
Pulse Sequences for Interventional MRI

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Abstract

Diagnostic MRI sequences aim to provide varied contrast mechanisms to increase the sensitivity and specificity of characterizing abnormal or degenerative tissue. Sequences are normally run in a “batch mode,” with each sequence being completed before another is begun. Interventional imaging sequences have numerous important differences from their diagnostic counterparts. First, they serve other roles besides providing imaging contrast. These include device visualization and tracking, 2D and 3D visualization of tissue near the device, and therapeutic monitoring. Pulse sequences for MRI-guided procedures are not run in batch mode and require interactive control. The sequences are interleaved and swapped in and out as the procedure demands. Finally, when latency is crucial, the design of these pulse sequences and their reconstruction algorithms must be constrained to minimize the time between the start of the acquisition and display of the reconstructed output. These differences create different requirements for pulse sequences and the way pulse sequences communicate with the rest of the scanner. Fortunately, recent developments in rapid contrast generation, k -space trajectory schemes, and interventional software environment platforms provide foundations for flexible configurations to meet the imaging needs of interventions. This chapter presents an overview of some of the methods used in designing and implementing pulse sequences for MRI-guided interventional procedures. The last section of the chapter describes platforms that integrate interactive control, acquisition, reconstruction, scan plane

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control, and visualization that significantly simplify the design of imaging capabilities for MRI-guided procedures.

1 Introduction

Advances in MRI are often led by the imaging disciplines that have the highest performance needs. Cardiac imaging, with its needs for rapid temporal resolution to capture physiological motion, breathing motion, and B_0 -induced inhomogeneity, is a prominent example. A similar argument can be provided for the impact on MRI for interventional procedures, with their needs for rapid imaging, rapid data processing, interactive control, varied image contrast, and visualization. For example, the need for rapid imaging in interventions fueled the renewal of interest in balanced steady-state free precession (SSFP) (Duerk et al. 1998). In this chapter, aspects of pulse sequence design useful in several areas of interventional imaging are discussed. Robust implementations of MRI-guided procedures also require system interface modifications that affect pulse sequences to provide interactive capabilities. System development environments that provide interactive capabilities are described at the end of this chapter.

2 Rapid Contrast

MRI has been able to generate quite high frame rates if image contrast is not a concern. However, soft-tissue contrast is often the motivation driving the creation of MRI-guided interventional procedures. The task then is to create high-performance rapid imaging sequences while still providing high image contrast.

The need for imaging speed, especially when capturing physiological motion or providing image-guided feedback on device location, often limits the range of pulse sequences available to the interventional developer. For example, sequences based on rapid acquisition with relaxation enhancement that provide highly desirable T2 contrast are often significantly too slow to use in an interventional setting. In general, generating the level of image contrast provided by diagnostic imaging methods, which normally take much more time, is difficult and some compromises are necessary.

Gradient-recalled steady-state imaging often provides the brunt of the imaging tools used in interventional MRI procedures. By simply switching schemes for setting the RF phase between excitations and echo location and making modest changes in gradient spoiling, one can quickly switch between heavily T1 weighted imaging [spoiled gradient (SPGR), fast low-angle shot (FLASH), T1 fast field echo (FFE)], mixed T1- and T2-weighted imaging [gradient-recalled acquisition in the steady state (GRASS), fast imaging with steady-state precession (FISP), FFE], T2*-weighted imaging, and fully balanced SSFP with T2/T1 weighting [fast imaging employing steady-state acquisition (FIESTA), true FISP, balanced FFE].

T1-weighted gradient-recalled imaging can be utilized to provide 3D vascular road maps or guide endovascular catheters with active catheter localization. Alternatively, imaging with T1-weighted sequences while periodically injecting small amounts of diluted contrast medium through catheters can highlight the catheter location without active localization. Delaying the echo time in gradient-recalled imaging provides T2*-weighting that sensitizes image contrast to the presence of drug therapies tagged with iron oxide particles.

T2 weighting is often desirable for visualizing tumors with positive contrast. Creating bright fluid images is often more desirable than viewing fluid as a void. Often, fully balanced SSFP sequences, even though they actually provide T2/T1 weighting, fulfill this need. If one desires to remove unwanted T1 contamination in balanced SSFP images, one can add unequally spaced 180° pulses to the balanced SSFP readout that switch the magnetization between states aligned with and inverted with respect to the static magnetic field (B_0) using T1-insensitive steady-state imaging (TOSSI) (Schmitt et al. 2011) (Fig. 1).

Balanced SSFP sequences often suffer from bright fat signal that is undesirable. Diagnostic methods such as iterative decomposition of water and fat with echo asymmetry and least-squares estimation (IDEAL) (Reeder et al. 2005), linear combination SSFP (Vasanawala et al. 2000), and fluctuating equilibrium SSFP (Vasanawala et al. 1999) essentially require redundant acquisitions, which make them less desirable for interventional imaging. Instead, methods that

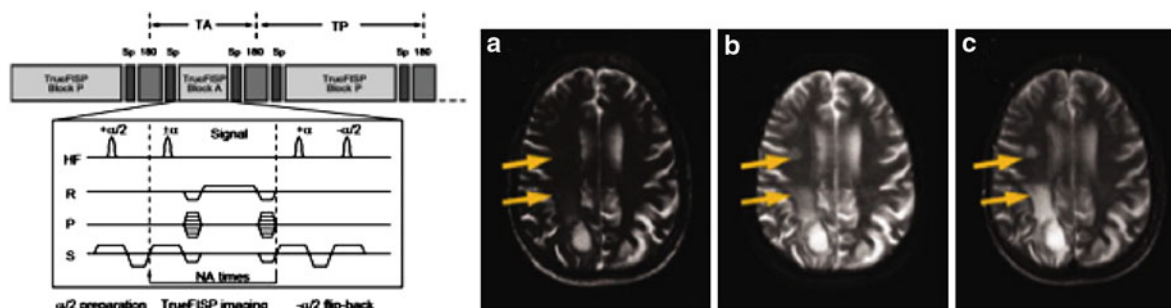


Fig. 1 Left: Balanced steady-state free precession (SSFP) readout (center block) is flanked by 180° pulses. Two readout blocks are created with differing magnetization by changing the interval between the 180° pulses. Right: **a** the standard balanced SSFP image shows bright cerebral spinal fluid (CSF) but poor gray matter/white matter contrast; **b** the T1-insensitive steady-state imaging (TOSSI) image shows much more improved gray

matter/white matter contrast by removing T1 contamination with less than 2-s scan time; **c** the reference turbo spin echo image shows similar contrast in 66-s scan time. *True FISP* true fast imaging with steady-state precession, *TA* Evolution time of anti-parallel orientation train of inversion pulses, *TP* Evolution time of parallel orientation train of inversion pulses, *NA* Number of acquisitions during TA

create nulls near the fat resonance using combinations of alternating repetition times and RF phase cycling are more likely to be useful (Leupold et al. 2006; Cukur and Nishimura 2008).

Interventional imaging often requires interleaving several imaging sequences, each of which has a different purpose. Examples include interleaving active tracking and continuous road map imaging for endovascular procedures. Although several methods have been proposed for interleaving these acquisitions, many interventionalists still separate road map imaging from active tracking. Another common interleaving of sequences occurs in biplane imaging.

Such interleaving requires some care in maintaining, or at least minimally disturbing, the magnetization steady states set up in each sequence. In general, maintaining the balanced SSFP steady state requires more care than maintaining the GRASS (gradient echo) steady state, with the RF-spoiled gradient echo sequence requiring the least effort. Methods to schedule the order of phase encoding such that the center of k -space is acquired when transients in the steady state have damped out (reverse centric) have proven effective in biplane imaging (Derakhshan et al. 2010).

Parallel imaging can also be used to reduce the acquisition time, but only if other imaging parameters (voxel dimension, field strength) provide a surplus of signal-to-noise ratio that can be traded off for imaging speed. Methods that are autocalibrating are generally preferable to methods that require prior coil sensitivity scans in interventional procedures.

