The use of susceptibility effects for tracking and encoding in MR-guided interventional procedures

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A thesis submitted in conformity with the requirements for the degree of Doctor of Philosophy
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University of Toronto

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Doctor of Philosophy, 2016
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Abstract

Magnetic Resonance Imaging (MRI) provides volumetric imaging with superior soft tissue contrast and is now standard-of-care for many diagnostic applications. A growing area is the use of MRI for minimally invasive procedures (a.k.a interventions) such as MRI-guided biopsy. In this thesis, the target procedure was placement of a drainage catheter in the upper urinary tract, the technical term for which is nephrostomy. Tracking and visualization of the interventional devices is crucial in achieving a successful outcome of the procedure. This thesis is a study of MRI methodologies to track interventional devices in MRI-guided procedures. A particular focus was the application of magnetic materials used to perturb the main magnetic field, an effect that was used to develop tools and techniques for tracking devices and making images at the device tip.

The experiments in Chapter 2 led to the development of a new technique using spectrally selective excitation, which is capable of simultaneously measuring the location and orientation of an interventional device. In Chapter 3, a new mechanism for tracking is proposed based on the magneto-optical effect in nickel nanoparticles, where fiber optics were used to deliver photons from a near infrared (NIR) laser. This development is intended to enable the incorporation of small tracking devices attached to the interventional tools, which can be pulsed on and off using a laser waveform. In Chapter 4, magnetic materials have been used to generate spatial encoding fields at the tip of a device, to generate endoscopic MR images.
These results have contributed to the development of tracking devices and to the advancement of MRI acquisition methods for passive tracking in MR-guided procedures, with an outlook towards clinical application in nephrostomy and beyond.
I would like to express my special appreciation and thanks to my advisor Dr. Charles H. Cunningham, you have been a tremendous mentor for me. I would like to thank you for encouraging my research and for allowing me to grow as a research scientist. Your advice on both research as well as on my career have been priceless. I would also like to thank my committee members, Dr. Graham Wright, Dr. Donald Plewes for serving as my committee members even at hardship. I also want to thank you for letting my defense be an enjoyable moment, and for your brilliant comments and suggestions, thanks to you.

A special thanks to my family. Words cannot express how grateful I am to my parents and siblings for all of the sacrifices that you’ve made on my behalf. I would also like to thank all of my friends who supported me in past few years.
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## Abbreviations

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<th>Description</th>
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<td>BW</td>
<td>Bandwidth</td>
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<tr>
<td>CC</td>
<td>Cross-correlation</td>
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<tr>
<td>CT</td>
<td>Computed tomography</td>
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<tr>
<td>FOV</td>
<td>Field of view</td>
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<tr>
<td>FA</td>
<td>Flip angle</td>
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<tr>
<td>FWHM</td>
<td>Full width of half maximum</td>
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<td>GRE</td>
<td>Gradient recalled echo</td>
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<tr>
<td>IVUS</td>
<td>Intravascular Ultrasound</td>
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<td>MRI</td>
<td>Magnetic resonance imaging</td>
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<td>MT</td>
<td>Magnetization transfer</td>
</tr>
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<td>MS</td>
<td>Magnetic susceptibility</td>
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<td>NIR</td>
<td>Near infrared</td>
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<tr>
<td>NEX</td>
<td>Number of excitations</td>
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<td>NMR</td>
<td>Nuclear magnetic resonance</td>
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<td>OCT</td>
<td>Optical coherence tomography</td>
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<tr>
<td>PNR</td>
<td>Peak-to-noise ratio</td>
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<tr>
<td>PRF</td>
<td>Proton resonance frequency</td>
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<tr>
<td>PCN</td>
<td>Percutaneous nephrostomy</td>
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<tr>
<td>SNR</td>
<td>Signal-to-noise ratio</td>
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<tr>
<td>SRF</td>
<td>Spatial response function</td>
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<td>SEM</td>
<td>Spatial encoding magnetic fields</td>
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<td>SSFP</td>
<td>Steady-state free precession</td>
</tr>
<tr>
<td>TR</td>
<td>Time of repetition</td>
</tr>
<tr>
<td>T/R</td>
<td>Transmit and receive</td>
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<tr>
<td>TE</td>
<td>Time of echo</td>
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x
Chapter 1

Background

1.1 Introduction and Motivation

Image-guided interventions provide a minimally invasive portal access to internal organs under different imaging modalities, such as fluoroscopy, Magnetic Resonance Imaging (MRI), computed tomography (CT), and ultrasound (US) [1].

