as in other clinical applications of gadolinium (Gd) based contrast agents, because Gd-chelates, while shortening both T1 and T2 (or T2\(^*\) in gradient-echo series), cause a greater change in T1 than in T2 or T2\(^*\).\(^2,0\) Gradient-echo imaging is used

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![Fig. 5.5](image-url)

**Fig. 5.5** (a) Precontrast and (b) postcontrast scans acquired without fat saturation and with the body coil as the receiver coil, rather than a dedicated bilateral breast coil. (c) The subtracted image (b minus a) shows artifacts in both breasts due to misregistration of pre- and postcontrast scans, making it difficult to distinguish enhancing lesions from background noise due to misregistration of unsuppressed fat between pre- and postcontrast series.
because it is most efficient in obtaining strong T1-weighting, while permitting high-resolution imaging. Gradient-echo imaging achieves strong T1-weighting by using a short TR value, a very short TE value, and a low flip angle that is matched to the TR value to obtain maximum SNR per unit time or maximum contrast-to-noise ratio (CNR) between lesion and background tissues per unit time. For 3D (volume) acquisitions, signal is acquired from a volume of tissue in position space and phase encoded to separate signal into two perpendicular directions, with frequency encoding separating signal in the third perpendicular direction. This can be thought of as filling a volume of signal data in 3D spatial frequency (or k-) space. A 3D Fourier transform (3DFT) is used to convert the signals collected in 3D k-space into individual voxels in 3D position space. This collection of signal from an entire volume of data, rather than just single planes of data as is done in 2D or planar imaging, adds efficiency to the 3D technique. Extremely short TR values are required in 3D imaging, however, to keep the scan times for the pre- and postcontrast T1-weighted series short, under 2 minutes per series in abbreviated breast MRI. For 3D acquisitions where TR values are extremely short (under 10 ms), gradient-echo flip angles are usually 10 to 15 degrees.

For 3DFT imaging, total acquisition time is

\[
T = (TR)(N_{pe})(N_{acq})(N_{slices})
\]

where \(TR\) is the pulse sequence repetition time, \(N_{pe}\) is the number of in-plane phase-encoding steps (to resolve signal into voxels along one in-plane direction, the phase-encoding direction), \(N_{acq}\) is the number of times each phase encoding step is repeated (which is almost always set to 1 in 3DFT imaging), and \(N_{slices}\) is the number of acquired slices within the 3D volume, which in 3DFT imaging is the number of phase-encoding steps used to separate the 3D volume into a second perpendicular direction, in this case the slice-select direction (Fig. 5.6). Frequency encoding applied during signal acquisition is used to separate the 3D volume into the third perpendicular direction, the frequency-encoding direction, but fortunately does not require an additional time factor to do so. In 2DFT (planar) imaging, \(N_{slices}\) is automatically set to 1 in the \(T_{total}\) formula, TR and flip angle are increased, and a multislice approach is used. Because in 3D imaging the additional factor, \(N_{slices}\), is large, typically 80 to 160 in breast MRI, very short TR values are needed in 3DFT imaging to achieve reasonably short total scan times per series. For example, a typical 3D acquisition has \(TR = 5\, ms\), \(N_{pe} = 320\) (the number of matrix elements in the phase-encoding direction), \(N_{acq} = 1\), and \(N_{slices} = 120\), so \(T_{total} = 5\, ms \cdot 320 \cdot 1 \cdot 120 = 192,000\, ms = 192\, s\), or 3 minutes, 12 s. This is still too long for a single pre- or postcontrast series in abbreviated breast MRI, so additional techniques compatible with 3DFT gradient-echo acquisitions are needed to further shorten acquisition times. These include the use of parallel imaging, partial-Fourier acquisitions, and slice interpolation, each of which is described below.

3DFT (volume) acquisitions have advantages over 2DFT (planar) acquisitions. One is a signal-to-noise advantage because 3DFT acquisitions collect signal from an entire volume of tissue that includes both breasts during each signal measurement, instead of from just a single plane of tissue as occurs in 2DFT imaging. This makes 3DFT acquisitions more efficient, but at the expense of requiring more total scan time (by the factor \(N_{slices}\)) to resolve the entire 3D volume into individual voxels. 2DFT acquisitions require resolving only a single preselected plane into individual voxels, using phase encoding to resolve signals in one in-plane direction and frequency encoding to resolve signals in the other perpendicular in-plane direction.

