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MR sequences and rapid acquisition for MR guided interventions

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Synopsis

Interventional MR uses rapid imaging to guide diagnostic and therapeutic procedures. One of the attractions of MR-guidance is the abundance of inherent contrast mechanisms available. Dynamic procedural guidance with real-time imaging has pushed the limits of MR technology, demanding rapid acquisition and reconstruction paired with interactive control and device visualization. This article reviews the technical aspects of real-time MR sequences that enable MR-guided interventions.

Keywords

Real-time MR imaging; MR image reconstruction; parallel imaging; non-Cartesian imaging

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Introduction

Interventional MR imaging is valuable for real-time dynamic procedural guidance and intra-procedural imaging during diagnostic or therapeutic procedures, including surgery, tissue biopsy, ablation therapy, endovascular procedures, and device placement. The flexibility of MR image contrast is appealing for procedural guidance; however, the demands of MR-guided interventions are unique and require a specialized environment.

Diagnostic MR imaging is well established in the clinic to provide high-resolution images with excellent soft tissue contrast, designed to assess pathological tissue and derive quantitative metrics. Typically, diagnostic MR imaging uses long scan times to generate the desired image contrast, and may require offline image reconstruction or processing.

Interventional MR imaging, on the other hand, demands much faster image acquisition, reconstruction, and processing. Furthermore, procedural guidance employs interactive parameter control and requires the simultaneous visualization of tissue and interventional devices (e.g., biopsy needles, guidewires, catheters, stents, occluders, forceps). Table 1 summarizes the differing demands of diagnostic and interventional MR imaging.

IMAGING PROTOCOLS

Interventional MR imaging encompasses pre-procedural imaging for planning, intra-procedural imaging to assess progress, and real-time imaging for dynamic procedural guidance. The preprocedural and intraprocedural imaging is used to assess anatomy, physiology, or pathology relevant to the procedure. Here, we focus primarily on the technical details of the fast real-time imaging used during dynamic procedural guidance.

Real-time imaging

Imaging efficiency is crucial during MR-guided interventions, especially when competing against established interventional modalities such as radiography and ultrasonography. Radiography generates approximately 15 frames/s with a pixel matrix of 1024×1024 . In comparison, 5-10 frames/s are used for MR imaging during procedural guidance with a much smaller pixel matrix of 128×128 or 144×192 . Real time imaging is not gated nor breath held, and the entire image is acquired in a single shot. Figure 1 shows the single shot real-time image acquisition running continuously, with multiple slices updating in rapid succession. Throughout an interventional procedure, slice geometry and image contrast are interactively controlled, either by interventionists in the MR suite or by operators in the control room.

For real-time procedural guidance, the challenge is to generate adequate tissue contrast and sufficient image quality in terms of signal-to-noise ratio (SNR) and artifact, while also maintaining short imaging times. Typically, balanced steady state free precession imaging (bSSFP) is used to accomplish this^{1,2}. bSSFP uses magnetization efficiently to provide high SNR with short repetitions times and the resulting images have T2/T1 weighted contrast with bright blood and fat signal and darker muscle tissue. Fully sampled bSSFP images can achieve a temporal resolution of 377 ms/image or 2.6 frames/s (echo time [TE]/repetition

time [TR] = 1.27/2.62 ms, matrix = 192×144). Using parallel imaging, the temporal resolution can be pushed to 94 ms/image or 10.6 frames/s (acceleration factor 4, see “Parallel Imaging” section below). Banding artifacts in bSSFP are not usually a concern for these short TR sequences; however, real-time bSSFP does suffer from undesirable bright signal from fat. If needed, fat suppression for bSSFP can be accomplished using radiofrequency (RF) cycling and TR alternation³⁻⁵. Real-time imaging can also be achieved using spoiled gradient echo sequences⁶, though used less frequently due to the lower SNR.

