

MULTIPLE FOV MR FLUOROSCOPY

by

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A thesis submitted to The Johns Hopkins University in conformity with the requirements
for the degree of Master's of Science in Engineering.

Baltimore, Maryland

September, 2000

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Abstract

In this work, a novel real-time imaging and visualization technique for magnetic resonance imaging that allows multiple FOV imaging on multiple receiver channels is presented. It was implemented online to provide real-time visualization of the placement of guidewires and catheters on a dynamic roadmap, during interventional procedures.

The loopless catheter antenna is an electrically active intravascular probe used for MR signal reception. It provides projection images where 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. Excitation with an MR pulse is followed by identical data acquisition over four available receiver channels. Our strategy is to acquire very narrow FOV images on the MR receiver channels that collect guidewire or catheter data to provide a very high frame rate; and to form large FOV images from receiver channels that collect anatomical data from standard imaging surface coils to provide a dynamic, frequently updated roadmap.

The fast gradient-echo pulse sequence was modified to acquire k-space data in a semi-bit-reversed order, such that reconstruction of half, quarter, or 2^{-n} FOV images is possible during acquisition. Small sets of k-space lines are acquired linearly, and interleaved with subsequent sets. Raw data are automatically transferred to a workstation directly connected to the MR scanner after acquisition of each such set. Each channel is then reconstructed independently at different fields-of-view, and frame rates, and later combined for display. A graphical user interface was written to allow control over reconstruction parameters, such as the spatial and temporal resolution, and whether the narrow FOV image needs to be unwrapped. The user interface also provides control over

display parameters, such as viewing a subset of the channels, color-coding, negation, as well as independent contrast and brightness controls.

This new technique provides reliable and robust MR guidance with very high frame rates up to 25 frames/second for a catheter or guidewire channel with very narrow FOV imaging, while roadmap channels are frequently updated once enough data for a large FOV image arrives, usually 1-2 frames/second.

Thesis committee members were Ergin Atalar, Andrew Derbyshire, and Elliot McVeigh.

Acknowledgements

I would like to acknowledge the many individuals who made my graduate school experience at Johns Hopkins University a challenging, rewarding, and pleasurable one.

In particular, I would like to express my gratitude to my thesis advisor Dr. Ergin Atalar for his generosity in providing the opportunity and resources for pursuing my research interest; and providing guidance with a delicate balance between independence and support, by knowing when to push further, and when to compliment. Thanks for getting me started on a great project, and seeing me through the end of it.

I am grateful to one of the finest researchers around, Dr. Andrew Derbyshire, my mentor, for his genuine interest in this work which made this project a reality; for enriching my thinking through his extensive knowledge, and many hours spent at the scanner; and for providing great answers to all the questions I could come up with. Thank you.

I would also like to thank Dr. Elliot McVeigh for the interest he has shown in the project, and for taking the time to read, and absorb the material to be presented.

Special thanks to Jean-Michel Serfaty, and Xiaoming Yang for making the animal experiments possible; Perry Karmarkar and Steven Tully for providing the antennae; Christopher Yeung, Ronald Ouwerkerk, Meiyappan Soleiyappan, Harald Quick for their time and insight, as well as the friendly support. It has been a pleasure working with all of you.

Finally, I want to thank my family; my parents Musa, Gülgün, my brother Erkin, and my love, for always supporting me in everything that I choose to do.

Funding for this research was provided by the National Institutes of Health, thanks to the people who believe science is important, and are willing to pay for it.

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List of Abbreviations

MR: Magnetic Resonance

MRF: MR Fluoroscopy

IVMRI: Interventional MR Imaging

IVUS: Intravascular ultrasound

2D: Two Dimensional

FOV: Field of View

SNR: Signal to Noise Ratio

TR: Repetition Time

TE: Echo Time

FGRE: Fast Gradient Echo

SPGR: Spoiled Gradient Echo

TG: Transmit Gain

FA: Flip Angle

Chapter 1

Introduction

1.1 MR in diagnosis of vascular disease

Intravascular magnetic resonance (MR) methods for imaging arterial walls with ultra high resolution for plaque characterization have been developed (Kantor 84, Hurst 92, Martin 92, Kandarpa 93, Atalar 96, Correia 97, Ocali 97, Quick 99). Intravascular MR compares favorably with the other clinical diagnostic methods of evaluating vascular disease, such as intravascular ultrasound (Martin 97) and angiography (Boos 96) in terms of superior contrast and resolution of the vessels and the vessel wall. The intravascular approach may also overcome the spatial resolution limitations of conventional MRI using external coils for which the upper bound of signal-to-noise ratio is limited (Ocali 98).

Ex vivo experiments have demonstrated the superiority of high resolution MR to intravascular ultrasound (IVUS) in terms of plaque characterization, due to the soft tissue contrast inherent to the MR experiment (Martin 97). IVUS helps determine lumenal dimensions and percent stenosis, and has real-time capability. However, several drawbacks remain. Poor contrast resolution and difficulty in positioning the device in the blood vessels limit evaluation of plaque composition (Waller 92). The imaging plane is limited to the aperture of the transducer, artifacts related to transducer angle relative to

the vessel wall occur, and it has variable resolution at different depths of view (Nissen 91).

The current clinical method of evaluating vascular disease with x-ray angiography also has several limitations. It provides detailed maps of the arterial lumen based on blood flow, where atherosclerotic disease is identified as a local narrowing. However, the severity of a stenosis can be grossly underestimated if the adjacent arterial segments are not disease free. In addition, arteries are known to enlarge in response to development of atherosclerotic plaque, and luminal narrowing may not occur until plaque comprises a large portion of the vessel diameter (Glagow 87). Early stages of disease may pass unidentified with x-ray angiography (Thomas 86). Additional concerns are limited tissue characterization, absence of multi-plane images, adverse reaction to contrast agents, and exposure to radiation.

Initial in vivo MR work based on external surface coils allowed imaging only of peripheral vessels such as the carotid, femoral, and popliteal arteries, due to the limited signal and depth penetration. Various intravascular coils have been designed to overcome these limitations. Studies on atherosclerotic animals showed that intravascular MR is feasible for detecting early atherosclerosis and characterizing more advanced plaque formations (Zimmerman 99). Due to its excellent soft tissue contrast MRI also provides means for characterizing plaque structure, such as differentiation between fibrous tissue and lipid deposits, and plaque classification (Correia 97).

By accurately quantifying plaque extent and characterizing plaque structure, intravascular high resolution MR can be an important addition to the concept of MRI in diagnosis and treatment of vascular disease.

1.2 Interventional Device Visualization in MR

For the success of interventional MR procedures, accurate visualization of the interventional device is important for its safe, correct placement. Currently the percutaneous placement of such intravascular MR devices is done 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. An MR Fluoroscopy technique to visualize interventional procedures with high spatial and temporal resolution is necessary.

Many innovative techniques have been developed to make MR guidance of interventional procedures possible. Two main approaches have been proposed: electrically passive localization, and electrically active localization.

Electrically passive methods are based on the physical characteristics of the catheter that are distinguishable on MR images. Visibility can be achieved through the magnetic susceptibility artifact, the signal void due to water displacement, or incorporation of contrast agents into the device (Kochli 94, Lenz 96, Bakker 96). The advantages of this approach are that no additional hardware is necessary, and visualization of the entire device is possible. However, the artifact allowing the visualization of the device usually depends on its orientation in the scanner, pulse sequence and imaging parameters. These methods have low temporal resolution, poor contrast requiring thin sections to avoid partial volume effects, with the added difficulty of tracking the plane of the catheter without active position information.

Electrically active methods, on the other hand, employ extra hardware for interventional device visualization. One approach is to incorporate a thin copper wire-loop into the catheter, through which a small current is passed (Glowinski 97). This current induces a local magnetic field resulting in an inhomogeneity in the main magnetic field. Localization of the catheter is achieved by this artifact, whose size can be varied with the amount of current passed through the wire.

The second active approach is to incorporate a small RF coil into the tip of the interventional device (Ackermann 86, Dumoulin 93). This RF coil detects the echo generated by a nonselective RF pulse followed by a readout gradient, which reveals its location. The three dimensional position of the catheter tip is encoded with four excitations. Thus, the coils position can be computed with a very high temporal resolution of up to 20 times per second, and displayed on a previously acquired roadmap image, allowing real-time device tracking. The drawback is that, the number of catheter points that can simultaneously be tracked, is limited by the number of receiver channels available on the scanner. Most advanced MR scanners are equipped with four to seven receiver channels, allowing four to seven points on the catheter to be tracked, which may not be sufficient for steering guidewires in the complex vasculature, or selectively targeting small vessels.

The third approach for active device tracking allows the visualization of the entire length of the catheter, regardless of its orientation. When an electrically coupled antenna is implanted on the catheter, the sensitive part of the catheter is seen as a bright line on MR images (McKinnon 96, Ocali 97). It provides high signal in a cylindrical region around the antenna, making not only the catheter, but also the vessel wall and

surrounding structures visible. The advantages of this active antenna design are numerous. MR signal originates solely from the catheter, and there are no obstacles to the catheter view. Signal is localized around the catheter, therefore, slice selection or adjustments to the imaging plane are not necessary (Atalar 98).

This latter type of active device visualization, specifically the loopless catheter antenna (Ocali 97) was chosen for the work presented in this study. The loopless antenna provides high signal in a cylindrical region along the entire length of the catheter, and can be used for imaging arterial walls with ultra-high resolution to evaluate vessel disease. However, the Multiple FOV MR Fluoroscopy technique can be applied to any interventional device that provides high signal on an MR image, such as solenoid, single/multiple loop antennae (Atalar 96, Ladd 97), or gadolinium coated catheters (Unal 98), and a 2ⁿ fold increase in frame rate of device visualization is possible.

1.3 Overview of Multiple FOV MR Fluoroscopy

Having presented the potential of MRI in diagnosing vascular disease and the methods currently available for interventional device visualization, it is apparent that a reliable and robust visualization technique for MR guided interventional procedures is necessary. Our purpose in this project is to provide high frame rate images of catheters and guidewires on a dynamic anatomical roadmap image, and this section will provide a summary of the key concepts that make Multiple FOV MR Fluoroscopy possible. Atalar demonstrated the increase in temporal resolution of interventional device visualization through narrow FOV imaging (Atalar 98). For this project, a novel data acquisition strategy to continuously acquire such narrow FOV images and allow roadmap image

formation was developed. This data acquisition method was implemented online with independent reconstruction of multiple channel data, and a combined display utility.

1.3.1 Background

1.3.1.1 Loopless Antenna

The method of interest for interventional device tracking in this project is the electrically active approach using the loopless catheter antenna. The loopless antenna design consists of a coaxial cable with an extended inner conductor, essentially a dipole antenna at the tip of a coaxial cable. The unbalanced currents on the coaxial cable of the catheter make possible the visualization of the whole length and structures within its diameter of sensitivity. Its sensitivity decays approximately inversely with radial distance from the catheter center axis. Both the signal and noise power detected by the antenna are very high, which allows placing the tuning and matching circuitry outside the body. Moreover, the electromagnetic properties of the antenna are virtually independent of its size, unlike coils that incorporate a loop. In short, the catheter antenna, which is essentially a wire, can be made very thin and flexible to allow insertion into very small tortuous vessels, while providing ultra-high resolution MR images with an extremely high sensitivity and SNR in its immediate vicinity (Ocali 97).

1.3.1.2 Narrow FOV Projection Imaging

On a projection image, one acquired without slice selection, the entire catheter and surrounding structures within its diameter of sensitivity appear bright. The image is essentially a white line on a dark background (Figure 1.1). The simplicity of a catheter projection image is the motivation for using a very narrow FOV acquisition in order to decrease imaging time, thus, increasing the temporal resolution of interventional device

tracking, as demonstrated by Atalar (Atalar 98). The accompanying decrease in SNR is tolerable due to the high sensitivity of the catheter probes. The image FOV is determined by the k-space sampling density. If only every 4th line of k-space were to be sampled, the FOV in the phase-encoding direction would decrease to 1/4th the full FOV, and temporal resolution would quadruple. Spatial resolution, which is determined by the size of the k-space region sampled, is also not compromised with this approach.



Figure 1.1: Projection image of a loopless catheter antenna placed in water. Imaging parameters are SPGR, 32 cm FOV, TR/TE = 5.2/1.5, 256x256 data acquisition matrix, 1 degree FA, 0 dB TG, no slice selection. Temporal resolution 1.3 s (5.2 ms x 256)

If the catheter is not perfectly straight, or if the frequency encoding direction is not parallel to the catheter axis, aliasing in the phase-encoding direction results in image overlap (Figure 1.2). Wrapping increases as the catheter aligns more with the phase-encoding direction. Due to the simple image geometry, unwrapping of these aliased images is possible with a simple row-scan algorithm, and the original catheter shape can be recovered. For successful unwrapping of an aliased catheter image, the view of one part of the catheter cannot overlap with the view of another part of the catheter. The FOV of each section of an image must be large enough to contain the catheter without overlap of catheter parts.

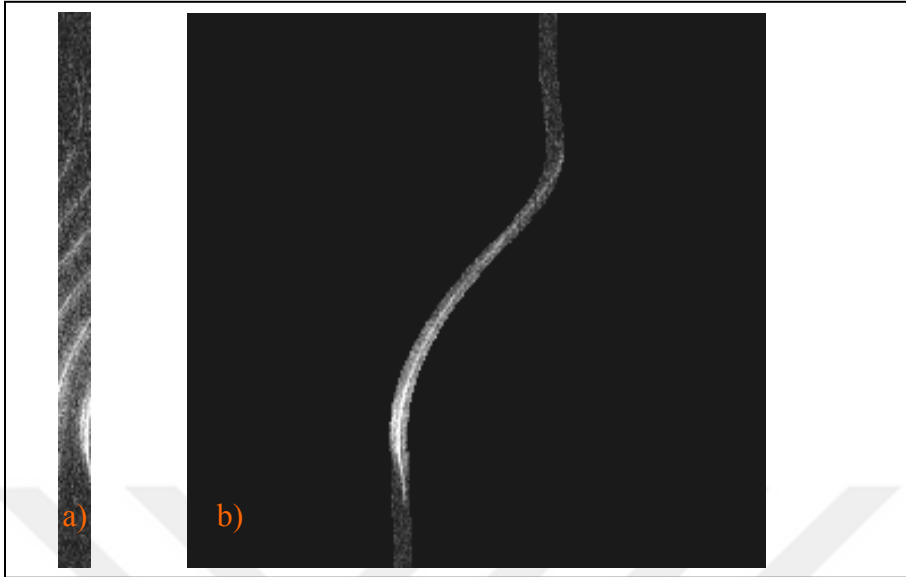


Figure 1.2: Narrow FOV image of a loopless catheter antenna. FGRE, 32cm FOV, TR/TE:5.2/1.5ms, 256x16 data acquisition matrix, 3 ° FA, 0dB TG, no slice selection.

Temporal resolution 81ms (5.2 ms $\times$ 16)

(a) Acquired narrow FOV image

(b) Image recovered by unwrapping (a)

1.3.2 Data Acquisition for Multiple FOV Imaging

An innovative data acquisition strategy was developed to simultaneously accomplish narrow FOV imaging of interventional devices and roadmap formation. The aim is to collect k-space data with the flexibility of reconstructing incoming data at the smallest FOV that contains the desired image, thus achieving the highest temporal resolution possible. K-space data is acquired such that reconstruction of half, quarter, or 2^{-n} FOV images is possible during acquisition. Views are ordered in a semi-bit-reversed fashion, rather than a purely linear acquisition on a Cartesian grid. Small sets of sparsely, but equally spaced m k-space lines are acquired linearly. Each such set produces a very narrow FOV image with a very high temporal resolution. The combination of

consecutive sets by interleaving the data allows increasing the FOV as necessary to avoid unrecoverable aliasing.

Note that the visualization technique described is not limited to the loopless catheter antenna. It can potentially be applied to any interventional device that provides high signal on a low background, such as solenoid or single/multiple loop antennas, gadolinium filled catheters, as long as unwrapping of the aliased images is possible. The phase-ordering scheme will be discussed in detail in Chapter 3.

1.3.3 Image Reconstruction In Multiple FOV MR Fluoroscopy

1.3.3.1 Multiple FOV's within One Image

The flexibility of our view ordering allows the formation of multiple FOVs within one image. If one part of the catheter aligns with the phase-encoding direction, or forms a loop, the frame rate of the entire movie does not need to be decreased to eliminate ambiguity in that part of the catheter image. Different parts of an object can be imaged with different frame rates, by varying the FOV in the phase-encoding direction as a function of position in the frequency encoding direction. By providing different frame rates at different locations, this approach maximizes the temporal resolution of each section of an image. Multiple FOVs within an image will be further discussed in Section 4.1.

1.3.3.2 Multiple FOV Imaging on Multiple Receiver Channels

Current device tracking methods employ two different approaches of providing an anatomical roadmap image for guidance. The first and more common approach is to plot the calculated catheter tip position, or superimpose dynamic catheter images onto a previously acquired still roadmap image. This roadmap is not necessarily the true

representation of the anatomy due to motion, and additional roadmaps may need to be acquired throughout the procedure. The second approach, mostly used with passive techniques, acquires the anatomical information and the interventional device as a single image. It is desirable to view the catheter with a high temporal resolution and the background anatomy with a high spatial resolution, which are conflicting demands with this approach. Due to the large amount of data that needs to be collected a compromise needs to be made between spatial and temporal resolution.

Most advanced MR scanners are equipped with at least four separate receiver channels. With the semi-bit-reversed phase order, besides viewing the catheter or guidewire at very high frame rates, we can also obtain a dynamic anatomical roadmap image using these four available receiver channels. In this case, excitation with the MRF pulse results in identical data acquisition over all channels. Two or three of the receiver channels can be dedicated to collecting anatomical data to provide a dynamic roadmap image, while the other channels can be connected to an active guidewire and/or an active catheter. The flexible view ordering allows reconstruction of images with different FOVs on different channels, where channels with a narrow FOV have a high frame rate, and channels with a large FOV have a slower frame rate. This permits reconstruction of guidewire or catheter images with a very high frame rate, independent of the roadmap images that are reconstructed with a large enough FOV to contain the patient.

Chapter 2

Background

MR images are reconstructed from k-space data, which is essentially the frequency spectrum of the image. Our strategy in the multiple FOV MR Fluoroscopy technique is to increase the frame rate of interventional device visualization by acquiring narrow FOV images. This corresponds to collecting fewer samples of the k-space data, thus reducing the imaging time, as demonstrated by Atalar (Atalar 98).

Several methods have been developed that acquire only part of the desired k-space in order to reduce scanning time. To obtain successful results, the acquisition strategy and the imaged object must be well matched. In the multiple-FOV MR Fluoroscopy method, high temporal resolution images of the catheter placement are obtained on a dynamic roadmap without compromising the spatial resolution of the catheter or roadmap images. K-space is sampled very sparsely, resulting in aliased catheter images, which are unwrapped to recover the original catheter shape. Multiple FOV MR Fluoroscopy can provide high frame rate visualization of any interventional device that provides high signal with a narrow sensitivity profile, such as single/multiple loop, or solenoid designs, gadolinium filled catheters, as long as the aliased catheter image can be unwrapped.

For MR signal reception, the loopless catheter antenna was used in this study. This chapter will provide the loopless catheter antenna design characteristics that allow for such narrow FOV imaging, specifically the narrow sensitivity profile, as well as an overview of other suitable interventional device designs. The narrow FOV approach used by Atalar for decreasing imaging time will be described, and a demonstration of the unwrapping algorithm will be presented.

2.1 Basic Principles of MR Imaging

In the context of MR imaging, the body can be visualized as a collection of tiny spinning charged spheres with small magnetic moments. In the presence of the main magnetic field of the scanner, these spins align with the direction of B_0 (longitudinal direction), and resonate at the Larmor frequency, ω_0 . $\omega_0 = \gamma B_0$, where γ is the gyromagnetic ratio, a known constant unique for each atom.

When the spins are excited with an RF pulse (B_1) tuned to their resonant frequency, they are tipped away from their equilibrium state, and are no longer aligned with B_0 . Upon turning the excitation off, relaxation back to equilibrium occurs, where the length of the magnetization vector does not remain constant over time. The behavior of the magnetization vector M is described by the Bloch equation:

$$\frac{d\vec{M}}{dt} = \vec{M} \times \gamma B_0 \hat{k} - (M_x \hat{i} + M_y \hat{j}) / T_2 - (M_z - M_0) \hat{k} / T_1$$

, where $\hat{i}$, $\hat{j}$, and $\hat{k}$ are unit vectors in the x, y, and z directions, M_0 is the equilibrium magnetization arising from the main field B_0 , and B includes the various magnetic fields applied to the sample. T_1 and T_2 are the time constants characterizing the return of the magnetization vector along longitudinal and transverse planes, respectively.

From Faraday's law of induction, the rotating magnetization vectors acting as RF sources, induce an electromotive force (emf) in an RF receiver coil. The induced emf is recorded as a time signal, called the Free Induction Decay (FID). Spatial localization is achieved by applying linear gradient fields in addition to B_0 , which leave the combined magnetic fields still pointing in the B_0 direction, but cause the field strength to change with x or y location. $G_x = dB_z/dx$ and $G_y = dB_z/dy$. If gradient fields are applied, then the frequency of precession is $\omega(x,y) = \gamma(B_0 + G_x x + G_y y) = \omega_0 + \gamma G_x x + \gamma G_y y$. Due to the applied gradient fields, the FID contains spatially dependent information about the amplitude distribution $m(x,y)$ of the excited spins that we wish to image, and a series of FID's are used to reconstruct the MR image.

The baseband signal equation can be derived as follows: The rotating magnetization vectors at each point in the excited region, let's say a plane, have some amplitude $m(x,y)$ and phase $\phi(x,y,t)$ depending on the applied magnetic fields. Assuming the receiver coil is uniformly sensitive over the plane, the total received signal is the sum of signals from each point in the plane:

$$s(t) = \int_x \int_y m(x,y) e^{-i\phi(x,y,t)} dx dy$$

By definition, frequency is the time rate of change in phase. We also know that frequency is proportional to B from the Larmor equation. Then:

$$d\phi(x,y,t)/dt = \omega(x,y,t) = \gamma B(x,y,t),$$

where $B(x,y,t) = B_0 + G_x(t)x + G_y(t)y$. G_x and G_y are the gradient fields we applied for the spatial localization of the MR signal from the excited region, and B_0 is the main magnetic field. Assuming the spins have an initial phase of zero, this leads to:

$$\phi(x,y,t) = \gamma \int_0^t B(x,y,\tau) d\tau$$

$$\phi(x,y,t) = \gamma \int_0^t B_0 d\tau + \gamma \left(\int_0^t G_x(\tau) d\tau \right) x + \gamma \left(\int_0^t G_y(\tau) d\tau \right) y$$

Demodulating to get rid of the carrier frequency term $\exp(-i\omega_0 t)$, the signal equation becomes:

$$s(t) = \int_x \int_y m(x,y) e^{-i[\gamma \left(\int_0^t G_x(\tau) d\tau \right) x + \gamma \left(\int_0^t G_y(\tau) d\tau \right) y]} dx dy$$

The 2D Fourier transform of $m(x,y)$ is defined as:

$$M(k_x, k_y) = \int_x \int_y m(x,y) e^{-i[2\pi (k_x x + k_y y)]} dx dy$$

We can see that:

$$s(t) = M(k_x(t), k_y(t)),$$

where $k_x(t) = \gamma/2\pi \left(\int_0^t G_x(\tau) d\tau \right)$, and $k_y = \gamma/2\pi \left(\int_0^t G_y(\tau) d\tau \right)$

This states that, at any given point in time t , $s(t)$ is the 2DFT of $m(x,y)$ at some spatial frequency $k_x(t)$, $k_y(t)$. The total recorded signal $s(t)$ maps directly to a trajectory through spatial frequency space (Fourier domain) as determined by the time integral of the applied gradient waveforms $G_x(t)$ and $G_y(t)$. This frequency domain in MR is called the “k-space”, and k_x and k_y are the spatial frequency variables.

(This section has been adapted from Nishimura 96)

2.2 Image FOV, Spatial Resolution and Temporal Resolution

Proper image formation depends on the appropriate coverage of the k-space by the trajectories given by $s(t) = M(k_x(t), k_y(t))$. The continuous time signal $s(t)$ is sampled along k_x , and k_y directions as shown in Figure 2.1a for 2DFT imaging. This rectangular coverage on a Cartesian grid allows convenient image reconstruction with a 2DFT. Δk_x and Δk_y are the sampling periods, while $k_{x\max}$ and $k_{y\max}$ are the highest spatial

frequencies sampled. Samples along k_x are collected during a single readout, while samples along k_y accumulate by repeating the excitation for different phase-encode steps. Thus, k_y and y directions will be referred to as the phase-encoding direction, while k_x and x will be referred to as the readout direction in the discussions that follow.

