INTRODUCTION

Early breast magnetic resonance imaging (MRI) studies conducted in the early to mid-1980s attempted to distinguish malignant breast lesions from benign lesions and normal breast tissues based on inherent tissue T1 and T2 values.\(^1,2\) Malignant breast lesions were found to have higher T1 and T2 values than normal breast tissues, but shorter T1 and T2 values than most benign breast lesions such as fibroadenomas.\(^3-5\) Significant overlap in both T1 and T2 values between benign and malignant breast lesions, however, discouraged the use of noncontrast breast MRI for cancer detection and diagnosis. It is now an accepted standard that high sensitivity to breast cancer requires contrast-enhanced breast MRI both without and with a gadolinium (Gd)-based paramagnetic contrast agent to identify enhancing lesions.\(^6-10\)

Most recent studies of contrast-enhanced breast MRI have reported sensitivities to breast cancer between 90% and 100%, depending on the subject cohort, imaging techniques, including other imaging tests performed along with breast MRI, and criteria for breast cancer.\(^11-16\)

One of the limitations of contrast-enhanced breast MRI has been the lack of standardized imaging protocols, contrast agent administration, image postprocessing, and image review. A major step forward has been standardization of breast MRI reporting terminology through publication of the fourth edition of the American College of Radiology’s (ACR’s) Breast Imaging Reporting and Data System (BI-RADS), which included reporting terminology for breast MRI and breast ultrasound, as well as mammography.\(^17\) Another recent step forward has been the initiation of the ACR’s Breast MRI Accreditation Program, which is described briefly at the end of this article.

This article provides specific recommendations for achieving high-quality breast MRI. Because of different MRI hardware, software, and scanning capabilities, it is not possible to achieve complete uniformity of protocols, but fairly specific guidelines for equipment requirements and scanning protocols are given for performing bilateral contrast-enhanced breast MRI with high spatial resolution, good temporal resolution, and high signal-to-noise ratio (SNR). This article describes the technical parameters needed to achieve...
consistently high-quality contrast-enhanced breast MRI.

**EQUIPMENT REQUIREMENTS**

MRI systems used for breast cancer detection and diagnosis should include (1) adequate magnetic field strength with good magnetic field homogeneity across both breasts, (2) adequate magnetic field gradients to permit fast gradient-echo imaging, (3) a bilateral breast coil enabling prone positioning, and (4) good fat suppression over both breasts.

These requirements are discussed individually.

**Adequate Magnetic Field Strength and Homogeneity**

The magnetic field strength of whole-body MRI systems ranges from 0.064 T to 8 T (1 T = 10,000 G; the naturally occurring magnetic field at the earth’s surface ranges from 0.25 to 0.65 G). MRI systems approved by the Food and Drug Administration (FDA) for clinical use have magnetic field strengths up to 3.0 T. Above a few tenths of a Tesla, image SNRs per voxel go up approximately linearly with magnetic field strength, if receiver coil design, voxel size, and imaging parameters other than field strength remain constant.18 Thus, higher magnetic field strength (B0) should provide higher SNR per voxel for breast imaging for the same pulse sequence, although the linear increase in SNR with field strength is moderated somewhat by the increase in tissue T1 values at higher field strengths. T1 values increase by about 20%, going from 1.5 T to 3.0 T.19

Most breast MRI is done on 1.5-T scanners, with a few sites performing breast MRI at 1.0 to 1.2 T and a growing number of sites performing breast MRI at 3.0 T. Although SNR per voxel is nearly doubled for the same pulse sequence at 3.0 T, compared with 1.5 T, 3.0 T systems have some additional technical challenges.20 It is more difficult to get uniform fat suppression on 3.0-T systems than on 1.5-T systems (Fig. 1). In addition, artifacts are often more pronounced at 3.0 T than at 1.5 T. Most importantly, higher-frequency radio waves used for tissue excitation are more highly attenuated. Because 3.0-T systems have double the resonant frequency of 1.5-T systems, the penetration of radiofrequency (RF) waves transmitted to excite breast tissues, the B1-field, is less uniformly distributed within breast tissues at 3.0 T because of greater absorption by external tissues. This causes nonuniformity of signal excitation and, thus, nonuniform measured signals.

