Multiple Field of View MR Fluoroscopy

Pelin Aksit, J. Andrew Derbyshire, Jean-Michel Serfaty, and Ergin Atalar

This work describes a real-time imaging and visualization technique that allows multiple field of view (FOV) imaging. A stream of images from a single receiver channel can be reconstructed at multiple FOVs within each image frame. Alternately, or in addition, when multiple receiver channels are available, image streams from each channel can be independently reconstructed at multiple FOVs. The implementation described here provides for real-time visualization of the placement of guidewires and catheters on a dynamic roadmap during interventional procedures. The loopless catheter antenna, an electrically active intravascular probe, was used for MR signal reception. In 2D projection images, the catheter and surrounding structures within its diameter of sensitivity appear as bright signal. The simplicity of the resulting images allows very-narrow-FOV imaging to decrease imaging time. Very-narrow-FOV images are acquired on MR receiver channels that collect guidewire or catheter data. These very-narrow-FOV images provide very high frame rate continuous, real-time imaging of the interventional devices (25 fps). Large-FOV images are formed from receiver channels that collect anatomical data from standard imaging surface coils, and simultaneously provide a dynamic, frequently updated roadmap. These multiple-FOV images are displayed together, improving visualization of interventional device placement. Magn Reson Med 47: 53–60, 2002. © 2002 Wiley-Liss, Inc.

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Intravascular magnetic resonance (MR) methods for imaging arterial walls with high resolution for plaque characterization have been reported by several investigators (1–8). Intravascular MR compares favorably with other clinical diagnostic methods of evaluating vascular disease, such as intravascular ultrasound (9) and angiography (10), with superior contrast and resolution of the vessels and the vessel wall.

To accurately and fully utilize these benefits, visualization of the interventional device placement and the surrounding anatomy, with frequent updates, is essential. Conventionally, percutaneous placement of the interventional devices is performed under x-ray fluoroscopy. This requires the presence of an x-ray fluoroscope and an MR scanner in the same room, with a compatible table. The disadvantages are that alternating between two imaging modalities can alter the position of the catheter, it is quite costly, and both the patient and the physician are exposed to harmful ionizing radiation. Therefore, an MR fluoroscopy technique to visualize interventional procedures with high spatial and temporal resolution is desirable.

Many innovative techniques have been developed to make MR guidance of interventional procedures possible. MR fluoroscopy (11,12) has been used to increase temporal resolution in both interventional (13,14) and noninterventional (15–17) procedures. Fluoroscopic methods are frequently combined with dedicated real-time reconstruction hardware or algorithms, and some researchers (18–20) have devised techniques to actively track the tip of interventional devices.

Previously, we developed a method that combines interventional device and anatomical roadmap images from multiple channels, during MR-guided interventional procedures (21). It has also been demonstrated by many researchers that narrow-FOV imaging can be used to decrease imaging time when the MR signal is localized to a small region of interest (22–24).

In the present study, the method of interest for interventional device tracking is the electrically active approach using the loopless catheter antenna design (7), which consists of a coaxial cable with an extended inner conductor. On a projection image, acquired without slice selection, the entire catheter and surrounding structures within its diameter of sensitivity appear as bright signal in a long, narrow, connected region. The loopless catheter antenna was the design of choice for this study because of its high signal concentrated in a narrow region around the catheter. However, several other interventional-device designs provide catheter images that may be compatible with the multiple-FOV MR fluoroscopy method, such as the flexible catheter antenna (5) and gadolinium-coated catheters (25).