3 Device Tracking

3.1 Active Tracking

A common approach to device localization during interventional procedures is the use of an active receiver microcoil on the device itself. This coil's limited spatial sensitivity allows measurement of localized signal information, which can be used to accurately determine device positioning.

Standard pulse sequences used in active device tracking employ a frequency-encoding gradient but no phase-encoding gradient. This produces a signal peak along the frequency-encoding direction that indicates the coordinate of the active coil along that axis. The pulse sequence is repeated multiple times with different axes chosen for the frequency-encoding direction to determine the three spatial coordinates of the device. Several factors can lead to nonideal signal profiles over the microcoil sensitivity volume, resulting in device localization being difficult.

Microcoil design limitations often prevent complete decoupling between the microcoil and the scanner's built-in receive coil, leading to a sensitive imaging volume larger than desired. To limit the sensitive volume to the microcoil's immediate vicinity, a dephasing gradient can be added to the tracking pulse sequence, applying a spatially dependent phase shift in directions orthogonal to the frequency-encoding axis (Dumoulin et al. 2010). This phase shift dephases

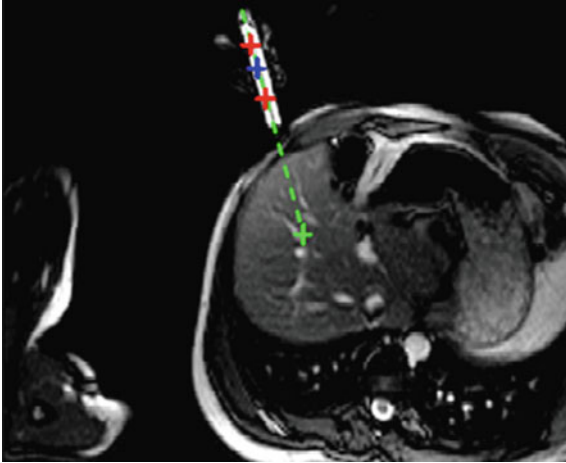


Fig. 2 Image slice aligned with a passive marker for needle trajectory guidance with the theoretical trajectory shown in *green*. The location and orientation of the slice are automatically aligned to the center of the passive marker and are updated every 0.9 s

signal over large spatial distances, but has a relatively small effect over the small volume of microcoil sensitivity. This results in suppression of unwanted signal due to coil coupling while retaining the desired microcoil-localization signal.

3.2 Passive Tracking

In cases where the technical demands and hazards of RF-induced heating of microcoils limit the potential uses of active device tracking, passive device tracking can be used instead. Passive devices create a positive or negative contrast because of their properties of their magnetic material and do not use a separate receiver coil.

Guidance of the needle trajectory for subcutaneous interventions has been performed using a cylindrical trajectory marker around the external trajectory axis (Maier et al. 2011). This marker is fluid-filled and MRI-visible. Two parallel slices through the marker are acquired using a FLASH sequence. As shown in Fig. 2, a 3D position-detection algorithm locates the marker in both slices and then calculates the current theoretical trajectory of a needle guided through the marker.

Intravascular catheter guidewires have been visualized using a combination of a passive double-echo pulse sequence technique and outer-volume suppression to achieve a total passive tracking acquisition time

of less than 0.5 s (Krafft et al. 2011). The double-echo pulse sequence generates a positive contrast image of a guidewire by applying a compensating gradient to correct for distortions of B_0 near the guidewire, which dephases signal from the surrounding volume. Outer-volume suppression enables reduced field of view (FOV) imaging in the phase-encoding direction without aliasing by applying a saturation pulse which saturates the spins outside the FOV.

Imaging of active devices and some passive devices is naturally done infrequently and these are therefore good candidates for acceleration using compressed sensing. Use of compressed sensing in real-time device tracking applications was previously avoided because of its noncausal attributes and time-consuming reconstruction methods. A method for generating causal, real-time imaging of an active device accelerated by compressed sensing has recently been implemented (Ouyang et al. 2011).

4 Non-Cartesian Trajectories

Using Cartesian trajectories in interventional imaging usually simplifies pulse sequence design, reconstruction design, and the requirements for reconstruction processing power and reduces the sensitivity to system instabilities and the sensitivity to patient-induced B_0 inhomogeneity. In short, Cartesian-based trajectories are the simplest way forward for getting an interventional application operational.

However, often non-Cartesian trajectories are worth the added effort because of the alignment of their strengths with the needs of interventional imaging. Non-Cartesian trajectories can offer increased performance in several ways, although not always simultaneously. Non-Cartesian methods can better utilize limited gradient hardware speed, improve the efficiency with which k -space is covered, decrease the sensitivity to motion, and improve flow properties. Increased performance is especially necessary when a quantitative image sensitive to some physiological parameter is required. Examples include thermometry and quantitative flow imaging. Quantitative imaging often requires sampling a larger-dimensional space, and thus imaging speed is at a premium.

Some non-Cartesian methods often offer a variable sampling trajectory where the center of k -space is sampled more often than higher spatial frequencies.

These sampling patterns support time-resolved imaging reconstruction algorithms that differ in performance, speed, accuracy, and complexity. In general, accuracy and performance generally improve with allowance for increased complexity and time within the reconstruction task.

However to produce real-time, causal imaging, reconstruction processing in interventional imaging is generally much simpler compared with application of the cutting-edge algorithms used in compressed sensing and constrained reconstruction. Sliding window reconstructions, where each displayed frame is built from the previous few interleaved trajectories, is a relatively easy method to produce time-resolved imaging without exceedingly complex acquisition or reconstruction strategies. Recently, causal algorithms with modest computing requirements have been proposed for utilizing compressed sensing to improve the frame rate (Ouyang et al. 2011; Sumbul et al. 2009). Fortunately, many of the hurdles for using non-Cartesian trajectories have diminished over the past decade. Computing power has grown much faster than 2D reconstruction processing needs, even when considering the added processing needed for many large, phased-array coils. The advent of imaging development platforms, such as the Interactive Front End (IFE) from Siemens (Lorenz et al. 2005), RTHawk from HeartVista (Santos et al. 2004), and the eXternal Control (XTC) interface from Philips (Smink et al. 2011), has simplified the data pathways for syncing driving gradient hardware with non-Cartesian k -space trajectories while informing the reconstruction algorithms of the trajectory path. These systems also provide standard methods for periodically computing B_0 maps that are necessary for robust imaging with many non-Cartesian methods.

The following sections describe non-Cartesian acquisition and reconstruction theory, loosely classified as spiral and radial trajectories. A brief summary of methods being utilized to provide consistent performance with non-Cartesian methods is provided as these trajectories are generally less robust to several system and patient-induced imperfections than Cartesian methods.

4.1 Non-Cartesian Trajectory Design

In MRI, data can be sampled in k -space on any 2D or 3D trajectory that the time-varying gradients and

safety regulations regarding peripheral nerve and muscle stimulation and tissue heating can support. Although the first MRI method proposed the acquisition of projections (Lauterbur 1973), spin-warp imaging on a Cartesian sampling grid (Edelstein et al. 1980) became the predominantly used trajectory. The acquisition of data on such a rectilinear grid is fairly robust to inhomogeneities in B_0 ; imperfections in the system cause geometric distortions but little degradation of the point-spread function.

In non-Cartesian acquisitions, these inhomogeneities introduce off-resonance effects which in turn cause blurring of the point-spread function. Advances in scanner hardware improved the field homogeneity, and alternative sampling patterns with nonuniform sampling densities were revisited. Projection imaging (Glover and Pauly 1992) is a computed tomography (CT)-like acquisition where each echo represents a radial line traversing through the center of k -space. This method offers good suppression of motion artifacts and allows imaging with very short echo times when the projections start in the center because they do not require any prewinding gradients. A disadvantage is the prolonged total imaging time to completely cover k -space because of the redundant oversampling of the central k -space region. This prolongation is mitigated by the time-resolved imaging capabilities provided by sliding window reconstructions. With spiral trajectories, k -space can be sampled with fewer excitations (Meyer et al. 1992), depending on the length of data acquisition one is willing to utilize with each repetition time. The sampling grids for these acquisition schemes are shown in Fig. 3.