Kuhl and colleagues21 compared contrast-enhanced breast MRI at 1.5 T and 3.0 T in the same group of 37 patients. Overall image quality scores were slightly higher and differential diagnosis of enhancing lesions was made with greater confidence at 3.0 T, as shown by larger areas under the receiver operating characteristic curve.21 The investigators pointed out, however, that technical problems exist at 3.0 T beyond those observed at 1.5 T.22 These included increased nonuniformity of transmitted B1 RF waves, particularly between left and right breasts with the larger field-of-view (FOV) used for transaxial scanning. This in turn led to reduced enhancement of lesions located in “low B1 areas.”22 The investigators used a 2-dimensional (2D) gradient-echo pulse sequence, pointing out that the adverse effects of low B1 areas on lesion enhancement should be reduced with the 3D (volume) sequences more commonly used in the United States because of the shorter repetition times (TRs) used in 3D imaging. Others have pointed out that B1 nonuniformities can be reduced by using 3D techniques, careful choice of flip angle to match the TR of the imaging sequence, smaller FOV (eg, sagittal rather than transaxial acquisitions), and optimized shimming of the acquisition volume.23

Another reason for performing breast MRI at magnetic field strengths of 1.0 T or greater, beyond higher SNR, is to ensure higher static magnetic field homogeneity over the entire imaged volume. High magnetic field homogeneity for breast imaging means that the static magnetic field strength (B0) should remain nearly constant across both breasts, including the chest wall and axillae. Because hydrogen nuclei in water and fat differ in resonant frequencies by 3.4 parts per million (ppm), the magnetic field homogeneity must be significantly less than 3.4 ppm to achieve good chemically selective fat suppression of hydrogen signals from fat, while preserving hydrogen signals from water.

The standard criterion to ensure that chemically selective fat suppression is effective is that the magnetic field strength should vary by less than 1 ppm over the entire volume of tissue being imaged. At 1.5 T, a nonuniformity of 1 ppm would amount to a magnetic field difference of 1.5 μT (microTesla), or a resonant frequency difference of 63.9 Hz, compared with the water-fat frequency difference of 224 Hz (3.4 ppm). The static magnetic field should be homogeneous to this level across a FOV 30 to 35 cm in diameter encompassing both breasts. This is generally not possible for low-field to midfield scanners (less than 1.0 T), and is a challenge even for high-field scanners, as the location...
of the breasts in prone-positioned breast MRI typically is not at the isocenter of the magnet. Instead, in most magnets, the breasts are below isocenter to allow prone positioning of the patient, with breasts in the breast coil, and to allow adequate space for the patient’s torso in the magnet bore.

A Bilateral Breast Coil with Prone Positioning

Bilateral imaging is recommended for the following reasons: (1) Clinical comparison of both breasts is as important in breast MRI as it is in mammography. Bilateral comparison helps identification of focal enhancement and helps prevent overcalling of physiologic enhancement, which tends to occur bilaterally, especially in premenopausal women and postmenopausal women on hormone replacement therapy.24,25 (2) Data from recent breast MRI studies indicate that when a breast cancer occurs, there is a 3% to 5% chance that breast MRI will detect a mammographically occult cancer in the contralateral breast.26–31 (3) Unilateral imaging in the transaxial or coronal plane can incur image wrap (or aliasing) artifacts from the contralateral breast, especially if phase encoding is set left-to-right (as it typically is in transaxial imaging), a bilateral breast coil is used, and the field-of-view is narrowed to include only the breast being imaged.

Bilateral breast imaging is typically performed using the body coil as the RF-transmit coil and a prone-positioned bilateral breast coil as the RF-receiver coil. A few systems, such as the Aurora breast MRI system, have bilateral breast coils serving as RF-transmit-receive coils. Modern breast coils, whether receive-only or transmit-receive coils, have multiple-channel elements. Bilateral breast coils currently have between 2 channels (1 channel for each breast) and 18 channels (9 channels for each breast). In multichannel coils, the received signals are recorded simultaneously using multiple amplifier and analog-to-digital converters. Multiple-channel receiver elements require a scanner capable of simultaneously recording multiple channels of data, so it is important to make sure that scanner hardware and software can accommodate the number of channels in the breast coil.

Multiple receiver channels were developed to boost coverage and signal uniformity, but acquired a single dataset for image reconstruction. A technique developed over the past decade, parallel imaging, modifies data acquisition so that different channels or sets of channels simultaneously acquire different datasets simultaneously.32,33 In parallel imaging, each coil element (or set of coil elements) acquires different pieces of the image simultaneously, or in parallel; then, more complex image reconstruction techniques are used to recombine the different partial datasets into planar images or volumes. Parallel imaging also requires a short prescan to map out the sensitivity profile of each coil element (or set of coil elements) on each patient. This is done in a separate acquisition on some scanners and within the parallel imaging pulse sequence itself on other scanners. Some parallel imaging techniques acquire multiple channels of data in physical space, whereas others acquire multiple channels of data in spatial frequency (or k-) space.32,33 Parallel imaging speeds image acquisition by a prespecified factor (eg, 2, 3, or 4), but requires longer for image reconstruction after all data have been acquired. Parallel imaging typically is done with an acceleration factor of 2, which cuts the acquisition time nearly in half. Use of higher-acceleration factors in breast imaging tends to cause image reconstruction artifacts (Fig. 2) and has been avoided in most clinical practices.
Adequate Magnetic Field Gradients