A second advantage of 3DFT acquisitions is that they always subdivide signals in the slice-select direction into contiguous slices with rectangular slice profiles (meaning that all tissues within each slice contribute to the measured signal), while 2DFT imaging typically has small gaps between individual slices and measured signals have Gaussian slice profiles (where signal is predominantly measured from the center of each slice and the signals from slice edges are reduced or lost), as shown in Fig. 5.7. A third advantage of 3DFT acquisitions is that with thin slices, isotropic voxels (meaning voxels with the same dimension in all three perpendicular directions) or nearly isotropic voxels can be acquired. This in turn enables orthogonal plane or MIP reconstructions of subtracted image data in virtually any orientation without loss (or in the case of nearly isotropic voxels, with only minor loss) of spatial resolution, which can permit better visualization of enhancing lesion margins. 2D or 3D acquisitions with thicker (e.g., 1.5-mm or greater) slices typically
are limited to reconstruction of MIP image projections along the slice-select direction to avoid the degradation of spatial resolution that occur in other planar or MIP projections, due to slice thickness significantly exceeding in-plane spatial resolution.

Fig. 5.6 Schematic illustrating 3D (volume) acquisition of slices in the transaxial (z-) plane. (a) The slice-select gradient (along the z-axis for a transaxial acquisition) is turned on to select a slab or volume of tissue to be excited by the radio frequency pulse transmitted into the patient after the patient’s breasts are placed as closely as possible to the isocenter of the magnet. (b) In a series of pulse sequence repetitions with different amounts of phase encoding applied along the z-direction, the excited slab is resolved into slices within the volume. The number of different slices resolved equals the number of slice phase-encoding steps. (c) For each slice phase-encoding step, the pulse sequence is repeated with a different in-plane phase-encoding gradient to resolve the volume of tissue into different planes in the in-plane phase-encoding direction, y. The number of in-plane phase-encoding views collected for each slice phase encoding equals the number of matrix elements in the y-direction. (d) During each signal readout, a frequency-encoding gradient is turned on in the third (in this case, x-) direction to resolve the volume into different voxels along the third orthogonal (x-) direction. The number of different time points at which signal is measured during each signal readout equals the number of matrix elements along the x-direction. The total imaging time for a 3D pulse sequence acquisition is $T_{total} = (TR) (N_{pe}) (N_{acq}) (N_{slices})$, where $TR$ is the gradient-echo pulse sequence repetition time, $N_{pe}$ is the number of in-plane phase-encoding steps (the number of matrix elements in the in-plane phase encoding direction), $N_{slices}$ is the number of slices acquired within the 3D volume, and $N_{acq}$ is the number of times the entire process is repeated, which is usually set to 1 in 3DFT acquisitions.
5.3.2 Methods to Speed 3D Gradient-echo Pulse Sequence Acquisitions

Parallel Imaging

To save additional time in either 2D or 3D acquisitions, parallel imaging can be used, assuming the scanner has the software and appropriate breast coils. This is apparent on the technologist scan menu. Many 3D gradient-echo series have parallel imaging options and some permit adjustment of the parallel imaging acceleration factor (AF). For example, if an AF of 2 is selected, scan time will be cut nearly in half for the same coverage, spatial resolution, and slice thickness. Parallel imaging does this by acquiring multiple segments of data simultaneously, speeding the data acquisition process. This acceleration of data collection can occur either in physical space (for instance, one segment of signal data being acquired on the left breast, while the other segment is simultaneously acquired on the right breast) or in k-space, where, for $AF=2$, every other line of signal data in k-space is sampled and intervening lines of data are interpolated. In either case, the sensitivity profiles of individual receiver coil elements are measured on that patient (which takes a small amount of additional time) to compensate for the simultaneously acquired or missing data and to enable reconstruction of parallel-acquired data into images. One prerequisite to parallel imaging is that the receiver coil must have at least as many channels as the AF. So for an AF of 2, the breast receiver coil must have at least two channels, but could also have a higher number of channels that get combined into two data segments.

In practice, parallel imaging with too high an AF leads to excessive artifacts in the resulting images (Fig. 5.8). Typically, at 1.5 T, AF is not set to be greater than 2, while at 3.0 T, the AF is not set to be greater than 3. Even with $AF=2$, the time for a full 3DFT acquisition would be reduced from 192 s (in the example of 3DFT image acquisition described above) to 81 to 110 s, a range that would be acceptable for each pre- or postcontrast series in an abbreviated breast MRI exam. The reason that a range of acquisition times is stated for an AF of 2 is that some manufacturers acquire the calibration scan that measures the sensitivity profiles of each coil channel as a separate pulse sequence when parallel imaging is prescribed, while other manufacturers acquire the calibration scan within the parallel imaging sequence itself. In either case, there is a small amount of additional time needed beyond the time estimated by dividing the original series scan time by the AF.