The trajectories can also be extended or combined into 3D acquisitions, for example, for truly 3D radial, 3D spiral, cone, stack of spheres, shells trajectory, cones, and spiral projection reconstruction. Hybrid 3D sampling patterns with non-Cartesian in-plane encoding and traditional Fourier slice encoding have also been implemented, predominantly for the sampling of imaging volumes of shorter dimensions in the through-plane direction. In general as in Cartesian imaging, 2D imaging is useful for tasks that must be done in real time, whereas 3D imaging is useful for covering wider territories where true real-time imaging is not crucial. More complete reviews of sampling patterns can be found elsewhere (Irarrazabal and Nishimura 1995).

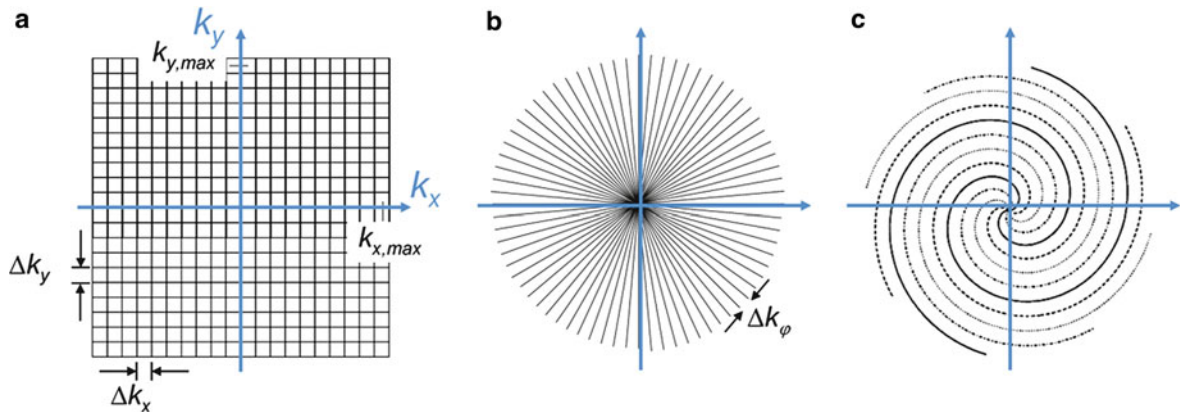


Fig. 3 Strategies for 2D k -space sampling. The spin-warp (a), radial sampling (b), and interleaved spiral (c) imaging trajectories are shown as examples of sampling patterns used in magnetic resonance angiography

4.2 General Considerations

4.2.1 Sampling Region

Sampling a cylinder of k -space saves 21.5% of the sampled space relative to a cube, whereas sampling a sphere will save 47.6% of the required samples. Whereas non-Cartesian trajectories can easily be tuned to cylindrical and spherical k -space regions, selection of phase-encoding and slice-encoding locations can achieve cylindrical sampling spaces.

4.2.2 Gradient Spoiling

As the readout direction changes throughout a non-Cartesian scan, some attention has to be given to the method that spoils the transverse signal in gradient-recalled sequences. Winding the magnetization to the same physical k -space location after each readout is generally a good way to remove variations in the transverse steady-state signal throughout the scan.

4.2.3 Field of View

In general, non-Cartesian trajectories are designed to sample along the readout direction at k -space intervals of $1/\text{FOV}$ as in Cartesian trajectories, where the FOV is the largest dimension of the acquired volume. Sampling along the readout dimension is constrained by the maximum slew rate achievable, and thus k -space sampling intervals often vary, especially at the beginning of the readout. Repetitions of the spiral or radial readout are then rotated in such a way to fill k -space. To provide full k -space sampling, enough repetitions of the model readout are needed such that

the space of the interleaves is less than $1/\text{FOV}$ in all areas of k -space.

4.2.4 Off-Resonance

The extent of phase accrued by off-resonance spins during each readout is directly proportional to the readout duration. The amount of effort needed to remedy off-resonance effects thus increases with readout duration. The sophistication of any needed off-resonance processing depends also of course on the amount of inhomogeneity present in the vascular territory of interest. Whereas the appearance of off-resonance varies with trajectory, off-resonance effects are generally manifested by blurring and signal dropout in non-Cartesian trajectories. Whereas radial acquisitions are less sophisticated than spiral trajectories, the short acquisition time required for each radial line or projection significantly limits off-resonance effects. The advent of dynamic magnetic resonance (MR) development systems can provide features that periodically acquire and reconstruct data to generate rapid B_0 maps, making it easier to implement and utilize different non-Cartesian sequences.

4.2.5 Flow Sensitivity

In general, trajectories whose first moment is small near the center of k -space have better flow properties (Nishimura et al. 1991). Trajectories for which the first moment changes smoothly as a function of k -space radius also are more robust in MR angiography. In general, trajectories which originate at the

center of k -space without previous slice encoding will have more advantageous flow properties.

4.2.6 Sampling Density

The sampling density of radial trajectories varies significantly with k -space radius, falling off as $\frac{1}{k(r)}$ for 2D radial trajectories and $\frac{1}{k^2(r)}$ for true 3D radial trajectories. Sampling some spatial frequencies more often at the expense of others has liabilities when trying to cover all of k -space rapidly and has deleterious effects on the signal-to-noise ratio compared with flat sampling trajectories (Tsai and Nishimura 2000). Oversampling lower spatial frequencies has significant advantages for interventional imaging, as it has shown advantages for representing time-resolved imaging (Korosec et al. 1996; Barger et al. 2002; Song and Dougherty 2004), motion artifact suppression, inherent field map and coil sensitivity generation, and constrained reconstruction methods (Mistretta et al. 2006; Johnson et al. 2008; Lustig et al. 2007).

Spiral waveform design initially emphasized efficient, rapid k -space coverage with flat sampling density (Irarrazabal and Nishimura 1995; Meyer et al. 1992), where only a minimum portion of the waveform was slew-rate-limited. Very simple methods that grid spiral data points to the nearest neighbor on an oversized Cartesian matrix have also been demonstrated (Oesterle et al. 1999) and simplify the density compensation computation. More recently, spiral design has incorporated variable sampling density to mitigate effects from aliasing from outside the FOV and to provide some of the advantages oversampling provides for representing time-resolved image volumes (Tsai and Nishimura 2000).

4.3 Spiral Trajectory Design

Although numerous implementations are possible, most spiral trajectories have been based on an Archimedes spiral. These trajectories follow the basis equation $k(t) = \lambda\theta(t)e^{-i\theta(t)}$. The desired gradient waveforms are given by derivative of the k -space trajectory and are scaled by the inverse of the gyromagnetic ratio. Linear functions of $\theta(t)$ lead to inefficient spirals with constant angular speed. The intuitive choice for efficient coverage of k -space would use a constant-velocity spiral where $\theta(t) = \sqrt{t}$.

As this choice is not realizable in regions of the spiral where the slew rate is limited, trade-offs in the formulation of $\theta(t)$ between constant angular speed and constant velocity were formulated by Bornert et al. (1999). Although a more accurate solution for optimal use of gradient slew rate was formulated by King et al. (1995), this solution required significant computation. A closed-form expression which produces images which are indiscernible from the optimal solution was provided by Glover (1999).

The complexity of the spiral trajectory has resulted in numerous publications reporting the use of computational power to create shorter trajectories, more accurate sampling density functions, and more powerful off-resonance correction methods. In many cases, simpler approximations can provide adequate performance for many MR-guided imaging applications. A good review of these trade-offs is provided by Block and Frahm (2005).

The advantages of spiral acquisitions grow with longer readout duration, although these increase problems with off-resonance. An example of the use of spirals to provide short echo times for visualizing cryoablation (Butts et al. 2001) in an MRI-guided procedure is shown in Fig. 4.