A magnetization preparation module can be programed into the pulse sequence such that it is toggled on/off interactively while the single shot acquisition runs continuously. For example, non-selective saturation pre-pulses can be added to the sequence to enhance gadolinium contrast while suppressing tissue signal (see “Device visualization” section below). Furthermore, flow-sensitive saturation preparations can produce dark-blood images to enhance the gadolinium contrast and preserve tissue signal⁷. Inversion pulses can be inserted into the real-time pulse sequence for infarct imaging⁸. Interactive color flow MRI using phase contrast has been implemented to rapidly visualize cardiac and vascular flow⁹. Virtual dye angiography uses volume selective saturation pulses that can be turned on/off interactively to provide flow visualization mimicking contrast angiography during endovascular procedures¹⁰ (Figure 2). These interactive magnetization preparation modules modify image contrast, as needed, throughout the procedure.

Parallel imaging

Parallel imaging can be used to accelerate real-time imaging by skipping some phase encoding lines throughout the acquisition and exploiting multi-channel signal receiver arrays during reconstruction for added spatial encoding. By eliminating some phase encoding steps, the resulting images are aliased. Parallel imaging describes a family of reconstruction techniques that allows us to recover images from the aliased ones. The two most common methods used in the clinic are SENSE¹¹ and GRAPPA¹². Excellent reviews of parallel imaging techniques are available^{13,14}.

Each point in an aliased single-coil image is a linear signal superposition with weights according to the coil sensitivity profile (Figure 3). SENSE reconstruction unambiguously unfolds aliased images by solving a linear system using knowledge of coil sensitivity maps calibrated at the beginning of the examination. TSENSE¹⁵ is an alternative algorithm which uses interleaved undersampling to integrate the coil sensitivity maps with the image acquisition. TSENSE is specifically designed for dynamic imaging and is critical for high frame rate imaging.

GRAPPA aims to regenerate the missing phase encoding lines from the raw k-space data using information about a given data point contained within the neighboring points in k-space. GRAPPA requires a fully sampled region of k-space known as the autocalibration signal that is used to calculate a kernel to regenerate all missing k-space points. GRAPPA is robust in cases in which sensitivity maps are difficult to generate, e.g., when the prescribed field of view is too small for the imaged object and there are regions with aliasing in the calibration data.

In general, acceleration rates of four are robustly used in a clinical setting, and both SENSE and GRAPPA reconstructions are available with vendor-supplied reconstruction software.

Compressed sensing algorithms show potential for further accelerating image acquisition¹⁶, but are currently limited in their application to interventional MR imaging by prohibitively long reconstruction times.

Efficient k-space trajectories

More efficient k-space trajectories can also be used to speed up acquisitions (Figure 4). Echo planar imaging (EPI) is an accelerated Cartesian acquisition whereby multiple phase encoding steps are acquired following an RF pulse. Using single shot EPI, the entire image can be acquired following a single RF pulse. EPI has found clinical utility for interventional and real-time applications¹⁷⁻²⁰.

Spiral imaging²¹ and radial imaging²² are examples of non-Cartesian acquisitions used for MR-guided interventions²³⁻²⁶. Spiral imaging is particularly attractive for high frame rate applications in the interventional MR imaging environment. Oversampling of the k-space center in non-Cartesian sampling patterns results in flow and motion insensitivity and robustness to aliasing artifacts. Either spoiled gradient echo or bSSFP contrast can be achieved with spiral and radial trajectories.

Spiral and radial k-space trajectories do not lie on a Cartesian grid and therefore, require samples to be interpolated onto a grid during image reconstruction in a process called “regridding”²⁷. Typically, regridding uses a Kaiser-Bessel kernel²⁸ or, in the case of nonuniform fast Fourier transformation²⁹, using least-squares design of interpolation coefficients. Since sampling density is nonuniform over k-space, density compensation is applied before regridding³⁰. Alternatively, radial acquisitions can be reconstructed using back projection methods originally designed for computed tomography reconstruction.

Parallel imaging can be combined with non-Cartesian acquisitions. Undersampling is achieved by omitting spiral interleaves or radial projections from the acquisition. For non-Cartesian trajectories, the resulting aliasing is irregular and classic SENSE unfolding is impossible. Instead, an iterative method, namely, conjugate gradient SENSE^{31,32}, is utilized to solve the linear system. GRAPPA reconstruction can also be performed with undersampled non-Cartesian data sets³³⁻³⁷; however, non-Cartesian GRAPPA schemes are currently incompatible with the interventional environment because of the large number of fully sampled calibration frames required. Approaches to reduce the number of calibration scans are under development³⁸. Readers are directed to reviews on non-Cartesian imaging for further details^{39,40}.