We have shown in an earlier work that the simplicity of a catheter projection image allows the use of a very narrow FOV acquisition to decrease imaging time (22). However, when a very-narrow-FOV imaging protocol is used, images may be wrapped (Fig. 1). This image overlap in the phase-encoding direction can be corrected using a standard unwrapping algorithm. Unwrapping is possible as long as the FOV in the phase-encoding direction is large enough to contain the catheter image without any overlaps where the catheter image crosses itself. The unwrapping algorithm extracts the geometric shape information from the catheter, and represents it as a set of skeletal points (Fig. 2b). The aliased image is then displayed according to these unwrapped catheter positions (Fig. 2c). The process of unwrapping the narrow-FOV image results in a nonrectangular FOV. This FOV itself will change shape to track the location of the catheter. In short, clinically useful catheter images can be extracted from an aliased narrow-FOV image (26,27).

When the simultaneous acquisition of the roadmap image is desired, the previously described very-narrow-FOV imaging technique is not compatible, because the image overlaps that occur when a surface coil is used cannot be
corrected. Here we propose a new data acquisition and image reconstruction technique to allow both very-narrow-FOV for tracking catheters with high frame rates and large-FOV for displaying the roadmaps.

In this work, the concept of merging data from surface coils and catheters was extended to incorporate the temporal resolution advantages of narrow-FOV imaging on interventional device channels, while simultaneously obtaining data for larger-FOV anatomical images.

**METHODS**

**Data Acquisition**

The aim of this data acquisition strategy is to fill the $k$-space in a flexible way to allow reconstruction of multiple-FOV images during acquisition. Very-narrow-FOV images are used to provide real-time catheter tracking with a high temporal resolution, and large-FOV images provide a dynamic roadmap image with a lower temporal resolution. While small subsets of the $k$-space provide fast updates of the catheter, the accumulation of such subsets should fill the $k$-space with appropriate data to reconstruct a full FOV roadmap image. Within each small set, the order of acquisition should be sequential to minimize ghosting of the linearly moving catheter image. It should also be possible to interleave consecutive sets and obtain a uniformly sampled $k$-space as data sets accumulate over time.

These requirements led to a partially bit-reversed phase-order. This concept can be illustrated using, for example, an $N_{full} = 256$ and $N_{set} = 8$, where $N_{full}$ is the total number of phase-encode lines and $N_{set}$ is the number of phase-encode lines in each small set. Eight bits are required to represent the phase-encode lines $0–255$ in binary form. Three of these bits are required to represent the sequential acquisition of phase-encode lines within each small set, $0–7$, while the other five bits keep track of the acquisition order of consecutive sets, $0–31$, in an interleaved fashion, as depicted in Fig. 3. In the eight-bit binary representation of the acquisition order of 256 phase-encoding lines, the three most significant bits will be incremented linearly within a set, and the five least significant bits are incremented in a bit-reversed fashion between consecutive sets.

Figure 4 depicts the order of acquisition for $N_{full} = 256$ and $N_{set} = 8$. On the right side of the figure, $k$-space lines representing the acquired data, possible larger groupings, and images reconstructed with different amounts of data are shown. Each set can be reconstructed and displayed individually, to produce one frame of catheter or guidewire images with $1/32$ the full FOV and $8 \times TR$ temporal resolution. Consecutive sets can be combined for an increased FOV at the expense of reduced temporal resolution. Although the temporal resolution of larger FOV images is lower, they can be updated as each new set is acquired at intervals of $8 \times TR$, providing a smoother transition over time (11).

**IMAGE RECONSTRUCTION**

**Multiple FOVs Within One Image**

The phase-encoding algorithm described permits imaging different parts of an object with different frame rates in real-time. The effective FOV required for successful unwrapping of the catheter or guidewire image depends on the diameter of the interventional device, its alignment, and overall image quality. If the catheter aligns primarily in the phase-encoding direction, the catheter image wraps onto itself, and the FOV needs to be increased until the ambiguity is resolved. However, if a part of the catheter forms a loop, or a pigtail shape, the frame rate of the entire movie does not need to be decreased to eliminate ambiguity in that part of the image. On a variable FOV image, the FOV in the phase-encoding direction may be varied as a function of position in the frequency-encoding direction.