4.4 Projection Imaging

More specific aspects of the design of various radial trajectories and their effects on point-spread functions are provided next.

4.4.1 Two-Dimensional Projection Imaging

In 2D projection imaging, each readout traverses through the center of k -space (Fig. 5). The sampling trajectory can be described in polar coordinates with a radial component k_r and an angle φ . A total of N_p repetitions are acquired with N_r samples and a sampling interval Δk_r along the readout direction. As the 1D Fourier transform of each repetition provides a projection of the object, the technique is also known a projection reconstruction.

Projection or radial imaging leads to a nonuniform sampling density. The radial sampling interval Δk_r supports an alias-free reconstruction of distance $D = 1/\Delta k_r$ along the readout. According to the Nyquist theorem, sampling with $\Delta k_{\varphi, \max} = \Delta k_r$, as shown in Fig. 6 produces isotropic resolution over a circular

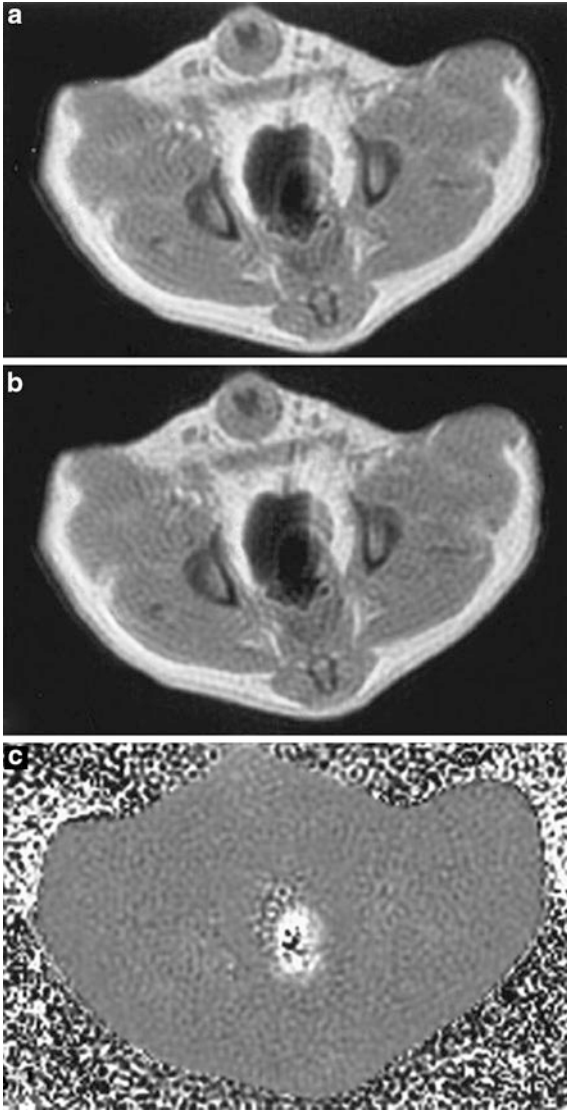


Fig. 4 In vivo results in the dog prostate after cryoablation. Spiral imaging provides the short echo times necessary to visualize the iceball with positive contrast. Echo times of 0.2 ms (a) and 1.2 ms (b) allow calculation of $R2^*$ (c). Elevated $R2^*$ is evident in the prostate

FOV with a diameter D . This optimal sampling requires

$$N_{p,opt} = \frac{\pi}{2} N_r \quad (1)$$

projections. More projections do not provide better spatial resolution or a larger FOV. In comparison, spin-warp imaging requires only N_r readouts ($2/\pi = 63.7\%$ less) for a squared FOV with identical resolution. This

decrease in sampling efficiency is due to the over-sampling of central k -space. It is important to note that any signal from outside the circular FOV causes data inconsistencies in the projections and results in streak artifacts in the image. The bandpass filter applied to the received signal limits the signal contributions along the readout direction but not perpendicular to it.

Although the repeated sampling of low spatial frequencies in each readout decreases the scan efficiency, radial sampling has very desirable properties for certain applications. Projection imaging is more robust to bulk motion because of the averaging effects from repeated sampling of the low spatial frequencies and more tolerable streak artifacts (Glover and Pauly 1992; Gmitro and Alexander 1993). The trajectory can be modified so that each projection starts at the k -space origin ($k_r = 0$) as shown in Fig. 6b. A free induction decay can then be acquired for the imaging of tissues with very short transverse relaxation times (T_2), such as in the lungs (Bergin et al. 1991; Gewalt et al. 1993). Projection imaging can also be advantageous to suppress displacement artifacts in flow imaging (Nishimura et al. 1991). Continuous and interleaved radial acquisitions have been proposed for dynamic imaging in studies of the joints (Rasche et al. 1995, 1999), catheter tracking in interventional MRI (Rasche et al. 1997; Shankaranarayanan et al. 2001), endovascular procedures (Buecker et al. 2000), and swallowing examinations (Zhang et al. 2012). The properties of angular undersampling for faster imaging have been explored in various studies and will be discussed next.

4.4.2 Undersampled 2D Projection Imaging

If the number of projections is decreased below $N_{p,opt}$, then the angular sampling interval $\Delta k_{\phi,max}$ exceeds the radial interval Δk_r and the high spatial frequencies are not sampled adequately. This leads to a reduced artifact-free FOV with diameter d . The ratio of the diameters of the reduced FOV (d) and the full FOV is given by

$$\frac{d}{D} = \frac{\Delta k_r}{\Delta k_{\phi,max}} = \frac{2 N_p}{\pi N_r}. \quad (2)$$

Figure 7 shows the point-spread functions for a fully sampled radial trajectory ($N_p = \pi/2 N_r$) and with a reduced number of projections ($N_p \ll \pi/2 N_r$).

In contrast to Cartesian acquisitions, where undersampling leads to coherent ghosts, undersampling creates a noise-like appearance. The property of

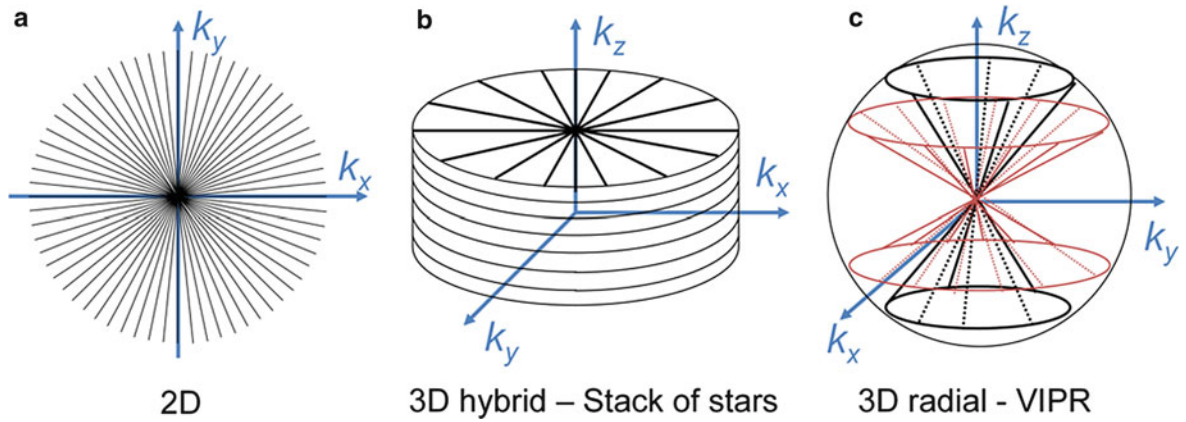


Fig. 5 Hybrid 3D projection reconstruction sequences use radial imaging in the through-plane direction (*left*) and fully sampled Fourier encoding in the slice direction. This ensemble

trajectory is often referred to as the stack of stars (*middle*). A truly 3D radial trajectory is shown on the *right*