For real-time catheter tracking, we are interested in the temporal resolution, the spatial resolution, and the FOV (field-of-view) of the reconstructed image. The temporal resolution in 2DFT imaging is determined by the number of phase-encode lines that are acquired and the length of time between consecutive excitations (TR, repetition time).

$$\Rightarrow \text{temporal resolution} = 1 / \text{frame rate} = \text{TR} * N_{\text{phase-encode}}$$

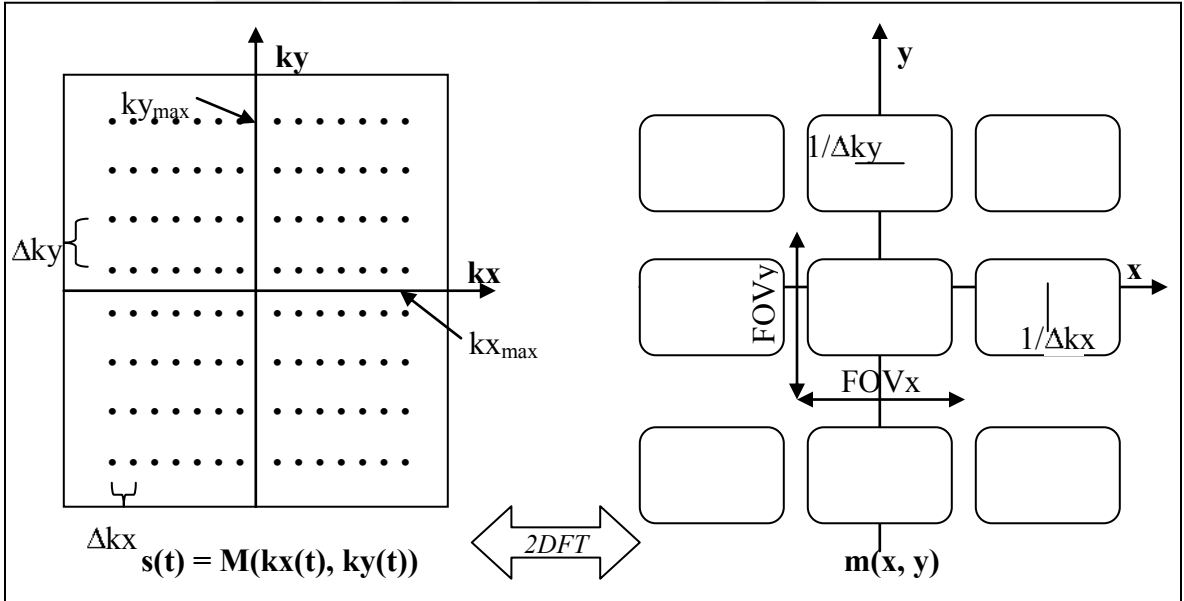


Figure 2.1: 2DFT imaging (a) k-space sampling (b) corresponding object and its replicas

The FOV of the resulting image depends on the sampling periods Δk_x and Δk_y . In MR, sampling, which is equivalent to multiplication by a 2D comb function, takes place in the spatial frequency domain. In the object domain, this corresponds to a convolution with a 2D comb function, which results in its replication at intervals of $1/\Delta k_x$

in the x direction, and $1/\Delta k_y$ in the y direction as shown in Figure 2.1b. The separation of these replicas determines the effective FOV in the image.

$$\Rightarrow \text{FOV}_x = 1/\Delta k_x, \quad \text{and} \quad \text{FOV}_y = 1/\Delta k_y$$

The spatial resolution of an image depends on the widths of the k-space coverage.

It is defined as (FOV/ number of sample points), thus, spatial resolutions δx , δy are:

$$\Rightarrow \delta x = \text{FOV}_x / N_{\text{readout}} = 1 / (\Delta k_x N_{\text{readout}}) \approx 1 / (2k_{x\text{max}}), \text{ for large } N (N > 100)$$

$$\Rightarrow \delta y = \text{FOV}_y / N_{\text{phaseencode}} = 1 / (\Delta k_y N_{\text{phaseencode}}) \approx 1 / (2k_{y\text{max}}), \text{ for large } N$$

For real-time catheter tracking, both a very high temporal resolution and spatial resolution are required. We know that:

$$\Rightarrow \text{temporal resolution} = \text{TR} \cdot N_{\text{phaseencode}}$$

$$\Rightarrow \text{spatial resolution} = \text{FOV}_y / N_{\text{phaseencode}}$$

In order to obtain high temporal resolution images, without sacrificing spatial resolution, one must cover a large area of the k-space with a small number of phase-encode steps. In other words, with this sparse k-space sampling, a smaller $N_{\text{phaseencode}}$ will accelerate the frame rate, while a smaller FOV will preserve high spatial resolution in the images. These changes correspond to narrow FOV imaging, which will be discussed in the next section.

2.3 Loopless Catheter Antenna Design and Sensitivity Profile

As mentioned earlier, the loopless catheter antenna consists of a dipole antenna at the tip of a coaxial cable (Figures 2.2 and 2.3). The unbalanced currents on the coaxial cable of the catheter make possible the visualization of the whole length and structures within its diameter of sensitivity. Figure 2.4 shows two projection images obtained using

the loopless antenna. It provides high signal in a cylindrical region around the catheter. The fast signal drop off from the catheter center allows very narrow FOV imaging without undesirable contributions from structures far from the catheter.

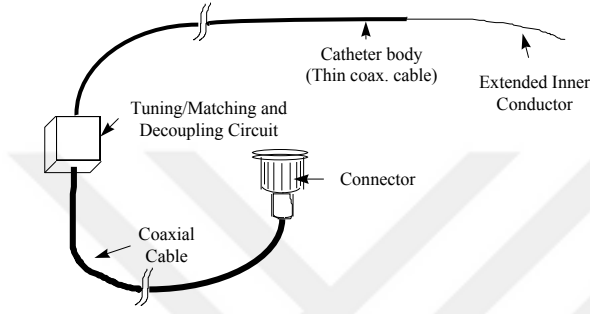


Figure 2.2: Schematic diagram of the loopless catheter antenna design

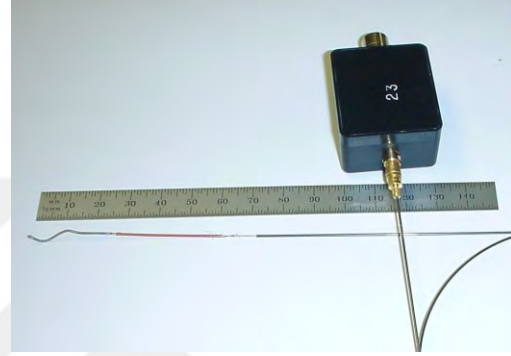


Figure 2.3: Photograph of the loopless antenna, and tuning and matching box

This section provides the theoretical basis and the experimental justification of the narrow sensitivity profile of the loopless catheter antenna.

The signal voltage in an MR experiment is given as:

$$V_s = \omega \mu H M, \quad (1)$$

where V_s is the signal voltage, M is the transverse magnetization of the voxel, H is the magnetic field generated by the coil in the transverse plane for unit input current. H is the only coil dependent parameter effecting signal voltage.

The root-mean-square (RMS) noise voltage is given by:

$$V_n = (4 k_b T R f_e)^{1/2}, \quad (2)$$

where k_b is the Boltzmann constant, T is the sample temperature, f_e is the effective pixel bandwidth. R is the real part of the input impedance seen from the input terminals of the coil. $f_e = 2 BW / N_x N_y NEX$, where N_x , N_y are the number of readout points and phase-

encoding steps, NEX is the number of image repetitions, and BW is the receiver bandwidth. The only coil dependent parameter that effects the noise voltage is R.

Combining (1) and (2):

$$\text{SNR} = V_s/V_n \Rightarrow \text{SNR} \propto H / (R)^{1/2} \quad (3)$$

In the catheter coil design (Atalar 96, Section 2.2), field cancellation occurs due to the interaction of the two opposing current branches of the loop, especially in close proximity. This reduces H, the transverse magnetic field, thus the signal-to-noise ratio. The loopless antenna design emerged from the desire to maximize $\text{SNR} \propto H/(R)^{1/2}$. A substantial increase in H is achieved when the conductors are separated. The loop design then evolves into a dipole. In this configuration, charges oscillate between the two tips of the antenna on the incomplete current path. R also increases due to the in stray electric field losses (Ocali 97).

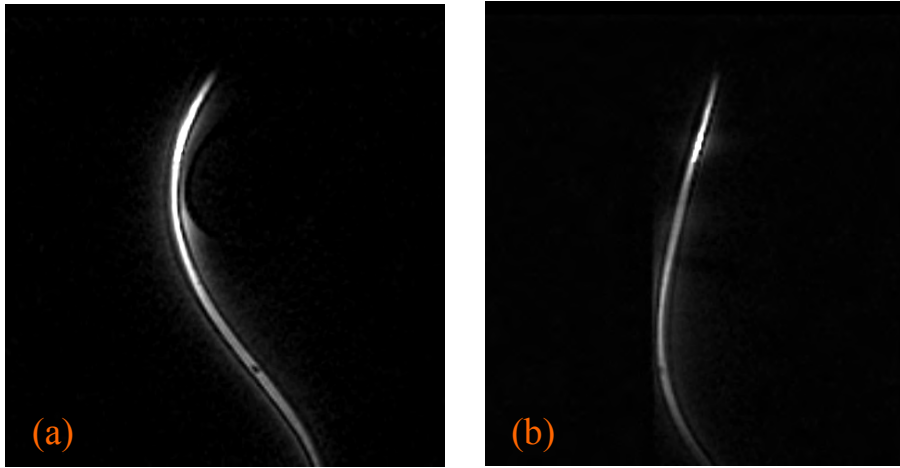


Figure 2.4: Projection images of the loopless catheter antenna, a) coronal, b) sagittal. Imaging parameters are SPGR, 32 cm FOV, TR/TE = 5.2/1.5, 256x256 data acquisition matrix, 1 degree flip angle, 0 dB transmit gain, no slice selection.

The transverse component of the H field is maximized when the catheter antenna is aligned parallel to the main magnetic field. This orientation provides the sharpest signal drop-off, which is desirable for very narrow FOV imaging without aliasing from objects distant to the catheter. The sensitivity profile of the loopless catheter antenna is shown in Figure 2.5. The sensitivity profile was obtained by imaging the antenna in a container of water with the following parameters: FGRE, 256x256 acquisition matrix, TR/TE:5.2/1.6ms, no slice selection, 62.5kHz receiver bandwidth. The profile was obtained from a perpendicular cross section of the catheter in an orientation parallel to the main magnetic field, and demonstrates the fast signal drop off from the catheter center.

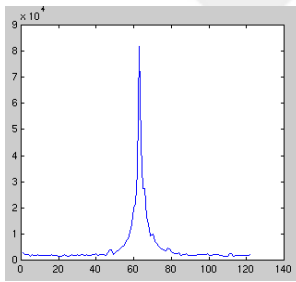


Figure 2.5: Sensitivity profile of the loopless catheter antenna. Signal intensity versus position (mm) plot shows sharp drop off in signal at locations away from the catheter center axis.

The loopless catheter antenna can be used to receive MR signal as a conventional surface coil, or both transmit and receive MR signal. There are two motivations for using the catheter probe as an RF pulse transmitter (Atalar 98). The first reason is that the phase of the sensitivity of the loopless antenna varies circumferentially. If an external coil is used for RF pulse transmission, and the loopless antenna is used for signal reception, the phase mismatch between the excited spins and the sensitivity results in signal cancellation. This can be avoided by transmitting with the catheter. Secondly, when the catheter is used in the transmit/receive mode, the amount of signal contribution from objects far from the catheter can be controlled by adjusting the power of the

transmit pulses. As the power increases, objects far from the catheter experience a larger flip angle, producing more MR signal. By adjusting the RF power level, the diameter of sensitivity can be very accurately adjusted. Figure 2.6 shows a comparison between the loopless antenna in the transmit/receive mode versus in the receive-only mode.

Ideally, for the multiple FOV method, the loopless antenna should be used in the transmit/receive mode. In this mode of operation signal from the antenna is higher, and contributions from objects far from the catheter can be minimized, providing the optimal conditions for reduced FOV imaging. Currently, the transmit/receive mode can only be used to view the catheter alone on a single channel of the MR scanner. When multiple channel data is required, the receive-only mode has to be used, due to the difficulty of defining the desired flip angle distribution when surface coils and/or the body coil, as well as the catheter are transmitting RF power simultaneously. In the receive-only mode, the sensitivity profile of the loopless catheter antenna is wider, and undesirable contributions from objects far from the catheter image can corrupt the narrow FOV catheter image. This issue will be further addressed in Chapter 7, Discussion.

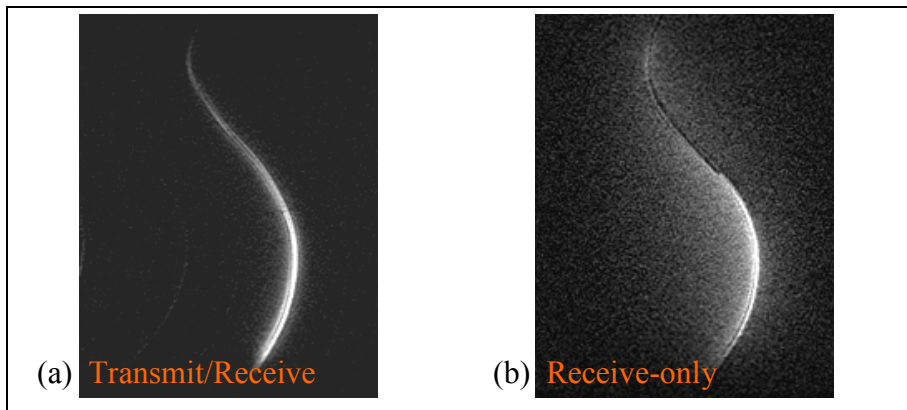


Figure 2.6: Transmit/receive versus Receive-only modes of the catheter antenna. Imaging parameters: FGRE, 32 cm FOV, TR/TE = 5.2/1.5ms, 256x256 data acquisition matrix, no slice selection, 15° FA, and 0dB TG in (a), 9° FA and 50dB TG in (b).

2.3.1 Other compatible antenna designs

Multiple FOV MR Fluoroscopy increases the frame rate of interventional device visualization by obtaining narrow FOV images of the interventional device to decrease the imaging time. These aliased narrow FOV images are later unwrapped, and displayed. The success of this technique depends on being able to obtain narrow FOV images of the interventional device that have not been corrupted by aliasing of structures far from the device. The loopless catheter antenna was the design of choice for this study, due to its high signal concentrated in a narrow region around the catheter. However, several other interventional device designs provide catheter images that are compatible with the Multiple FOV MR Fluoroscopy method, such as solenoid, single/multiple loop antennae (Atalar 96, Ladd 97), or gadolinium-coated catheters (Unal 98). As alternatives to the loopless catheter antenna, one active and one passive method of interventional device visualization will be briefly discussed in this section.

The electrically active catheter coil design (Atalar 96) is a flexible, long and narrow receiver coil placed on the tip of a catheter. The coil is composed of two parallel conductors that are shorted at one end, and the other end is used as the terminal of the coil. This elongated coil design allows multi-slice high resolution imaging of small vessels. Similar to the loopless antenna design (Ocali 97), the entire length of the catheter is visible on a projection image. The signal received by the catheter coil is very high in the vicinity of the conductors and drops as the distance from the conductors increases. The area of interest on an MR image is limited to a few centimeters around the catheter, beyond which signal levels are very low. Thus, the catheter coil can also be used with the multiple-FOV method to increase the frame rate of its visualization.

Another example of an interventional device that can benefit from the multiple FOV approach for real-time visualization is the gadolinium-filled catheter (Unal 98). This electrically passive technique exploits the T_1 shortening effect of the paramagnetic contrast agent, Gadolinium DTPA, for catheter visualization. Commercially available catheters are filled with 4% to 6% Gd-DTPA solution, making the entire length of the catheter visible on an MR image. The background signal is usually suppressed by using a projection dephaser, and the entire length of the catheter appears as a bright line on a dark background, making it suitable for the multiple FOV approach for an increased temporal resolution.

Other single/multiple loop, or solenoid antenna designs (Ladd 97) which provide similar images where the catheter appears as a thin bright line on an MR image, can also be used with the multiple FOV technique to increase the frame rate of their visualization. The temporal resolution can be increased by a factor m in each case, where m is the reduction in FOV where unwrapping of an aliased catheter image is still possible.

2.4 Narrow FOV Imaging

MR images are reconstructed from k-space data, which is essentially the frequency spectrum of the image by a two dimensional Fourier transformation. The extent of k-space sampling determines the spatial resolution of the resulting image (k_{max}), while the density of sampling determines the FOV of the resulting image (Δk). Very narrow FOV imaging corresponds to a scarce sampling of the k-space (Figure 2.7), where fewer phase-encode lines of the k-space are acquired. Narrow FOV imaging can be used to decrease imaging time, thus increase temporal resolution as demonstrated by Atalar (Atalar 98). The relationship between the image FOV and the frame rate is

demonstrated in Figure 2.8. For instance, if 256 k-space lines are to be acquired for an image with a 32 cm FOV, the temporal resolution will be $256 \times TR$, and the spatial resolution will be $32\text{cm}/256$. If only every other line of the k-space is sampled, the FOV will only be 16cm, however, the temporal resolution will double to $128 \times TR$. Note that the spatial resolution of the resulting image, which stays at $32\text{cm}/256$, is not compromised with this approach.

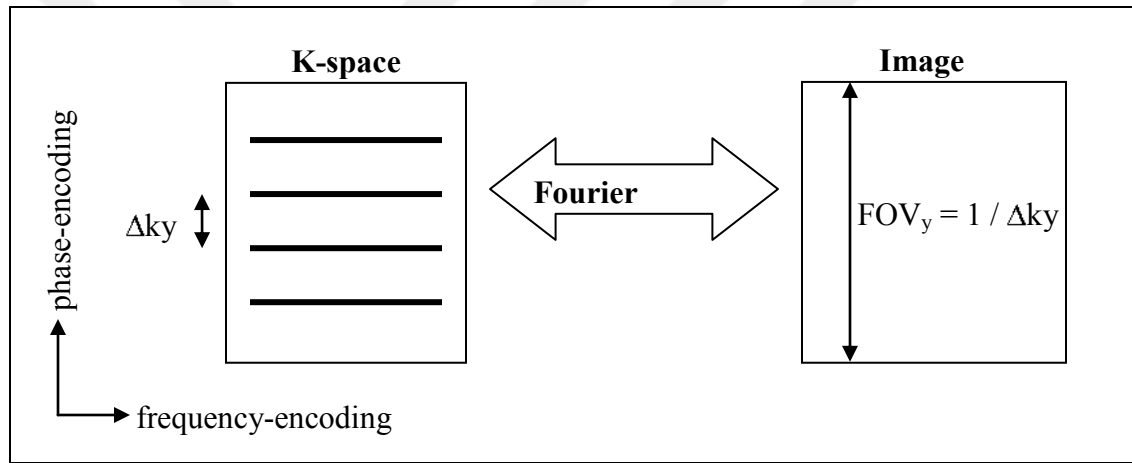


Figure 2.7: Relationship between k-space sampling density and FOV.

The sampling theorem states that, the sampling rate must be at least twice the maximum frequency in the signal. Otherwise the resulting image will show aliasing and image overlap.

With full FOV imaging, the Nyquist criterion is met. The spatial frequency space is densely sampled, and no image overlap results. When the Nyquist criterion is not met, due to narrow FOV imaging with few phase encode lines, replicas of the original image are placed closely and overlap, and aliasing occurs (Figure 2.8). The catheter is represented by the yellow line, and aliasing increases with decreasing FOV.

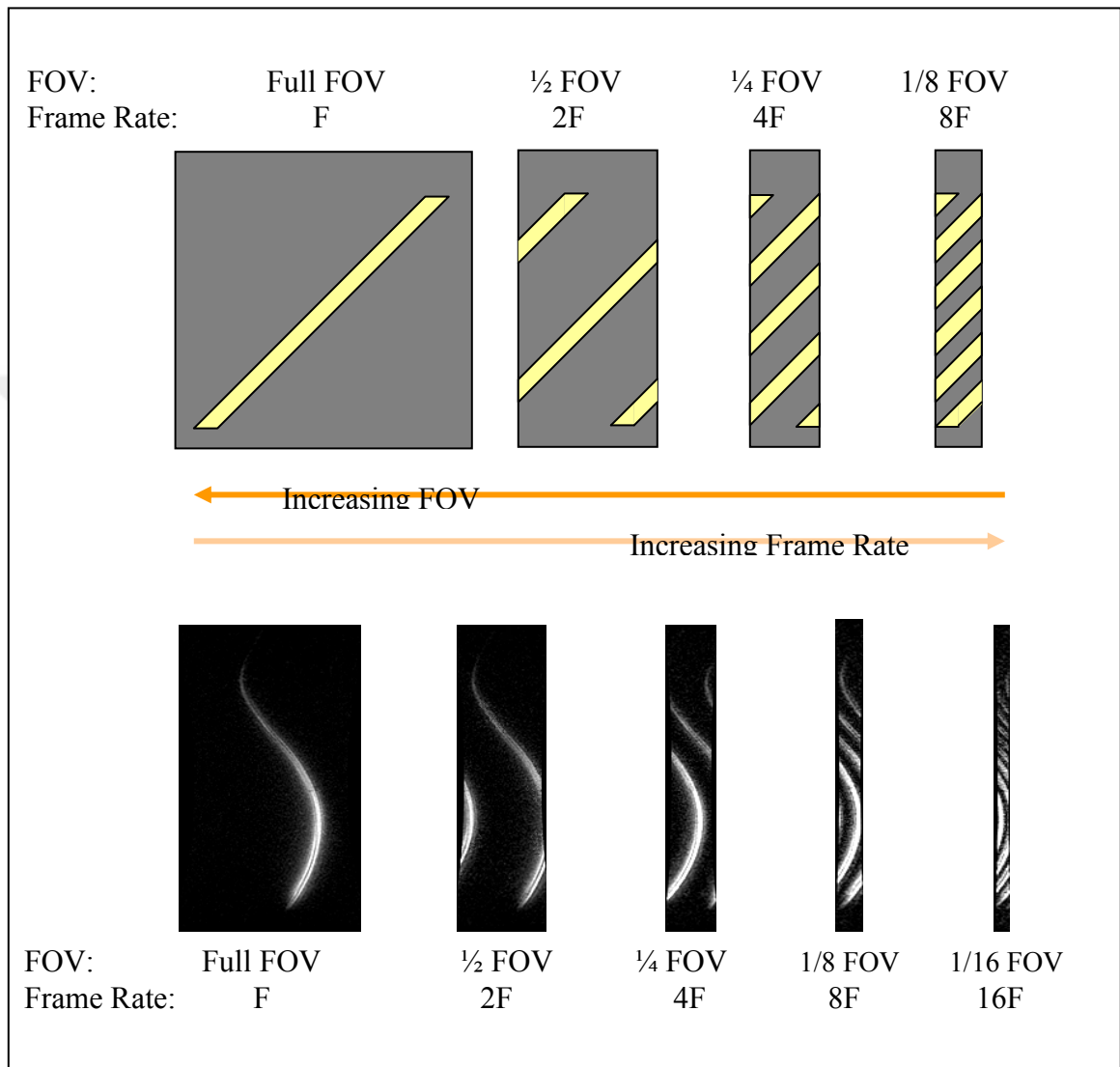


Figure 2.8: FOV versus speed

Although additional encoding would be necessary for complex structures such as the brain or the heart, this aliased image is appropriate for catheter visualization due to the simple image geometry. For any given image line in the phase-encode direction, the catheter has a very narrow profile, and therefore the Nyquist criterion can be satisfied with very sparse sampling. The narrow FOV images in Figure 2.8 FOV show aliasing, however, there is no overlap in the signal from the actual catheter. This image overlap in

the phase-encoding direction can be corrected using a standard unwrapping algorithm (i.e. Matlab, MathWorks, Inc), which will be discussed in Section 2.3.

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. Wrapping becomes more significant as the catheter bends away from the frequency encoding direction, towards the phase encoding direction. If the catheter and surrounding structures are viewed in λ pixels, the maximum amount that the catheter can bend away from the frequency encoding direction is $\cos^{-1}(\lambda/\text{FOV}_y)$, where FOV_y denotes the FOV in the phase-encoding direction (Atalar 98). Derivation of this limit is graphically depicted in Figure 2.9.