While reducing scan time by a factor nearly equal to the AF, there is an SNR penalty for using parallel imaging. SNR is reduced by a factor equal to the square root of the AF, so for $AF=2$, SNR is reduced by a factor of 1.41, resulting in an SNR that is approximately 71% of that without using parallel imaging (Fig. 5.8). For $AF=3$, SNR would be reduced to approximately 58% of that without parallel imaging, although this degree of SNR loss would be compensated by the nearly doubled SNR at 3.0 T compared to 1.5 T, all other acquisition parameters being equal.
Partial-Fourier Acquisitions

A second way to speed image acquisition in either a 2D or a 3D imaging series is partial-Fourier imaging. In partial-Fourier imaging, instead of acquiring every phase-encoding view in $k$-space, only a fraction of phase-encoding views (between 0.5 and 1.0, with 1.0 being equivalent to a full dataset acquisition) is acquired; the remaining phase-encoding views (those containing the higher spatial frequency components rather than those comprising the strongest contrast weighting, which occurs at the lower spatial frequency components) are created assuming a symmetry between signal values in positive and negative phase-encoding views (Fig. 5.9). If a partial-Fourier acquisition is done with the phase factor set to 0.75 (or 75%), then three-fourths of the phase-encoding views (half of those in the + or − spatial frequency direction) are assumed based on phase symmetry. Since acquiring different phase-encoding views is the most time-consuming aspect of either a 2D or a 3D acquisition, setting the phase factor to 0.75 shortens the overall acquisition time to 75% of its original duration. SNR is reduced in proportion to the square root of the phase factor, so for a phase factor of 0.75, SNR would be proportional to the square root of 0.75, or to 87% of the SNR without partial-Fourier imaging. Half-Fourier imaging assumes that nearly half of $k$-space is filled by assuming that negative phase signal values are equal to positive phase signal values (with slight oversampling of the low spatial frequency elements near zero along the $k_y$-axis), resulting in nearly a factor of 2 time savings, with an approximate 40% loss in SNR. In 2D imaging, partial-Fourier imaging can be applied only in the single phase-encoding in-plane direction. In 3D imaging, partial-Fourier techniques can be applied to either the in-plane or the slice-select direction. Application of partial-Fourier techniques in the slice-select direction is the basis of slice interpolation, discussed next.

Slice Interpolation

A third way to speed 3D gradient-echo acquisitions is to use slice interpolation. The principle behind slice interpolation is that a contiguous set of slices equal to about half of those needed is acquired, each with double the slice thickness needed (to cover the full prescribed volume), while intervening slices are interpolated (Fig. 5.10). For example, in a 3D axial series where 16-cm coverage is needed to include all breast tissue in the slice-select or z-direction (along the direction of the static magnetic field in a solenoidal magnet) and slices every 1 mm are desired, rather than acquiring 160 1-mm slices, 80 (or 81) 2-mm slices would be acquired and 80 intervening slices would be interpolated from the 81 originally acquired slices. This would result in 160 slices spaced every 1 mm apart. The interpolated slices are constructed pixel by pixel by averaging the signals from the two adjacent pixels, one from each slice, on opposite sides of the interpolated slice.
When slice interpolation is used, it is important to distinguish between \textit{acquired} slice thickness and \textit{interpolated} slice spacing. It is also important to understand how to prescribe slice interpolation on different scanners. On GE Healthcare scanners, slice interpolation is invoked by prescribing ZIP2 on the scan option menu. In the example above, 80 (or 81) 2-mm slices would be prescribed and acquired, but 160 slices, each spaced 1-mm apart, would be reconstructed in the final 3D volume.

Other manufacturers require the technologist to prescribe slice interpolation differently. For example, in 3D gradient-echo acquisitions on Siemens MRI systems, the slice interpolation factor is set on the pulse sequence menu by setting the “slice resolution,” which can be varied between 0.5 (or 50\%), which is roughly equivalent to ZIP2 on GE and is essentially half-Fourier techniques applied in the slice-select phase-encoding direction and 1.0 (or 100\%, which means no slice interpolation). The acquired slice thickness is multiplied by this factor to get the interpolated slice spacing. For example, if the pulse sequence is set up to acquire 2-mm thick slices over a range of 16 cm (160 mm) with a “slice resolution” of 0.6 (or 60\%), the system will acquire 80 2-mm-thick slices, but will reconstruct and present the reader with 133 images spaced 1.2-mm apart (2.0 mm × 0.6 = 1.2 mm) covering the same total range in the slice-select direction. One odd feature of the Siemens system, unlike other manufacturers, is that when slice interpolation is selected by the operator, the slice thickness indicated on images (and in the Digital Imaging and Communications in Medicine [DICOM] header in the “slice thickness” location) is the \textit{interpolated} slice spacing, not the \textit{acquired} slice thickness. All other manufacturers report the \textit{acquired} slice thickness in the slice thickness field when slice interpolation is used.