Figure 5 demonstrates this concept. Data is acquired in small sets of eight $k$-space lines, and is 1D Fourier transformed to spatially resolve it in the readout direction. The top and bottom portions of the data are reconstructed with...
another 1D Fourier transform in the phase-encoding direction, and displayed at a high frame rate, with a narrow FOV. The reconstruction of the middle portion of the data is delayed, however, until the next set of eight k-space lines are acquired. The top and bottom portions of this second set are also reconstructed and displayed at a high frame rate, with a narrow FOV. The middle portions of groups comprising two sets of eight k-space lines are combined, doubling the apparent density of sampling in this section of the image. These lines are reconstructed with a 1D Fourier transform with twice the size in the phase-encoding direction, yielding a section on the image with twice the FOV of the top and bottom portions (Fig. 5).

### Multiple-FOV Imaging on Multiple Receiver Channels

Many commercially available MR scanners are equipped with at least four receiver channels that can simultaneously acquire image data from separate receiver coils according to a prescribed pulse sequence. Using these available scanner resources, the multiple-FOV MR fluoroscopy method can provide a dynamic anatomical roadmap image in addition to viewing interventional devices at very high frame rates.

In the multiple-FOV method, two or three channels collect data through phased-array surface coils for a roadmap image, while the other channels are connected to an active guidewire and an active catheter. Data are acquired with the modified phase order described in the data acquisition section, and automatically transferred to a workstation directly connected to the MR scanner after acquisition of each small set of k-space lines.

An application on the workstation controls the processes for data transfer, reconstruction, and display. A graphical user interface allows control of all user preferences. The reconstruction options include 1) whether the channel is on or off; 2) the spatial resolution of the incoming data (which must be matched with the scanner console); 3) the desired FOV on the channel (thus, the frame rate); 4) whether unwrapping is necessary; and 5) the unwrapping rate for narrow-FOV images. The display options include 1) independent contrast/brightness adjustment of each channel; 2) an option to negate any channel (common radiologist practice for viewing the surrounding anatomy); and 3) an option to view each channel on a different color map, to easily distinguish between multiple interventional devices and signal from blood vessels.

Each channel is then reconstructed independently at the specified FOVs. The resulting images from multiple channels are combined using the square root of the sum of squares of independent channels. The resulting frame rate on each channel depends on the TR during acquisition and the number of phase-encode lines used for reconstruction. The refresh rate of the display is that of the fastest channel.

Unwrapping of the images from particular channels may be performed if required. For catheter and guidewire images, the degree of image overlap depends on the apparent width of the interventional device, and its orientation with respect to the phase-encoding direction. The unwrapping algorithm is typically activated for images with a 1/4 FOV or less. When aliasing in the phase-encoding direction is allowed, there is an ambiguity in the absolute position of the catheter. The resulting images may be shifted by exactly p*Nset, any multiple (p) of the number of phase-encode steps in the small data set (Nset). This ambiguity is resolved at the commencement of scanning. As soon as enough data arrives to form a full FOV image, the correct catheter position is calculated, automatically determining p. All other images are registered using the information
from previous frames, and ambiguity is eliminated. Once unwrapping is complete, the narrow-FOV image is rearranged to place the catheter on the center of the displayed image.

Typically, data is collected in sets of \( N = 8 \) \( k \)-space lines; this allows temporal resolution up to \( 8 \times \) TR on guidewire and catheter channels. For an acquisition with a TR of 5 ms, this provides a frame rate of 25 frames/s. For the roadmap channels, it is necessary to wait for sufficient data to form a large FOV image to avoid aliasing of the complex structures of the surrounding anatomy. For a full FOV (256 phase encodes), this provides a temporal resolution of 1.28 s (0.78 frames/s) for anatomical images.