Keyhole imaging

Keyhole imaging increases apparent frame rate by reconstructing consecutive images combining newly acquired data with data acquired from previous frames^{23,41,42}. Non-Cartesian sampling patterns are better suited to keyhole reconstruction because the center of k-space is reacquired with each interleaf/projection and a full range of spatial frequencies is contained in both new and old data.

Fast reconstruction

For interventional applications, fast acquisition is only valuable if it is paired with fast reconstruction. MR system vendors provide the capability to reconstruct Cartesian images with standard parallel imaging in real-time. However, for more complicated reconstructions (e.g., non-Cartesian imaging, complex parallel imaging schemes, or iterative reconstructions), additional reconstruction tools are necessary.

Imaging with large coil arrays is computationally costly. A number of algorithms have been proposed for coil selection⁴³⁻⁴⁵ to choose the most suitable subset of coils for reconstruction and array compression⁴⁶⁻⁴⁹ to combine channels for reducing reconstruction time. Most notably, principle component analysis has been used for array compression without the need for coil sensitivity maps⁴⁷.

Graphics processing unit (GPU) accelerated computing has been used for significant improvements in reconstruction speed, has been applied to advanced 3D reconstructions⁵⁰, parallel imaging^{51,52}, and nonuniform FFT⁵³. An 85-fold acceleration compared with a state-of-the-art 64 bit central processing unit⁵³ has been reported. GPU-accelerated computing requires additional hardware that may not be available on the vendor-supplied reconstruction system.

An open source software package (Gadgetron, <http://gadgetron.github.io>⁵⁴) for medical image reconstruction has recently been made available. The software contains standard reconstruction tools, iterative solvers, and GPU components. It is designed to run on either the local MR computer, an external workstation, or on multiple nodes in a distributed computing environment⁵⁵. Images may be piped directly to the host computer for online display. This framework permits complicated reconstructions in a reasonable time frame with image display on the MR imaging host computer.

Device Visualization

Real-time device visualization is also necessary during MR imaging-guided. Passive visualization uses the intrinsic material properties. For example, metallic devices (e.g., biopsy needles, guidewires, stents, occluders) create a signal void on real-time MR images because they distort the local magnetic field, leading to local signal dephasing⁵⁶. Signal voids can be emphasized using long TE gradient echo sequences. Alternatively, positive contrast techniques in which the metallic device appears bright compared to the background signal can be used for real-time visualization^{57,58}. Non-metallic devices, such as plastic catheters, can be made visible on real-time imaging by filling the lumen with gadolinium^{59,60} although it restricts use of the lumen to deliver other devices or agents. Air, gadolinium, or carbon dioxide filled balloon catheters have been used to guide right heart catheterization under MR-guidance^{61,62}. Saturation pre-pulse modules are used to enhance gadolinium contrast of catheter devices (Figure 5).

Active visualization involves the embedding of receiver electronics into the device such that a unique device signal can be collected separately from the other receiver coils. This unique signature can be overlaid in color onto anatomical images in real-time⁶³. Alternatively, device tracking uses fast device localization and graphically represents the device position

on previously acquired images⁶⁴ (Figure 6). Point source coils can be localized rapidly using three orthogonal echoes⁶⁵ whereas spatially extensive coils can be localized using three orthogonal projection images (non-slice-selective). The unique signature of active devices can also be exploited for automatic repositioning of the imaging slice when the tip moves out of plane^{66,67}. Accurate localization of devices is essential for effective procedural guidance.