Slice interpolation can only be invoked on 3D acquisitions where, by definition, contiguous slices are acquired. Slice interpolation is completely analogous to applying partial-Fourier techniques in the second phase-encoding direction, the slice-select direction, in 3D acquisitions. Unlike the other techniques mentioned earlier for reducing scan time, slice interpolation has no SNR penalty. In fact, it boosts SNR by acquiring thicker slices—SNR is higher because there are more hydrogen nuclei per voxel in thicker slices—and then interpolating the signal in intervening slices. The only penalty in using slice interpolation is that there is more partial volume effect in acquired slices, since...
they are thicker. In reconstructed images in orthogonal or angled planes or MIPs, however, slice interpolation is preferable to reducing scan time by merely using thicker slices without interpolation.

Abbreviated breast MRI can make use of one, or any combination of all three, of these time-saving techniques to ensure that the T1-weighted pre- and postcontrast series are each less than 2 minutes in duration.

5.3.3 Inclusion of All Breast Tissues

While inclusion of all breast tissues seems an obvious requirement in any screening exam, it is surprising how frequently the FOV is improperly selected in dynamic breast MRI exams. The goal is to include all breast tissues, to make sure no breast cancers are excluded in the screening exam, while restricting the FOV as much as possible based on body habitus. This is particularly important in the axillae, where breast cancers often occur (▶ Fig. 5.11 a). Technologists sometimes exclude superior breast tissue in an effort to keep scan times reasonable, since more slices in 3D acquisitions means longer scan times, while keeping slice thickness reasonably small to avoid excessive partial volume effects.

On the other hand, sites frequently fail to adjust the FOV and slice range to the body habitus of the woman being examined (▶ Fig. 5.11 b-d). Since pixel size is determined by the FOV and the in-plane matrix in each in-plane direction (the frequency-encoding and phase-encoding matrix) by the formula

\[
\text{Pixel Size (freq)} = \frac{\text{FOV(freq)}}{\text{number of matrix elements in the frequency-encoding direction}}
\]

\[
\text{Pixel Size (phase)} = \frac{\text{FOV(phase)}}{\text{number of matrix elements in the phase-encoding direction}}
\]

it is important that the FOV be adjusted to each patient to be as small as possible while including all breast tissue.

5.3.4 In-Plane Spatial Resolution (Pixel Size) of 1 mm or Less

Submillimeter in-plane spatial resolution has been shown by Kuhl et al to increase both the positive and negative predictive values of breast MRI compared to lower spatial resolution acquisitions.23 The fine morphologic detail enabled by submillimeter acquisitions yields better visualization of spiculations, irregular lesion margins, and rim enhancement sometimes masked by lower spatial resolution imaging. To achieve submillimeter in-plane spatial resolution, the matrix size in both frequency-encoding and phase-encoding directions should be greater than the FOV in units of mm. For example, when a square 320-mm FOV is used, the acquisition matrix should be 320 × 320 (resulting in 1.00 mm × 1.00 mm pixels) or greater (resulting in submillimeter pixels). Keeping the FOV as small as possible while including all breast tissue improves spatial resolution by minimizing pixel size for a given acquisition matrix size (▶ Fig. 5.12).
5.3.5 Acquired Slice Thickness of 3 mm or Less

Slice thickness is important because it sets the limit on the smallest lesions that can be imaged without slice partial volume effects, which decrease lesion contrast. While thicker slices may not impair the conspicuity of high-contrast focal lesions that enhance dramatically, thicker slices can compromise the detection of smaller, low-contrast lesions. To image a lesion of a given diameter without partial volume effects (which would decrease its contrast relative to surrounding tissues), slice thickness of half the lesion’s diameter or less should be used. For example, to avoid partial volume effects of a 5-mm enhancing lesion, a slice thickness of 2.5 mm or less should be used (▶ Fig. 5.13 a,b). Limiting slice thickness is particularly important in minimizing partial volume effects on diffuse, non–mass-like enhancing lesions, such as those often occurring in DCIS, which have lower lesion-to-background contrast on postcontrast and subtracted images (▶ Fig. 5.13 c).

Both the in-plane spatial resolution requirement of pixel sizes being 1.0 mm or less and the slice thickness requirement of 3.0 mm or less are specified in the American College of Radiology (ACR) Breast MRI Accreditation Program for dynamic breast MRI and are reasonable quality requirements for abbreviated breast MRI.24

Fig. 5.11 Examples of improperly selected field of view (FOV). (a) Unilateral exam where a 15 cm × 15 cm FOV was selected, leading to exclusion of axillary tissue. (b) Sagittal exam where an excessive 28 cm × 28 cm FOV was used where a 20 cm × 20 cm FOV would have been more appropriate, as shown in (c). (d) Postcontrast transaxial series where an excessive 36 cm × 36 cm FOV was used. A smaller FOV (e.g., 30 cm × 30 cm) would have included all breast tissue and resulted in improved spatial resolution for the same acquisition matrix.