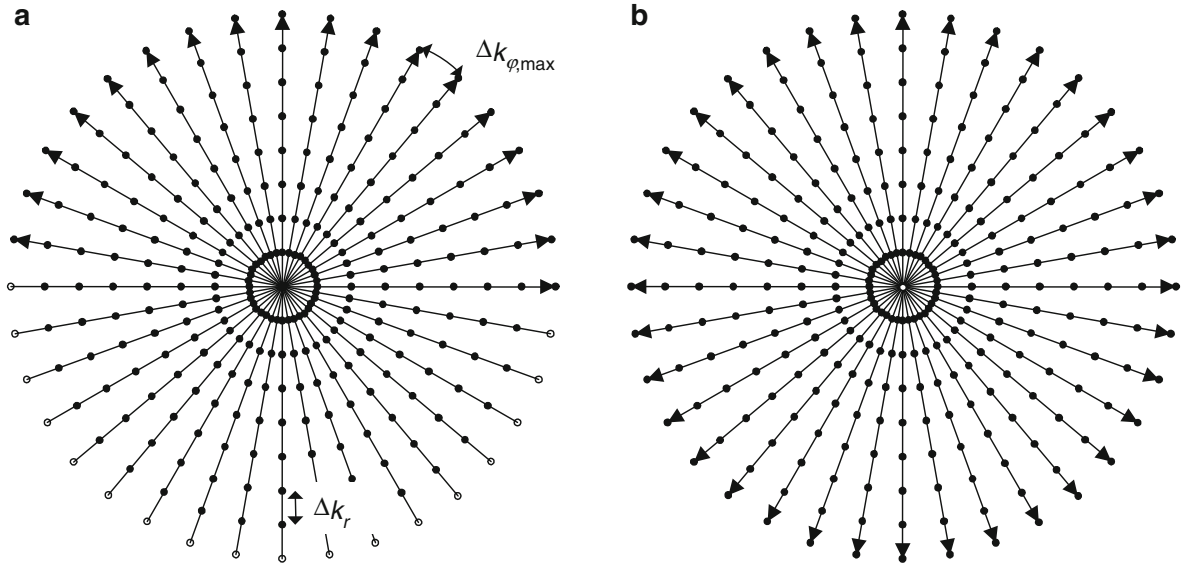


Fig. 6 Radial sampling schemes for projections from $-k_{r,max}$ to $+k_{r,max}$ (a) and from 0 to $+k_{r,max}$ (b). The starting point for each readout is shown as an *unfilled circle* and the end point is

shown by an *arrowhead*. Both schemes are characterized by a constant radial sampling interval Δk_r and a maximum angular sampling interval $\Delta k_{\phi,max}$

spreading the artifact at a distance from the object is used for interventional MRI with a large static FOV and a reduced dynamic FOV (Scheffler and Hennig 1998; Weiss and Rasche 1999). In another approach, Shimizu et al. (1998) purposefully undersampled at very high ratios to utilize the streaks for tracking the tip of a biopsy needle.

5 General Reconstruction of Non-Cartesian Acquisitions

In principle, there are many methods to reconstruct images from data acquired along non-Cartesian trajectories. In practice, methods that grid the acquired

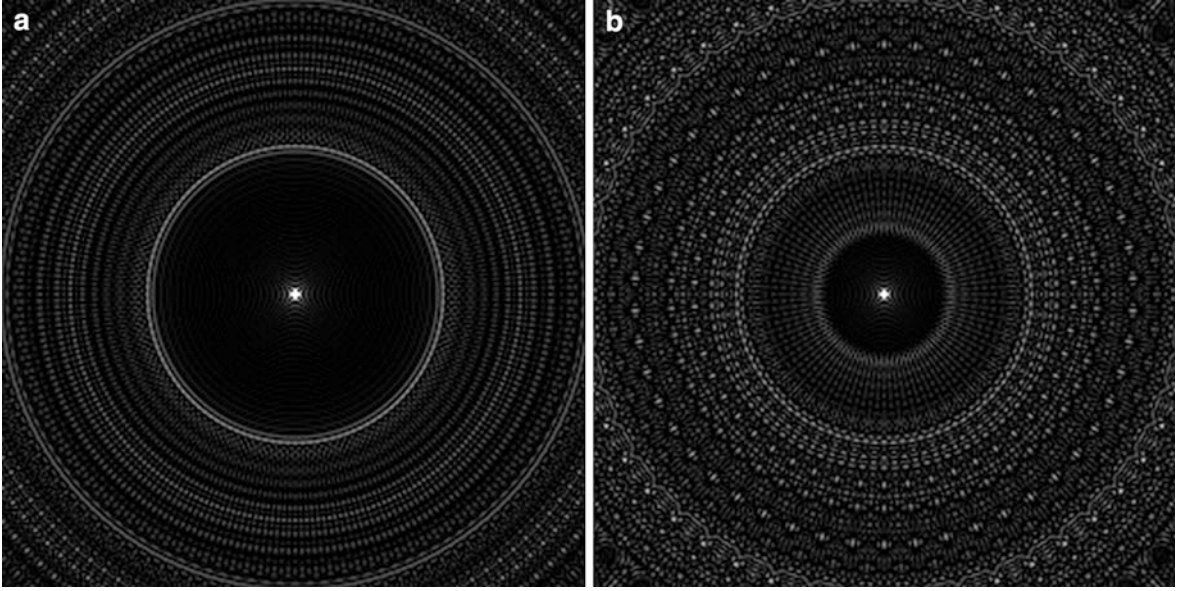


Fig. 7 Radial sampling with adequate sampling ($N_p = \pi/2N_r$) results in a symmetric point-spread function with an artifact-free circular field of view inside the first lobe (a) up to $r = 1/\Delta k_r$. Angular undersampling causes streak artifacts outside a

reduced FOV with a smaller diameter as shown for an undersampling factor of 2.5 in b. (Modeled after Scheffler and Hennig 1998)

data onto a Cartesian grid are most utilized in MR-guided imaging procedures. A more thorough description of gridding is provided in Bernstein et al. (2004).

5.1 Filtered Backprojection

Filtered backprojection was developed for CT and is an approximate implementation of the inverse Radon transform. In CT, the projections are directly measured and in MRI the projections are obtained by an inverse 1D Fourier transform of each radial line along k_r . Whereas filtered backprojection is still used extensively in CT, it has been widely replaced by gridding in radial MR image reconstruction.

5.2 Reconstruction by Gridding

MR data acquired along non-Cartesian trajectories are generally reconstructed with a process known as gridding (O'Sullivan 1985). If one represents $M(k)$ as the continuous Fourier transform of the magnetization of the object $m(x)$ and the k -space sampling points as

$S(k)$, sampled data are represented as $M_s(k) = M(k)S(k)$. Gridding interpolates the sampled MR data onto a Cartesian grid using a convolution kernel, $C(k)$, after compensating for differences in sampling density, $\rho(k)$, using the expression

$$M_c(k) = \left[\frac{M_s(k)}{\rho(k)} * C(k) \right] \cdot III(k).$$

Several methods exist to compute the sampling density function $\rho(k)$. A simple operation is simply $\rho(k) = S(k) * C(k)$, whereas a more accurate iterative approach is given by Pipe and Menon (1999). The reconstructed image volume is then

$$m_c(x) = \frac{1}{c(x)} \text{FTFT}_{3D}^{-1}(M_c(k)),$$

where x describes the position in the object domain. The resultant image must be divided by the Fourier transform of the convolution kernel, as convolution in one domain results in multiplication in the other.

The choice of the interpolation kernel is ultimately a trade-off between precision and speed and can have great impact on the reconstruction results (Jackson et al. 1991). In general, good convolution kernels act

over a quite localized region of k -space. As one tries to reduce error due to interpolation with wider kernels, the time necessary for interpolation grows rapidly. A simple and fast convolution kernel is the triangular window with a total width of two k -space samples. A popular kernel with more precise interpolation is the Kaiser–Bessel window. Dale et al. (2001) introduced the use of a precalculated lookup table that allows very rapid gridding in real-time applications. Gridding introduces some errors from the interpolation process but has been shown to produce images with better spatial resolution than filtered backprojection (Lauzon and Rutt 1998).