Accurate tracking of interventional tools plays a major role in the successful outcome of image-guided procedures, especially for percutaneous nephrostomy, which is one of the most common image-guided interventions. Nephrostomy is a surgical procedure where a catheter is placed in the upper urinary tract to relieve a blockage and to deliver therapeutics.

Various imaging modalities have been used to guide the interventional tools in nephrostomy, including bedside ultrasound, X-ray fluoroscopy, and computed tomography. However, there are several cases where MRI has been proven to be additionally beneficial [2–4].

The research objective of this thesis is to develop interventional tools and techniques for improving the feasibility of MRI-guided nephrostomy procedures. Accurate tracking using low-cost devices and limited hardware complexities is essential in achieving this end.
Magnetic resonance imaging has been established as a valuable tool in the diagnosis and treatment of diseases. Because of the soft-tissue contrast and the spatial resolution provided by this modality, as well as its capability of imaging multiple planes in arbitrary orientations, MRI is often described as showing exquisite anatomical detail. These images are routinely obtained in standard applications of MRI, which include studies of the central nervous system and the musculoskeletal system.

For this reason, MRI-guided interventions have been recognized as a promising technology for improving current diagnostic and surgical standards of care, and thus, these interventions have become the subject of clinical research over the past decade. In addition to providing anatomical images with superior soft-tissue contrast, MRI is capable of rapidly tracking interventional tools during nephrostomy procedures without the ionizing radiation associated with the conventional fluoroscopy [5]. These properties render MRI particularly advantageous compared to other imaging modalities.

1.2 Physical basis for MRI

The MR signal comes from hydrogen nuclei, which possess a spin angular momentum ($\vec{J}$). Eq. 1.1 shows the relationship between the spin angular momentum and the magnetic dipole moment ($\vec{p}$),

$$\vec{p} = \gamma \vec{J}$$

(1.1)

where $\gamma$ is the gyromagnetic ratio of the nuclei (e.g., $\gamma_{\text{proton}} = 4258 \text{Hz/G}$). Once subjected to an external magnetic field, a torque will be applied to the dipole moment, aligning it with the direction of the main field. The summation of the magnetic dipole moments gives rise to a net magnetic moment, referred to as net magnetization ($\vec{M} = \sum \vec{p}$) per unit volume.

If an external field exerts a torque on the net magnetization, tipping it away from the direction of the main field, it will precess around the main field (longitudinal axis) in

1The explanations here are adopted from classical nuclear magnetic resonance (NMR) phenomena, which do not include the quantum aspects of NMR theory.
CHAPTER 1. BACKGROUND

A motion similar to the rotation of a top. The frequency at which the net magnetization precesses around the external field is called the Larmor frequency. Eq. 1.2 shows the precessional frequency as a function of the external magnetic field,

$$\omega = \gamma B_0$$

where $B_0$ is the strength of the external magnetic field.

A radiofrequency (RF) pulse is then applied in order to rotate the magnetization into the transverse plane. The components of the magnetization in the transverse plane and the remaining portion in the longitudinal axis are called the transverse magnetization and the longitudinal magnetization, respectively.

Once the RF pulse is turned off, the transverse magnetization will precess around the external magnetic field. During this precession, the transverse magnetization will decay to zero with a time constant called $T_2^*$, giving rise to MR signals; simultaneously, the longitudinal magnetization will return to its equilibrium state with a time constant called $T_1$. The time constants, along with the density of the hydrogen atoms, play a crucial role in generating the contrast in MR images because different tissues have different MR properties. The spatial location of the magnetizations from different locations in the body/object can be encoded within the MR signals prior to, or during, the recording of the signals. Section 1.3 will explain these spatial-encoding techniques in detail.