Fig. 5.12 (a) Precontrast and (b) postcontrast transaxial series where an excessive 36 cm × 36 cm field of view (FOV) was used, leading to an excessive pixel size of 1.24 mm in the horizontal (phase-encoding) direction. A smaller FOV (e.g., 29 cm × 29 cm) could have included all breast tissues and resulted in submillimeter pixels, which in turn would have yielded improved detail in enhancing lesion margins.
5.3.6 Adequately High SNR to Detect Small (2–3 mm Diameter) Vessels in MIP Reconstructed Images

Being able to see small enhancing vessels (2–3 mm in diameter) on MIP images is an excellent surrogate for the ability to display small enhancing lesions on subtracted or MIP images. Failure to demonstrate relatively small blood vessels on subtracted or MIP images gives the radiologist little confidence that small, subtle enhancing lesions will be well demonstrated, when present.

There are a number of ways that the pre- and postcontrast series in abbreviated breast MRI can fail to detect small vessels and small enhancing lesions. One way is to have technical problems that compromise SNR in either the pre- or postcontrast series (Fig. 5.14). Excess noise in either series will make subtracted images, and MIP images based on subtracted images, excessively noisy, which will mask small vessels and small or diffusely enhancing lesions.

Another way that abbreviated MRI can miss enhancing lesions is to make voxels so small that SNR is compromised. This can occur either by using too high a matrix, so that in-plane pixels size is excessively small, or by using slices that are too thin. Both will compromise SNR to the point that enhancing lesions become difficult to detect (Fig. 5.15). Another way to compromise SNR is to fail to select the breast coil as the receiver coil, which will result in the system defaulting to the body coil as the receiver and lower SNR by a factor of 5 to 10 compared to dedicated breast coils (Fig. 5.2). Lack of fat suppression or poor fat suppression in pre- and postcontrast series and misregistration of pre- and postcontrast images can lead to excessive structured noise in subtracted and MIP images, also leading to inadequate detection of small vessels and enhancing lesions (Fig. 5.16).

▶ Fig. 5.3 shows an example of adequate SNR in pre- and postcontrast images, along with reasonable fat suppression, good image registration, and high (submillimeter) spatial resolution, leading to subtracted and MIP images that show small vessels...
Fig. 5.15 Precontrast (a), postcontrast (b), and subtracted (c) images of one slice of a 3D transaxial acquisition revealing a small, enhancing invasive ductal carcinoma adjacent to the chest wall in the central left breast (shown on the right in each image). The acquisition was done with such a high matrix and such thin slices that signal-to-noise ratio per voxel was compromised, leading to low conspicuity of the enhancing cancer in subtracted images and no visibility of either the enhancing cancer or small vessels within the breast in the maximum intensity projection reconstruction of the entire volume (d).

Fig. 5.16 Precontrast (a), postcontrast (b), and subtracted (c) images of one slice of a 3D transaxial acquisition revealing an enhancing lesion in the lateral aspect of the left breast (shown on the right). While pre- and postcontrast images have adequate signal-to-noise ratio, signal nonuniformities and lack of fat suppression in the acquired images resulted in significant structured noise in subtracted images and the maximum intensity projection (MIP) image (d) reconstructed from the 3D stack of subtracted images, obscuring the enhancing lesion and all but the largest vessels within the breast in the MIP image.

and enhancing lesions well. Fig. 5.17 provides an example of adequate SNR and reasonable fat suppression, but borderline spatial resolution, leading to subtracted and MIP images that display strongly enhancing focal lesions well, but lack the visibility of smaller vessels in MIP images that suggest that more subtle, non-mass-like lesions will be well detected in MIP images. The main point is that the ability to detect small vessels in MIP images adds confidence that small or weakly enhancing lesions also will be detected, while consistent lack of small vessels in MIP images suggests that improved acquisition techniques are needed to successfully perform abbreviated breast MRI.
5.4 Contrast Agent Considerations

All MRI contrast agents affect tissue signals by shortening T1 (the spin-lattice relaxation time) and T2 (the relevant spin-spin relaxation parameter in a spin-echo series) or T2* (the relevant spin-spin relaxation parameter in a gradient-echo series). Most clinical studies using Gd-based MRI contrast agents perform T1-weighted imaging instead of T2- or T2*-weighted imaging because Gd-based agents have a stronger fractional effect on shortening T1 values than on shortening T2 or T2* values.20 Another reason is that shorter T1 in T1-weighted imaging causes enhancing lesions to be brighter, making detection easier, while shorter T2 or T2* in T2- or T2*-weighted imaging causes lesions to become darker, making detection more difficult in a background of heterogeneous breast tissue.6 Consequently, this discussion focuses on the effects of Gd-based contrast agents on T1, although it important to note that any paramagnetic MRI contrast agent shortens both T1 and T2 or T2*.