Magnetic field gradients are produced by additional coils that generate magnetic fields (each pointing along the static magnetic field, B₀) that intentionally vary the magnetic field strength linearly along each of the 3 perpendicular axes: x, y, and z (Fig. 3). Gradient fields are switched on and off rapidly during each repetition of the pulse sequence to spatially resolve the source of signal by briefly altering the precessional frequencies of hydrogen nuclei at different locations as a function of x, y, or z location. The knocking noise heard from MR units as they scan is due to magnetic field gradients being turned on and off.

Two parameters characterize the performance of magnetic field gradients: (1) Maximum gradient strength, expressed in milliTesla per meter (mT/m), plays a role in determining how small voxels can be made. Modern MR scanners have magnetic field gradient strengths of up to 50 mT/m. (2) Gradient rise times describe the time interval needed for a magnetic gradient to go from zero to maximum strength, which in turn determines

Fig. 2. The same slice of sets of 3D gradient-echo sequences acquired on the same volunteer using a 3.0-T scanner without (A) and with (B–D) parallel imaging using acceleration factors (AF) of 2 to 4. All were acquired with the same FOV, matrix, and slice thickness. Total acquisition time for each 3D gradient-echo acquisition covering both breasts was 102 seconds without parallel imaging (A), 57 seconds with AF = 2 (B), 42 seconds with AF = 3 (C), and 34 seconds with AF = 4 (D). Note the presence of parallel imaging reconstruction artifacts in (C) and (D) (arrows). As a result of more artifacts with higher AF values, most sites performing parallel imaging in breast MRI use an AF of 2.

Fig. 3. The x, y, and z gradients shown within the bore of a solenoidal magnet. The main static magnetic field, B₀, points along the z direction, which is also the direction of the magnetic field of each gradient; x, y, or z gradients, when applied, alter the strength of the magnetic field pointing along the z-axis as a function of the x, y, or z direction, respectively.
how quickly pulse sequences can be performed. The shorter the gradient rise time, the shorter TR and echo time (TE) can be made in 2D or 3D gradient-echo imaging. Modern MR scanners have gradient rise times as short as 200 microseconds, yielding TR values as short as 4 ms and TE values as short as 1 ms. Generally, to achieve adequate spatial resolution and short enough imaging times in 3D gradient-echo imaging, TR needs to be shorter than 10 ms and TE needs to be less than 4 ms. In gradient-echo sequences without or with incomplete fat suppression, TE should be carefully chosen to minimize chemical-shift artifacts.33

**Good Fat Suppression Over Both Breasts**

In 2D or 3D MRI, fat suppression is typically achieved by applying a frequency-selective 90° saturation pulse that acts only on the hydrogen nuclei in fat (Fig. 4). In 2D imaging, this saturation pulse is applied to each slice at the start of each pulse sequence repetition; in 3D (volume) imaging, this saturation pulse is applied to the entire volume of tissue within the RF-transmit coil. If applied uniformly across both breasts, the fat suppression pulse effectively eliminates fat signal from signal measured during the pulse sequence repetition that follows. Fig. 5 shows examples of T1-weighted bilateral breast MRI without fat suppression, with good fat suppression, and with incomplete fat suppression. Fat suppression is useful in contrast-enhanced breast MRI because it reduces the signal from fat in both precontrast and postcontrast scans. In postcontrast scans, lack of fat suppression makes it more difficult to separate enhancing breast lesions from fat, because fat and enhancing breast lesions have similar signal intensities. In subtracted images (postcontrast images minus precontrast images), even a small amount of motion or misregistration between precontrast and postcontrast scans causes structured noise artifacts that complicate interpretation and, in some cases, simulate enhancing lesions. Good fat suppression in both precontrast and postcontrast images minimizes the structured noise of misregistration artifacts in subtracted images, allowing detection of smaller enhancing lesions or non-masslike lesions with greater reliability.

**PULSE SEQUENCE REQUIREMENTS**

Beyond good equipment, high-quality breast MRI requires the use of appropriate pulse sequences. Breast MRI pulse sequences should include several noncontrast series, performed before contrast administration, along with a multiphase series of pulse sequences applied just before and several times after contrast agent administration. Recommended pulse sequences include (Fig. 6):

1. Scout images obtained in transaxial, sagittal, and coronal planes.
2. A T1-weighted non-fat-saturated series obtained bilaterally, including axillae and chest wall, to distinguish fat from water-based tissues including fibroglandular tissue, cysts, lymph nodes, and other benign lesions, muscle, and cancers.