Interactive Environments

Interactive environments permit graphical control for real-time imaging, allowing flexibility to modify imaging parameters according to the needs of the procedure without stopping to change the pulse sequences or imaging protocol.⁶⁸ The entire infrastructure of the MR imaging system, reconstruction pipeline, and interactive control is outlined in Figure 7. Features found in interactive environments are:

- Graphical slice positioning
- Multi-slice display
- Real-time modification of scan parameters (eg, slice thickness, acceleration factor, magnetization preparation modules)
- Roadmap visualization in combination with real-time imaging or device tracking
- Color visualization of device channels and device tracking

Video 1 demonstrates the modification of imaging parameters in an interactive environment. Interactive imaging environments are available from each of the major MR system vendors: Siemens Medical Solutions offers the Interactive Front End (Siemens Corporate Research, Princeton, NJ) for investigational use; General Electric Medical Systems (Waukesha, WI) offers MR Echo; and Philips Healthcare (Best, Netherlands) offers eXTernal Control (XTC)⁶⁹. In addition, third party software can provide interactive imaging and visualization. RT Hawk (HeartVista, Menlo Park, CA)⁷⁰ runs from a “stub” pulse sequence and offers dynamic switching between pulse sequences and reconstruction algorithms. Vurtigo (Sunnybrook Health Sciences Centre, Toronto, ON, Canada)^{71,72} is open source software for visualization of interventional procedures.

Furthermore, automatic device-scanner interaction has been demonstrated, where temporal resolution and field of view are automatically increased when fast device motion is detected⁷³. Automatic slice positioning for device visualization can also be incorporated in the interactive environments^{66,67}.

PEARLS, PITFALLS AND VARIANTS

Artifacts

Pushing the limits of temporal resolution can result in unwelcome artifacts in images. Parallel imaging acceleration comes at the expense of a loss in image SNR:

$$SNR_{\text{parallel imaging}} = \frac{SNR}{g \sqrt{R}}$$

where R is the acceleration factor and g ($g \geq 1$) is the coil geometry factor (or “g-factor”) related to the properties of the coil receiver array. The coil geometry in relation to the slice position is constantly changing throughout an MR-guided intervention; therefore, it is useful to interactively modify acceleration factor to tradeoff image quality and imaging speed for a given slice geometry.

EPI, spiral imaging, and radial imaging methods are also susceptible to image artifacts. Specifically, ghosting artifacts are common in EPI images caused by system imperfection which result in the misalignment of odd and even echoes during the bipolar readout. Standard reference lines are used to correct this by default, but users should be aware of the potential for residual ghosting. EPI is also susceptible to distortions caused by magnetic field inhomogeneity.

Off-resonance manifests as blurring in non-Cartesian images that can be eliminated in a composite image from multi-frequency reconstruction⁷⁴. Non-Cartesian k-space trajectories are also sensitive to distortions caused by errors in gradient waveforms. A number of methods have been developed to retrospectively measure the true spiral k-space trajectories for distortion correction during image reconstruction^{75,76}. Similarly, gradient delays can be measured and compensated for within the pulse sequence for correction of radial images⁷⁷. Unfortunately, these additional measurements are impractical in the interventional MR imaging setting. Recently, the gradient system impulse response function (GIRF) has been demonstrated for predicting true gradient waveforms from the nominal waveforms prescribed in the pulse sequence⁷⁸, and a real-time framework using GIRF trajectory correction and interactive off-resonance reconstruction for de-blurring has been applied for distortion correction of spiral images⁷⁹.

RF-induced heating

Device safety can also be a concern in the interventional MR imaging environment. Energy deposited by the RF pulses used for imaging can cause heating in conductive devices that may lead to tissue damage. Eddy currents in conductive devices will create slight heating⁸⁰. Resonating RF waves along long metallic guidewires can generate dangerous levels of heating (up to 70°C)⁸¹⁻⁸⁴. RF induced heating is inversely proportional to TR and is proportional to the square of the flip angle. bSSFP imaging uses flip angles of 40°- 60° and short TRs. Gradient echo (low flip angle) and non-Cartesian (long TR) imaging methods use lower RF energy. Patient safety must be considered when using metallic devices in the MR imaging environment.