**EXPERIMENTS AND RESULTS**

All experiments were performed on a GE 1.5T CV/i MRI scanner (GE Medical Systems, Waukesha, WI) using the modified fast gradient-echo (FGRE) pulse sequence (four receiver channels). The real-time data transfer, reconstruction, and display programs were run on a SunUltraII Workstation (Sun Microsystems, Mountainview, CA) directly connected to the scanner by a high-bandwidth data bus (Bit3 Corporation, St. Paul, MN). Experiments were monitored on an in-room display (Aydin Displays, Horsham, PA). The FGRE pulse sequence was modified to acquire data continuously using the proposed phase-encoding technique. In addition, the real-time process (RTP) within the FGRE was programmed to make data available on the workstation side. These modifications provide the workstation software with a signal that announces the completion of the acquisition of a new small set of \( k \)-space data.

Figure 6 provides an experimental demonstration of the increase in temporal resolution of an image by narrowing its FOV. The phantom consisted of an S-shaped tube that was immersed and fixed in a bath of saline solution. The catheter was retracted while data was being acquired with the modified phase order. The tip of the catheter is at the top of the images, and it exits from view at the bottom. The modified FGRE pulse sequence was used for the acquisition of 256 lines of \( k \)-space data (TR/TE = 5/1.6 ms, FOV = 25 cm, 256 \( \times \) 256 partial NEX). The following images were reconstructed from different groupings of the same data (Fig. 6).

The full FOV image reconstructed by combining all 32 sets of eight \( k \)-space lines shows ghosting due to the motion of the catheter during the data acquisition time of 1.28 s (0.78 frames/s). Alternatively, these same 32 small sets can be combined into two temporally contiguous time frames, each comprising the union of 16 small sets (the even and odd lines of the \( k \)-space), providing two 1/2-FOV images. The temporal resolution doubles (1.56 frames/s), and the large jump of the catheter is visible in the unwrapped frames. When every fourth line of \( k \)-space is used for reconstruction, four distinct frames with 1/4 the FOV are obtained from the same amount of data that forms a full FOV image. The temporal resolution quadruples, and more details of the catheter motion are visible on the unwrapped frames (3.12 frames/s). The same trend continues with 1/8 and 1/16 the FOV. The very narrow nonrectangular FOV very smoothly follows the catheter as it moves out of view at 6.14 frames/s and 12.5 frames/s, respectively.

**Multiple FOVs Within One Image**

Figure 7 shows results obtained from a phantom experiment by reconstructing different FOV sections on catheter images. The phantom consisted of a bath of saline solution; a small cup was placed into the saline bath, and the catheter was looped around it. The tip of the catheter is at the bottom of the images. The catheter was retracted toward the top of the images while data was being acquired with the modified phase order.

The full FOV image shows significant ghosting and blurring due to the motion of the object during the imaging time of 1.28 s (TR = 5 ms, 256 \( \times \) 256 matrix). The variable FOV images (Fig. 7) were reconstructed at different FOVs regionally, as described in the Image Reconstruction section. The central section containing the loop required a larger FOV than the tip or tail of the catheter, and thus had a lower temporal resolution. The center of the image was updated 3.12 times/s at the smallest FOV that contains the loop, without affecting the tip and tail of the catheter, which were still imaged with a very narrow FOV at 25 frames/s.

**Multiple-FOV Imaging on Multiple Receiver Channels**

Multiple-FOV imaging on multiple receiver channels allows tracking of more than one interventional device on a dynamic roadmap image, and was demonstrated in an
animal study. In this experiment, an active guidewire and an active catheter were inserted through the carotid artery of a dog and advanced to the aortic arch and heart under MR guidance. Two phased-array surface coils were used to provide anatomical roadmap images.