![Fig. 4. Schematic of the resonant frequency difference between hydrogen nuclei in fat and hydrogen nuclei in water at 1.5 T. The MR scanner's center frequency is tuned to the resonant frequency of hydrogen nuclei in water. When fat-saturation is selected, a saturation pulse is applied with a narrow frequency range to cancel the signal from hydrogen nuclei in fat molecules, which resonate at about 220 Hz (1 Hz = 1 cycle per second) lower frequency than the hydrogen nuclei in water at 1.5 T. At 3.0 T, the frequency shift between hydrogen fat and water peaks doubles to about 440 Hz.](image)
3. A T2-weighted fat-saturated series obtained bilaterally to distinguish cysts from solid lesions. A STIR (short inversion time [TI or τ] inversion recovery) series can be used in place of a T2-weighted series (Fig. 7) if TI is set correctly to minimize the signal from fat. A TI of about 180 ms at 1.5 T, or about 215 ms at 3.0 T, should do a good job of suppressing fat signal. Good fat-suppression is important in either T2-weighted or STIR images, so that the brightest tissues in the image are fluid-filled cysts or blood vessels.

4. A multiphase 3D Fourier transform (3DFT or volume acquisition) gradient-echo T1-weighted pulse sequence acquired once before and multiple times after contrast agent administration, preferably with chemically selective fat-suppression, is used to identify the vascular bed and any enhancing lesions in the breast. T1-weighting is achieved by setting the pulse sequence TR short relative to the T1-values of tissues being imaged, setting the TE as short as possible, and using a flip angle that is relatively small and based on the TR value to optimize SNR.

Any modern MR scanner should be able to deliver the first 3 pulse sequences without difficulty. Scout images acquired in all 3 perpendicular planes are routine and should take less than 1 minute to acquire and display. Both T1-weighted non-fat-saturated and T2-weighted fat-saturated series can be obtained using accelerated spin-echo sequences, called fast spin-echo (FSE) or turbo spin-echo (TSE) sequences, in a time of less than 3 to 4 minutes for each series. If parallel imaging with an acceleration factor of 2 can be applied to these noncontrast series, scan times can be decreased by nearly a factor of 2, to about 2 minutes per series.

The key pulse sequence for breast cancer detection and lesion characterization is the multiphase 3D gradient-echo T1-weighted series acquired before and several times after MR contrast-agent
Stronger gradients permitting very short TR and TE values, along with multichannel coils and scanner software permitting parallel imaging, have sped multiphase acquisitions, allowing improved spatial resolution by using a higher matrix (that is, more phase-encoding and frequency-encoding steps), while maintaining adequate SNRs per voxel. This has enabled breast MRI to meet all of the spatial resolution and temporal resolution goals listed below when proper pulse sequence techniques are used.

The important features of a contrast-enhanced multiphase T1-weighted series are as follows:

1. Consistent Gd-chelate contrast agent administration based on patient mass or weight: 0.1 mmol/kg, followed by a 20-mL saline flush.
2. Bilateral acquisition with prone positioning.
3. A multiphase 3D gradient-echo T1-weighted pulse sequence (1 precontrast and multiple postcontrast series extending at least 6 minutes after contrast injection).

administration. Stronger gradients permitting very short TR and TE values, along with multichannel coils and scanner software permitting parallel imaging, have sped multiphase acquisitions, allowing improved spatial resolution by using a higher matrix (that is, more phase-encoding and frequency-encoding steps), while maintaining adequate SNRs per voxel. This has enabled breast MRI to meet all of the spatial resolution and temporal resolution goals listed below when proper pulse sequence techniques are used.
4. Adequately thin slices of 3 mm or less.
5. Pixel sizes of less than 1 mm in each in-plane direction.
6. Phase-encoding direction chosen to minimize artifacts across the breasts.
7. Total acquisition time for each series (or “phase” of the multiphase series) of 1 to 3 minutes.
8. Adequate SNR to visualize small enhancing vessels on 3D maximum intensity projection (MIP) images.

Each of these items is described in more detail in the following sections.