Relaxivity describes a Gd-based contrast agent’s effect on T1 and T2 (or T2*) per molar concentration of contrast agent.25 The relaxation rate (R1) is the inverse of T1, so the shorter the T1, the larger the R1, and the brighter the enhanced tissue on a T1-weighted sequence. The relaxivity (r1) is the change in R1 per unit concentration of contrast agent. Since T1 is measured in seconds, R1 is measured in inverse seconds and r1 in inverse seconds per millimole per liter of contrast agent (L/(mmol × s)). r1 is determined by measuring R1 at various contrast agent concentrations and calculating the change in R1 for a change in agent concentration (the slope of the plot of R1 vs. Gd-chelate concentration). A higher relaxivity corresponds to a greater ability of an agent to enhance tissues or fluids that take up contrast agent per unit concentration of agent.

Currently, seven general-use extracellular fluid Gd-based agents are used for breast MRI (▶ Table 5.1).26,27,28,29,30,31,32 Only one, Gadavist, has been FDA approved specifically for breast MRI as an indication.31 The other extravascular agents shown in ▶ Table 5.1 are approved for central nervous system applications, but are used off-label for breast MRI. The five FDA-approved “conventional” agents (Magnevist, ProHance, Omniscan, OptiMark, and Dotarem) are similar in terms of their concentrations and their relaxivities, which are in the range of 3.6 to 4.7 L/(mmol × s) at 1.5T. Of this group, gadoterate (Dotarem) was approved most recently, in 2013. This agent has properties similar to the other conventional agents and an r1 relaxivity of 3.6 L/(mmol × s).

Fig. 5.17 Precontrast (a), postcontrast (b), and subtracted (c) images of one slice of a 3D transaxial acquisition revealing an enhancing lesion in the left breast (shown on the right). Acquisition parameters included fat suppression, with in-plane pixel sizes of 1.03 mm in each direction, resulting in good signal-to-noise ratio, but with lesion margins and vessels slightly less well resolved in subtracted images and the maximum intensity projection image (d) than in the case shown in ▶ Fig. 5.3, where submillimeter pixels were used.
Gadobenate (MultiHance) and gadobutrol (Gadavist) have been FDA approved since 2004 and 2011, respectively. They differ from the five conventional agents in their physicochemical properties. All of the general-purpose Gd-based contrast agents are formulated at molar concentrations (or molarities) of 0.5 mol/L (or mmol/mL) except gadobutrol (Gadavist), which is formulated at a molar concentration of 1.0 mol/L or 1.0 mmol/mL. This means that instead of administering 1 mL per 11 lb of body weight, which is the labeled dose of the six other agents, Gadavist should be administered at a labeled dose of 1 mL per 22 lb of body weight (see two paragraphs below for a more detailed explanation).

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<table>
<thead>
<tr>
<th>Brand name (Manufacturer), FDA approval date</th>
<th>Generic name</th>
<th>Approved indications</th>
<th>Approved dose</th>
<th>Molarity (M/L)</th>
<th>Protein interaction</th>
<th>r1 relaxivity (L/mmol s) at 1.5 T in plasma at 37 °C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Magnevist (Bayer Healthcare), 1988</td>
<td>Gadopentetate dimeglumine</td>
<td>CNS, extracranial/extraspinal tissues, body (excluding heart) in adults and children ≥ 2 y of age</td>
<td>0.1 mmol/kg</td>
<td>0.5</td>
<td>None</td>
<td>3.9–4.1</td>
</tr>
<tr>
<td>ProHance (Bracco), 1992</td>
<td>Gadoteridol</td>
<td>CNS in adults and children ≥ 2 y of age; extracranial/extraspinal tissues in adults</td>
<td>Adults: 0.1 mmol/kg 2nd dose of 0.2 mmol/kg up to 30 min after first dose</td>
<td>0.5</td>
<td>None</td>
<td>4.1</td>
</tr>
<tr>
<td>Omniscan (GE Healthcare), 1993</td>
<td>Gadodiamide</td>
<td>CNS, extracranial/extraspinal tissues, body (excluding heart) in adults and children 2–16 y of age</td>
<td>0.1 mmol/kg</td>
<td>0.5</td>
<td>None</td>
<td>4.3</td>
</tr>
<tr>
<td>OptiMARK (Covidien), 1999</td>
<td>Gadoversetamide</td>
<td>CNS, liver in adults</td>
<td>0.1 mmol/kg</td>
<td>0.5</td>
<td>None</td>
<td>4.7</td>
</tr>
<tr>
<td>MultiHance (Bracco), 2004</td>
<td>Gadobenate dimeglumine</td>
<td>CNS in adults and children ≥ 2 y of age; MRA in adults with known or suspected renal or aortoiliacocalvascular disease</td>
<td>0.1 mmol/kg</td>
<td>0.5</td>
<td>Weak</td>
<td>6.3–7.9</td>
</tr>
<tr>
<td>Gadavist (Bayer Healthcare), 2011</td>
<td>Gadobutrol</td>
<td>CNS in adults and children ≥ 2 y of age; breast for detection and characterization of disease, assessment of local extent of disease, and guidance for localization and biopsy</td>
<td>0.1 mmol/kg</td>
<td>1</td>
<td>None</td>
<td>5.2</td>
</tr>
<tr>
<td>Dotarem (Guerbet), 2013</td>
<td>Gadoterate meglumine</td>
<td>CNS in adults and children ≥ 2 y of age</td>
<td>0.1 mmol/kg</td>
<td>0.5</td>
<td>None</td>
<td>3.6</td>
</tr>
</tbody>
</table>