1.2.1 Susceptibility

Similar to the intrinsic angular momentum of protons, electrons have an intrinsic spin. If there are unpaired electrons in an atom when subjected to an external magnetic field, the generated magnetic moment is much larger than the nuclear magnetic moment. Magnetic susceptibility (MS) ($\chi$) refers to the degree of interaction between a material and the external magnetic field. MS includes contributions from electrons as well as a relatively insignificant contribution from the nuclear magnetization (mentioned in the previous section). Different materials exhibit different interactions with the external field, based
on their atomic structure. The internal magnetic flux \( B \) is given by Eq. 1.3 as a function of the external field \( H \) and the induced magnetic moment in the material \( M \),

\[
B = \mu_0(H + M)
\]

where \( \mu_0 \) is the magnetic permeability of free space. For linear materials, the magnetization \( M \) is proportional to the applied magnetic field \( H \) using Eq. 1.4.

\[
M = \chi H
\]

The susceptibility \( \chi \) in Eq. 1.4 can be negative \( (\chi < 0) \), positive \( (\chi > 0) \), or zero \( (\chi = 0) \) for diamagnetic, paramagnetic, and non-magnetic materials, respectively. For ferromagnetic materials, the susceptibility value is much larger than 1 \( (\chi >> 1) \). Susceptibility is often expressed in parts per million (ppm). For example, the susceptibility of water is \( \chi_{\text{water}} = -9.05 \times 10^{-6} \) or \(-9.05 \) ppm [6].

Maxwell's equations can be solved to calculate the external field changes due to the presence of magnetic materials. The implication of this calculation is that the magnetic field adjacent to the material will deviate from the value of the main applied field. Considering Eq. 1.4, an opposing magnetization will be induced in diamagnetic materials, which will reduce the surrounding magnetic field. Conversely, paramagnetic materials will increase the surrounding magnetic field.

Equation 1.5 shows the spatial distribution of the magnetic field deviation outside of a sphere with susceptibility shifted \( \Delta \chi \) from the surrounding material in an external magnetic field \( B_0 \).

\[
\delta B_d(r) = \lim_{a \to 0} \left\{ \frac{\Delta \chi B_0}{3} \left( \frac{a}{r} \right)^3 (3 \cos^2 \phi - 1) \right\}
\]

where \( \phi \) is the azimuthal angle in cylindrical coordinates, \( a \) is the radius of the sphere, and \( r \) is the spatial location with respect to physical axes. Figure 1.1 visualizes the magnetic field deviations that are caused by the presence of a sphere made with a nickel ball with a radius of 5 mm and a magnetic susceptibility of 600 ppm in an external 1.5T magnetic field along the Z axis. The surrounding material was assumed to be water with
a magnetic susceptibility equal to -9.05 ppm [6]. The colour represents the frequency offsets from the on-resonance frequency.

Figure 1.1: The field changes due to a sphere (Δχ = 600 ppm) with a radius of \( a = 5\,\text{mm} \) in a 1.5T external magnetic field. The colour represents the frequency offsets from the on-resonance frequency. The contour interval is 5 Hz.

The susceptibility of human tissues is usually small (\( \|\chi\| \ll 1 \)). Most tissues exhibit diamagnetic properties [6].

**Ferromagnetic materials**

Ferromagnetic materials are materials with high susceptibility values (\( \chi \gg 1 \)). They have permanent domains of aligned spin angular momentum, which can be several microns in size [6]. Spins in these domains are aligned with each other, thus producing a very strong macroscopic magnetization in the domain that exists without an external field. In equilibrium, these domains are randomly oriented, thus producing zero net magnetization. Once placed in an external magnetic field, they will align with the external field and produce a large magnetization along this direction. Residual magnetization will remain upon removal of the external field, due to the hysteresis effect.

Ferromagnetic materials can be categorized into soft and hard materials. Soft materials refer to materials that have weak saturation magnetization and occupy a small area inside the hysteresis curve. Hard materials have strong saturation magnetization
and occupy a large area inside the hysteresis curve, leading to strong remnant magnetization after the removal of the external field.

The large magnetization generated by the external field of ferromagnetic materials can result in substantial applied forces that can be a safety issue, especially in metallic implants that are not MR compatible.

The susceptibility of ferromagnetic materials depends on their temperature. The Curie temperature is the temperature at which a ferromagnetic material becomes paramagnetic. In moving towards this temperature, the magnetic domains become increasingly smaller, to the extent that an external field is aligned with a smaller number of the spins, thus generating a much smaller magnetization [7].