**Gd-chelate Contrast Agent Administration: 0.1 mmol/kg Followed by 20 mL of Saline**

Although MR contrast agents are not labeled specifically for breast cancer detection, use of an appropriate contrast agent is essential for high sensitivity to breast cancer. There are 6 MR contrast agents that are FDA-approved for use in the brain and spine (Table 1) and suitable for breast MRI. All are labeled for a recommended dose of 0.1 mmol/kg of patient body mass. All but one (Gadavist, Bayer Healthcare Pharmaceuticals Inc, Wayne, NJ, USA) are packaged in
a concentration of 0.5 mmol/mL; therefore, in terms of packaged contrast agent volume, the recommended dose is 0.2 mL per kg of body mass. For example, a 140-pound woman has a body mass of 64 kg (140 lb/2.2 lb/kg = 64 kg) and her administered dose of MR contrast agent packaged at a concentration of 0.5 mmol/mL should be 13 mL (64 kg * 0.2 mL/kg = 12.8 mL) rounded to the nearest milliliter. A simple rule to follow to administer label-recommended doses of 0.1 mL/kg of body mass for agents packaged at 0.5 mmol/mL is to inject 1 mL (or 1 cubic centimeter, cc) of contrast agent for every 11 pounds of body weight. Using the previous example, a 140-lb woman should receive 140 lb * (1 mL/11 lb) = 13 mL of Gd-chelate contrast agent.

Gadavist is packaged at a higher concentration of 1.0 mmol/mL, so half a much Gadavist should be administered for a given body mass to achieve a dose of 0.1 mmol/kg of body mass. A simple rule with Gadavist is to administer 1 mL (or 1 cc) of agent for every 22 lb of body weight.

Contrast agent should be administered with a controlled flow rate (most sites use a rate of 2 mL per second) followed immediately by a bolus of 20 mL of saline administered at a similar rate. This is best done with an dual-headed MR-compatible power injector that can administer both contrast agent and saline flush sequentially at controlled flow rates.

### Bilateral Acquisition with Prone Positioning

Prone positioning in a dedicated bilateral breast coil positions the breasts pendently and reduces breast motion due to respiration and cardiac pulsation.

### Table 1

Gd-chelated contrast agents approved for central nervous system indications in the United States (and used for breast cancer detection)

<table>
<thead>
<tr>
<th>Agent</th>
<th>Common Name</th>
<th>Molecular Weight</th>
<th>Molarity, mol/L</th>
<th>Viscosity, cP, 37°C</th>
<th>Relaxivity $\alpha_1$, L/(mmol·s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Magnevist (Gd-DTPA)</td>
<td>Gadopentetate dimeglumine</td>
<td>938</td>
<td>0.5</td>
<td>2.9</td>
<td>4.1–4.9</td>
</tr>
<tr>
<td>Prohance (Gd-HP-DO3 A)</td>
<td>Gadoteridol</td>
<td>559</td>
<td>0.5</td>
<td>1.3</td>
<td>4.1–5.4</td>
</tr>
<tr>
<td>Omniscan (Gd-DTPA-BMA)</td>
<td>Gadodiamide</td>
<td>574</td>
<td>0.5</td>
<td>1.4</td>
<td>4.3–5.4</td>
</tr>
<tr>
<td>Optimark (Gd-DTPA-BMEA)</td>
<td>Gadoversetamide</td>
<td>662</td>
<td>0.5</td>
<td>2.0</td>
<td>4.7</td>
</tr>
<tr>
<td>Multihance (Gd-BOPTA)</td>
<td>Gadobenate dimeglumine</td>
<td>1058</td>
<td>0.5</td>
<td>5.3</td>
<td>6.7–9.7</td>
</tr>
<tr>
<td>Gadavist</td>
<td>Gadobutrol</td>
<td>605</td>
<td>1.0</td>
<td>5.0</td>
<td>5.2</td>
</tr>
</tbody>
</table>

Viscosities are measured in centipoises (cP) at 37°C (viscosity of water is 1.002 cP); Relaxivity $\alpha_1$ is the relaxation rate (the inverse of relaxation time, $T_1$) per unit concentration of agent and is expressed in mmol/L$^{-1}$·s$^{-1}$ or L/(mmol·s). Magnevist (Bayer Healthcare Pharmaceuticals Inc, Wayne, NJ, USA); Prohance (Bracco SpA, Milan, Italy); Omniscan (GE Healthcare, Princeton, NJ, USA); Optimark (Mallinckrodt Inc, St. Louis, MO, USA); Multihance (Bracco SpA, Milan, Italy).

Because the patient is supported by the coil at the sternum, lateral chest, and above and below the breasts, most respiratory and cardiac motion affects chest tissues posterior to the breasts. Any motion between precontrast and postcontrast scans or during scanning causes misregistration in subtracted breast images. By positioning the patient comfortably and by properly instructing the patient before the multiphase T1-weighted series (rather than during the series, such as just before administration of contrast agent), the MR technologist can help minimize breast and patient motion. Keeping total scan time reasonably short (20 minutes or less) will also help decrease patient discomfort and motion during scanning.