Abbreviations: CNS, central nervous system; FDA, Food and Drug Administration.
The relaxivities of these agents reflect the most important clinical differences among them, given that relaxivity is the property most closely related to the signal intensity difference between enhanced lesions and background tissues relative to background noise levels (the CNR of enhancing lesions). Compared to the five conventional agents, gadobutrol (Gadavist) has an r1 relaxivity approximately 20% higher than the other agents (*Table 5.1), while gadobenate (MultiHance) has nearly twice the relaxivity of conventional agents, approximately 6 to 8 L/mmol/s at 1.5 T, depending on the methods used to measure relaxivity. This higher relaxivity has been demonstrated to increase signal-to-background-noise ratios relative to conventional agents by 25 to 35%, an increase similar to that found when going from a single dose to a double dose of a conventional Gd-based contrast agent. Importantly, gadobenate (MultiHance) has been shown to provide better diagnostic performance for MRI than the other Gd-based contrast agents in a variety of applications, including central nervous system (CNS), liver, vasculature, and breast cancer detection.

Some practical issues with regard to contrast agent administration include that all agents should be administered according to the labeled doses, all of which are based on patient body mass and all of which have label recommended doses of 0.1 mmol/kg. The volume of agent to be drawn and administered is slightly more complicated because MRI contrast agents are prepared in two different molar concentrations. Molar concentration is the number of moles of Gd-chelate per liter of solution (mol/L), or molarity. All approved extracellular MRI agents have a molarity of 0.5 mol/L (or millimol per milliliter, mmol/ml) except Gadavist, which has a molarity double that of the other agents, of 1.0 mol/L (or mmol/ml). Therefore, to deliver a labeled dose of 0.1 mmol/kg of body mass, all extracellular MRI contrast agents except Gadavist should be administered based on body mass at a dose of 2 mL per 10 kg or based on body weight, at a dose of 1 mL for each 11 lb, since 1 kg of mass weighs 2.2 lb in Earth’s gravity. Gadavist, on the other hand, should be administered at 1 mL/10 kg, which equals 1 mL for each 22 lb of body weight, due to its higher molarity of 1.0 mol/L.

All MR contrast agent administrations should be followed by a flush of at least 20 mL of 0.9% saline solution to ensure that contrast agent is washed out of the tubing and arm and into the woman’s major vessels. Since the goal of abbreviated breast MRI is to perform imaging quickly, all administrations of contrast agent and saline flush should be done with a power injector. This is particularly important with higher viscosity agents such as MultiHance.

A practical issue in contrast agent use is that higher relaxivity agents cause greater lesion enhancement for the same labeled dose of contrast agent. If computer-aided display (CAD) systems with color maps are used, it may be necessary to raise the threshold at which color is added to display the presence of rapidly enhancing lesions when a higher relaxivity agent is used. Additionally, because dynamics are not captured in abbreviated breast MRI, there will be no color distinction between rapidly enhancing lesions that have continuous uptake, plateau, or washout. It remains to be seen how breast MRI CAD systems will accommodate abbreviated breast MRI protocols.

In the proof-of-concept study by Kuhl et al, both contrast agent and saline flush were administered at a rate of 3 mL/s. A rate of at least 2 mL/s is advisable to ensure rapid administration of agent and rapid appearance of agent in the postcontrast series. Like conventional breast MRI, the technologist should always check for the presence of contrast agent in the heart and great vessels in postcontrast images to ensure that contrast agent was successfully injected into the patient’s circulatory system. If contrast agent is not seen in postcontrast images, the technologist should check the injection site for a hematoma-like bulge of contrast agent in the woman’s arm or for a pool of contrast agent on the table under her arm. If this occurs, the woman will need to be re-imaged after appropriate needle placement.