As sampling in one domain connotes replication in another, the choice of interpolation kernel also determines the amount of aliasing error that will result from adjacent image replicates in the image domain. In practice, this problem primarily affects tissue at the edge of the FOV. As the focal point in interventional imaging is usually not at the edges of the image, this problem is of less concern in vascular imaging. Intentionally gridding the data onto a finer grid artificially creates a larger FOV in which the replicates are further apart. Known as overgridding, this process reduces aliasing error. Although initially many workers overgridded by a factor of 2, recent work with improved interpolation kernels demonstrates minimal error with overgridding factors as small as 1.25 (Beatty et al. 2005). As an alternative to the computation involved in interpolating data onto a Cartesian grid, it is possible to simply use nearest-neighbor interpolation with a significantly enlarged Cartesian k -space matrix (Oesterle et al. 1999).

5.3 Image Degradation Due to k -Space Sampling Errors

The effects of uncompensated system delays and eddy currents lead to sampling errors between the theoretical k -space sampling locations and the actual k -space location. These errors manifest themselves as some type of blurring in non-Cartesian methods, particularly as one moves away from the center of the image. These errors are much more benign in Cartesian imaging, as the errors are predominantly the same along each phase-encoding acquisition. As these delays and eddy currents can change simply with gradient coil heating, measuring these errors quickly

without a service procedure is essential for robust imaging.

5.3.1 Linear Eddy Current Correction

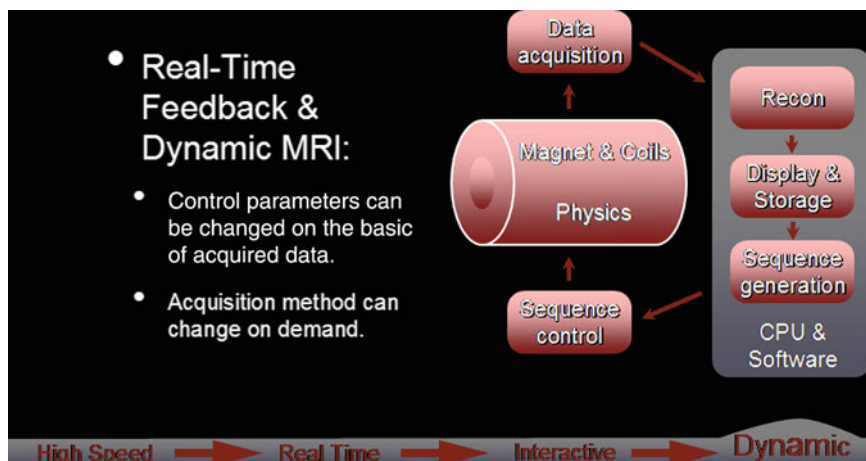
Multiecho, echo-planar, and non-Cartesian trajectories place high demands on the gradient hardware, leading to increased induction of eddy currents. For conventional Cartesian acquisitions, these trajectory errors may be ignored as the resulting phase shift across the single readout direction is constant and does not appreciably affect image quality. This is not the case for multiecho, echo-planar, and non-Cartesian trajectories, especially those employing bipolar readouts.

Methods to compensate for these errors fall into two categories: system characterization and k -space measurement. System characterization methods model the gradient system as a linear system and then determine a modulation transfer function to relate the theoretical input waveforms and the actual waveforms that are created. Although this method seems well suited for interventional imaging as the characterization needs to be performed only once, most correction schemes use k -space measurement algorithms because of drift and other complexities in MR systems.

One can further break down k -space measurement methods into methods that use numerous self-encoding pulses of different amplitudes prior to examining the readout gradient and methods that exploit localized signal. Owing to the limitations in speed provided by these methods, methods that exploit the phase of localized signal are gaining interest. These measurement methods can be performed periodically when scan planes and imaging trajectories are changed during an interventional procedure, similar to periodic acquisitions of B_0 maps.

If a small slice off-isocenter is excited and then subjected to the readout gradient under consideration, the entire test slice develops a phase that is the integral of the applied gradient field. The integral, of course, is proportional to the k -space trajectory. These measurements can be repeated on each physical gradient, or logical gradient if an oblique slice is being imaged. This method, suggested by Duyn et al. (1998), can be performed in only a few excitations and thus can be performed easily prior to each patient scan. The difference in actual k -space trajectory and the ideal trajectory can then be provided to the reconstruction.

Fig. 8 Vendors first designed MRI systems primarily for diagnostic purposes where data flows one way as a pipeline. The interactive, dynamic nature of interventional MRI requires that all subsystems be able to communicate simultaneously in both directions



5.3.2 Off-Axis Imaging

Timing errors in the hardware demodulator may lead to phase errors if one uses real-time frequency demodulation to center at a location off-isocenter. The easiest way to mitigate this problem, if one can increase the data acquisition rate, is to avoid using the real-time frequency demodulation feature and to do all demodulation digitally during reconstruction. This solution doubles the input data rate for the reconstruction processing chain, but the size of the k -space matrix need not double. A fractional overgridding factor of only 1.25 is often sufficient (Beatty et al. 2005). If one cannot simply increase the receiver bandwidth of an acquisition as if one is at the isocenter, then one must take care to account for these errors. A method to quickly measure the actual delay and produce the necessary phase shifts is provided in Jung et al. (2007).

6 Dynamic MR Systems

Typically much of the needed MR-guided imaging infrastructure is consistent from application to application. Although interventional tasks differ widely, they all require similar basic capabilities of scanner control, data acquisition and reconstruction, geometry transformations, 3D visualization, and interactive user interfaces. To accomplish dynamic MRI capabilities, two-way communication is required between several hardware and software subsystems, as shown in Fig. 8. However, MRI system manufacturers first designed their architectures to operate as a one way pipeline. Thus real-time MRI requires a significant

investment in designing, coding, and testing an alternative data and control pathway for research sites and the imaging tools needed to translate new MRI-guided procedures into widespread clinical use through commercialization.

Tight coupling of the software that controls a manufacturer's acquisition, reconstruction, control, and visualization systems often requires intimate knowledge of many software modules before researchers and developers can build image-guided capabilities. Even with that knowledge, building a flexible system that can support device visualization, rapid real-time imaging at lower resolution, and high-resolution imaging with multiple contrast mechanisms has proven challenging. Traditionally, these tasks have been provided by different pulse sequences, and switching between pulse sequences required a large overhead cost in terms of time and complexity. Fortunately, MRI system manufacturers and some third-party providers are providing alternative real-time imaging development environments that pave the way for this paradigm shift from one-way data pipelines to interactive, two-way MRI systems.

There are some predominant interventional tasks that real-time systems may be designed to perform. Road map and real-time images may be visualized together in the same geometry, typically including biplane views. When a catheter or needle is being tracked, the tip may be colorized and overlaid on anatomical images. Scan plane and pulse sequence parameters may be changed in real time. Some specialized tasks that real-time systems can accomplish include automatic, dynamic image parameter adjustment in response to active device

Fig. 9 Interactive Front End (Siemens) user interface for general MRI-guided navigation. (Courtesy of Christine H. Lorenz, Siemens Corporate Research, Baltimore, MD, USA)