A 3D Fourier Transform Gradient-Echo T1-Weighted Pulse Sequence

T1-weighted pulse sequences are used in contrast-enhanced breast MRI because Gd-chelates, while shortening both T1 and T2, cause a greater fractional change in T1 than T2 (or T2*).34 In gradient-echo imaging, T1 weighting is achieved by using a short TR, very short TE, and a relatively low flip angle that is matched to the TR.33 For 3D Fourier transform (3DFT) imaging, extremely short TR values are used to keep the scan times for each phase of the multiphase series reasonably short, ideally 3 minutes or less. Although 2DFT pulse sequences acquire image data from a single plane at a time, 3DFT pulse sequences acquire image data from an entire volume at a time. Multislice 2DFT imaging typically has small gaps between individual slices, with Gaussian slice profiles. The 3DFT imaging acquires contiguous slices within the 3D volume, with rectangular slice profiles, so that no signal gaps occur between slices.

For 3DFT imaging, total acquisition time is $T_{total} = (TR)(N_{pe})(N_{acq})(N_{slices})$, where TR is the basic pulse sequence repetition time, $N_{pe}$ is the number of in-plane phase-encoding steps to resolve signal in a single in-plane direction, $N_{acq}$ is the number of times each phase encoding step is repeated (usually set to 1 in 3DFT imaging), and $N_{slices}$ is the number of slices, which equals the number of phase-encoding steps used to separate the 3D volume in the third (slice-select) direction (in 2DFT imaging, $N_{slices}$ is automatically set to 1). It is because of this additional factor, $N_{slices}$, which can be as high as 160 with slices comparable in thickness to the in-plane pixel size (isotropic voxels), that gradient-echo sequences with very short TR values are needed in 3DFT imaging. The 3DFT sequences have a signal-to-noise advantage over 2DFT sequences because signal is acquired from the entire excited volume of tissue, including both breasts, instead of from just a single plane, at each signal measurement. The 3DFT sequences require more phase-encoding steps (by the factor $N_{slices}$) to resolve not just a plane of tissue, but an entire volume of tissue, into individual voxels.

Adequately Thin Slices of 3 mm or Less

Slice thickness sets the limit on the smallest lesion that can be imaged without slice partial volume effects decreasing lesion contrast. Although slice thickness may not impair visualization of high-contrast lesions that enhance dramatically, it can play an important role in the detection of low-contrast lesions. To image a low-contrast lesion of a given diameter without partial volume effects, which would decrease its contrast relative to surrounding tissues, a slice thickness of half the lesion’s diameter or less should be used. For example, to be sensitive to a low-contrast 5-mm enhancing lesion, a slice thickness of 2.5 mm or less should be used (Fig. 8). Thin slices are particularly important for minimizing partial volume effects on diffuse, non-masslike enhancing lesions, such as those sometimes associated with ductal carcinoma in-situ (DCIS) (Fig. 9).33

Pixel Sizes of Less than 1 mm in Each In-Plane Direction

Pixel sizes smaller than 1 mm in each in-plane direction can be achieved by selecting an acquisition matrix (number of phase-encoding and frequency-encoding steps) that exceeds the FOV (in mm) in both the phase-encoding and frequency-encoding direction. For example, in transaxial imaging with a $30 \times 30$ cm (300 $\times$ 300 mm) FOV, an acquisition matrix of $300 \times 300$ or greater should be used. If a $384 \times 384$ matrix were used for this FOV, each in-plane pixel would be $(300 \, \text{mm})/384 = 0.78 \, \text{mm}$ in each direction, which would give excellent spatial resolution. Submillimeter in-plane pixels are important for good lesion margin visualization, which helps distinguish benign from malignant enhancing lesions based on their morphology.35

Phase-Encoding Direction Chosen to Minimize Artifacts Across the Breasts

A primary cause of image artifacts (structured noise in MR images), is patient motion, including cardiac and respiratory motion. These motion artifacts propagate across the image in the in-plane phase-encoding direction, regardless of the direction of motion in the patient.33 Therefore, it is essential to orient the in-plane phase-encoding direction to minimize artifacts across the breast. For
sagittal plane acquisitions, the phase-encoding direction should be head-to-foot (or superior-inferior) (Fig. 10). For transaxial plane acquisitions, phase-encoding should be oriented left-to-right to ensure that cardiac and respiratory motion obscure a minimal amount of breast tissue (Fig. 11). For coronal plane acquisitions, phase encoding can be either left-right or head-to-foot, as cardiac and respiratory motion will not propagate across the breasts in either in-plane direction.