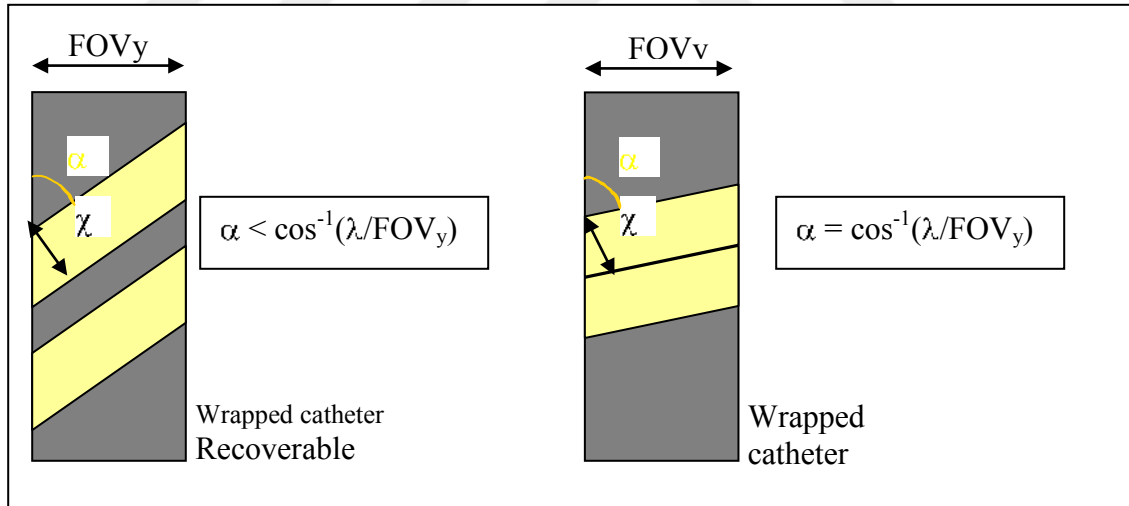


Figure 2.9: Limiting condition for successful unwrapping

2.5 Unwrapping Narrow FOV Images

The aliased narrow FOV images contain all of the catheter information, however, are wrapped in the phase-encoding direction (Figure 2.10a). Although wrapping seems to be a problem, it can be corrected with a simple unwrapping algorithm. The unwrapping algorithm extracts the geometric shape information of the catheter, and represents it as a set of skeletal points (Figure 2.10b). The aliased image is then displayed according to these unwrapped catheter positions (Figure 2.10c).

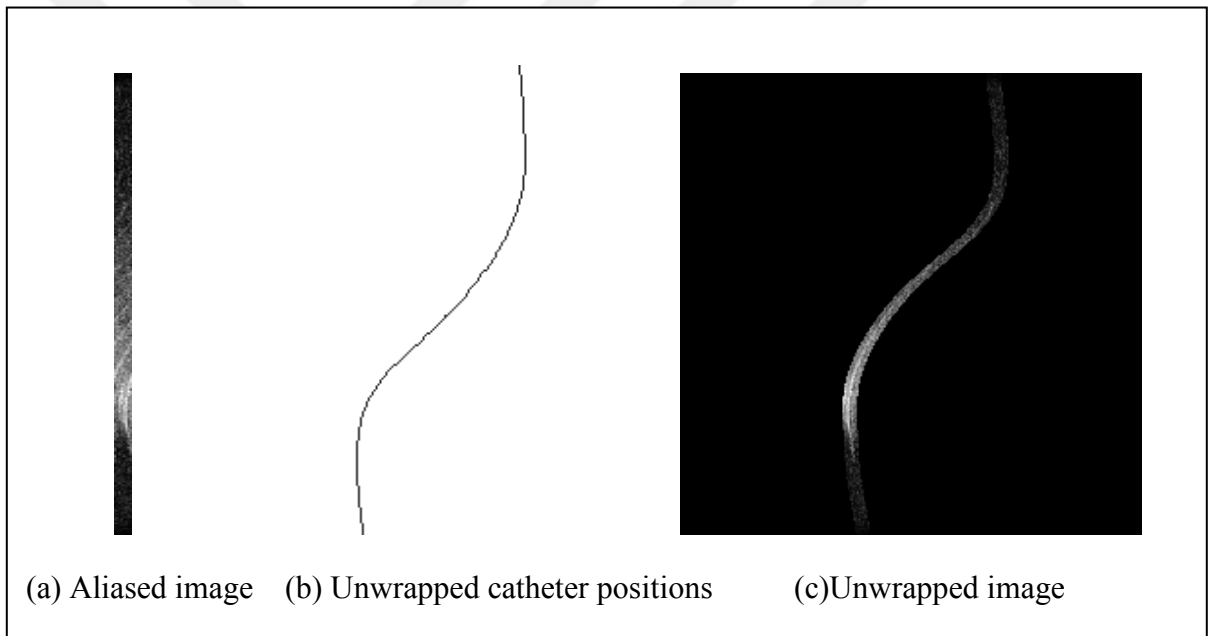


Figure 2.10: Unwrapping of an aliased catheter image. The aliased narrow FOV image in (a) is scanned row by row to extract a set of skeletal points that represent the actual catheter shape (b). The narrow FOV image is then displayed according to the catheter shape dictated by the list of skeletal points, resulting in the unwrapped catheter image shown in (c).

A simple row-scan algorithm is used for extracting the geometric information from the projection images of the catheter antenna. Each row of the image is systematically scanned to locate the position of the catheter, and if there is a substantial discontinuity between successive catheter points, a correction is made to the position

information. For each row of the image, the signal with a significant width (band) is identified as the catheter. Signal is differentiated from background noise by applying a threshold equal to 15% (default) of the maximum signal amplitude, which can be easily changed on the user interface if necessary. The mean position of the signal band is recorded as the catheter position for that row, and compared with the catheter position of the preceding row. If there is a discontinuity of more than half the narrow FOV the catheter is assumed to have wrapped, and so the position is shifted by one FOV in the appropriate direction. Once the entire image is scanned, the rough catheter position information is low pass filtered to eliminate any high frequency variations, errors caused by the background noise. Figure 2.11 shows the representative input and output of the unwrapping algorithm, and the pseudo code is given below:

Unwrapping algorithm:

```
/* y increases along frequency encoding direction */
/* x increases along phase encoding direction */
/* signal(x,y): signal intensity at (x,y) in the aliased catheter image */
/* threshold: signal intensity level to distinguish noise from catheter signal */
/* x_n: number of points in row (along phase-encoding direction) where the catheter signal exceeds threshold */
/* x_sum: sum of the positions in a row where the catheter signal exceeds threshold */
/* catheter_position_temp: approximate position of catheter within narrow FOV image */
/* unwrap_n: multiple of FOV's that the catheter position needs to be corrected for */
/* catheter_position: array containing approximate position of catheter in a full FOV image */
```

```
for (each row y of the image){
    x_n = 0;
    for (each pixel x on the row){
        x_sum = 0;
        if (signal(x,y) > threshold){
            x_sum += x;
```

```

        x_n++;
    }
}
catheter_position_temp = x_sum / x_n;
unwrap_n = round((catheter_x(previous) - catheter_x_temp)/FOV);
catheter_position(current_row) = catheter_position_temp
                                + FOV * unwrap_n;
}
catheter_x = low_pass_filter(catheter_x);

```

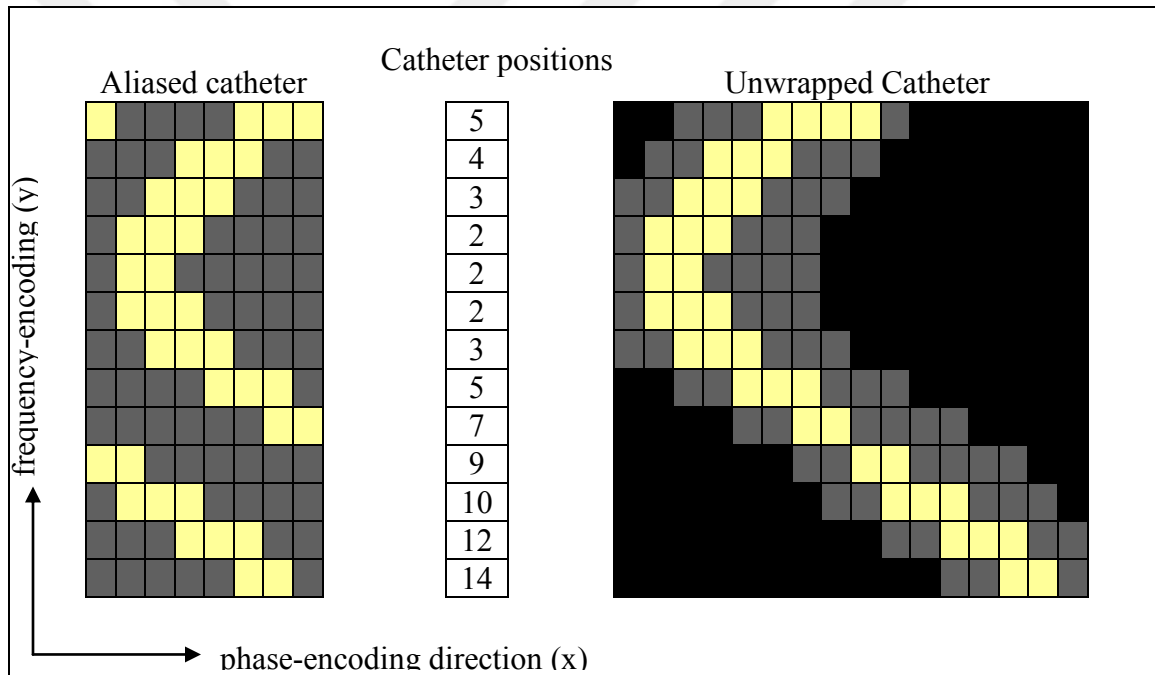


Figure 2.11: Implementation of the unwrapping algorithm. The aliased catheter image is scanned to find the catheter positions in each row of the frequency encoding direction. These catheter positions are corrected by unwrapping, if a considerable discontinuity is encountered in successive catheter positions. Then, a narrow FOV around these catheter positions is displayed. (The catheter points that exceed threshold are shown in yellow, while the acquired FOV is shown in gray indicating background noise or catheter signal below threshold. Black squares indicate zero points)

The narrow FOV image is then displayed according to these unwrapped catheter positions such that the actual catheter itself will always appear as a bright band in the center of the displayed FOV (Figure 2.10c). The process of unwrapping the narrow FOV image results in a non-rectangular FOV. This FOV itself will change shape to track the location that the catheter is in. As an analogy, this catheter-tracking FOV can be thought of as a spotlight following the performer (our catheter) as she moves across the stage. The actual catheter shape is not jeopardized by the failure of the unwrapping algorithm, but merely “poorly illuminated”.

The row-scan algorithm described is fast and simple, but requires the SNR to be high enough such that noise and signal can be discriminated by thresholding.

2.6 Stabilizing the Unwrapping Algorithm

The fast and simple approach chosen for unwrapping narrow FOV images highly depends on the quality of the received image. As the FOV is reduced to increase temporal resolution, SNR in the image decreases with decreasing acquisition time (Section 2.2.3). Furthermore, if the sensitivity of the catheter does not decay very rapidly, the catheter picks up signal from objects far away from the catheter. The last factor that contributes to the degradation of image quality is cross-talk between multiple receivers, that is, the data collected on the catheter channel is actually the catheter data additively corrupted by the data from other receiver channels with a complex coupling coefficient. The negative effect of all these factors increases when the FOV is narrowed.

The minimum FOV required for successful unwrapping of an aliased catheter image, depends on the apparent width of the interventional device, its orientation with respect to the phase encoding direction (Section 2.3), and these factors affecting the

image quality (namely, the SNR, sensitivity profile, and coupling). Even if the image quality is not very good, and the unwrapping fails for very narrow FOV images, stabilization is possible due to the nature of our phase encoding.

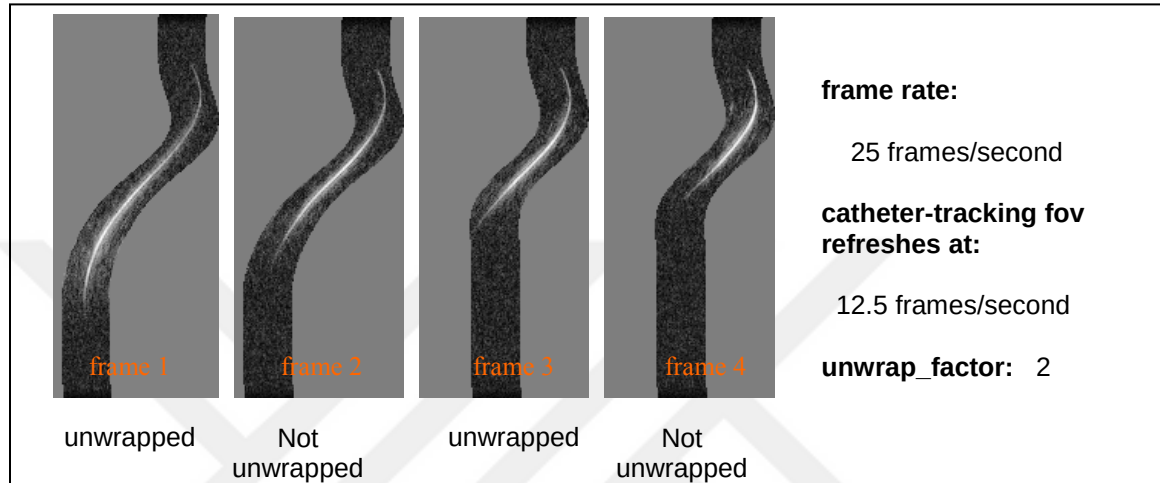


Figure 2.12: Demonstration of using an unwrap_factor of "2"

Although only small sets of k-space data are used to form high frame rate catheter images, the accumulation of any power of two such sets allows forming a larger FOV image. If unwrapping fails to find the exact catheter shape from a narrow FOV image, a larger FOV image can be reconstructed in the background, and the catheter shape can be extracted from this less aliased image with higher SNR. For instance, if it not possible to successfully unwrap $1/16^{\text{th}}$ FOV images, $1/8^{\text{th}}$ FOV images can be unwrapped in the background. This implies that the frame rate of the catheter image will still be that of a $1/16^{\text{th}}$ FOV image, however, the shape of the narrow FOV displayed around the catheter will change twice as slowly. This factor by which the refresh rate of the image, and the surrounding FOV differ by is called unwrap_factor. Figure 2.12 shows a $1/32^{\text{th}}$ FOV catheter image, unwrapped using information from $1/16^{\text{th}}$ FOV images. The catheter is viewed with a frame rate of 25 frames/second, but the shape of the FOV around the catheter changes only 12.5 times per second. The frames that were unwrapped with the

correct information appear in the center of the non-rectangular FOV (frames 1 & 3), and the frames that use the position information from previous frames are not exactly in the center, but still within the non-rectangular FOV of the previous image (frames 2&4).

Using an image with ($\text{unwrap_factor} \times \text{desired FOV}$) for determining the catheter shape, greatly improves the stability of the unwrapping algorithm. The assumption of this approach is that within unwrap_factor frames, the catheter should still be somewhere within the FOV displayed by the first frame. And again if this assumption fails, the catheter will not be incorrectly placed, but “poorly illuminated” in some parts. In addition, the narrow FOV to be displayed can be shifted by a few pixels by setting the center-offset option on the user interface (that was written for the online implementation of Multiple FOV MR Fluoroscopy), to achieve the optimal visualization.

2.7 Resolving the Catheter Position Ambiguity

Another issue that needs to be addressed with narrow FOV images is the ambiguity in the catheter position (Figure 2.13). The unwrapping algorithm provides us with the correct catheter shape, however, there are m distinct locations that this catheter can be placed when the FOV is $1/m$ th the full FOV. The unwrapping algorithm cannot determine the absolute position of the catheter along the phase-encoding direction.

The catheter image may be shifted by $(N_y \cdot m)$ pixels, where N_y is the number of phase-encoding steps, and m is an integer. The value of m can be determined from a full FOV image, and all subsequent images can be registered using the position information from previous time frames.

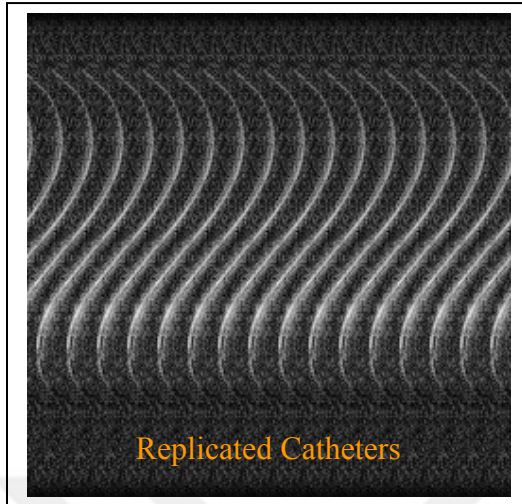


Figure 2.13: Catheter position ambiguity in narrow FOV images. This image is the result of sampling every 16th line of k-space. There are $m=16$ distinct locations that the actual catheter might be at.

Although only small sets of k-space data are used to reconstruct high frame rate catheter images, the accumulation of such small sets allows forming a full FOV image, where there is no ambiguity in the catheter position. In the first few seconds within the beginning of data acquisition, unwrapped catheter images are displayed on the center of the display. Once enough data for a full FOV catheter image is available, this image is reconstructed in the background, and its center of mass is calculated. This center of mass is compared to the center of mass of the narrow FOV image, revealing the value of m . All subsequent catheter positions are registered according to this correct catheter information and the center ambiguity is resolved. Chapter 5 provides a thorough explanation of these algorithms, within the implementation of the reconstruction stage of the project.

Chapter 3

Data acquisition

3.1 Semi-Bit-Reversed Phase Order

The aim of our data acquisition is to fill the k-space in a flexible way to allow reconstruction of multiple FOV images during acquisition. Very narrow FOV images will be used to provide real-time catheter tracking with a high temporal resolution, and large FOV images will 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. A semi-bit-reversed phase order was developed to accomplish the reconstruction of half, quarter, or 2^{-n} FOV images during acquisition.

Small sets of equally spaced k-space lines are acquired linearly. Each such set comprises sufficient data to reconstruct a narrow FOV image. A data set can also be interleaved with subsequent sets such that after acquisition of any power of 2 small sets, all k-space lines will be equally spaced with uniform sampling. The number of k-space lines in each small set depends on the minimum FOV required for successful unwrapping of a catheter image. The size m of each small set should be a divisor of the total number

of phase encode lines, such that the result is a power of 2, to allow multiple FOV imaging: $N_{pe} / m = 2^n$, $n = 0, 1 \dots \log_2(N_{pe}/m)$,

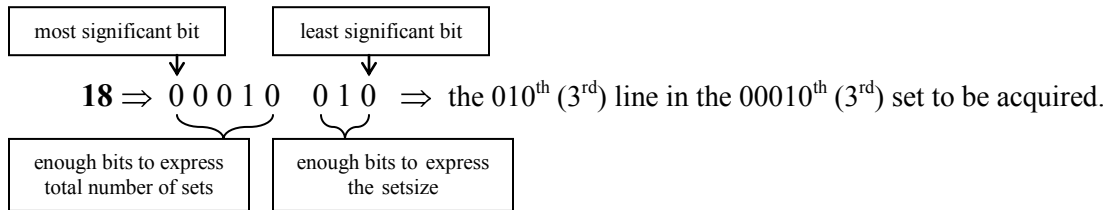
where N_{pe} is the total number of phase-encode lines, and m is the number of phase-encode lines in each set. After the acquisition of 2^n ($n = 0, 1 \dots \log_2(N_{pe}/m)$) small sets, all k-space lines will be equally spaced, and the combination can be used to form an image with 2^{-n} the full FOV and 2^n higher frame rate. For instance, for a 256 line k-space acquisition, we determined the smallest number of k-space lines necessary for successful unwrapping of a catheter image to be 8 lines. This results in a 32 fold reduction in FOV, and 32 fold increase in the frame rate of the catheter movie. Each such set of 8 lines is sufficient to produce one frame of a very narrow FOV catheter movie. Consecutive sets can also be joined to increase the FOV of the entire movie, or a section of the movie. The acquisition of 8 k-space lines per set also provides a conveniently sized “data packet” and data rate (interrupts/second) for data transfer/communication and processing, and frame rate, as will be further discussed in Chapter 5, Implementation.

Each small set of k-space lines is acquired linearly to minimize ghosting of the linearly moving catheter image. Consecutive sets are then interleaved to provide maximum equal spacing among the collected data over time. This leads to a semi-bit reversed phase-order, which will be explained with the example of 256 phase-encode lines. When k-space is acquired in sets of 8 lines, there are a total of 32 such sets per image. In the following figure, the *acquisition order* column represents the order of acquisition of the k-space lines in time, while the *phase order* column represents the actual k-space lines to be acquired in that order. The *linear* column shows the binary numbers progressing from 0 to setsize in a set, while the *bit-reversed* column shows the

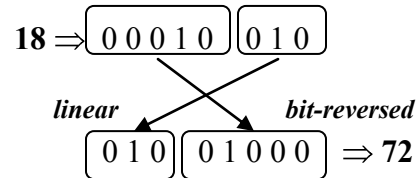
bit-reversed binary numbers progressing from 0 to the number-of-sets = (total phase-encodes/ setsize). The phase-order for the acquisition of any line can be found by combining these two columns as shown below.

<i>acquisition order</i>		<i>linear (within each set)</i>	<i>bit-reversed (among sets)</i>		<i>phase order</i>	
1 st line		000	00000		0	
2 nd line		001	00000		32	
3 rd line	zero	010	00000		64	
4 th line		011	00000		96	
5 th line		100	00000		128	
6 th line	to	101	00000		160	
7 th line		110	00000		192	
8 th line		111	00000		224	
9 th line		000	10000		16	
10 th line		001	10000		48	
11 th line	zero	010	10000		80	
12 th line		011	10000		112	
13 th line		100	10000		144	
14 th line	to	101	10000	zero	176	
15 th line		110	10000		208	
16 th line		111	10000		240	
17 th line		000	01000	to	8	
18 th line		001	01000		40	
...		...	...		...	
...		...	...		...	
...		...	...		...	
255 th line		110	11111		223	
256 th line		111	11111		255	

For instance, the 19th line to acquire can be found as follows:



Then the gradient strength should be:



$\Rightarrow$ The 19th line to be acquired is phase encode line 72.

Figure 3.1 depicts the order of acquisition for 256 k-space lines acquired in small sets of 8 (The first 8 sets, and the last set are shown). On the right side of the figure, k-space lines representing the acquired data, and possible larger groupings are shown. While each small set is acquired linearly to avoid ghosting of the linearly moving catheter image, consecutive sets are interleaved to provide maximum equal spacing among the collected data over time. Images reconstructed with different size groupings of this data are also shown in Figure 3.1. Each set can be reconstructed and displayed individually, to produce one frame of catheter or guidewire images with $1/32^{\text{nd}}$ the full FOV. In this case, the temporal resolution is $8 \cdot \text{TR}$. Two consecutive sets can be combined to create an image with $1/16^{\text{th}}$ the FOV, if 8 k-space lines are insufficient for successful unwrapping. Temporal resolution in this case is $16 \cdot \text{TR}$. Similarly, larger groupings can

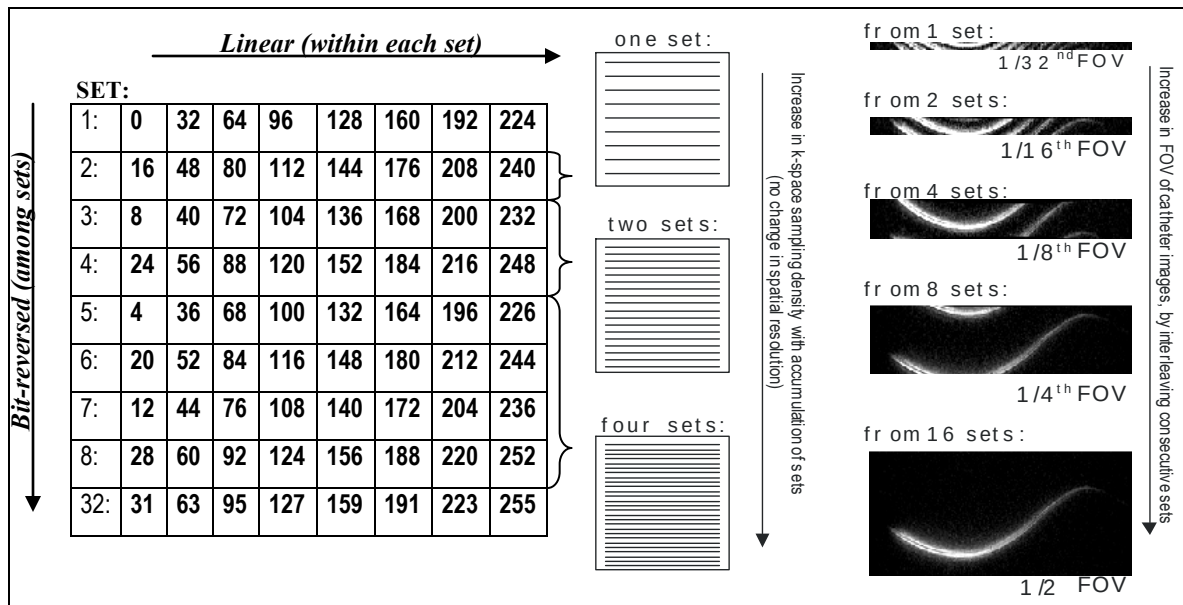


Figure 3.1: Semi-bit-reversed phase order for 256 lines of k-space, acquired in sets of 8; larger groupings of k-space lines; and the increase in FOV achieved by combining multiple such sets (Imaging parameters: FGRE, 256x256 data matrix, 7° FA, 50dB TG, 62.5kHz BW, TR/TE: 5.6/1.6msec, no slice selection)

be used to create images with a larger FOV at the expense of reduced temporal resolution.