Figure 8 shows one frame of the movie recorded during the procedure, as well as the outputs of individual receiver channels. Channel 0 was connected to the catheter, and channel 1 was connected to the guidewire. Channel 2 was the surface coil on the anterior side of the

![Figure 6: Demonstration of the reduction in FOV and corresponding increase in temporal resolution, with the modified phase order. The top row is a full FOV image of the catheter moving inside the phantom, acquired in 1.28 s (0.78 frames/s). In the second row the same data was used to form two 1/2-FOV images, with twice the frame rate. The same data is rearranged to form 1/4-, 1/8-, and 1/16-FOV images to increase the frame rate in the succeeding rows. As the FOV becomes narrower, temporal resolution increases, and in the last row the catheter is tracked at 12.5 frames/s.](image)

![Figure 7: Variable FOV images of a looped catheter. (Imaging parameters: FGRE, data matrix = 256 × 256, flip angle = 4°, TG = 0 dB, BW = 62.5 kHz, TR/TE = 5.1/1.6 ms, no slice selection, arrow represents the frequency-encoding direction). Frame rate of tip and tail of catheter = 25 frames/s. Frame rate of loop = 3.12 frames/s.](image)
animal, and channel 3 was the surface coil on the posterior side.

Both active interventional devices were used in the receive-only mode of operation, and their diameter of sensitivity extends to several centimeters away from the device. The minimum FOV required to avoid aliasing of image components far away from the catheter or guidewire was 1/8 the full FOV (Fig. 8). In this case, the catheter and guidewire data were reconstructed after the acquisition of two sets of eight k-space lines, providing a frame rate of 11.6 frames/s for catheter tracking while maintaining a better than 3 mm in-plane resolution. The reconstruction of the anatomical surface coil data from channels 2 and 3 was delayed until the k-space was filled to form a full FOV image. The frame rate of the roadmap formed by the surface channels was 1.45 frames/s. As expected from MRI theory, the anatomical images benefit from increased signal-to-noise ratio (SNR) due to the much longer imaging time, when data from the two receiver coils (anterior and posterior) are combined using standard phased-array magnitude image reconstruction. Corresponding movies can be seen on our web page (31).

The Fastest Fourier Transform of the West (FFTW) library (30) was used for image reconstruction, providing performance that ranges from 20 ms for a 256 × 256 transform to under 1 ms for a 256 × 8 transform (in-place transforms with four-byte data). The unwrapping algorithm behaves similarly to the Fourier transform in terms of time consumption. Larger-FOV images (128 × 256) are unwrapped in up to 8 ms, while unwrapping of narrow-FOV images is on the order of 2 ms. Due to the nature of the unwrapping algorithm, it is not possible to provide its exact timing behavior. The amount of work performed by the algorithm depends strongly on the appearance of the interventional device on the MR image. Finally, the other operations performed during reconstruction increase the total time by around 10 ms, depending on the options chosen by the user. Image display latency is also determined by image size and other user preferences (100–140 ms).

**DISCUSSION**

The combination of our new data acquisition strategy with real-time reconstruction and display allows visualization of multiple interventional devices simultaneously with a dynamic roadmap image, and provides continuous online feedback to the interventionist. The advantages of the technique, and some problems we faced are discussed below.

The loopless antenna design incorporated into the guidewire or catheter results in MR signal originating from the entire length of the device. Projection imaging avoids the need for slice selection or adjustments to the imaging slice, since the entire length of the catheter is visible at all times on a projection image. There are no obstacles to the catheter view, and it ensures that there is no buckling or folding of the device.

By combining the strengths of the loopless antenna design with a continuously updated roadmap image, all registration problems that may result from motion during the procedure are eliminated, as well as the need to periodically reacquire new roadmap images. Many fast sequences run at higher frame rates by using a reduced spatial resolution. However, for interventional device tracking, it is essential that the catheter be viewed with the highest spatial and temporal resolutions possible. The multiple-FOV MR fluoroscopy method achieves high frame rates by decreasing the FOV rather than the spatial resolution. The reconstruction of each channel data is independent of the other channels, and the FOV can be controlled by the user to achieve the best frame rate, according to a tolerable amount of reduction in the FOV. The strength of this method is that during an experiment the resulting frame rate and FOV on each channel are controlled by the reconstruction algorithm, not the data acquisition.