Fig. 8. Relationship between slice thickness and minimum lesion size seen without, or with minimal, partial volume effects. (A) Slice thickness equals lesion size. In this case, partial volume effects will significantly decrease lesion conspicuity. (B) Slice thickness equals one-half lesion size. Regardless of alignment of lesion and slices, in this case at least one slice will display the lesion without, or with minimal, partial volume effects.

Fig. 9. Surveillance breast MRI in a 48-year-old woman revealed a subtle non-masslike enhancement in a linear/ductal, clumped pattern (arrows) in sagittally acquired and axially reconstructed images. Core biopsy demonstrated the enhancing lesion to be intermediate-grade cribriform and solid DCIS. Note that the enhancing lesion appears less sharp in the reconstructed axial image than in the originally acquired sagittal image because the slice thickness of acquired sagittal images exceeded the in-plane pixel size. The thicker slices in sagittally acquired images lowered spatial resolution in the slice-select direction, which is left-to-right in the reformatted axial images. (Courtesy of Robyn Birdwell, MD, Breast Imaging, Brigham and Women’s Hospital, Boston, MA.)
Phase-encoding is usually chosen to be head-to-foot for coronal imaging because it requires less spatial coverage than left-to-right, minimizing the number of phase-encoding steps needed.

**Total 3DFT Acquisition Time for Each Series of 1 to 3 Minutes**

The temporal resolution required for breast MRI is determined by the time course of contrast agent uptake. Peak contrast enhancement in malignant lesions typically occurs between 60 and 120 seconds after injection. It is important to capture contrast uptake at or near its maximum with one of the postcontrast series. To do that, it is also important to know that 3DFT acquisitions have maximum contrast-weighting at the low spatial frequency acquisitions (ie, when the center of k-space is being acquired), which occurs at one-third to one-half of the total pulse sequence acquisition time, depending on the manufacturer (eg, 3DFT sequences on Siemens acquire the center of k-space about one-third of the way into the full acquisition, on GE and Philips at half-way through sequence acquisition). The goal is to select imaging parameters that place the maximum contrast-weighting of the first postcontrast series at or near the time of peak contrast agent uptake. Assuming that peak enhancement of breast lesions occurs 90 seconds after contrast injection, you would like the center of k-space of the first postcontrast series to occur 90 seconds after the end of contrast injection. If you were using a Siemens 3DFT series with a 2-minute series scan time, peak contrast-weighting would occur at one-third of 2 minutes, or 40 seconds, into the series.

![Fig. 10.](image1) Sagittal plane imaging with the phase-encoding direction: (A), correctly selected in the head-to-foot (or superior-inferior) direction, and (B), incorrectly selected in the anterior-posterior direction. Note in (B) that motion and wrap artifacts are propagated across breast tissue.

![Fig. 11.](image2) Transaxial plane T2-weighted FSE images identically acquired, except with the phase-encoding direction: (A) correctly selected in the left-to-right direction, and (B) incorrectly selected in the anterior-posterior direction. In (B), cardiac and respiratory motion artifacts propagate across breast tissue.
so to properly time the center of k-space 90 seconds after injection, you would wait 50 seconds after injection to begin the first postcontrast series. If you were using a GE or Philips scanner with, for example, a 3-minute 3DFT acquisition, peak contrast weighting would occur at the midpoint of the series, so you would begin scanning immediately after injection to place maximum contrast-weighting at 90 seconds.

Typically, a precontrast series and several postcontrast series are acquired with identical acquisition parameters so that subtractions of precontrast from postcontrast images reveal only temporal changes. Thus, all precontrast and postcontrast series should be identical. A single precontrast scan should be acquired, followed immediately by contrast agent injection. A pause of scanning during, and perhaps after, contrast injection might be needed, depending on the calculation outlined previously to place peak contrast at the center of k-space of the first postcontrast series. Then, several postcontrast series should be acquired without pauses or delays between them, extending so that the last measurement samples the center of k-space at least 6 minutes after the end of contrast injection. This is done so that the detailed shape of the time-enhancement curve can be determined for any significantly enhancing lesions. Contrast agent uptake is best characterized by dividing enhancing lesions into 3 categories: continuous uptake (Type 1), plateau (Type 2), or washout (Type 3) (Fig. 12). Kuhl and colleagues demonstrated in a study of 266 enhancing lesions (101 breast cancers) that only 6% of lesions with Type 1 curves were malignant, 65% of lesions with Type 2 curves were malignant, and 87% of lesions with Type 3 curves were malignant. Characterizing lesions by both their morphology, which does not require multiple postcontrast time points, and their time-enhancement curve shape, which does, adds specificity to contrast-enhanced breast MRI. Thus, it is important to collect data with adequate temporal resolution, and adequate duration, to accurately capture the time course of lesion enhancement.