Timing of contrast agent administration relative to pre- and postcontrast image acquisition is particularly important in abbreviated breast MRI since only a single postcontrast series is performed. To proceed as quickly as possible, contrast agent administration should begin immediately after the precontrast series ends. It is advisable for the technologist to prepare the patient for contrast administration prior to starting the precontrast series, not between pre- and postcontrast series, both to minimize the total procedure time and to avoid startling the patient and causing misregistration between pre- and postcontrast image sets.
A final timing consideration is whether the single postcontrast scan should be started just as contrast agent administration begins (without a gap), as contrast administration ends (which would lead to a variable gap of 5–10 s depending on the quantity and rate of administration), or after an additional timing gap to ensure that adequate time has elapsed to maximize detection of enhancing lesions in the single postcontrast series. The proof-of-concept study conducted by Kuhl et al began the postcontrast series after injection of contrast agent at a rate of 3 mL/s and during injection of the saline flush at the same injection rate. Based on a typical body weight of 150 lb (or 68 kg of body mass), a 0.5 molarity agent would require 14 mL of injected agent, while a 1.0 molarity agent (Gadavist) would require 7 mL of injected agent. At a rate of 3 mL/s, this would require about 5 s for injection of the 0.5 molarity agent or 2.5 s for the 1.0 molarity agent, causing a minimum delay between the end of the precontrast and start of the postcontrast series.

Previous work has determined that peak lesion enhancement relative to background parenchymal tissue signal occurs at 1 to 2 minutes after contrast injection for focal lesions and slightly longer than 2 minutes after injection for non–mass-like lesions. Maximum contrast weighting of 3D gradient-echo pulse sequences occurs about one-third of the way into the scan for Siemens systems and about halfway into the scan for most other manufacturers’ systems. Maximum contrast weighting occurs midway through most multislice 2D acquisitions, as well. Without adding a time gap between the end of injection and the start of the postcontrast series, a postcontrast series of 1- to 2-minute duration, as recommended for abbreviated breast MRI, will have peak contrast weighting at 20 to 60 s from the start of the sequence. The proof-of-concept study for abbreviated breast MRI had a series scan time of 80 s, with peak contrast weighting occurring about 40 s after the end of contrast injection, resulting in a high sensitivity to breast cancer. This suggests that with manufacturers’ current pulse sequences, scans of 80- to 120-s duration per series beginning just after completing rapid injection of contrast agent and bolus should have sufficient enhancing-lesion-to-background contrast to reveal enhancing cancers. It would be optimal in abbreviated MR imaging to have specially designed pulse sequences that place the peak contrast weighting of the pulse sequence (the low spatial frequency acquisitions at or near the center of k-space) at or near the end of the abbreviated pulse sequence, so that maximum contrast weighting would occur at about 90 s after the end of contrast injection. However, this would require MRI manufacturers to provide specialized fast gradient-echo series for abbreviated MRI, customizing peak contrast weighting to be at or near the end of the series, so that the time of peak uptake in most enhancing cancers (about 90 s after end of injection) would occur at the peak contrast weighting of the series, rather than one-third to halfway into the series.

5.5 Open Questions about Abbreviated Breast MRI

Abbreviated breast MRI has been examined in only a few studies, the most complete of which has been conducted by Kuhl et al, at a single institution using 2D multislice gradient-echo imaging without fat saturation and with breast immobilization. Their study showed that their approach to abbreviated breast MRI has the same high sensitivity, high specificity, and high negative predictive value as a full diagnostic breast MRI study. Other more recent studies of abbreviated breast MRI using 3D techniques have focused on sensitivity rather than specificity, recall rates, and negative predictive values, so the effects of 3D fat-suppressed techniques on these additional important measures of accuracy beyond sensitivity are yet to be determined. It is also yet to be determined whether breast immobilization has a significant effect on the accuracy of abbreviated breast MRI, although these subsequent studies using 3D fat-suppressed techniques without breast immobilization suggest that sensitivity is not reduced significantly with fat suppression without breast immobilization. Another open question is whether the accuracy of abbreviated breast MRI shown by Kuhl et al will be maintained with a broader range of acquiring MRI sites using a variety of acquisition techniques and across a wider range of interpreting physicians.

5.6 Conclusion

Abbreviated breast MRI, like dynamic breast MRI, depends on having adequate SNR, high in-plane spatial resolution to better resolve lesion margins, thin slices to minimize partial volume effects, good breast positioning to minimize patient motion and image misregistration, and consistent contrast agent administration based on body mass or
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weight, with the additional requirement of completing imaging in less than 5 minutes. Careful attention to the visibility of small vessels in MIP reconstructions of subtracted series gives the radiologist confidence that abbreviated breast MRI will detect not only larger strongly enhancing focal lesions, but also more subtle non-mass-like lesions, providing high sensitivity to breast cancers including in situ cancers.

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