To reconstruct the full FOV image without sacrificing spatial resolution, all of the k-space needs to be sampled. This means acquisition of all 32 sets of 8 k-space lines, resulting in a temporal resolution of $256 \cdot TR$ between frames. Although the temporal resolution of this large image is $256 \cdot TR$, large FOV images can be updated as each new set is acquired at intervals of $8 \cdot TR$ providing a smoother transition over time.

3.2 Pulse Sequence Modification

The fast-gradient-echo pulse sequence was modified to acquire data in the semi-bit-reversed phase order. A control variable “myorder_lin” activates our method of phase ordering when set to 1. The phase-ordering table, view_table, was recalculated with the following modification:

```
/* view_table: the phase-encoding table */
/* rspviews: the total number of phase encode lines to be acquired */
/* setsize: number of phase encode lines in each set to be acquired linearly, currently 8 */

if (myorder_lin){
    view_table[0] = 1;
    m = rspviews / setsize;    /* #of sets to acquire = offset between lines within a set */
    add = m;
    for (i=0; i<setsize; i++)
        view_table[i] = view_table[i-1] + add;
    while (add > 1){
        k = m / add;
        add = add / 2;
        for (j=0; j<k; j++)
            for (n=0; n<setsize; n++)
                view_table[(j+k)*setsize + n] = view_table[j*setsize + n] + add;
    }
}
```

The algorithm begins by finding the order of the first set of k-space lines to be acquired. All lines in the first set are separated by $rspviews/setsize = number\ of\ sets\ to\ acquire$. Using our example of 256 lines acquired in sets of 8, the density of k-space sampling is $add = 256/8 = 32$ at this point. When the second set is acquired, the distance between the lines will be half of this number ($add = add/2$). So, the second set of k-space lines is offset from the first set by 16 lines. When the third and fourth sets are acquired, the density of k-space lines doubles again, and all lines will again be separated by $add = add/2$ at this point. So the third and fourth sets are offset from the first two sets by 8 lines only. The algorithm continues until all lines are offset by 1, that is, the k-space has completely been filled.

3.3 Demonstration

This section experimentally demonstrates the increase in temporal resolution of an image by narrowing its FOV. The phantom consists 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 semi-bit-reversed phase order. The tip of the catheter is at the top of the images, and it exits the view at the bottom.

The modified fast gradient echo pulse sequence was used for the acquisition of 256 lines of k-space data (TR/TE : 5/1.6ms, FOV: 24cm, 256x256 partial NEX). The following images were reconstructed from different groupings of the same data (Figure 3.2). The full FOV image was reconstructed by combining all 32 sets of 8 k-space lines. It shows ghosting due to the motion of the catheter during the data acquisition time of 1.28 seconds, which corresponds to a frame rate of 0.78 frames/second. When the same data is split into even and odd lines of the k-space, two ½ FOV images are obtained. The

temporal resolution doubles, and the large jump of the catheter is visible in the unwrapped frames. The frame rate of these two frames is 1.56 frames/second (temporal resolution = 640ms). When every 4th line of k-space is used for reconstruction, we obtain four distinct frames with 1/4th the FOV 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 (temporal resolution 320ms, frame rate = 3.12 frames/second). The same trend continues in the cases with 1/8th and 1/16th the FOV. The very narrow non-rectangular FOV, very smoothly follows the catheter as it moves out of our view at 6.14 frames/second (temporal resolution = 160ms), and 12.5 frames/second (temporal resolution = 80ms), respectively.

In short, as we use fewer phase-encode lines for each frame and narrow the FOV, temporal resolution increases and the catheter motion becomes very smooth. The time it requires to construct a full FOV image is determined by the TR of the pulse sequence that is used, and N_{pe}, the total number of phase-encode lines to acquire. N_{pe} depends on the desired spatial resolution, and FOV for full FOV anatomical images (Section 2.3.2). From the multiple FOV MR Fluoroscopy point of view, this time is constant. Increase in the frame rate of catheter images is achieved by reconstructing narrow FOV images from fewer phase encode lines, where $\text{Frame Rate} \propto 1/\text{FOV}$. This relationship is depicted in Figure 3.3.

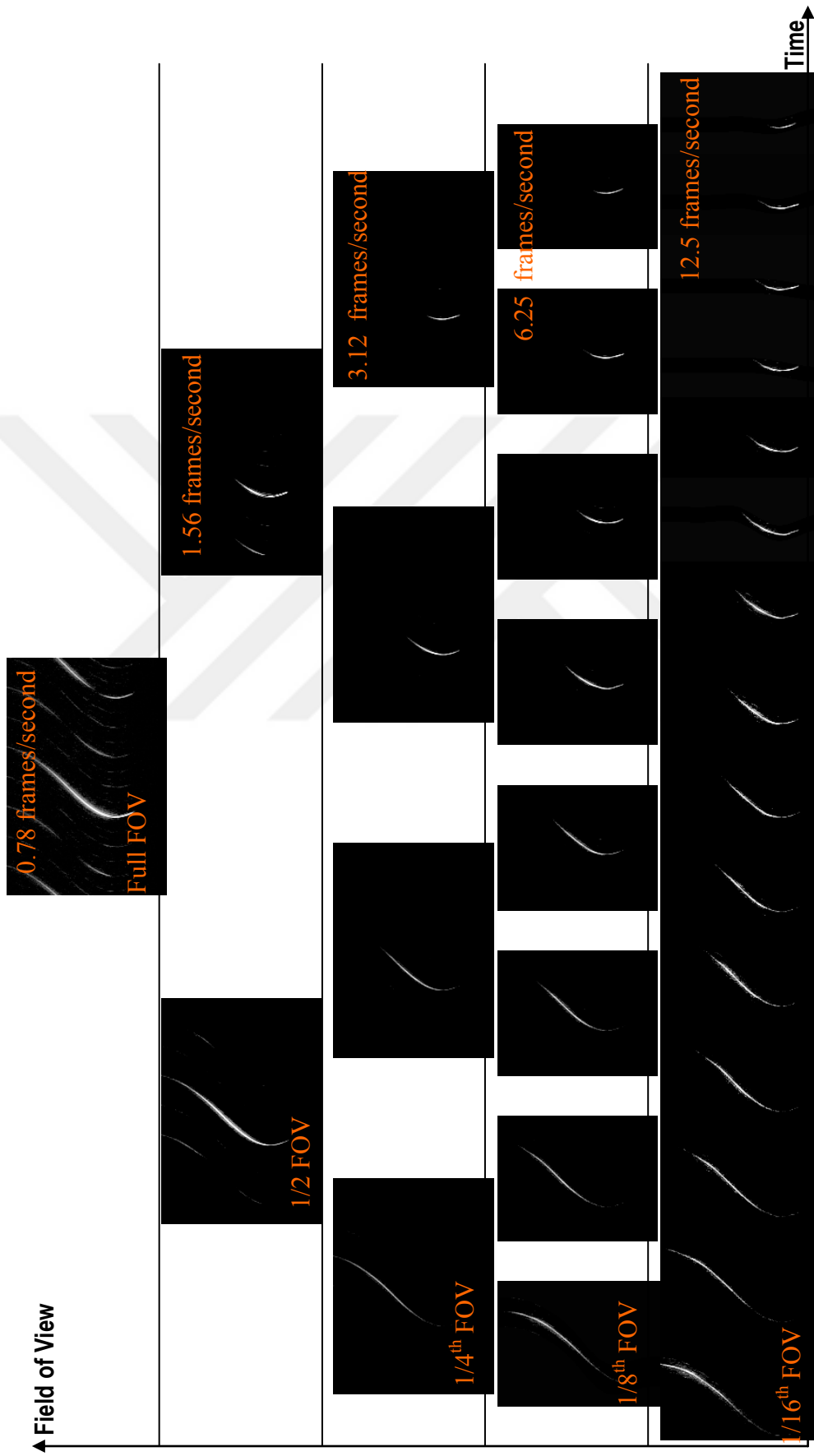


Figure 3.2: Demonstration of the reduction in FOV and corresponding increase in temporal resolution, with the semi-DIT-reversed phase order. The top row is a full FOV image of the catheter moving inside the phantom, acquired in 1.28 seconds (0.78 frames/second). The second row the same data was used to form two $\frac{1}{2}$ FOV images, with twice the frame rate. The same data is rearranged to form $\frac{1}{4}$ th, $\frac{1}{8}$ th and $\frac{1}{16}$ th 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/second.

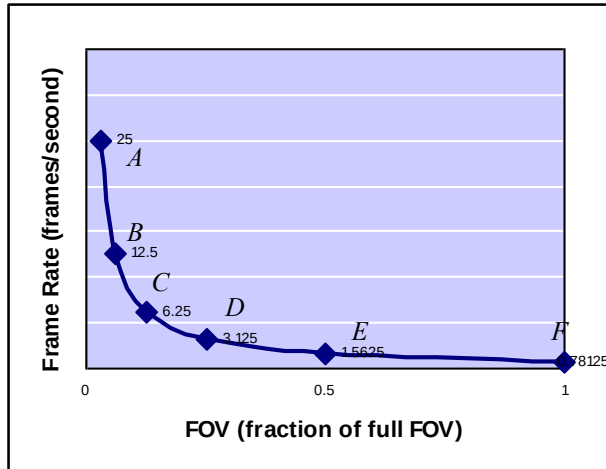


Figure 3.3: Frame Rate versus FOV

Figure 3.4 summarizes the relationship between the frame rate and FOV for the acquisition of a total of 256 phase encode lines at a TR of 5 msec. The full FOV is 24 cm, resulting in an in-plane spatial resolution of less than 1mm. Images with a smaller FOV can be reconstructed by using fewer phase-encode lines without compromising the spatial resolution. The factor of increase in frame rate by using the multiple FOV method is also shown in the last row of Figure 3.4.

	F	E	D	C	B	A
FOV	Full	$\frac{1}{2}$	$\frac{1}{4}$	$\frac{1}{8}$	$\frac{1}{16}$	$\frac{1}{32}$
# phase-encode lines used	256	128	64	32	16	8
Frame Rate (frames/sec)	0.78	1.56	3.12	6.25	12.5	25
Acceleration	x1	x2	x4	x8	x16	x32

Figure 3.4: Relationship between Frame Rate and FOV (Assuming imaging parameters: TR = 5msec, Npe = 256, FOV(full) = 24cm, Spatial resolution = 0.9375mm in plane)

By immediately reconstructing, unwrapping and displaying k-space data, we will be operating at point A on catheter and guidewire channels (Figure 3.3). This point is chosen to provide catheter images at a “clinically useful” frame rate, and when 8 k-space

lines are acquired per set at $TR = 5$ msec, corresponds to column A of Table 3.1. The frame rate on catheter and guidewire channels will be 25 frames/second. If this FOV is too narrow for unwrapping, two consecutive sets of 8 k-space lines can be combined, moving the operating point to B (Figure 3.3), column B (Table 3.1). The roadmap channels need to have a larger FOV, so the operating point F has a lower frame rate (Column F, Table 3.1). All k-space lines are used for image reconstruction, resulting in a frame rate of 0.78 frames/second. Note that, in all of the frames, the in plane spatial resolution is less than 1mm.

Higher frame rates than those shown in Table 3.1 can be achieved by optimizing the type of pulse sequence used. Currently, with the fast gradient echo (FGRE) sequence, the acquisition time of each k-space line is around 5 msec. However, TR 's in the range of 1-1.5 msec can be achieved by using a faster sequence such as the interleaved echo planar (EPI). This 3 to 4 fold decrease in acquisition time will permit 3-4 fold increase in frame rate of anatomical roadmap images, or 3-4 fold increase in FOV of catheter images, making unwrapping simpler. This issue will be further discussed in Chapter 7, Discussion.

Chapter 4

Reconstruction

4.1 Multiple Fields of View within one Image:

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, 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. The phase-encoding algorithm described in Chapter 3, allows imaging with different the frame rates for different parts of the object.

In 2DFT Imaging, signal readout occurs in the presence of a constant gradient field. Applying a gradient G in the readout direction when the FID is recorded, corresponds precisely to acquiring a projection in which the line integrals are along the phase-encode direction. The presence of G means that upon Fourier transforming the recorded FID signals, each frequency component corresponds to the contributions from the spins at some x location (along readout). At a particular x , all spins along y (along phase-encoding) oscillate at the same frequency, and sum to give the value at that frequency bin.

Normally, a 2D Fourier Transform of the k-space data is used to reconstruct the image. This is equivalent to applying a 1DFT in the readout direction, followed by another 1DFT in the phase-encoding direction. The first Fourier transform rearranges k-space data along the readout direction such that data becomes spatially resolved in that direction, and there is a one-to-one correspondence between distinct portions of the catheter image, and the 1D Fourier transformed data. At this point, we can either reconstruct a constant FOV catheter image with the data that is available, or we can reconstruct a variable FOV catheter image, by regionally increasing the FOV of certain catheter sections. On a variable FOV image, the FOV in the phase-encoding direction varies as a function of position in the frequency encoding direction.

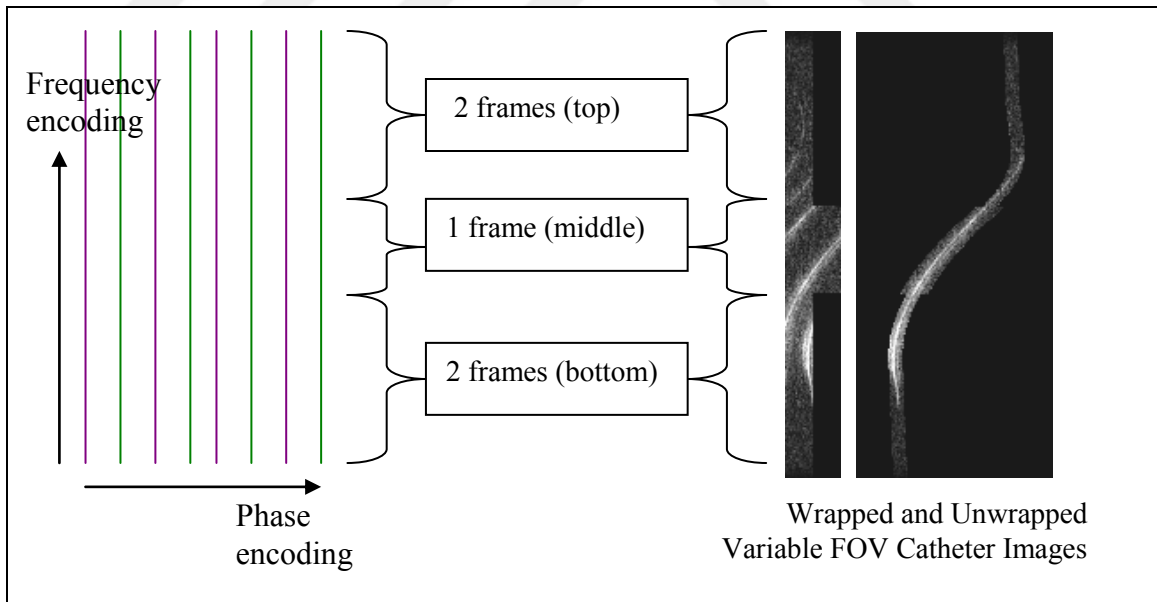


Figure 4.1: Demonstration of regionally increasing the FOV of a catheter image. Green and purple lines represent the k-space data that has been 1D Fourier transformed in the frequency encode direction. Purple set is the first set of data, and the green set is the second set. For the top and portions of the catheter image, each set is separately reconstructed. The middle portion is reconstructed from the combination of the two sets of data, thus has half the temporal resolution.

Figure 4.1 demonstrates this concept. Data is acquired in small sets of 8 k-space lines, and 1D Fourier transformed to allow mapping between k-space data and the desired image. 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 8 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 8 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 (Figure 4.1).

The flexibility of reconstructing different parts of an image with different FOVs is useful in cases where one part of the catheter requires a larger FOV than all other parts, which would normally slow down the entire movie. Results obtained from a phantom experiment demonstrating the utility of forming multiple FOVs within a single image are shown in Section 6.1.

There are some additional issues that need to be addressed, when reconstructing multiple FOV's within a single image: different SNR's in different portions of the image, and how to join sections with different FOV's.

When comparing images acquired with the same pulse sequence and identical spatial resolution, SNR is dictated by the total readout duration:

$$\text{SNR} \propto (\text{measurement time})^{1/2}$$

In a multiple FOV image, a different number of phase-encode lines are used for reconstructing different FOV portions of an image. The total measurement time is directly proportional to the number of phase-encode lines used to form the image. Then: $\text{SNR} \propto (\text{Npe})^{1/2}$, and different sections of the image not only have different frame rates, but also different SNR's. For instance in Figure 4.1, the middle portion of the image used 16 phase-encode lines, while the narrow FOV top and bottom portions were reconstructed with only 8 phase-encode lines. The total acquisition time was halved in order to double the frame rate. This leads to $\sqrt{2}$ decrease in SNR in the narrow FOV sections.

It is important to provide uniform signal across a multiple FOV image, despite the variation in SNR across its different sections. This is accomplished by scaling the image intensity of narrow FOV sections to match the signal intensity of the large FOV sections. This scaling factor depends on the ratio of the acquisition times of the sections, thus the number of phase encode lines used, and can be formulated in terms of the relative FOV's as follows:

$$\begin{aligned} \text{SNR} \propto (\text{Npe})^{1/2} & \Leftrightarrow \text{SNR} \propto (\text{FOV})^{1/2} \\ \Rightarrow \text{scaling factor} &= (\text{FOV}_{\text{wide}} / \text{FOV}_{\text{narrow}})^{1/2} \end{aligned}$$

Scaling the narrow FOV portions by this factor produces an image with uniform signal intensity, and non-uniform noise. Figure 4.2 demonstrates the effect of correctly scaling the signal intensity. The larger FOV central section of the image had higher signal than the narrow FOV portions, and visualization is greatly improved after scaling.

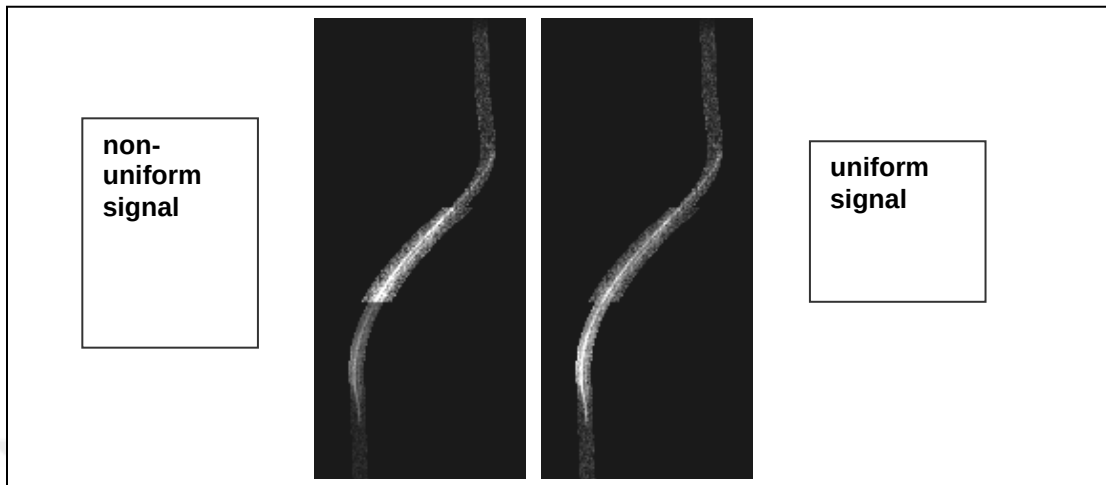


Figure 4.2: Image intensity correction on multiple FOV images

The second issue of how to join sections with different FOV's arises when the large FOV section of a variable FOV image contains a loop. The normal procedure for narrow FOV imaging is to reconstruct a narrow FOV image with few phase-encode lines, unwrap it (Section 2.4), find its center of mass from a full FOV image (Section 2.5), and display the catheter according to this corrected catheter position information. In this scenario, the catheter is always displayed in the center of the FOV (Figure 4.3a).

When a part of the catheter forms a loop, a shape unresolvable by unwrapping, the FOV of that section needs to be increased until the catheter image shows no aliasing, and the ambiguity is resolved. The sections of the catheter without a loop are treated like regular narrow FOV images. Each of the “loopless” sections are reconstructed and unwrapped independently, the correct catheter position is found from a full FOV image, and displayed at a high frame rate. The catheter in these sections is in the center of the displayed FOV. On the other hand, the section containing the loop is reconstructed with a large enough FOV to contain the loop, and slower frame rate. However, it is not clear

which part of the image this large FOV section should display. By definition, the loop cannot be unwrapped, so the relative catheter position information is lacking (Figure 4.3 b, c, and d). The narrow and wide FOV sections may be joined in several different ways. By default the large FOV is placed along a straight line joining the narrow FOV sections. The loop may or may not be at this position, and if it is not, the FOV must be shifted manually until the catheter is correctly displayed.

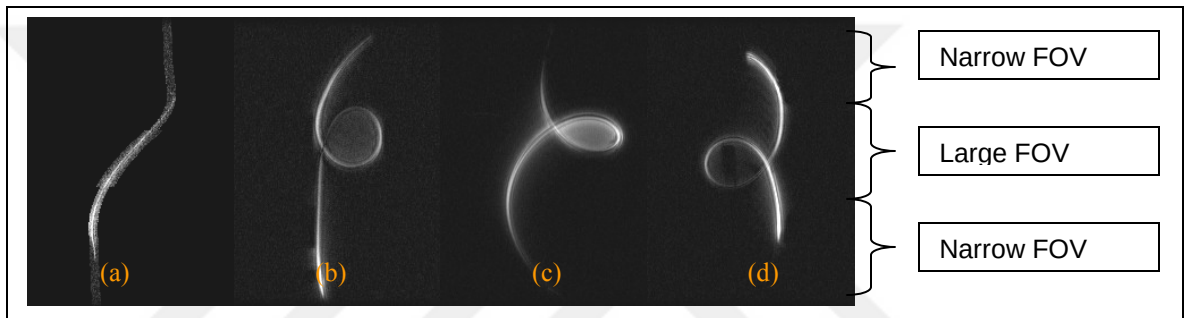


Figure 4.3: Joining sections to form variable FOV images.

(a): "Loopless" case, catheter is displayed in the center of the FOV found by unwrapping.

(b), (c), and (d): "Looped" cases: The loop may be slightly on the right (b), on the far right (c), on the left side (d), or anywhere else with respect to the rest of the catheter. Absolute catheter position information is lacking on the large FOV sections containing loops.

4.2 Multiple FOV Imaging on Multiple Receiver Channels

There are two common approaches employed by current device tracking methods of providing an anatomical roadmap image for guidance. The first and more common approach is to plot the calculated catheter tip position, or superimpose dynamic catheter images onto a previously acquired still roadmap image. Due to motion, this roadmap is not necessarily the true representation of the anatomy due to motion, and additional roadmaps may need to be acquired throughout the procedure. The second approach, mostly used with passive interventional device tracking techniques, acquires the anatomical information and the device as a single image. For proper guidance, it is

desirable to view the catheter with a high temporal resolution and the background anatomy with a high spatial resolution. These are conflicting demands when acquiring a single image for two very different information contents. A compromise needs to be made between spatial and temporal resolution, due to the large amount of data that needs to be collected.

The Multiple FOV MR Fluoroscopy method can provide a dynamic roadmap image, without compromising spatial resolution, or temporal resolution of interventional device images. Most 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. Besides viewing the catheter or guidewire at very high frame rates, we can also provide a dynamic roadmap image using these available scanner resources. In this case, excitation with the MRF pulse results in identical data acquisition over all channels. 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.