The reconstruction of each channel runs as a separate process, and the user interface allows independent manipulation of each channel view throughout the procedure for better visualization. For instance, during a balloon angioplasty the guidewire view is not necessary or desirable during balloon inflation and can be turned off, whereas during device placement it is necessary to visualize both the guidewire and catheter. Color-coding of channels also proves to be useful for ease of distinction when signals from multiple devices and blood overlap (Fig. 8).

Implementation of variable FOVs within an image addresses an important limitation of the method of unwrapping narrow-FOV images. If certain portions of the catheter

**Fig. 8.** Multiple-channel, multiple-FOV images of a dog. (Imaging parameters: modified FGRE pulse sequence, TR/TE = 5.6/1.6 ms, FOV = 36 cm, flip angle = 7°, acquisition matrix = 256 × 128 (partial NEX), no slice selection, arrow represents the frequency-encoding direction). Frame rates of the FOVs: catheter, 11.6 frames/s (1/8); guidewire, 11.6 frames/s (1/8); roadmap, 1.45 frames/s (full).
align with the phase-encoding direction, more data will be needed for successful unwrapping. The maximum allowable angle that the catheter can be from the readout direction depends on its apparent width (w), and the FOV in the phase-encode direction. If this angle exceeds \( \cos^{-1}\left(\frac{w}{\text{FOV}}\right) \), the catheter image wraps onto itself and the FOV needs to be increased, resulting in a lower temporal resolution. Implementing variable FOVs within an image solves this problem by allowing the frame rate to be reduced only for problem sections of the image, without affecting the frame rate of the other parts of an image, and eliminates the need to slow down the entire movie. Interactively rotating the scan plane to align the catheter with the readout direction is a possible solution. However, due to the anatomy (vessels mostly aligned with the S/I direction), the criterion described above may be violated at only short segments of the catheter. A regional solution was needed, and was addressed by the variable-FOV approach within individual images.

The FOV can be narrowed as long as unwrapping of the aliased catheter image is possible and SNR permits. With narrow-FOV imaging, the catheter image is allowed to wrap onto itself, and the FOV can be narrowed if all signal from the catheter is contained within each line along the phase-encoding direction of the narrow-FOV image without aliasing. If a catheter has a wide sensitivity, or there is significant coupling between the catheter and other receiver coils, and the catheter picks up signal from a wide area, the FOV should not be decreased beyond the point where these other components in the image start to cause aliasing. Otherwise, overlaps cause intensity variation between successive frames, and UNFOLDing (29) may be necessary.

Although in the present work the application of this technique for vascular use was stressed, it may also be adapted to other interventional MR-guided approaches, such as in interstitial methods for tracking a biopsy needle (28).

The problems faced by the multiple-FOV MR fluoroscopy method are the limitations on the minimum FOV sufficient for unwrapping, and that eliminates intensity variations. Fortunately, there is a lot of room for improvement in the type of pulse sequence used. Simply eliminating features not used by catheter tracking in the FGRE pulse sequence, or transferring our pulse sequence modifications to a more efficient sequence, such as interleaved echo planar imaging (EPI), will permit a three- to four-fold increase in FOV on catheter and guidewire channels, without a penalty in temporal resolution, or a three- to four-fold increase in temporal resolution of the roadmap channel(s), without compromising spatial resolution.

CONCLUSIONS

A new data acquisition and reconstruction technique for guiding MR catheter placement has been implemented. Multiple-FOV MR fluoroscopy allows for FOV variations within a single image stream, as well as across multiple receivers. Using multiple MR receiver channels allows visualization of the catheter with high frame rates and a narrow FOV (25 frames/s, 1/32 of acquired), as well as of the surrounding anatomy with fairly frequent updates and large FOV (0.78–1.56 frames/s, full or half FOV). Combined with high-resolution interventional MRI, this technique may aid in the diagnosis of vessel disease, and in monitoring of intravascular procedures.

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