Initially, European breast MRI protocols emphasized the need for high temporal resolution, on the order of 1 minute per series, to gain specificity. A subsequent article by Kuhl and colleagues demonstrated that temporal resolution could be relaxed to approximately 2 minutes without sacrificing specificity, especially if that added time was used to improve spatial resolution to submillimeter in-plane pixels. More recent work by Gutierrez and colleagues indicated that 3-minute temporal resolution was adequate to correctly characterize time-enhancement curve shapes when the center of k-space was properly positioned at approximately 90 seconds after contrast administration.

![Fig. 12. Three time-enhancement curve types typical of enhancing breast lesions. Lesions with Type 1 curves have continuous uptake and have the lowest probability of malignancy. Lesions with Type 2 curves enhance by at least 80% to 100% from their noncontrast signal values and then demonstrate a plateau behavior, not gaining or losing signal appreciably from their peak value. Lesions with Type 2 curves have a moderate (40%–70%) suspicion of malignancy. Lesions with Type 3 curves have rapid uptake of contrast within 3 minutes of administration, then washout, and have a high (60%–80%) suspicion of malignancy. Images producing these curves were acquired every 60 seconds (1-minute temporal resolution), with the first postcontrast series acquired 90 seconds after contrast administration.](image-url)
injection. In addition, the longer acquisition time per series of 3-minute acquisitions captured greater peak signal than 90-second acquisitions. Based on their findings, it appears that 3-minute temporal resolution is adequate to add specificity by correctly characterizing time-enhancement curve shapes. Going faster than 1 minute per series fails to add additional information about curve shape and decreases SNR per series.\textsuperscript{33,39} Other studies, such as Schnall and colleagues,\textsuperscript{39} have shown that, although curve shape and degree of lesion enhancement are important, lesion morphology assessment is an even more important factor in the overall accuracy of breast MRI for cancer detection.

**Adequate SNR to Visualize Small Enhancing Vessels on 3D Maximum Intensity Projection Images**

The ability to visualize small enhancing vessels (2–3 mm in diameter) on maximum intensity projection (MIP) images is a good surrogate for the ability to visualize small enhancing lesions on subtracted or MIP images. Failure to see relatively small blood vessels on subtracted or MIP images gives the radiologist little confidence that small or subtle enhancing lesions would be detected, if present. Fig. 13 provides examples of good, SNR-deficient, and SNR-starved MIP images: good, marginal, and poor-quality breast MR images.

**THE ACR BREAST MRI ACCREDITATION PROGRAM**

The ACR Breast MRI Accreditation Program (BMRAP) began accrediting facilities that perform breast MRI on May 10, 2010. As of May 1, 2013, 1264 facilities have been accredited, with 72 facilities under review. The repeat rate for facilities is 20%. Like other ACR accreditation programs, the BMRAP includes requirements for personnel (radiologists, MRI technologists, and medical physicists), equipment, quality assurance, and

![Fig. 13. Sagittal MIP images demonstrating varying degrees of quality in terms of displaying enhancing vessels (and lesions, if they were present): (A) good MIP image, where small vessels are clearly displayed; (B) marginal MIP image, where visibility of small vessels is limited due to low SNR; and (C) poor MIP image, due to extremely low SNR and no display of large or small vessels. The poor image quality in (C) gives low confidence that this scan would demonstrate a small or diffusely enhancing lesion, if present.](image-url)
accreditation testing based on submission of clinical images to assess breast MRI scanning protocols and clinical image quality. A complete discussion of the BMRAP is beyond the scope of this article, but is available at http://www.acr.org/Quality-Safety/Accreditation/BreastMRI, including a complete list of BMRAP requirements, a Breast MRI Clinical Image Quality Guide, and complete procedures for applying for ACR Breast MRI Accreditation.

SUMMARY

Current MRI systems are capable of meeting the stringent technical requirements of performing multiphase T1-weighted contrast-enhanced scanning with high in-plane spatial resolution (≤1 mm pixel sizes), thin slices (≤3 mm thick), adequate temporal resolution (1–3 minutes), bilateral breast coverage, and adequate SNR to detect small or diffusely enhancing breast lesions. Careful attention to breast MRI equipment selection and breast MRI protocols is required to achieve all of these requirements simultaneously. The ACR’s BMRAP provides a peer-review system for validating that breast MRI personnel, equipment, quality-control procedures, scanning protocols, and image quality are adequate to perform high-quality breast MRI.

REFERENCES