POINTS FOR THE REFERRING PHYSICIAN

Interventional MR imaging is different from diagnostic MR imaging and the successful implementation of MRI-guided interventions requires a highly specialized environment and

trained staff beyond that requisite for a standard diagnostic MR imaging⁸⁵. Features found in the interventional MRI environment are:

- A real-time interactive scanning environment
- Devices that are visible and safe with MR imaging
- In-room image display used during dynamic procedural guidance
- Audio communication between interventionists and scanner control room
- Physiology and hemodynamic monitoring (higher quality than that used for diagnostic MR imaging) to monitor the patient throughout the procedure.
- An emergency patient bailout strategy (usually to an adjoining suite using an intermodality transfer table).

SUMMARY

A wide variety of different pulse sequences may be used during MR-guided interventions. The choice among these is dictated by the pathology and type of procedure. One common requirement for all MR-guided interventions is that both imaging and reconstruction must be fast. The capability now exists to perform very fast imaging with interactive control, permitting real-time procedural guidance with MRI. With the technology in place, many more clinical applications of this promising tool can be expected in the future.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Key Points

- Interventional MR imaging requires a specialized environment and work-flow.
- Rapid image acquisition and rapid image reconstruction are a prerequisite for all MR-guided interventions.
- High frame-rate real-time imaging can be achieved for dynamic procedural guidance.
- Imaging can be accelerated using parallel imaging or efficient k-space trajectories.
- MR imaging can enable simultaneous device and tissue visualization.

Figure 1.

Diagram of real-time acquisition running continuously with multiple slices updating in rapid succession. Image contrast is changed using an optional magnetization preparation module and interactive parameter control. bSSFP, balanced steady-state free precession.

Figure 2.

Virtual dye angiography uses volume selective-saturation pulses to saturate blood signal locally. The difference images between saturation off/on (black and white images) are used to produce a color-flow map. Ao, aorta; LA, left atrium; LV, left ventricle. (From George AK, Faranesh AZ, Ratnayaka K, et al. Virtual dye angiography: flow visualization for MRI-guided interventions. *Magn Reson Med* 2012;67(4):1013–21; with permission.)

Figure 3.

Illustration of SENSE reconstruction using undersampled aliased images and coil sensitivity maps. (Data from Collins DL, Zijdenbos AP, Kollokian V, et al. Design and construction of a realistic digital brain phantom. IEEE Trans Med Imaging 1998;17(3):463–8; and BrainWeb: Simulated Brain Database. Available at: <http://brainweb.bic.mni.mcgill.ca/brainweb/>.)

Figure 4. Spin warp Cartesian imaging (a) compared to more efficient k-space trajectories: echo planar imaging (b), spiral (c), and radial k-space trajectories (d).

Figure 5.

Right heart catheterization using real-time MR imaging to navigate a gadolinium-filled balloon (arrows) between cardiac chambers. Standard real-time bSSFP (A, C, E) and real-time bSSFP with saturation prepulses (B, D, F) are depicted. Saturation prepulses are used to isolate the gadolinium-filled balloon signal. (From Ratnayaka K, Faranesh AZ, Hansen MS, et al. Real-time MRI-guided right heart catheterization in adults using passive catheters. *Eur Heart J* 2013;34(5):380–9; with permission.)

Figure 6.

(A–D) Active microcoil (arrow) tracking of brachytherapy stylet. The trajectory (yellow points) represents consecutive tracking positions overlaid on previously acquired images. (From Wang W, Dumoulin CL, Viswanathan AN, et al. Real-time active MR-tracking of metallic stylets in MR-guided radiation therapy. *Magn Reson Med* 2015;73(5):1810; with permission.)

Figure 7.

Overview of the imaging infrastructure that could be used in the interventional MR imaging environment. The device data can be isolated during reconstruction to permit the color overlay of the device signal in the interactive environment. Dashed lines represent optional components. GPU, graphics processing unit.

Table 1

Differences between diagnostic and interventional MRI

Diagnostic MRI	Interventional MRI
Scans run in “batch mode”	Interactive environments are used to modify real-time imaging on-the-fly
Long scan times	Short scan times
Offline reconstruction and post processing is possible	Reconstruction and processing must be performed with low latency
Image quality is of utmost importance	Imaging speed is of utmost importance
Used for anatomical imaging	Used for anatomical imaging and device imaging
Cardiac and respiratory gating and/or breath holding can be used to compensate for motion	Real-time MRI is not gated or breath-held