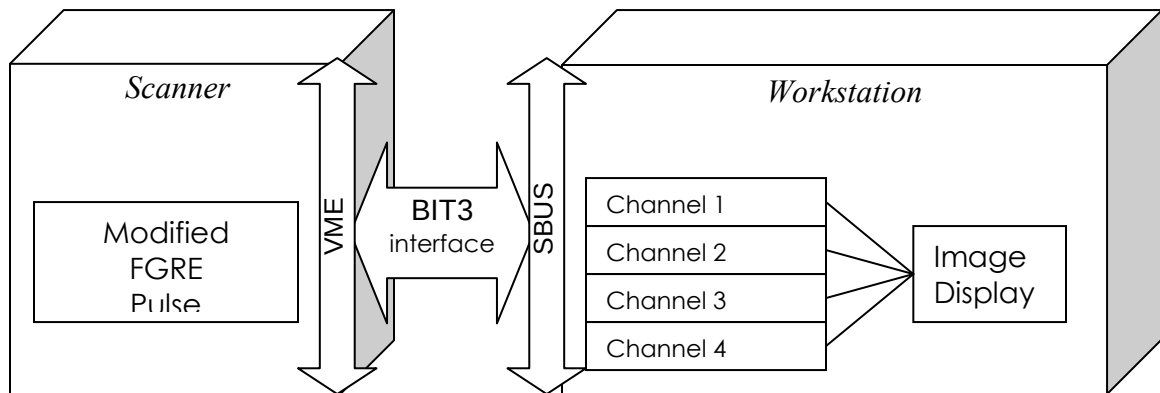


Figure 4.4: Data flow from scanner to workstation

In the Multiple FOV method, raw data is acquired with the semi-bit-reversed phase order (described in Chapter 3), and automatically transferred to a workstation directly connected to the MR scanner after acquisition of each small set of k-space lines. Each channel is then reconstructed independently at different FOVs, and later combined for display on the workstation (Figure 4.4).

This section will introduce the modifications that were made to gain access to data immediately after acquisition, and present an overview of the workstation software design. The complete implementation is presented in detail in Chapter 5, for the interested reader.

On the scanner data is acquired continuously. The fast gradient echo pulse sequence (FGRE) was modified as described in Chapter 3, to acquire data in a semi-bit-reversed fashion. The view ordering table was recalculated to acquire phase-encoded data in small interleaved sets. Sets of 8 equally spaced k-space lines are acquired linearly, followed by another set of 8 lines, interleaved with the previous set. In addition to changing the phase-ordering table, the Real-time Process (RTP) within the FGRE was programmed to make data available on the workstation side. This implementation, and a comparison with regular scanner operation can be found in Chapter 5. At the end, 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, and a data structure containing important information about the new set of data. This includes its location in memory, and the part of the k-space to which it belongs.

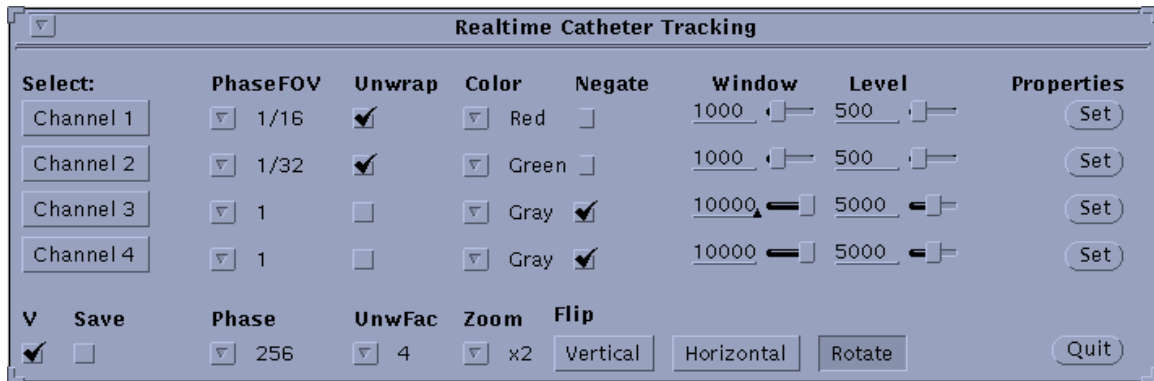


Figure 4.5: The graphical user interface for the workstation software

On the workstation, the application *realtime.c* was written to control the processes for data transfer, reconstruction, and display. The graphical user interface controlling all user preferences is shown in Figure 4.5. The reconstruction instructions include whether the channel is on/off, spatial resolution of the incoming data (needs to be matched with the scanner console), desired FOV on the channel (thus, the frame rate), whether unwrapping is necessary, the unwrap_factor for narrow FOV images, and the threshold level and center offset for unwrapped catheter images.

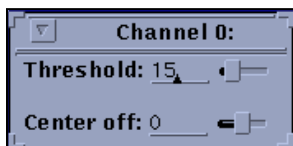


Figure 4.5 continued: Properties Tab (shown for the first channel, default values)

The resulting frame rate on each channel depends on the TR during acquisition and the number of phase encode lines used for reconstruction. Then the resulting images from multiple channels are combined for display. The user has independent control over the contrast/brightness of each channel, as well as the option to negate any channels as is common practice for radiologists in viewing the surrounding anatomy. Each channel can

also be color coded for easily distinguishing between multiple interventional devices and signal from blood vessels. A magnification factor of up to x4 for a full screen display is also available on the GUI as an option. While the program is executing, the “V”erbose, and “S”ave options allow the user to view timing parameters, state changes (mostly used for debugging); and to save each image on each channel, respectively.

A schematic diagram of the control mechanism and associated data buffers is provided in Figure 4.6. On the workstation, *realtime.c* controls the transfer, reconstruction, and display of the incoming data.

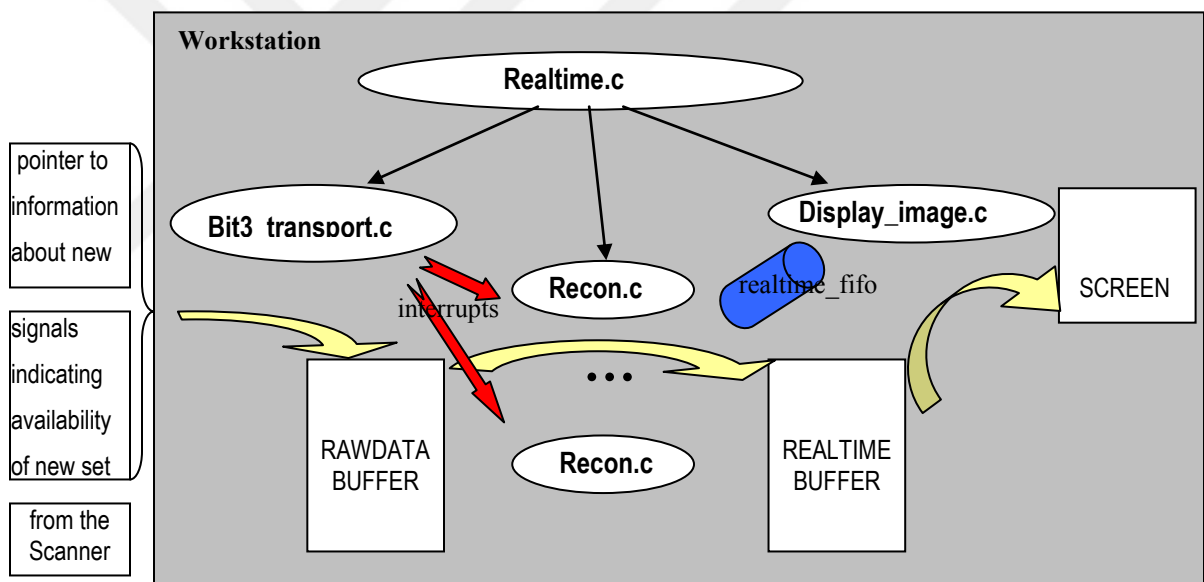


Figure 4.6: Flow of control and of data on the workstation. Yellow arrows indicate the flow of data, while red arrows represent Unix signals, and the blue pipe is a Unix fifo.

Realtime begins by activating the data transfer to the workstation, *Bit3_transport*. The sole purpose of *Bit3_transport* is to handle the transfer/copying of data over the bit3 interface from scanner memory to the workstation. The pulse sequence modifications signal the workstation with information regarding the arrival of a new small set of

k-space data, and its location in memory. *Bit3_transport* receives this signal, and copies the newest set of 8 k-space lines into a circular data buffer on the workstation. In addition, *Bit3_transport* receives information from the user interface regarding which channels are on, the process ID's of the corresponding recon processes, and the number of phase-encode lines expected by each recon (obtained from the FOV and spatial resolution information on the GUI). *Bit3_transport* keeps track of how many small sets of data arrived, and signals the reconstruction process of each active channel if enough data for that channel is available.

The reconstruction processes (*Recon.c*) are idle until informed by *Bit3_transport* of the arrival of enough data, for the desired FOV, on that channel. Once activated, each *recon* accesses the newest data from the rawdata buffer, and reorders it according to the semi-bit-reversed phase order. A 2D Fourier Transform is applied to reconstruct an image. A magnitude-squared image is calculated from the real and imaginary images obtained from the Fourier Transform. Unwrapping of the resulting image is performed if specified to do so. 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 (Chapter 2) is typically turned on for images with $1/4^{\text{th}}$ FOV or less, with an appropriate *unwrap_factor* chosen by the user. 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 $n \cdot N_y$, any multiple (n) of the number of phase encode steps (N_y). As soon as enough data to form a full FOV image arrives, this ambiguity is resolved, by calculating the correct catheter position, automatically determining n . 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. If an `unwrap_factor` of more than one is used, the catheter may not exactly be in the center of this FOV, but the center-offset of the image can be set on the GUI, to shift the displayed FOV around the catheter. The reconstructed (and unwrapped) image is then placed into an image buffer, and the display program (*display_image.c*) is signaled to refresh the image.

The resulting images from multiple channels are combined using the square root of the sum of squares of independent channels. The display process performs the contrast/brightness adjustment and negation on each channel. The combined image is also color coded and magnified if specified to do so. The refresh rate is that of the fastest channel.

Since we decided to collect data in sets of 8 k-space lines, this allows temporal resolution up to $8 \cdot TR$ (TR: repetition time) on guidewire and catheter channels. For an acquisition with a TR of 5msec, this leads to a frame rate of 25 frames/second. Roadmap channels should be reconstructed only after enough data for a FOV that contains the subject, is available. For a full FOV, (with 256 phase encode lines) this means 32 sets of 8 k-space lines need to be acquired and reconstructed, yielding a temporal resolution of 1.28 seconds (0.78 frames/second) for anatomical images.

Multiple channel multiple FOV images obtained from phantom and animal experiments are presented in Chapter 6, Results.

Chapter 5

Implementation

These sections address the interested reader who may be using this software, and involves details of the implementation. Others may skip to Chapter 6, Results, without any loss of generality or understanding of the general principles of the Multiple FOV MR Fluoroscopy method.

Most 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. Besides viewing the catheter or guidewire at very high frame rates, we can also provide a dynamic roadmap image using these available scanner resources. In this case, excitation with the MRF pulse results in identical data acquisition over all channels. 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.

Raw data acquired with the semi-bit-reversed phase order (Chapter 3) is automatically transferred to a workstation directly connected to the MR scanner after acquisition of each set of 8 k-space lines. Each channel is then reconstructed independently at different FOVs, and later combined for display.

This chapter will describe regular scanner operation and the modifications that were made to gain access to data immediately after acquisition, as well as the user-interface that controls reconstruction and display of the acquired data.

5.1 Scanner Software

5.1.1 Scanner Operation

All experiments were conducted on a GE 1.5T CV/I MRI (GE Medical Systems, Waukesha WI) scanner directly connected to a SunUltraII Workstation (Sun Microsystems, Mountainview CA) scanner by a high bandwidth data bus (Bit3 Corp., St.Paul MN). A block diagram of regular scanner operation is provided in Figure 5.1.

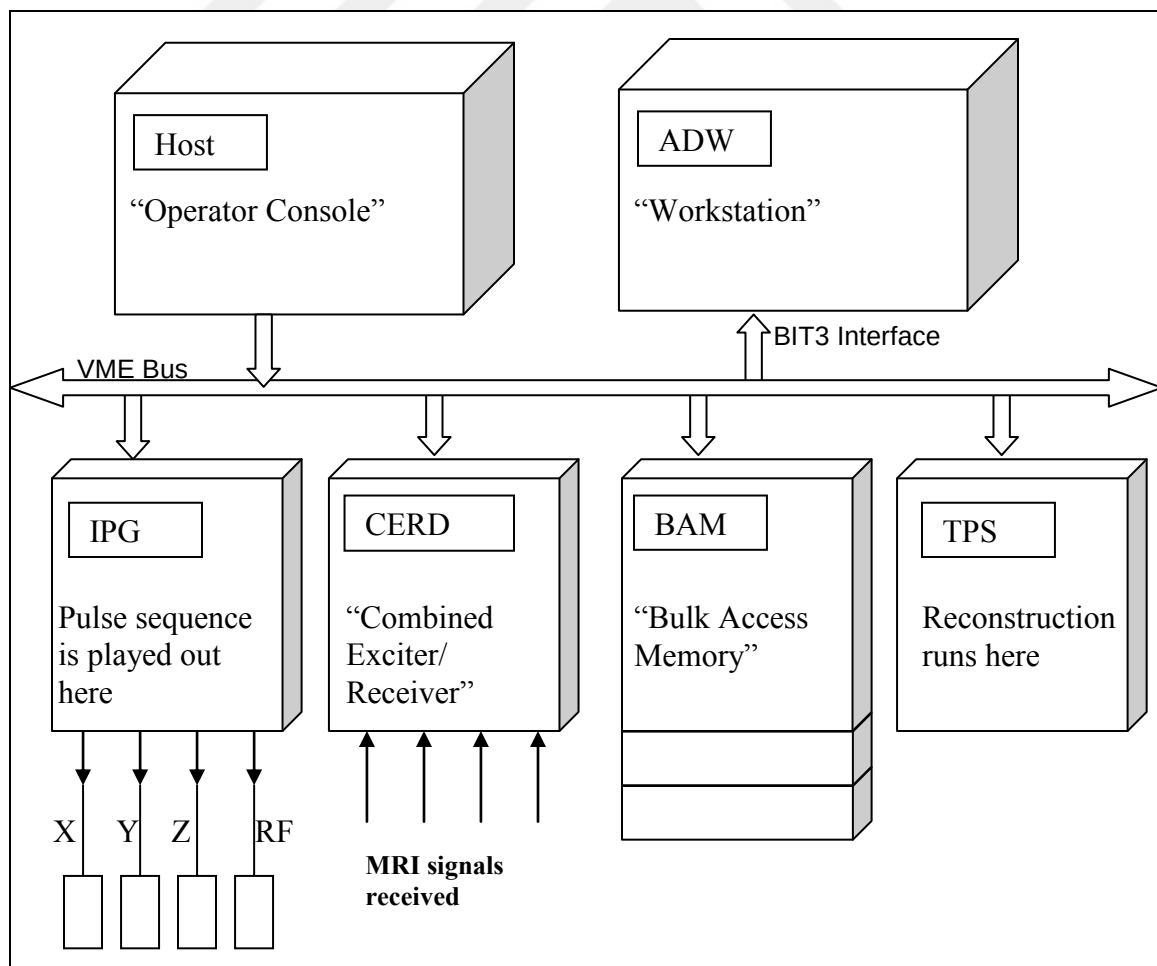


Figure 5.1: Scanner Block Diagram

The pulse sequence and imaging parameters are prescribed on the Operator Console, the Host, which is a Silicon Graphics Workstation. The IPG is where the pulse sequence runs once it is downloaded, and the Gx, Gy, Gz gradients and RF pulse are generated. The IPG deals mostly with playing out the pulse sequence, while the CERD, BAM and TPS deal with the signal reception, storage, and reconstruction. All of these components are connected to each other and the host via the VME bus. The CERD, combined exciter/receiver, receives MR signal, stores it in the BAM, bulk access memory, and signals the TPS with information regarding where the data is located and how it is to be reconstructed. Reconstruction is normally performed by a dedicated subsystem known as the AP (Array Processor, actually GxPowerPC chips, not a true AP) on the TPS, and the resulting images are displayed again on the host.

5.1.2 Data Transfer Protocol

Having described normal scanner operation, let's look at the changes that were made to export data from the scanner. The reconstruction, unwrapping and display programs run on a SunUltraII Workstation, the ADW1, directly connected to a GE 1.5T CV/I MRI scanner by a high bandwidth data bus, the BIT3 interface (Figure 5.2).

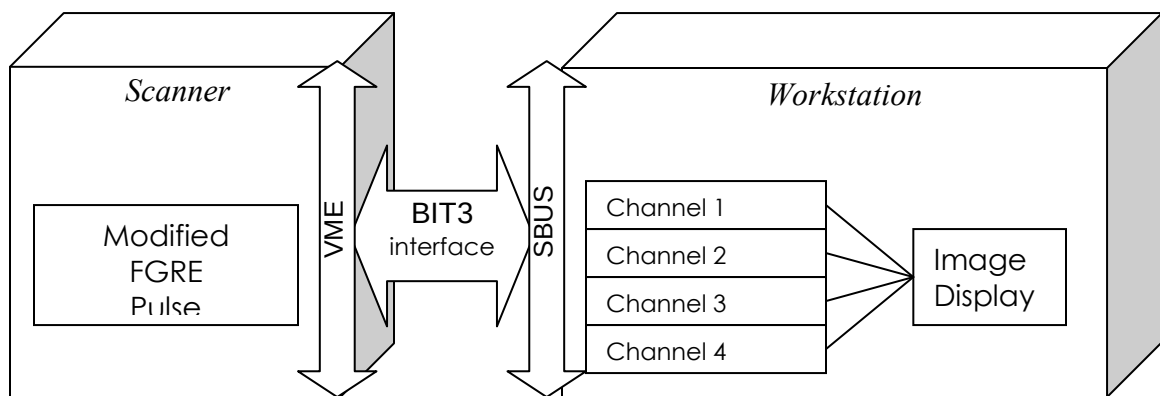


Figure 5.2: Data flow from scanner to workstation

All code was written in the C programming language. XView was used to create a user-interface and display utility.

On the scanner data is acquired continuously. The fast gradient echo pulse sequence (FGRE) was modified as described in Chapter 3, Data Acquisition. The view-ordering table was recalculated to acquire phase-encoded data in a semi bit-reversed fashion. Sets of 8 equally spaced k-space lines are acquired linearly, followed by another set of 8 lines, interleaved with the previous set. In addition to changing the phase-ordering table on the pulse sequence, the Real-time Process (RTP) was programmed within the FGRE, to make this data available on the workstation side. This option of making data available to the workstation, was named the Data Transfer Protocol (DTP) and is activated by the control variable “dtp_nsets”, when set to 32. Figure 5.2 summarizes the strategy of our data transfer protocol; to make data available on the workstation for our own reconstruction and display programs. Figure 5.3 provides details of how this is accomplished.

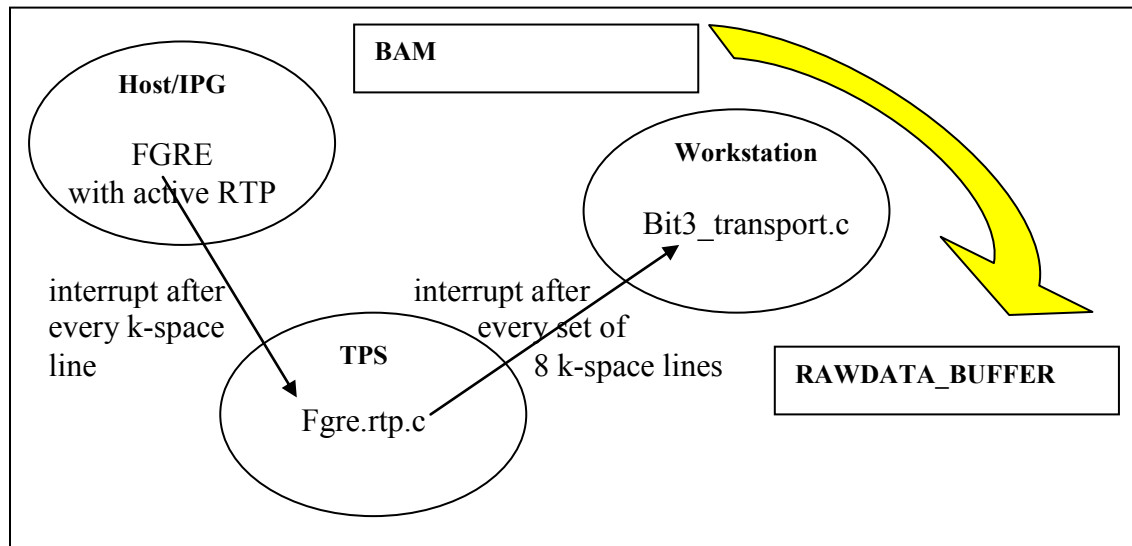


Figure 5.3: Data Transfer Protocol, flow of control and data (large arrow).

The entire sequence of events from acquisition, to data transfer, reconstruction and display was designed to be interrupt driven to avoid any delays or inconsistencies in timing associated with polling.

The pulse sequence and imaging parameters are prescribed on the host in the usual way. IPG provides the appropriate excitation and gradients, and CERD receives the corresponding MR signal. At this point the changes we made to the RTP become effective (Figure 5.3). Once the data is saved on the BAM, rather than signaling the TPS for a standard reconstruction, the RTP signals the TPS to run our TPS code, *fgre.rtp.c*, instead. This program, *fgre.rtp*, receives an interrupt from the CERD after acquisition of each k-space line. This program also has access to information regarding where in the BAM memory data was placed, as well as pulse sequence specific information in the *ntp_header*. This *ntp_header* is formed within the FGRE, and contains information about the acquisition, such as the x, y, z, resolution, data size, size of the linear sets of data, number of such sets to be acquired, and the number of receivers collecting data. When the first RTP interrupt is received by *fgre.rtp*, this code creates another header with this information and the location of the data, and an index indicating how many sets of data have been acquired. It then places a pointer to this new header in a “well known location” on TPS memory (0x3fff00). The 256 bytes of TPS memory from 0x3fff00 to 0x400000 are not currently used by normal Signa operations. On subsequent interrupts, the task of the *fgre.rtp* code is to count the number of interrupts it received, and increment the corresponding index of the header. When the desired number of k-space lines have accumulated, currently 8, an interrupt is generated over the Bit3 to alert the

workstation of the availability of a new set of data. The `fgre.rtp.c` code is summarized in the block diagram of Figure 5.4.

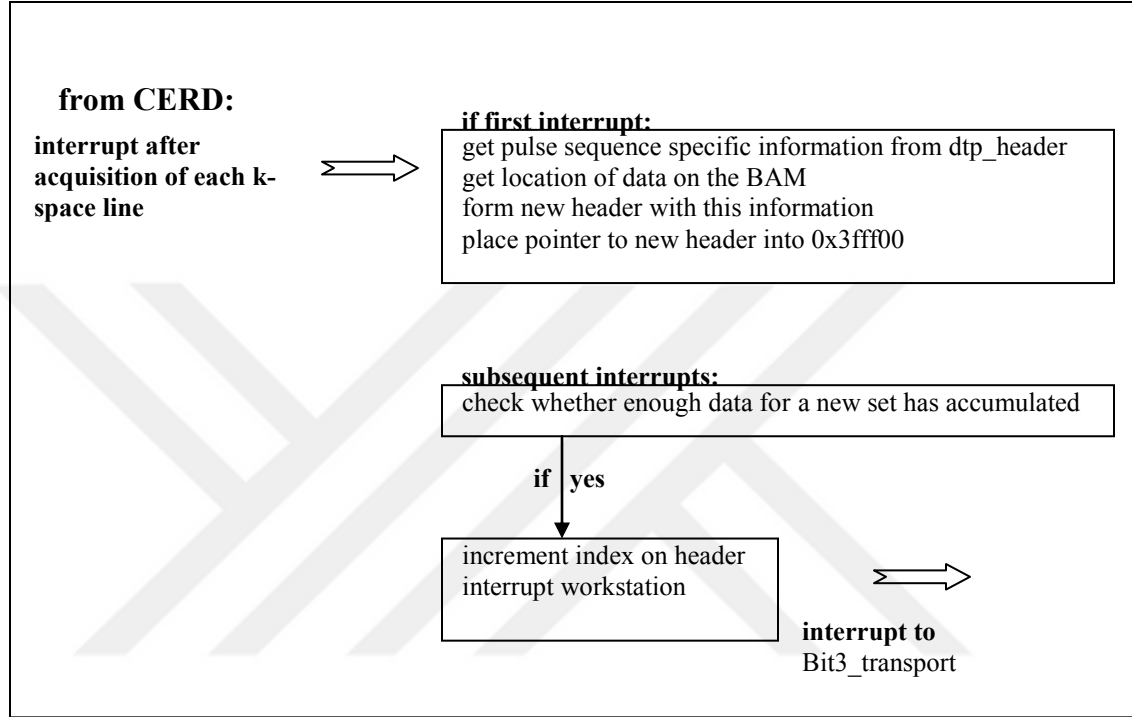


Figure 5.4: `Fgre.rtp.c`

5.2 Workstation Software

5.2.1 Overview

So far, the operations on the scanner side were described; namely, data acquisition with the semi-bit-reversed phase order, and the modifications that were made to make data available on the workstation side. At this point, the information available on the workstation side is the “well known” memory location on TPS, and periodic interrupts over the Bit3 announcing the arrival of a new small set of k-space lines.