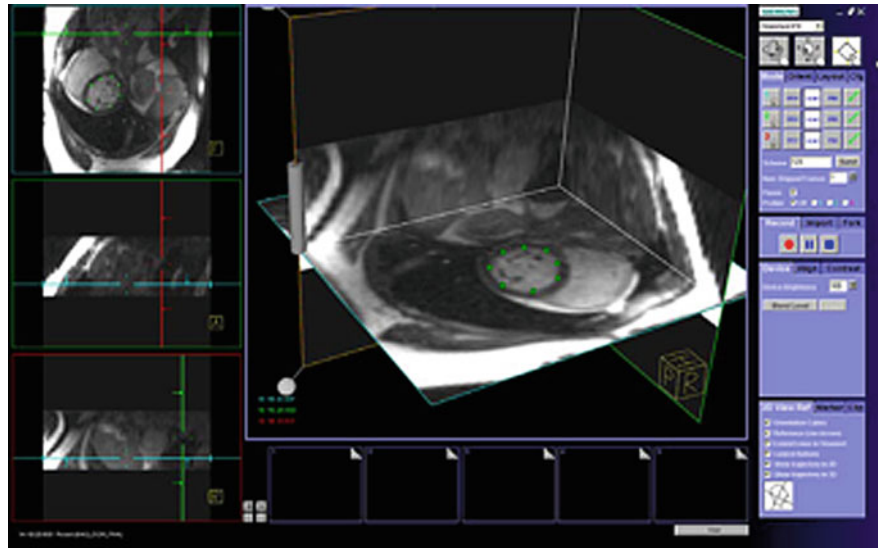
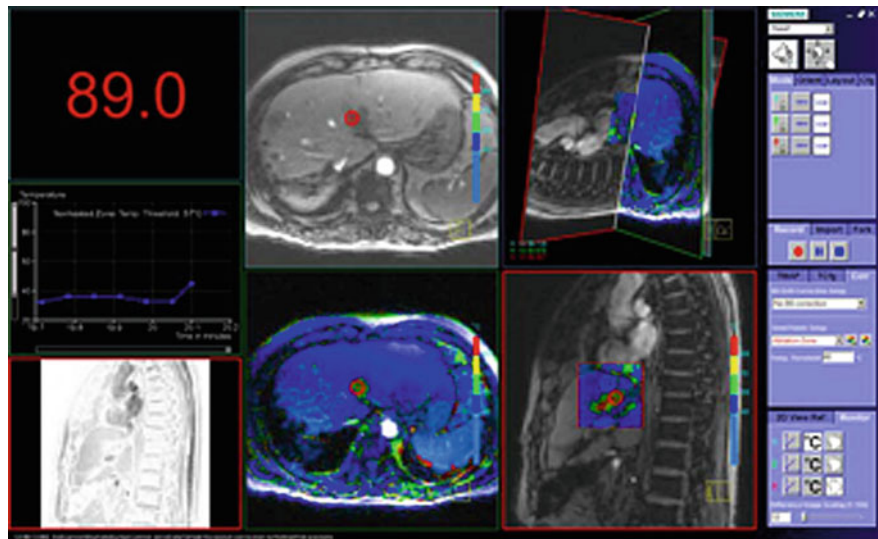


Fig. 10 Interactive Front End (Siemens) user interface for MRI-guided thermal monitoring of a liver procedure. Various planes, with both grayscale and temperature monitoring views, are provided to guide the procedure. (Courtesy of Christine H. Lorenz, Siemens Corporate Research, Baltimore, MD, USA)



movement (Zhang et al. 2000) and, more recently, passive device movement (Patil et al. 2009). Systems designed specifically for real-time interventional applications are advantageous because they enable rapid and flexible incorporation of these specialized tasks.

The Siemens IFE has arguably the longest development track record in promoting a real-time, interactive environment for MRI-guided procedures. The IFE is a prototype designed to work together with an interactive real-time pulse sequence (BEAT_IRTTT) to enable navigation during interventional or other real-time examinations. It replaces common scanner

controls with a graphical control, and allows the user to focus on navigation rather than on manipulating classic MR parameters. The IFE has three modes: (1) real-time navigation (Fig. 9), (2) planning for percutaneous procedures, and (3) thermal visualization (Fig. 10). In the navigation mode, the user can superimpose previously acquired DICOM images or segmented 3D objects along with real-time images in order to aid navigation. In the planning mode, the user can plan trajectories in three dimensions for needle-based procedures and verify that no critical structures are in the path. The planning mode also transfers determined scan planes to the real-time navigation mode to reduce

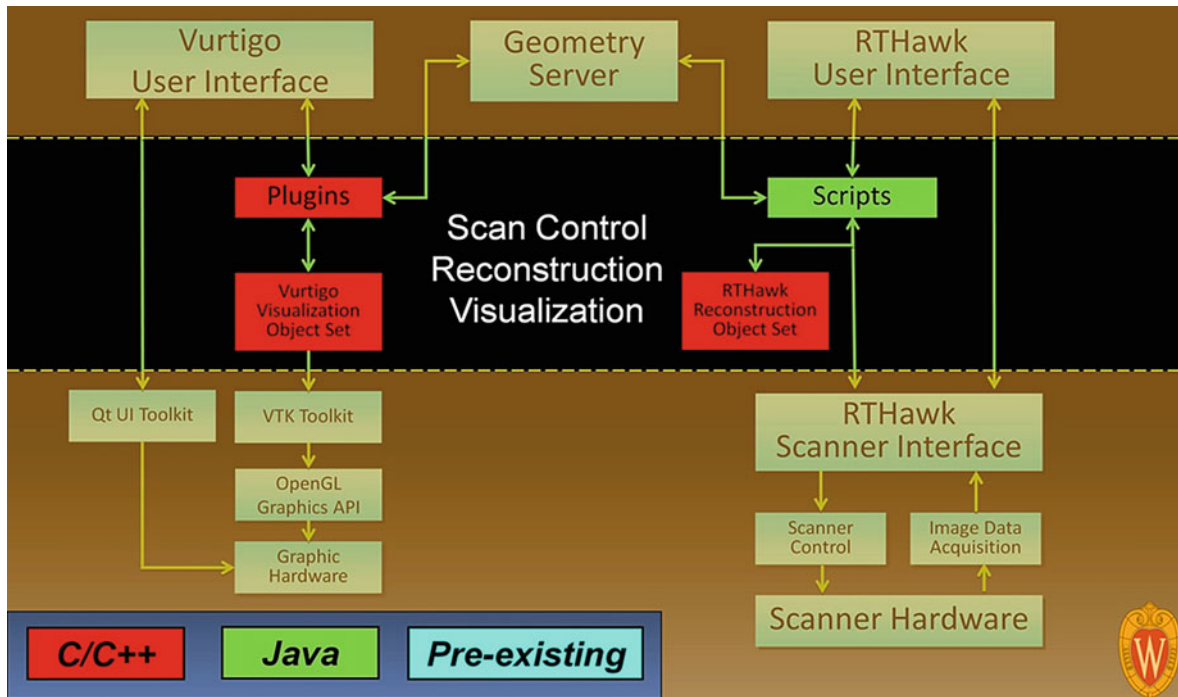
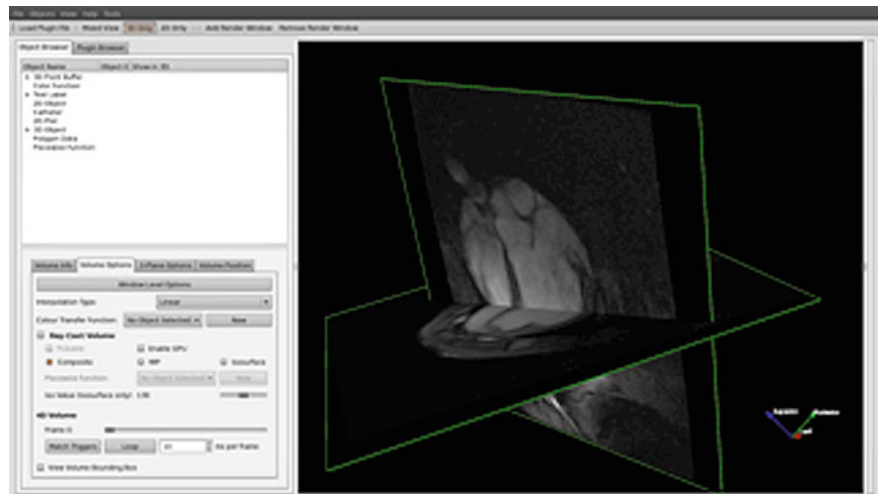


Fig. 11 Most interventional research and development platforms require extensive software development of user interfaces (*above the first dotted line*) and scanner control and reconstruction software at the vendor-specific level (*below the*

second dotted line). The RTHawkTM/Vurtigo platform allows one to work at the conceptual level (*between the dotted lines*), allowing concentration on the actual application

Fig. 12 Biplane visualization of an imported 3D data set depicting a porcine heart on the Vurtigo platform. Multiple coregistered data sets can be visualized and manipulated simultaneously



the preparation time. The thermal visualization mode works together with a proton resonance frequency based thermography sequence to allow the user flexibility in evaluating thermal parameters, including setting points for thermal monitoring, peak temperature alerts, and a variety of display methods.