1.3 Signal acquisition and image formation

1.3.1 Selective excitation

In order to acquire a signal, the longitudinal magnetization has to be tipped into the transverse plane using an RF pulse. If the RF pulse is tuned to the Larmor frequency of the nuclei that are being imaged, a resonance phenomenon will occur where the $\vec{B}_1$ field is stationary with respect to the precessing spins, thus tipping the magnetization into the transverse plane. If the magnetization is denoted by $\vec{M}$ in the associated rotating frame, the rotation of the magnetization due to the $\vec{B}_1$ is calculated using a Bloch equation shown in Eq. 1.6.

$$\frac{d\vec{M}}{dt} = \gamma \vec{M} \times \vec{B}_1 \quad (1.6)$$

Under Eq. 1.6, the magnetization will rotate in a plane perpendicular to the $\vec{B}_1$ field (shown in Figure 1.2a). However, if the RF pulse is tuned to a frequency different from the precessional frequency of the magnetization (by $\Delta\omega$), then the magnetization will rotate about a $\vec{B}_\text{eff}$ (shown in Figure 1.2b).

Every RF pulse has a certain tip angle profile with respect to frequency (as shown in the $M_{xy}$-frequency plots in Figure 1.3). Magnetizations precessing with frequencies
outside the sensitive region of the RF pulse will not be affected by the RF pulse. In the presence of a linearly varying magnetic field (gradient), magnetizations will precess with spatially dependent frequencies according to Eq. 1.2. Therefore, some magnetization will be placed in the sensitive range of the RF pulse and excited while others are shifted outside the range, thus not excited. As a result, a slab of magnetization at a known spatial location can be excited. This phenomenon is referred to as spatially-selective excitation (shown in Figure 1.3a). In Figure 1.3a, the magnetic field gradient along the X axis is used to generate linearly varying magnetic fields in order to only excite a range of magnetization in space.

However, when spatially selectivity is not required (e.g., spectroscopy or fat suppression), linear gradients are not used. If the RF pulse is tuned to a specific range of
Figure 1.3: Spatially selective (a) and spectrally selective excitation (b). In (b), the magnetization precessing at frequencies between dashed lines will be excited by the RF pulse.

frequencies (in the absence of gradients), only the magnetizations that are precessing in that range will be excited. This process is referred to as spectrally selective excitation. An example of spectrally selective excitation is shown in Figure 1.3b. Here, the magnetization precessing at frequencies between the dashed lines will be excited by the RF pulse.

1.3.2 Fourier encoding

In magnetic resonance imaging, the acquired signal is a summation of small signals arising from individual spins at different locations. Without employing a signal-localization technique, the recorded signal will reveal no information about the location of individual spins within the object. Therefore, it is necessary to develop a technique to encode spatial information into the signal. Once the signal is recorded by a receiver coil, a mathematical transformation can be performed on the signal to decode this spatial information.
Conventionally, spatial encoding has been performed by employing gradient coils that are mounted on the scanner bore. These coils generate magnetic fields that vary linearly along each of the three spatial axes of the laboratory frame (X, Y, and Z), as shown in Figure 1.4 for the X and Y axes. The local spatial distribution of the spins can be determined from the frequency and phase contents of the recorded MR signal. Eq. 1.7 shows the relationship between the precession frequency ($\omega$) of a spin and its location (in this case, the X axis) while the gradient field is switched on. $\omega_0$ is the Larmor frequency at $B_0$, $\gamma$ is the proton gyromagnetic ratio, $G$ is the amplitude of the applied gradient field, and $x$ is the spatial location along the X axis.

$$\omega = \omega_0 + \gamma Gx$$  \hspace{1cm} (1.7)

After selecting a slice using an appropriate RF pulse and slice-selective gradients, the spatial information within the slice must be encoded in the MR signals. In-plane spatial encoding is conventionally performed by using the gradient fields along two orthogonal axes, known as phase-encoding and frequency-encoding directions. Frequency- and phase-encoding directions are defined based on the imaging plane (sagittal, coronal, axial, and oblique). Figures 1.4a and 1.4b show phase-encoding and frequency-encoding gradients along y and x directions, respectively, for an axial slice.