On the workstation side, the application *realtime.c* runs, that controls the processes for the transfer, reconstruction, and display of the incoming data. A schematic diagram of the control mechanism and associated data buffers is provided in Figure 5.5.

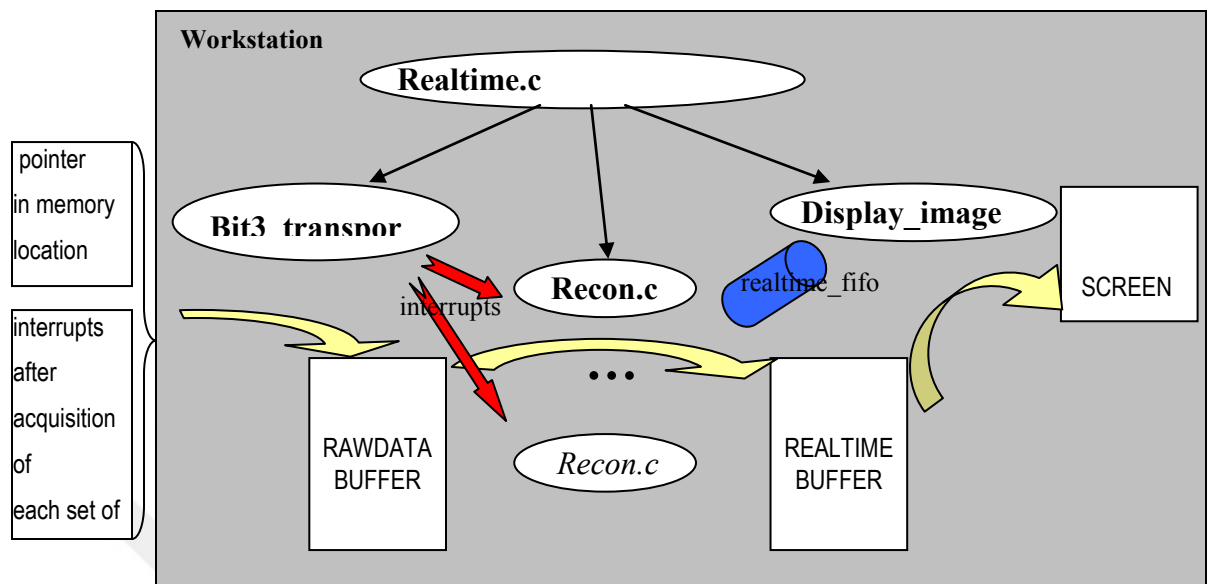


Figure 5.5: Flow of control and of data on the workstation. Yellow arrows indicate the flow of data, while red arrows represent Unix signals, and the blue pipe is a Unix fifo.

Bit3_transport.c is responsible for transferring data from scanner BAM (bulk access memory) to the RawdataBuffer on the workstation. The RawdataBuffer consists of a header (*rawdata_header*) that contains information about the most recently acquired set of data, the process ID's of the reconstruction process for each channel, and a circular data buffer with enough space to hold a full FOV image from each active receiver channel. The *recon.c*'s are the reconstruction processes (that also handle unwrapping) of each channel. They access rawdata from the RawdataBuffer, and place the reconstructed images in the RealtimeBuffer. The RealtimeBuffer consists of a header for each channel (*realtime_header*) with reconstruction and display preferences from the user, and a circular buffer with enough room to hold 20 images for each channel.

Display_image.c is the program responsible for combining and manipulating the multiple channel images.

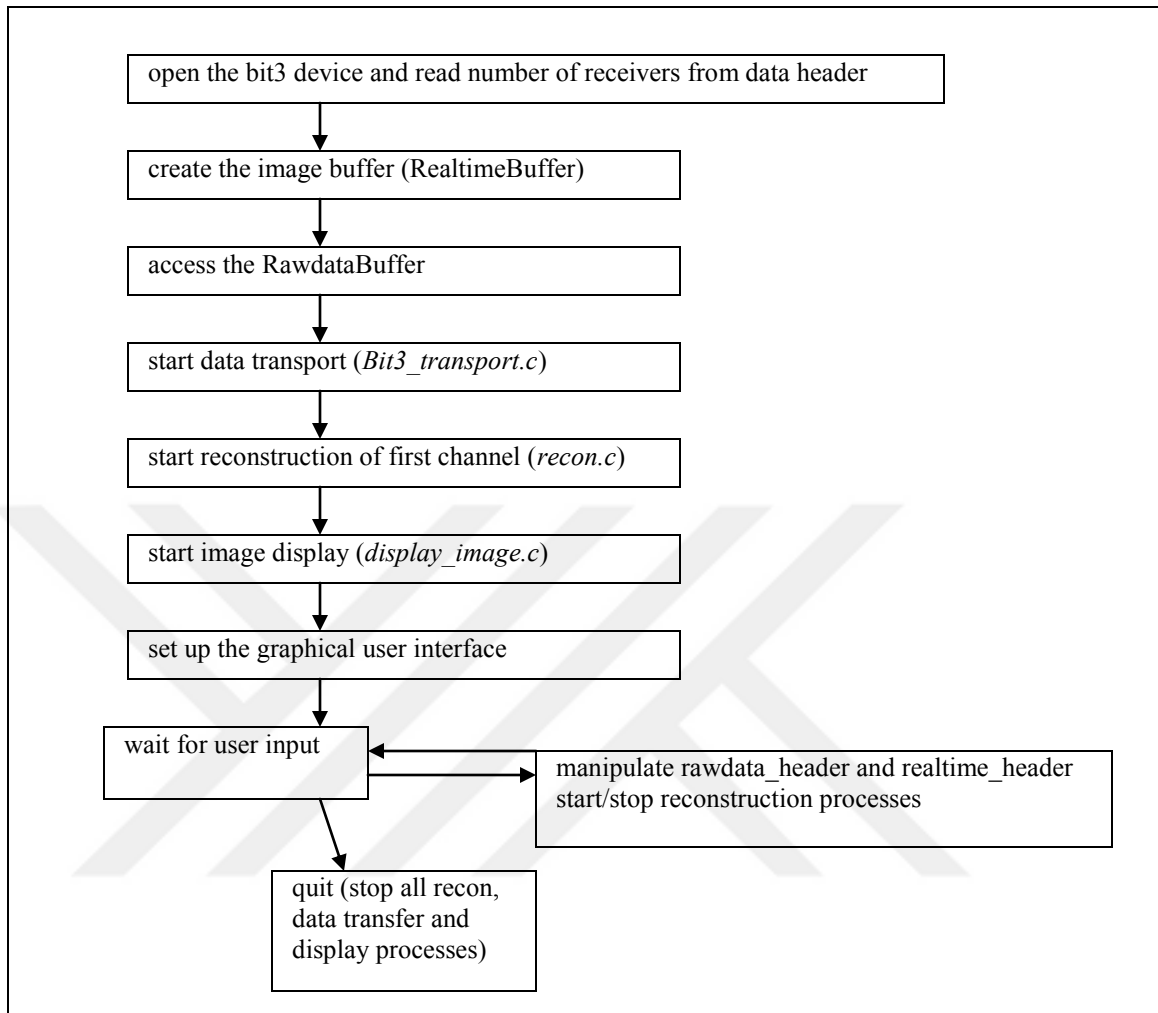


Figure 5.6: Realtime.c

Realtime starts by obtaining the number of active receiver channels from the `dtp_header` in TPS memory (Figure 5.6). This header is accessed via the “well known” memory location in the TPS, which contains a pointer to it. It then activates the *Bit3_transport*, *recon* for the first channel, and the *display* programs. *Bit3_transport* transfers 8 lines of k-space to the `RawdataBuffer` immediately after their acquisition, increments the corresponding index indicating which set of data was transferred on the `rawdata_header`. It then sends an interrupt to each of the active *recon* processes, for which enough data for the desired FOV image has accumulated. The process id’s of the active *recon*’s are written into the `rawdata_header` by *realtime* after their activation. Each

of the *recon* processes obtain data from the RawdataBuffer, reconstruct it according to user preferences, place the final image into the RealtimeBuffer, and increment the corresponding index indicating which image was reconstructed last. The *recon* and *display* programs communicate via a first-in-first-out pipe, *realtime_fifo* (Figure 5.5). Each *recon* that has finished, writes the channel number that it is reconstructing into this pipe. The display program waits for messages to become available on the *realtime_fifo*, and whenever a new image is available, displays it according to user specifications on that channel.

A summary of the *realtime.c* code is provided in the block diagram of Figure 5.6.

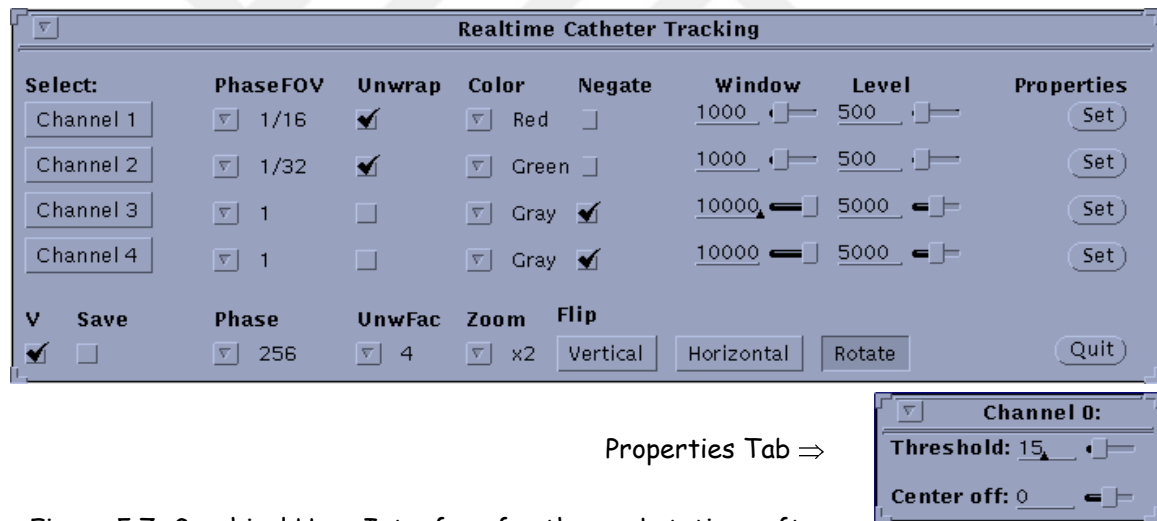


Figure 5.7: Graphical User Interface for the workstation software

The user-interface controlling the reconstruction and display programs is shown in Figure 5.7 for four receiver channels. All the user preferences regarding reconstruction and display are communicated to the *recon* and *display* programs via headers in the RealtimeBuffer (Figure 5.5). *Realtime* continues to receive user input, and manipulates these headers accordingly, and starts/stops *recon* processes as desired. Each channel has its own *realtime_header*, which contains reconstruction instructions, as well

as display instructions which will be discussed in the next section. The reconstruction instructions include: whether the channel is on/off, spatial resolution of the incoming data (needs to be matched with the scanner console); desired FOV on the channel (thus the frame rate); whether unwrapping is necessary; and the threshold level, and center-offset for unwrapped catheter images. By default all channels are reconstructed at full FOV initially, since the *realtime* process does not know which channels the guidewire and catheter are physically connected to. Once found, catheter and guidewire data can be reconstructed immediately after acquisition. The resulting frame rate depends on the TR during acquisition, and the number of phase encode lines used for the reconstruction. Since we decided to collect data in sets of 8 k-space lines, this yields a temporal resolution of 40ms (25 frames/second) for guidewire and catheter channels (TR = 5ms, 256 phase encodes). Roadmap channels should be reconstructed only after enough data for a FOV that contains the subject is available. For a full FOV, this means 32 sets of 8 k-space lines, yielding a temporal resolution of 1.28 seconds for anatomical images.

5.2.2 Bit3_Transport

Realtime starts by activating the data transfer to the workstation, *Bit3_transport.c*. The sole purpose of *Bit3_transport* is to handle the transfer/copying of data over the Bit3 from Signa BAM to the workstation.

The interrupts generated by *fgre.rtp.c* (running on the TPS) after the acquisition of each set of 8 k-space lines are caught by our data transfer program, *Bit3_transport* on the workstation. *Bit3_transport* opens the bit3 device from the workstation side. The device drives supplied for the Bit3 adapter support interrupts. A hardware interrupt sent at the

bit3 adapter is passed to the Unix process controlling the device *-Bit3_transport-* as a software interrupt, a Unix signal.

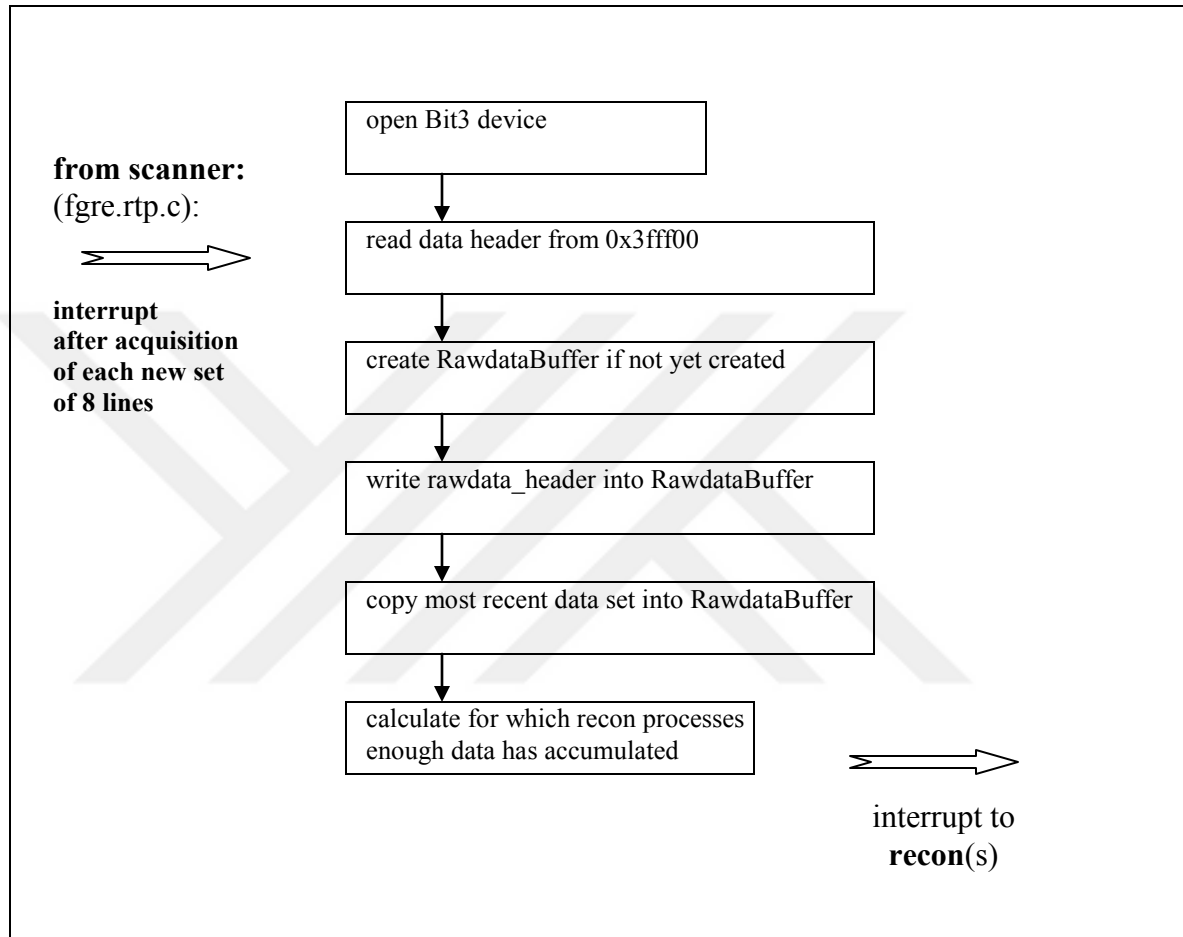


Figure 5.8: Bit3transport.c

Upon receiving an interrupt, *Bit3_transport* begins by obtaining information about the incoming data from the “well known location” 0x3fff00 on TPS memory. A pointer to the data header was placed into this location by the *fgre.rtp* code (Figure 5.4). Once the validity of this header is confirmed, the size of data that needs to be moved is calculated, and placed into a memory mapped buffer file on the workstation memory (RawdataBuffer). The RawdataBuffer consists of a header that contains information about the most recently acquired set of data, the process ID’s of the reconstruction

processes for each channel, and a circular data buffer with enough space to hold a full FOV image from each active receiver channel. The header consists of a “magic” number (to validate its source), x, y, z resolutions, data size, number of total sets that will form a full FOV image, and an index that indicates which of these sets is the most recently acquired one. This index is very important since it indicates which part of k-space this newest set belongs to, and will be used by the reconstruction processes. The `rawdata_header` also contains information about the spatial resolution (total number of phase encode lines), and the FOV desired on each channel. This information is updated by *realtime.c* according to user preferences, and used to calculate how many k-space lines are expected by the recon processes of each channel. The rawdata header index is the indicator of whether enough data is available to reconstruct the desired FOV image on a certain channel or not. Enough data for a certain channel is available if the index is a multiple of $(\text{phase_encodes}/\text{YRES}*\text{FOV})$, where phase-encodes is 256, 128, or 64, indicating the spatial resolution of the incoming data, YRES is the size of each linear set of k-space lines, currently 8, and FOV is the desired FOV for that channel (1/2, 1/4, 1/8, ..., but not less than that achievable with one linear set of k-space lines). If enough data is available for a recon, it will be informed, and if not left idle until enough data is available. Once each set of data is placed into the RawdataBuffer, *Bit3_Transport* sends interrupts to active *recon* processes with enough data to inform them about the availability of their data.

A summary of *Bit3_transport* is provided in the block diagram of Figure 5.8.

5.2.3 Reconstruction

Each *recon* process starts by gaining access to the RawdataBuffer, and RealtimeBuffer, and then waits until an interrupt from *Bit3_transport* is received indicating the arrival of enough data to reconstruct the next image.

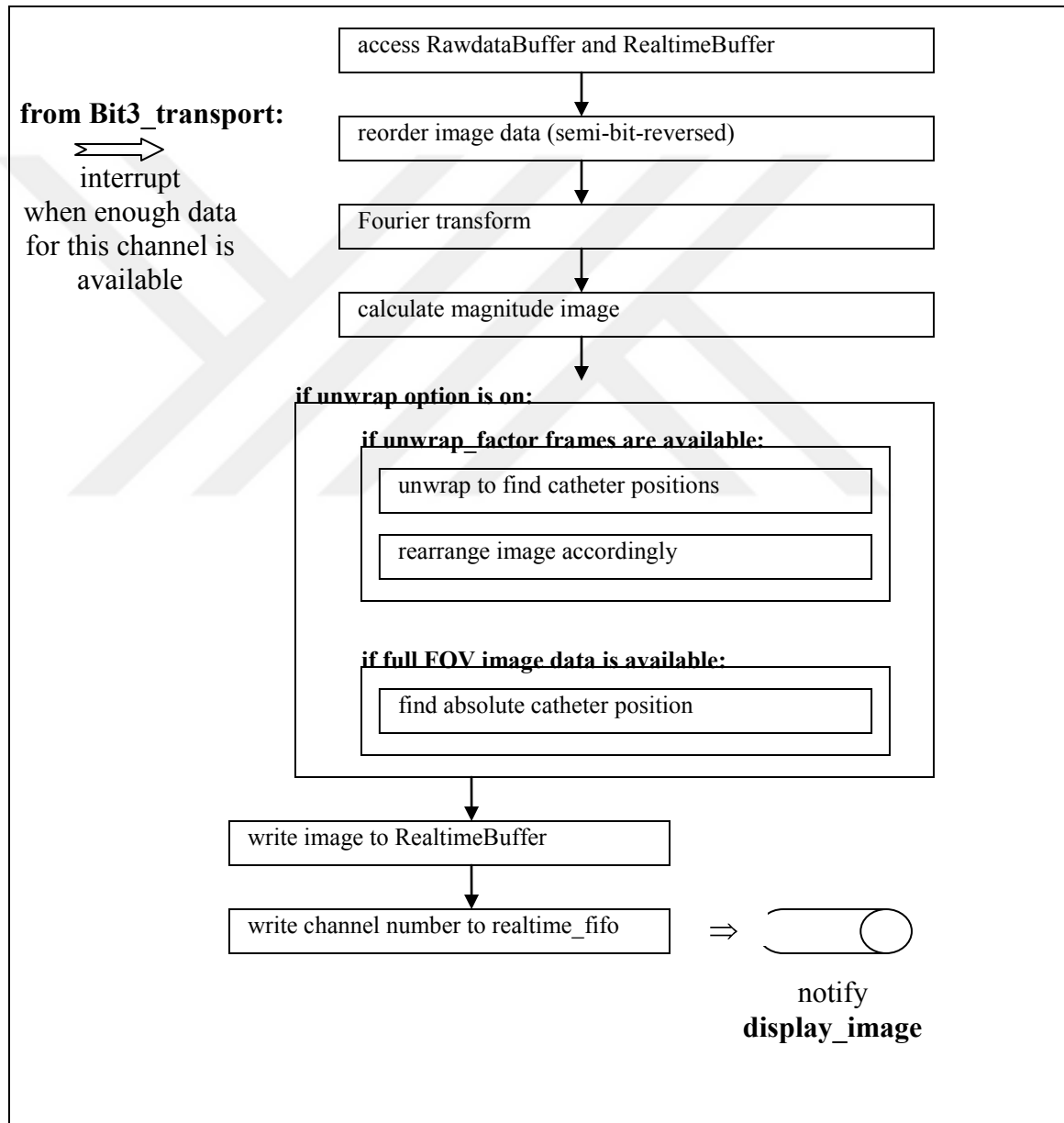


Figure 5.9: Recon.c

If enough data is available, the incoming data is reordered according to the semi-bit-reversed phase order, and placed into the correct locations in the data matrix. A2D Fourier Transform is applied to reconstruct the data. The FFTW, “Fastest Fourier Transform in the West” was used for its superior speed to allow operation under realtime constraints (MIT 96). The square of the magnitude image is formed from the real and imaginary images obtained from the FFTW. The square root is not calculated at this stage since the display program combines images from multiple channels using the square root of the sum of the squares, and redundant calculation is avoided.

Unwrapping of the resulting image is performed if specified to do so. For catheter and guidewire images, degree of image overlap depends on the apparent width of the interventional device, and its orientation with respect to the phase-encode direction. The unwrapping algorithm is typically turned on for images with $1/4^{\text{th}}$ the full FOV or less. If the image quality is not good enough to successfully unwrap the desired FOV image, the variable “unwrap_factor” can be set on the user interface. Then, larger FOV images (desired FOV * unwrap_factor) will be reconstructed and unwrapped in the background, and this information will be used to unwrap the narrower FOV images without reducing the frame rate of catheter visualization (Section 2.4). By default, the value of “unwrap_factor” is “1”, that is, each reconstructed image is unwrapped to find the correct catheter shape. If image quality does not allow unwrapping of each frame individually a larger unwrap_factor should be used. Unwrap factor is set to the value 2^k ($k=0, \dots, \log_2(\text{total phase encodes}/\text{setsize})$) with an appropriate k , and unwrapping is delayed until the 2^{kth} frames data arrives. At this point, enough data exists to reconstruct an image with a larger FOV (unwrap_factor*desired FOV). Unwrap_factor frames worth of

data are reordered, Fourier transformed, and a magnitude image is formed in the background. The catheter shape is extracted from this large FOV, high quality image by unwrapping, and used to display the next `unwrap_factor` frames with narrow FOV catheter images. Note that when aliasing is allowed in the phase-encode direction, there is an ambiguity in the absolute position of the catheter. The resulting image may be shifted by $(n \times N_y)$ pixels, where N_y is the number of phase-encode steps, and n is any integer number. To solve this ambiguity, as soon as enough data to form a full FOV image arrives, a full image is created, and the absolute catheter position is calculated. This automatically determines the value of n . All of the other images can be registered using the information from previous frames, and the ambiguity in the catheter position is eliminated. Once unwrapping is completed, the aliased image is rearranged according to the unwrapped catheter positions.

The reconstructed (and unwrapped) image is then placed into the `RealtimeBuffer`, the corresponding image index in the `realtime_header` for the channel is incremented, and the channel number of the completed *recon* process is written into the `realtime_fifo`, so that the display can be refreshed.

The *recon.c* code is summarized in the block diagram of Figure 5.9.

5.2.4 Display

The resulting images from multiple channels are combined for display using the square root of the sum of squares of independent channels. The user has independent control over the contrast, and brightness of each channel, as well as the option to negate any channel, as is common practice for radiologists in viewing the surrounding anatomy.