A third-party platform is based upon two integrated, extensible software architectures—the RTHawkTM system for real-time acquisition and reconstruction and the Vurtigo system for interactive visualization. RTHawkTM (HeartVista, Palo Alto, CA, USA) is an extensible software platform that permits a variety of

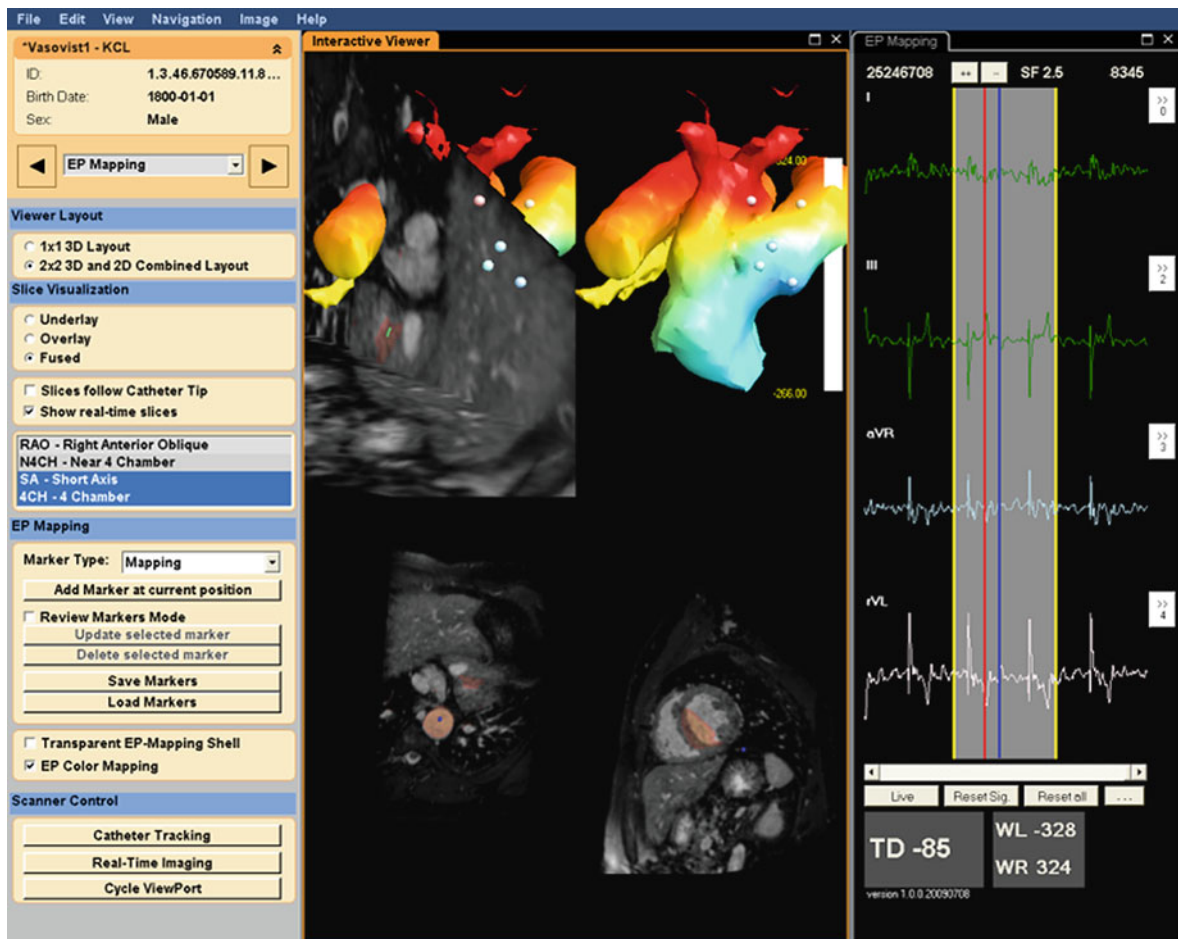


Fig. 13 The eXternal Control (Philips) interface used to guide a cardiac electrophysiology procedure

pulse sequences, acquisition trajectories, and reconstruction techniques to be easily developed and interleaved (Santos et al. 2004). The acquisition is controlled by a “stub” pulse sequence program that runs on the scanner computer and communicates with a program running on another computer. Acquired raw data (views) are tagged with contextual information and are sent to this external program for processing. The reconstruction, user interface, and visualization are written in JavaScript as a pipeline of processing blocks. A base set of blocks is included with the RTHawkTM system, and additional blocks can be programmed as C++ objects and added through a “plug-in” architecture. The user interface is built upon the Qt open-source cross-platform application. Vurtigo (Sunnybrook Health Sciences Centre, Toronto, Canada) is an open-source visualization system that simplifies simultaneous display

and interaction with multiple 3D and 2D data sets (Pintilie et al. 2009). Its user interface is built upon Qt, with visualization based on VTK (Kitware, Clifton Park, NY, USA), with OpenGL rendering. It has built-in support for object types, including 3D volume images, 2D slice images, device positions, and 3D point meshes, with multiple rendering modes for each.

These objects can be manipulated manually by the end user and automatically by user-developed plug-ins, which are written as C++ objects. Plug-ins can implement a variety of behaviors, anything from automating simple object interaction up to the “workflow” for entire interventional applications. The implications for developing MRI-guided procedures with such a platform are depicted in Fig. 11. An example of visualization of prior acquisition of a porcine heart using Vurtigo is shown in Fig. 12.

The XTC interface (Smink et al. 2011), developed by Philips, provides fast, low-latency, low-jitter access to output data, and flexible access and update of scan parameters during scanning. The system is applicable to all scan protocols; the only restriction is a slight increase in the minimum echo time. Communication with the reconstruction and scanner processes on the host and interfaces to a networked application is accomplished using a minimalistic CORBA interface which uses TCP/IP as the transport layer.

XTC has been successfully used in four applications. In the Philips Sonalleve MR–high-intensity focused ultrasound platform, CE-approved for the ablation of uterine fibroids, the planning console selects remotely the required scan protocols, downloads DICOM data sets for planning subsequent scans and high-intensity focused ultrasound therapy, starts and stops dynamic scanning, retrieves reconstructed image data as they become available, and calculates temperature maps from the data. The second example is the RealTI open-source environment available for Windows and Linux. A graphical user interface is developed in IDL and a wrapper layer is made in C++. The environment forms a modular framework and is now used for MRI-guided thermotherapy and local drug delivery applications and gene expression. Available modules include motion tracking, thermometry, and T1 and T2* calculation. The third example is the MRI-guided electrophysiology (EP) project, in which the MR–EP application integrates MR images from the scanner with cardiac EP data from an EP recorder (Fig. 13). The last example is the XTC datadumper, written in C#, which can be used as an example project for other standalone prototype applications. It runs on an external Windows PC and dumps all retrieved images and other data sources in files on the local hard disk. It reads continuously local geometry update files and sends this information, when necessary, to the scanner.

7 Conclusion

Designing and implementing all of the imaging needs for an MRI-guided procedure can sometimes be seen as a time-consuming, challenging task. Determining the imaging performance requirements beforehand (resolution, frame rate, FOV, etc.) will determine where shortfalls in performance require more sophisticated procedures and algorithms. Where performance

needs are moderate, more conventional imaging methods can be utilized. The advent of dynamic MR-guided imaging development platforms provides tool-boxes of commonly used imaging procedures that significantly reduce the scope of the overall imaging development task.

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