The ability to color code each channel using grayscale, red, green, or blue values is also provided. This is very helpful in for distinguishing the various contributors to increased signal level in the vasculature. For instance in an MR guided angiography procedure, the contrast agent in the vessels, the guidewire, and catheter all contribute to the signal, however, simple distinction between them is provided with the color-coding of data from separate channels.

All the user preferences regarding display are communicated between the user interface and *display* via the *realtime_headers* of the *RealtimeBuffer*.

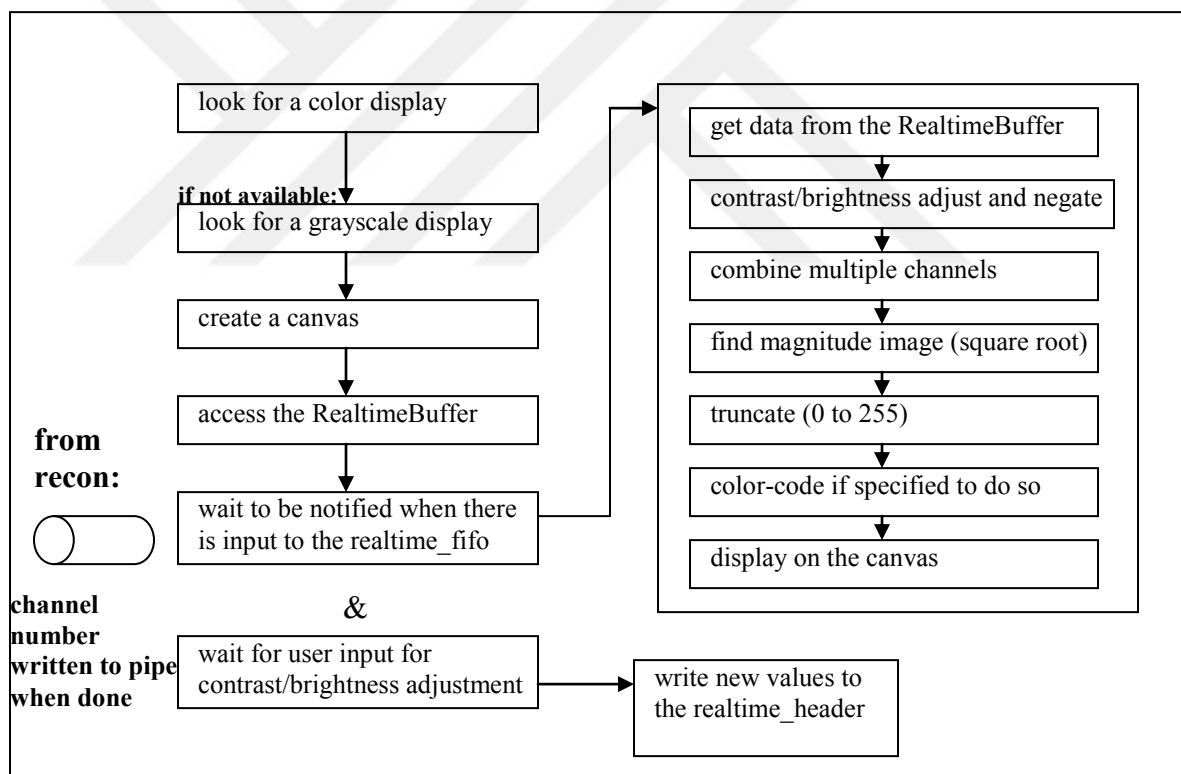


Figure 5.10: Display.c

Display.c begins by checking the available Visuals on the current display. A TrueColor visual with 24-bit depth is desirable to allow color-coding of images from different channels. If color is not available on the screen, a PseudoColor visual with 8-bit display depth is accepted. If neither are available, the program aborts.

If a color or grayscale display is found, a canvas is created for displaying images, the program gains access to the RealtimeBuffer, and opens the `realtime_fifo`. *Display* is idle until any of the *recon* processes writes their channel number on the `realtime_fifo`. As mentioned previously, the Multiple-FOV MR Fluoroscopy technique is interrupt driven from beginning to end. When there is new data on the `realtime_fifo` from any of the recon processes, *Display* gets notified, and proceeds to refresh the current image. A block diagram of *Display.c* is provided in Figure 5.10.

The displayed combined image is the square root of the sum-of-squares of contrast/brightness adjusted (negated) images from each individual channel.

$$\text{Combined Image}^2 = \sum_i \text{Channel}_i^2 \quad (i = 0 \text{ to number of receivers})$$

The contrast and brightness values to be used for each channel are obtained from the `realtime_header` of the corresponding channel. Figure 5.11 shows the mapping between signal intensities and corresponding pixel values for a certain channel.

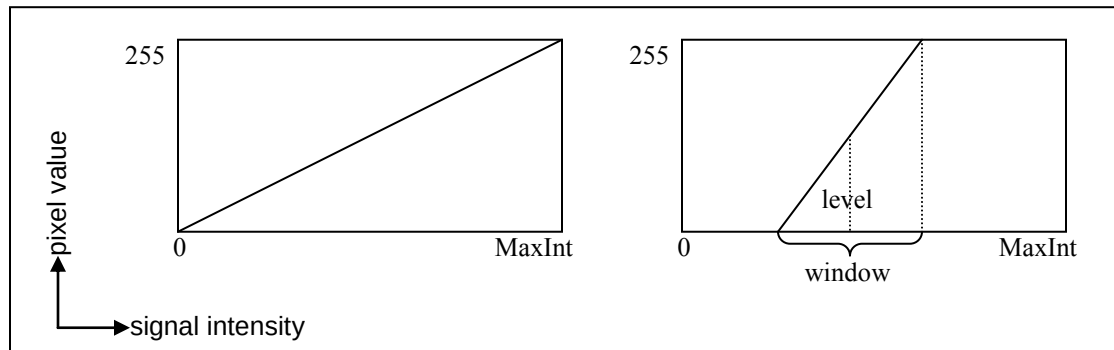


Figure 5.11: Colormap manipulation. Default, and Contrast/brightness adjusted

The window variable determines the width of the window, which will be assigned the pixel values 0 through 255, while the level variable determines the median around which this window is placed. With some channels collecting roadmap data and others collecting catheter or guidewire data, there is a very wide range of signal intensities that

need to be displayed. The window and level variables on the user interface, are thus allowed values ranging from 1 to 10000 (WinMax), and 0 to 10000 (LevMax) respectively. The default window value is 10000, and default level value is 5000, which provide the linear mapping shown in Figure 5.11. MaxInt is a constant number on the order of the maximum intensity value expected on any of the channels.

Note that the images placed in the RealtimeBuffer by the *recon* processes are magnitude “squared” images, not magnitude images. So, the window/level adjustment on these images will be performed using the “square” of the desired window/level values, and pixel values will be allowed a range up to 255^2 , at this point.

The contrast and brightness adjusted pixel value for each non-zero point on every active channel is found as follows:

$$\text{window} = (\text{window} * \text{MaxInt} / \text{WinMax})^2;$$

$$\text{level} = (\text{level} * \text{MaxInt} / \text{LevMax})^2;$$

$$\Rightarrow \text{pixel_value} = 255^2 * (\text{signal} + (\text{window}/2 - \text{level}))/\text{window}; \text{ (DSP 92)}$$

If negation of the channel is specified the colormap is inverted in the same operation as follows:

$$\Rightarrow \text{pixel_value} = 255^2 * (-\text{signal} + (\text{window}/2 + \text{level}))/\text{window}; \text{ (DSP 92)}$$

8 bits are available to display the signal intensity at each pixel (8 bit grayscale values on a monochrome system, and 8 bits for each of the colors, red, green, and blue, on a color system). So signal intensities must be scaled to values between 0 and 255 for each channel following the contrast and brightness adjustments. Note that the image passed to the *display* is the magnitude squared image to avoid a redundant square root

calculation in *recon*, and will be scaled between 0 and 255^2 at this stage. The square root will be calculated after data from multiple channels is combined. So, this contrast/brightness adjusted pixel value is truncated between 0 and 255^2 :

```
pixel_value = (pixel_value > 2552) ? 2552 : ((pixel_value < 0) ? 0 : pixel_value);
```

The last operation applied before data from multiple channels is combined is color-coding. Color coding is available only on systems that support a 24 bit screen depth. As shown in Figure 5.12, 8 bits are available for each color. For a grayscale image, all three colors are assigned the same pixel value.

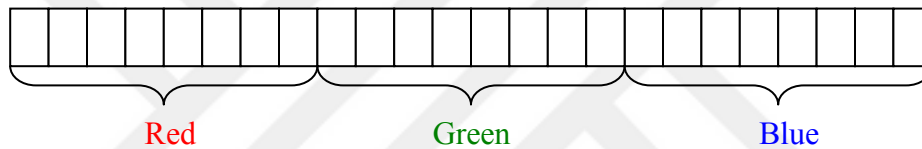


Figure 5.12: Color allocation on a 24-bit display

Finally, the images from each active channel are combined to form the final image, by adding the contrast/brightness adjusted, truncated, color-coded, square pixel values, and calculating the square root. The final image displayed according to the magnification factor, zoom, specified on the user interface. The default is to display the image as it is, however, an enlargement of x4 in both dimensions is available.

Chapter 6

Results

This chapter will present the results obtained on some phantom and animal experiments with the multiple FOV MR Fluoroscopy method. All experiments were performed on a GE 1.5T CV/i MRI scanner (GE Medical Systems, Waukesha WI) using the modified fast-gradient-echo pulse sequence. The realtime data transfer, reconstruction and display programs ran on a SunUltraII Workstation (Sun Microsystems, Mountainview CA) directly connected to the scanner by a high bandwidth data bus (Bit3 Corporation, St.Paul MN).

Section 6.1 demonstrates the formation of multiple FOV's within one image, while Section 6.2 presents multiple channel multiple FOV images from an animal experiment. Timing information on the performance of the realtime code is provided in Section 6.3.

6.1 Multiple Fields of View within one Image:

The flexibility of reconstructing different parts of an image with different FOVs is useful in cases where one part of the catheter requires a larger FOV than all other parts, which would normally slow down the entire movie.

Figure 6.2 shows the results we 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 towards the top of the images while data was being acquired with the semi-bit-reversed phase order.

The full FOV image (Figure 6.1) shows significant ghosting due to the motion of the object during the imaging time of 1.28 seconds ($TR = 5\text{ms}$, 256×256 matrix). The variable FOV images (Figure 6.2) were reconstructed regionally at different FOVs as described in Section 4.1. The central section containing the loop required a larger FOV than the tip or tail of the catheter, thus has a lower temporal resolution. The center of the image was updated 3.12 times/second 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, and 25 frames/second frame rate.



Figure 6.1: Full FOV image of a looped catheter. (Imaging parameters: FGRE, 256×256 data matrix, 4° FA, 0dB TG, 62.5kHz BW, TR/TE: 5.1/1.6msec, no slice selection)

frame rate = 0.78 frames/second

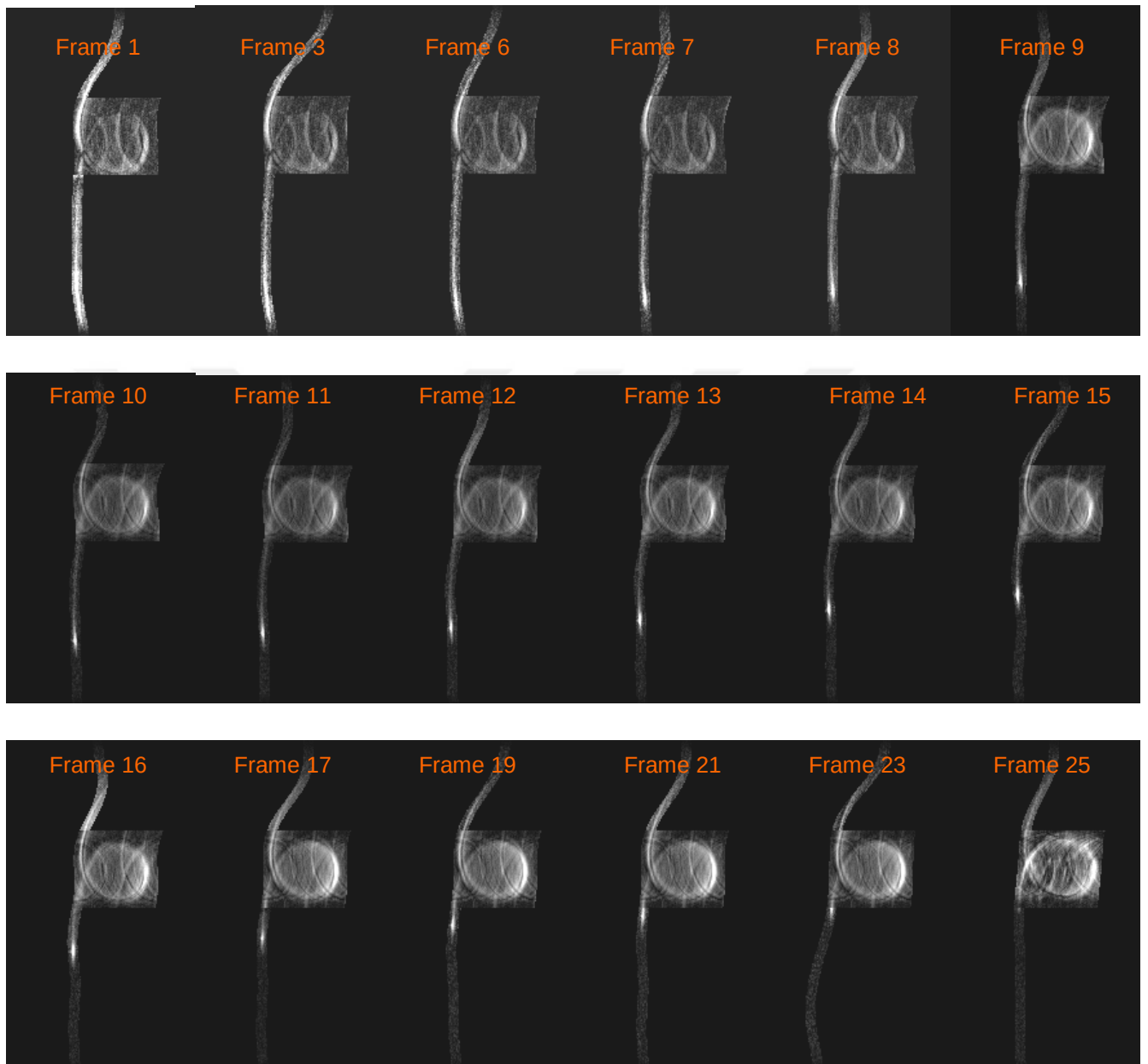


Figure 6.2: Variable FOV images of a looped catheter, reconstructed with the same data as the full FOV image shown in Figure 6.1.

Imaging parameters: FGRE, 256x256 data matrix, 4° FA, 0dB TG, 62.5kHz BW, TR/TE:

5.1/1.6msec, no slice selection

frame rate of tip and tail of catheter = 25 frames/second

frame rate of loop = 3.12 frames/second

6.2 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 during 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.

Figures 6.3 and 6.4 show one frame of the movie recorded during the procedure, as well as the outputs of individual channels. Channel 0 was connected to the catheter, and channel 1 was connected to the guidewire. Channel 2 is the surface coil on the anterior side of the animal, and channel 3 is the surface coil on the posterior side.

Both active interventional devices are used in the receive-only mode of operation, and their diameter of sensitivity extends to several centimeters away from device. The minimum FOV required to avoid aliasing of image components far away from the catheter or guidewire was 1/4th the full FOV (Figure 6.3). In this case, the catheter and guidewire data was reconstructed after the acquisition of four sets of 8 k-space lines, providing a frame rate of 5.8 frames/second for catheter tracking, while maintaining less than 3mm in-plane resolution. The reconstruction of the anatomical data 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 is 1.45 frames/second. Also note the increase in SNR of the anatomical images, when data from the two receiver coils (anterior and posterior) are combined.

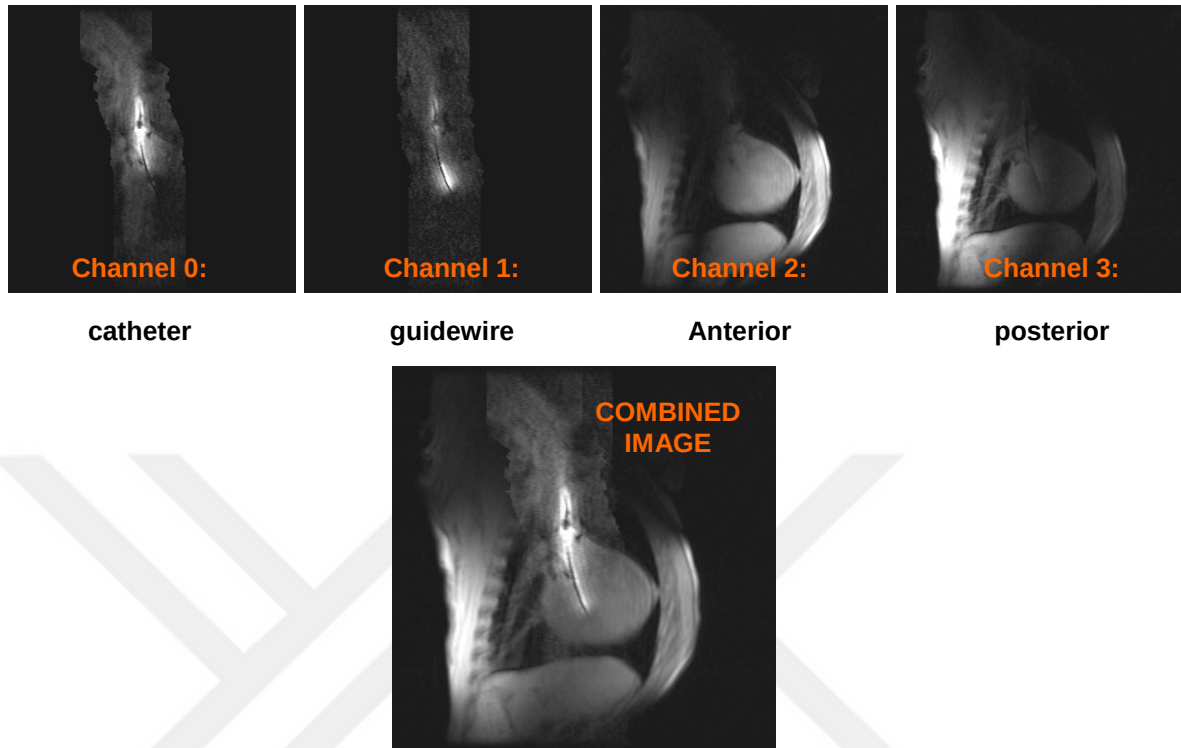


Figure 6.3: Multiple Channel, Multiple FOV images of a dog.

(Imaging parameters: modified fast-gradient-echo pulse sequence, TR/TE: 5.6/1.6ms, FOV: 36cm, Flip Angle: 7°, 256x128 acquisition matrix (partial NEX), no slice selection.)

On the user interface:

Phase:128; phaseFOV: "1/4th" on channels 0 and 1, "full" on channels 2 and 3;

Unwrap: "ON" for channels 0 and 1; with unwrap_factor:2;

Frame Rates:

Catheter:	5.8 frames/second
Guidewire:	5.8 frames/second
Roadmap:	1.45 frames/second

FOV's:

1/4 th
1/4 th
Full

The frame rate of catheter and guidewire visualization can be increased, by further decreasing the FOV on these channels. The experiment was repeated using 1/8th FOV images on channels 0 and 1. This increases the frame rate of interventional device visualization to 11.6 frames/second. In this case, some aliased signal effected the catheter and guidewire images, and caused a phenomenon called "flashing" in the

resulting movie. Corresponding images are shown in Figure 6.4, and the flashing issue will be further addressed in Chapter 7, Discussion.

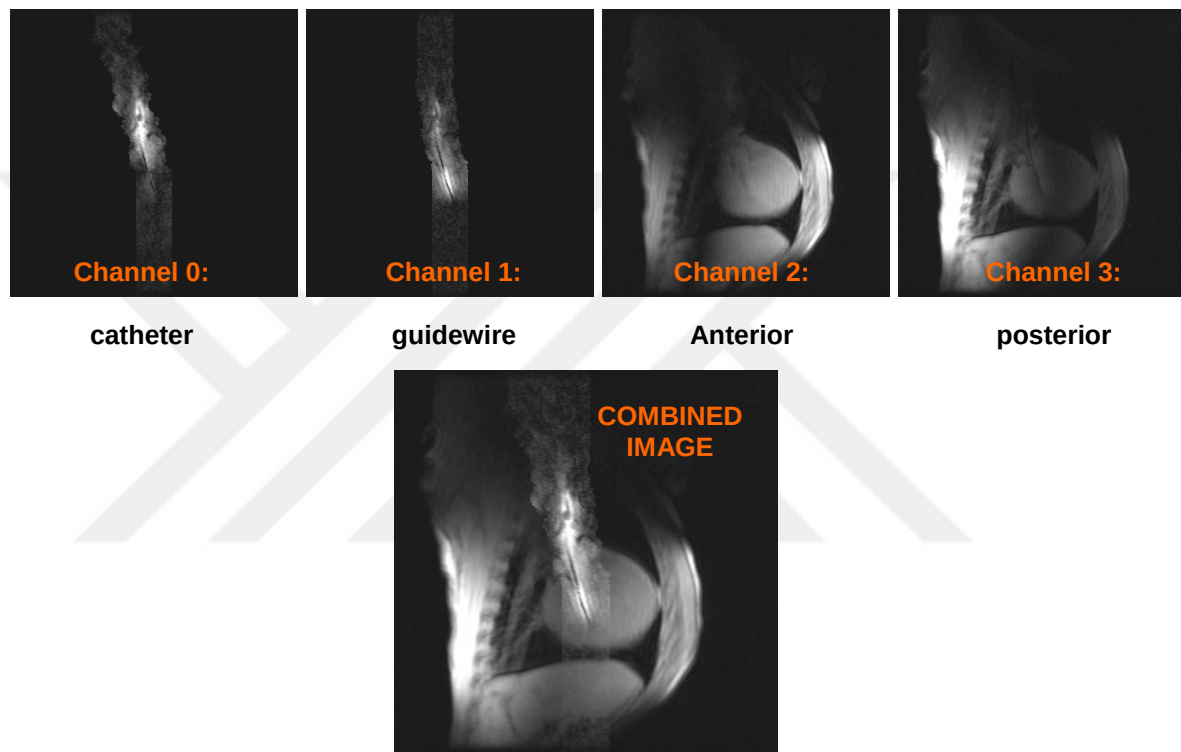


Figure 6.4: Multiple Channel, Multiple FOV images of a dog.

(Imaging parameters: modified fast-gradient-echo pulse sequence, TR/TE: 5.6/1.6ms, FOV: 36cm, Flip Angle: 7°, 256x128 acquisition matrix (partial NEX), no slice selection.)

On the user interface:

Phase:128; phaseFOV: "1/8th" on channels 0 and 1, "full" on channels 2 and 3;

Unwrap: "ON" for channels 0 and 1; with unwrap_factor:2;

Frame Rates:

Catheter:	11.6 frames/second
Guidewire:	11.6 frames/second
Roadmap:	1.45 frames/second

FOV's:

1/8 th
1/8 th
Full

Corresponding movies can be seen on our web page (IVMRI), and the appendix CD.

6.2 Timing:

The components of the real-time implementation of the Multiple FOV MR Fluoroscopy technique are data transfer from the scanner to the workstation (Bit3_transport.c), reconstruction and unwrapping of images (Recon.c) and display (Display_image.c). The following performance statistics were obtained on a SunUltraII workstation (Sun Microsystems, Mountainview CA).

Bit3_transport is responsible for copying rawdata from scanner memory into the workstation memory, and sending signals to appropriate reconstruction processes when their data is available. It receives interrupts from the scanner immediately after the acquisition of each small set of k-space data. The frequency of these signals depends on the TR during acquisition, and the number of phase-encode lines in each small set. Its performance is measured by the percentage of these interrupts that it successfully handled. When Bit3_transport fails to respond to an interrupt a certain set of k-space lines will never be transferred to the workstation.

At a TR of 5msec and when each small set consists of 8 k-space lines, Bit3_transport receives interrupts every 40msec, and successfully services around 99.8% of them.

Recon is where an image is formed, and unwrapped if necessary. Each recon is idle until enough data to reconstruct a desired FOV image arrives in the workstation memory. When signaled by Bit3_transport, recon performs several operations, the most time consuming of which is Fourier Transforming the incoming data, followed by unwrapping. The frame rate on each channel determines the amount of time available for the reconstruction process of that channel

The performance of the FFTW (“Fastest Fourier Transform of the West”, Frigo M, Johnson SG) code used for various size 2DFT’s is depicted in Figure 6.5(for in-place transforms with 4 byte data). It is most time consuming for large FOV images, but becomes very fast for narrow FOV images, and does not pose a limitation on the overall performance of our method.

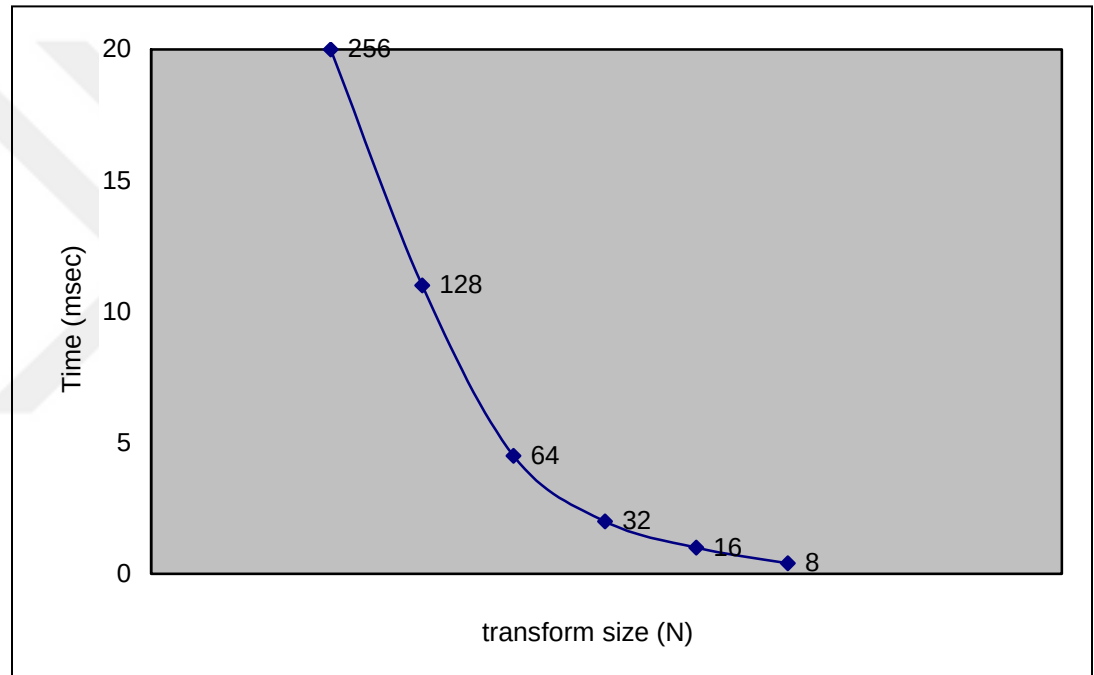


Figure 6.5: Reconstruction speed, with various size Fourier transforms (FFTW). The transform sizes used are: (256 readout points) $\times$ (N), where N denotes the number of phase-encode lines used for a particular FOV image. (The horizontal axis uses a logarithmic scale)

In addition to Fourier transforming the rawdata, recon also performs unwrapping of narrow FOV images. The unwrapping algorithm behaves similar to the Fourier Transform in terms of time consumption. Larger FOV images (128 $\times$ 256) are unwrapped in up to 8msec, but unwrapping of narrow FOV images is much faster (~3msec). 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 by recon, increase the total time by around 10msec, depending on the options chosen by the user.

Overall, processing large FOV images takes much longer than narrow FOV images. Fortunately, large FOV images need to be processed rarely, as opposed to high frame rate, narrow FOV images. And the realtime implementation of the Multiple FOV MR Fluoroscopy technique can form over 40 frame/second full FOV images, and over 100 frame/second narrow FOV images.

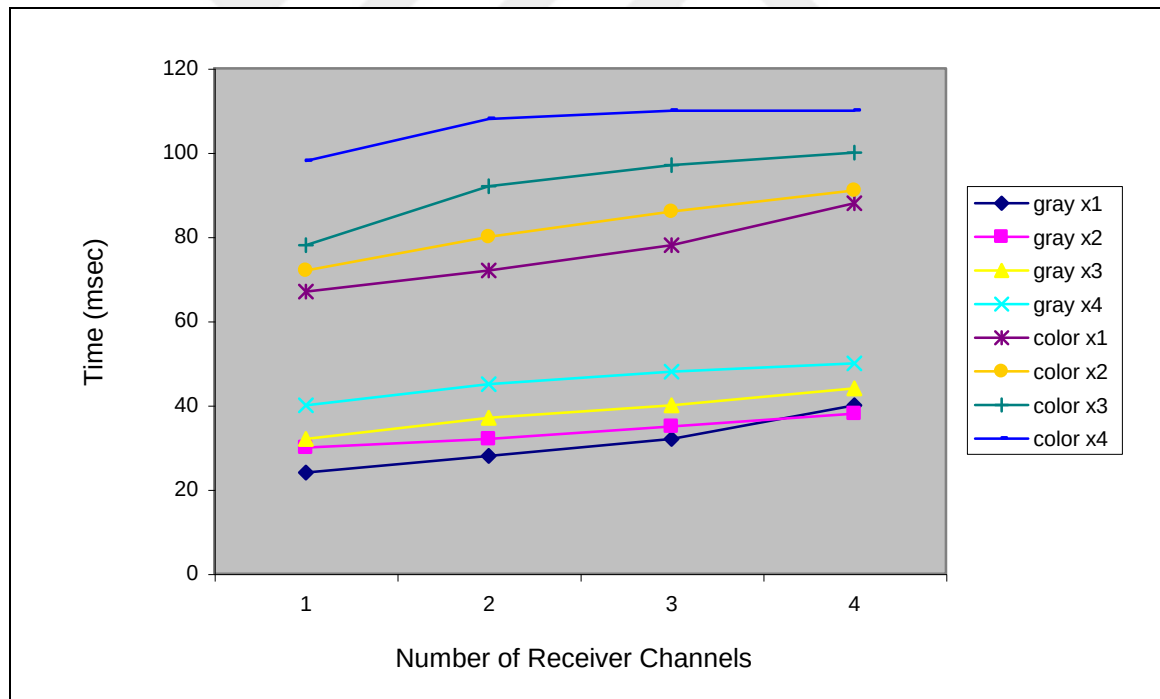


Figure 6.6: Performance of the image display as a function of active channels.
Comparison of color-coded versus grayscale displays, and the effect of the magnification

Display, which combines multiple channel images is currently the slowest performing portion of the realtime code. Its performance is greatly degraded by color coding as shown in Figure 6.6. It can provide frame rates up to 25 frames/second on grayscale images, but color-coding on all four channels combined by a x4 magnification in both directions decreases the achievable frame rate to 8-10 frames/second. Currently, it does not limit the frame rate of our catheter tracking on a dynamic roadmap, however, if a faster pulse sequence is used or all options provided on the user interface are needed simultaneously, further optimization of the code should be pursued.

Finally, note that the entire implementation of Multiple FOV MR Fluoroscopy, from data transfer to display, is interrupt driven, and closely tied to indices of the most recent data or image. If any section of the code fails to meet the timing demands, some frames of the catheter/guidewire or roadmap movies will be not be serviced, in an attempt to provide the latest information. This approach, dropping old frames in favor of newer ones, is more useful for safely guiding interventional procedures.

Chapter 7

Discussion

We have presented a powerful method for increasing the frame rate of interventional device visualization, and providing a dynamic anatomical roadmap image for guidance. The method was implemented online, and the software can perform in real-time at over 50 frames/second for single channel, and over 25 frames/second for multiple channel input. It involves the acquisition of phase-encoded data with the semi-bit-reversed phase order; data transfer from the scanner to a workstation; independent reconstruction and unwrapping of multiple channel images; and display of the combined image with some options.

7.1 Spatial Resolution versus Frame Rate

The achievable frame rate on MR images depends on the desired spatial resolution and FOV, as explained in Section 2.2.3. 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 is viewed with as high a spatial and temporal resolution as possible. The multiple FOV MR Fluoroscopy method achieves high frame rates by decreasing the FOV instead of the spatial resolution.

Currently, most interventional procedures are performed under x-ray guidance. The motion of an interventional device can be clearly viewed due to the very high spatial

resolution (100-200 microns) achievable with x-ray, in real-time (30 frames/second). Despite the lower spatial resolution of MR images (in plane), the slice selection capability may provide better definition of the surrounding anatomy than x-ray, where for instance, in an anterior/posterior view, everything from the ribs to the spine contribute to signal. Blood vessels can be viewed with the injection of contrast agents in both modalities. However, the biggest advantage that MR offers over x-ray is the capability to view high resolution cross sectional images of vessels, providing valuable information that is not available with x-ray (Chapter 1).

Our aim is to provide reliable MR-guidance to interventional procedures although the spatial resolution achievable with MR becomes limited by temporal resolution requirements. Near-microscopic (close to 50 micron) resolution is achievable on MR images that take several minutes to acquire, however, interventional device tracking experiments are performed at spatial resolutions in the order of 1mm. It is not wise to further lower the spatial resolution of MR images for an increased frame rate.

So in our method, the FOV of catheter images is narrowed and real-time temporal resolution is achieved. The combination of high frame rate, intermediate spatial resolution images with detailed anatomical roadmap images, and finally high resolution cross sectional images, which are not available on x-ray, makes MR guidance of interventional procedures a more realistic alternative.

7.2 FOV versus Frame Rate

For a constant spatial resolution, FOV and frame rate are inversely proportional (Section 2.2.3). The reduction in FOV by a factor m , increases the frame rate by m , while resulting in an aliased catheter image. The FOV can be narrowed as long as

unwrapping of the aliased catheter image is possible. The minimum FOV required for successful unwrapping of a catheter image depends on the apparent width of the device, its orientation with respect to the phase-encoding direction (Section 2.2.3), and overall image quality. There are three major factors that degrade the image quality of catheter images and impose a limit on the minimum FOV that can be used: SNR, sensitivity of the catheter, and coupling between multiple receiver coils.

7.2.1 Decrease in SNR:

For images acquired with the same pulse sequence and identical spatial resolution, SNR is dictated by the total readout duration. In other words, SNR is proportional to the square root of the FOV of the image in our case (Section 4.1). Each time the temporal resolution is doubled by halving the FOV of a catheter image, SNR decreases by a factor of $\sqrt{2}$. This decrease in SNR is tolerable due to the very high sensitivity of catheter probes within their immediate vicinity (Atalar 98). For extreme cases, where the FOV is very narrow and the SNR of the image is too low to allow successful unwrapping, an option was provided to reconstruct larger FOV, higher quality images in the background (unwrap_factor), and stabilize the unwrapping algorithm (Section 2.4). So, the issue of SNR drop with narrow FOV images has been handled, and does not pose a serious limitation to the algorithm.

7.2.2 Catheter Sensitivity:

The motivation for pursuing the narrow FOV approach to increase temporal resolution was the narrow sensitivity profile of the catheters and guidewires. On a projection image, the devices appear as a bright line on a dark background. Large FOV images that meet the Nyquist criterion for the entire catheter in order to avoid aliasing,

have large redundant areas of no signal in them (Figure 1.1). With narrow FOV imaging, the Nyquist criterion is not met for the entire catheter, and it is allowed to wrap onto itself. However, the Nyquist criterion is met regionally (for each line along the phase-encoding direction), and the FOV can be narrowed as long as the catheter image does not wrap onto itself. How sharply the sensitivity of the catheter decays determines the apparent width of the catheter, and poses a limit on the minimum FOV that satisfies the Nyquist criterion within each single line of a catheter image.

Ideally, for the Multiple FOV method, the antennae should be used in the transmit/receive mode. In this mode of operation, signal from the antennae is higher than in receive-only mode, and contributions from objects far from the catheter can be minimized, providing the optimal conditions for narrow FOV imaging (Section 2.2). In this case, the sensitivity declines very rapidly with radial distance from the catheter, and there is no corrupting signal beyond about 1cm from the catheter (Figure 2.4). On the other hand, in the receive-only mode, the catheter is sensitive to a much wider area, and these undesirable contributions from objects far from the catheter can corrupt the narrow FOV images (Figure 2.5 provides a comparison of T/R and receive-only images). In the receive-only mode, the sensitivity of the catheter can extend to several centimeters away from the center axis, and minimum FOV required to meet the Nyquist criterion in each line of the image is quite large, limiting the maximum achievable frame rate. The transmit/receive mode of operation can only be used when imaging on a single channel of the MR scanner. The receive-only mode has to be used when multiple channel data is desired. The MRI scanner used is equipped with a single preamplifier for the transmission of RF pulses, and multiple channels can only be used for reception of MR

signal. This stems from the difficulty of defining the desired flip angle distribution when multiple coils (surface coil and/or body coil as well as catheter) are transmitting RF power simultaneously. A temporary fix for this limitation is to use a still roadmap image with the high temporal resolution catheter movie from a single channel of the scanner. The more desirable approach, however, is to use a faster pulse sequence, where the FOV of the catheter images does not need to be very narrow to obtain a 25-30 frame/second movie, and will be addressed in Section 7.3.

7.2.3 Coupling between multiple receiver coils:

As explained in the previous section, for narrow FOV imaging, we would only like to see the immediate vicinity of the catheter on our images. The frame rate can then be increased by reducing the FOV of the catheter image. When there is coupling between the different coils (surface/body) and interventional devices (catheter/guidewire), the catheter image may contain some information from all of the other coils as well. For instance; $catheter_channel = catheter_image + [(surface_coil_image) * coupling_coefficient]$.

From the narrow FOV imaging perspective, this is very similar to the case where the catheter has a wide sensitivity profile, and picks up signal from objects far from it. The minimum FOV to avoid aliasing is much larger in this case, since we need not only worry about aliasing of signal from the actual catheter, but also of signal from other receiver channels. Decoupling the transmit (body) and receive (surface/ body/ catheter/ guidewire) coils is accomplished by the insertion of a PIN diode shunt to the coaxial cables that transmit the MR signals to the preamplifier. Incorporating isolation circuits in all the RF coils used may improve performance by minimizing the coupling between them. Another approach may be to subtract out other channels' contributions from the

catheter images, but is not practical due to the wide range of coupling coefficient (complex) values possible.

7.2.4 Flashing

If the FOV of the catheter images is reduced despite corruption from coupling with other channels, or a wide sensitivity profile, a condition called “flashing” occurs in the resulting catheter movie.

The phase order in the Multiple FOV MR Fluoroscopy method is such that small linear phase-encode lines are acquired interleaved with one another. For instance, if 40 phase-encode lines are used for an image, the next image will be reconstructed from the next group of 40 phase-encode lines, which is shifted in the spatial frequency domain, with respect to the previous group. The amount by which the k-space lines are shifted is not constant, but varies in order to create a uniformly sampled k-space for large FOV images. Shifting the acquired k-space lines in the phase-encoding direction, corresponds to adding linear phase that changes along the phase-encoding direction to the image:

$$f(k_x, k_y - k_{y_0}) \Leftrightarrow F(x, y) e^{-i 2\pi k_{y_0} y}$$

In the normal case, with a narrow sensitivity profile catheter image, narrow FOV imaging results in aliasing, which can be corrected by unwrapping. There is a linear phase on each frame, which changes due to the shifting of k-space lines in the spatial frequency domain. Since there are no contributions to the catheter image from objects far away (zero), this amounts to the addition of zero (with a different phase) to the catheter images.

However, if a catheter has a wide sensitivity and picks up a wide area, or there is significant coupling between the catheter and other receiver coils, the FOV should not be

decreased beyond the point where these “other” components in the image start to cause aliasing. One may be tempted to decrease the FOV as long as unwrapping of these corrupt images is still successful, however, the effect is more drastic.

The Nyquist criterion will not even be met regionally, and objects far from the catheter with a certain amount of phase, will be added to (aliased on) the catheter image. The amount of phase on these other objects is different in each frame, since a set of k-space lines shifted by a different amount are used to reconstruct successive images. On each frame, a catheter and the effects of aliasing from other objects (with a certain phase) will be visible. Each individual frame may be acceptable to the user, however, successive frames will all have a catheter onto which the other objects with different phases have been aliased. Sometimes, the phase is such that the aliasing is additive, sometimes negative, but it constantly varies from frame to frame. When a movie of such successive frames is viewed this is visible as a flickering in the background of the catheter, or “flashing”. In summary, if the catheter picks up objects from far away (either due to wide range of sensitivity, or coupling to other receive coils) the maximum frame rate achievable becomes limited by this “flashing”, even if the image can be successfully unwrapped. The contributions from objects distant to the catheter usually have much lower intensity than the signal from the catheter, and unwrapping is possible, however, “flashing/flickering” is very distracting and should be avoided by increasing the FOV.

Increasing the FOV to avoid flashing, decreases the frame rate considerably. Another approach to eliminating flashing is similar to the UNFOLD technique described by Madore (Madore 99). The flashing and the catheter motion occupy different sections of the temporal frequency spectrum. Flashing is a very high frequency phenomenon, and

changes are quite drastic between the frames while the catheter data changes much slower, and occupies lower temporal frequencies. An appropriate filter can be experimentally determined, that successfully distinguishes these two components, and discards the effects of flashing. This is essentially low pass filtering, and slightly smearing consecutive images over time. A certain percentage of the temporal frequencies will be allocated to catheter data while the rest is discarded. This operation will also decrease the frame rate depending on the percentage of data discarded. A 10-20% decrease in frame rate by UNFOLD'ing is more acceptable than a 50% decrease each time the FOV is doubled, to avoid aliasing. Using a faster pulse sequence will also eliminate flashing by allowing the use of larger FOV catheter images, and still provide a very high frame rate as shown in the next section.

7.3 Pulse Sequence Optimization

Most of the problems faced by the Multiple FOV MR Fluoroscopy method are the limitations on the minimum FOV that is sufficient for unwrapping, and that eliminates flashing. Our goal is to display high frame (25 frames/second) catheter images on a dynamic anatomical roadmap image.

Currently, frame rates up to 25 frames/second are achievable with only very narrow FOV imaging, which makes unwrapping more challenging, and causes additional problems due to flashing. For instance, for a 256x256 data matrix, 32 cm full FOV, and a typical TR of 5msec, the FOV must be reduced to 1cm for a 25 frame/second catheter movie.

Fortunately, there is a lot of room for improvement of the type of pulse sequence used. Simply eliminating the redundancies, features not used by catheter tracking, in the

fast gradient echo pulse sequence (FGRE), or transferring our pulse sequence modifications to a more efficient sequence such as the interleaved echo planar (EPI) will allow repetition times in the range of 1-1.5msec. Therefore, the same MRF technique applied to these sequences reduces the effective acquisition time 3-4 fold. This decrease in acquisition time will permit a 3-4 fold increase in FOV on catheter and guidewire channels, without a penalty in temporal resolution; and a 3-4 fold increase in temporal resolution of the roadmap channel(s), without compromising spatial resolution. For instance for the same parameters mentioned above (256x256 data matrix, 32 cm FOV), a 25 frame/second catheter movie can be obtained by reducing the FOV to 4cm (instead of 1cm), and the full FOV roadmap images will have a frame rate of 3 frames/second (instead of 0.78 frames/second).

7.4 Computation Speed

The timing parameters for data transfer, reconstruction and display are given in Chapter 6, Results. The data transfer fails to update around 0.2% of the acquired sets immediately after their acquisition. Reconstruction can keep up with over 50 frames/second, and four channel data can be displayed at 25 frames/second (no color coding) to 12 frames/second (color coding on each channel, and maximum magnification). Under current conditions, the performance of the software poses no limitations on the frame rate of catheter guidance on a dynamic roadmap, however, optimization of the display is necessary to allow utilization of all available options on the graphical user interface simultaneously.

Chapter 8

Conclusion

We have combined an innovative data acquisition strategy with a real-time reconstruction and display that allows visualization of multiple interventional devices, simultaneously with a dynamic roadmap image, providing continuous online feedback to the interventionist. The advantages are numerous:

The loopless antenna design incorporated into the guidewire or catheter results in MR signal originating from the entire length of the device. Compared to active tip tracking methods visualizing up to four points on the tip of a catheter (Ackermann 86, Dumoulin 93) this increases the potential clinical usefulness of MR guidance, and allows ensuring that there is no buckling or folding of the device. Furthermore, slice selection or adjustments to the imaging slice are not necessary since the entire length of the catheter is visible at all times on a projection image, and there are no obstacles to the catheter view.

By combining the strengths of the loopless antenna design with a continuous updated roadmap image, all registration problems are eliminated that may result from motion during the procedure, and the need to periodically reacquire new roadmap images is eliminated. Compared to some passive tracking techniques that collect anatomical and interventional device data together (Kochli 94, Lenz 96, Bakker 96), in our method there is no compromise between the spatial resolution of the roadmap view and the temporal

resolution of the interventional device views. The reconstruction of each channel data is independent of the other channels, and the field of view can be controlled by the user to achieve the best frame rate, according to a tolerable amount of reduction in the FOV.

The reconstruction of each channel runs as a separate process, and the user interface allows independent manipulation of each channel view during certain stages of a 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, and the color-coding proves to be very useful in distinguishing between the connected devices.

The limitation of the reduced FOV imaging is that if certain portions of the catheter align with the phase encoding direction, more data will be needed for successful unwrapping. The maximum allowable angle that the catheter may be from the readout direction depends on its apparent width (w), and the FOV in the phase-encode direction. If this angle exceeds $\cos^{-1}(w/\text{FOV})$, the catheter image wraps onto itself and the FOV needs to be increased, resulting in a lower temporal resolution. This problem can be solved by implementing variable FOV's within an image. This allows reducing the frame rate only on problem sections of the image without effecting the frame rate of the other parts of an image, and eliminates the need to slow down the entire movie.

The next step to be taken in this study is to optimize the type of pulse sequence used. Using a faster sequence such as the interleaved echo planar, the effective TR for data acquisition can be considerably lowered. This 3 to 4 fold decrease in acquisition time can be used to increase the temporal resolution of the roadmap channel, and increase

the FOV on the catheter or guidewire channels, so that catheter alignment in the phase-encode direction will be much less of a problem.

In conclusion, we have implemented a new data acquisition and reconstruction technique for guiding MR catheter placement. Multiple-FOV MR Fluoroscopy allows for a variable frame rate within one movie, 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/sec, $1/32^{\text{nd}}$ of acquired), as well as the surrounding anatomy with fairly frequent updates and large FOV (0.78 to 1.56 frames/sec, full or half FOV). This eliminates an important obstacle for interventional MRI by providing a robust and reliable tool for device tracking and visualization. Combined with high resolution interventional MRI, this technique may aid in diagnosis of vessel disease, and monitoring of intravascular procedures.

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Aselsan Military Electronics Industry, Antalya, Turkey

- **Technical Intern** 1995, 1998
Collaborated with engineers in upgrading transceiver models.

patent/invention

Multiple FOV MR Fluoroscopy,
E. Atalar, P. Aksit, A. J. Derbyshire,
Provisional Patent Application 60/193,401 filed March 29, 2000.

publications

Multiple FOV MR Fluoroscopy,
P.Aksit, J.A.Derbyshire, J.Serfaty, E.Atalar.
Submitted to MRM.

Simultaneous Acquisition of Multiple FOV Images For Real Time Catheter Tracking, P.Aksit, J.A.Derbyshire, E.Atalar.
ISMRM 8th Scientific Meeting, Denver 2000

Real Time Catheter Visualization Using Simultaneous Fluoroscopic MR Imaging on Multiple Channels at Multiple Fields of View,
P.Aksit, J.A.Derbyshire, J.Serfaty, E.Atalar.

World Congress on Medical Physics and BME, Chicago 2000

Real Time Catheter Tracking on a Dynamic Roadmap,
P.Aksit, J.A.Derbyshire, J.Serfaty, E.Atalar.
Submitted to the RSNA 86th Scientific Assembly and Annual Meeting.

Toward MRI-Guided Coronary Catheterization: Visualization of Guiding Catheters, Guidewires, and Anatomy in Real Time.
J.Serfaty, X.Yang, P.Aksit, H.H.Quick, M.Soleiyappan, E.Atalar.
Submitted to JMRI

MRI Guided Coronary Interventions,
J.Serfaty, X.Yang, H.H.Quick, P.Aksit, J.A.Derbyshire, E.Atalar.
Submitted to AHA for the Young Investigator's Award

Toward Coronary Interventions,
J.Serfaty, X.Yang, P.Aksit, M.Soleiyappan, E.Atalar,
Submitted to the RSNA 86th Scientific Assembly and Annual Meeting

honors

Turkish Student Association, Vice President, Purdue University (1997 and 1998)
Turkish Ministry of Education Scholarship, coverage of education expenses
International Society of Magnetic Resonance in Medicine, student member
ETA KAPPA NU, Electrical Engineering Honor Society
PHI SIGMA RHO, Sorority for Women in Engineering and Technology
Academic Honors and Dean's List

skills

Computer:

Unix/C, multi-process, realtime & event driven software,
GUIs X Window System and Xview toolkits
Turbo Pascal
Matlab
MS Office Apps: Word, Excel and Powerpoint

Language:

Native speaker of Turkish, limited German.
Strong verbal, written, and interpersonal skills.