The MRI signal will then be a summation of the magnetization using Eq. 1.8a,

$$s(t, \vec{k}(t)) = \int M_{\perp}(\vec{r}) e^{-i2\pi \vec{k}(t) \cdot \vec{r}} d^3r$$  \hspace{1cm} (1.8a)

$$k_x = \frac{\gamma}{2\pi} \int_0^t G_x dt \quad k_y = \frac{\gamma}{2\pi} \int_0^t G_y dt \quad k_z = \frac{\gamma}{2\pi} \int_0^t G_z dt$$  \hspace{1cm} (1.8b)

where $M_{\perp}(\vec{r})$ is the transverse magnetization at location $\vec{r}$, and $\vec{k}(t) = (k_x(t), k_y(t), k_z(t))$ is the spatial frequency being sampled at time $t$. In Eq. 1.8b, $G_x$, $G_y$, and $G_z$ are the strengths of the magnetic field gradients along the corresponding physical axis. The sampled signal is the value at a point $k_x(t)$, $k_y(t)$, and $k_z(t)$ in a domain called $k$-space. In comparing Eq. 1.8a with the Fourier transform definition, the $k$-space is the Fourier transform of the object in the image domain, and $s(t, \vec{k}(t))$ is the coefficient for the Fourier basis. Therefore, the image is the inverse Fourier transform of the acquired $k$-space.
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Figure 1.4: Spatial encoding gradient fields: (a) phase-encoding gradient along \( y \) axis (b) frequency-encoding gradient along \( x \) axis. \( \omega \) and \( \phi \) represent the precessional frequency and accumulated phase under the magnetic gradients, respectively.

In simple slice-selective 2DFT imaging (e.g., gradient recalled echo, aka GRE), each line of \( k \)-space is acquired in one repetition time (TR). This process is referred to as the \( k \)-space trajectory. When all the lines are acquired, the image will be reconstructed by taking the inverse Fourier transform. Figure 1.5a illustrates the GRE pulse sequence, and Figure 1.5b shows the \( k \)-space trajectory of this pulse sequence.

In order to acquire images with different contrasts, many pulse sequences have been designed with different \( k \)-space trajectories, acquisition timing (TR and TE), and magnetization manipulation blocks (such as fat saturation).

In Eq. 1.8a, it was assumed that any change in precession frequency was due to the linear gradients applied for spatial encoding. However, magnetic-field uniformity in MRI scanners will be perturbed as a result of the presence of magnetic material in the main field. This field perturbation results in additional accumulated phase. Eq. 1.9 shows the additional phase term.

\[
s(t, \vec{k}(t)) = \int M_{\perp}(\vec{r}) e^{-i2\pi \vec{k}(t),\vec{r}} e^{-i\phi(\vec{r},t)} d^3r
\]

(1.9)
where $\phi(\vec{r}, t)$ is the accumulated phase due to the off-resonance at location $\vec{r}$ at sampling time $t$. The effect of the magnetic-field perturbations in MRI images is called susceptibility artifacts, which often manifest themselves as signal losses and geometric distortions in some MRI images due to the $\phi(\vec{r}, t)$ term.

### 1.3.3 Nonlinear encoding

In conventional clinical MRI applications, spatial encoding has been done using linearly varying magnetic gradients (linear spatial encoding fields). However, in recent years, non-linear spatial encoding fields have gained more attention as they can provide some interesting features such as tailored field-of-view (FOV) for a target application, reduced peripheral nerve stimulation (PNS), and standalone MRI probes [8–11].

The general form of the MRI signal is given in Eq. 1.10. In this equation, $s(t, \phi(t))$ is the signal value at time $t$, and $\phi(\vec{r}, t)$ is the phase accumulation due to the spatial encoding fields, the off-resonance sources, and the initial phase term due to excitation.

$$s(t, \phi(t)) = \int M_{\perp}(\vec{r})e^{-i\phi(\vec{r}, t)}d^3r$$ (1.10)

The implication of the linear gradient is that the term $\phi(\vec{r}, t)$ is simplified to $2\pi\vec{k}(t)\cdot\vec{r}$, which is a Fourier basis. Therefore, the sampled signals will be the Fourier coefficients.
CHAPTER 1. BACKGROUND

However, for non-linear spatial encoding fields, $\phi(\vec{r}, t)$ will not be a Fourier basis. Therefore, the spatial encoding requires a new regime of image reconstruction. Eq. 1.11a shows the discretized version of Eq. 1.10.

$$\tilde{s}_i(t_j) = E_{ij}\tilde{M}$$  \hspace{1cm} (1.11a)

$$E_{ij} = \begin{bmatrix} e^{-i\phi_i(\vec{r}_1,t_j)} & e^{-i\phi_i(\vec{r}_2,t_j)} & \cdots & e^{-i\phi_i(\vec{r}_{MN},t_j)} \end{bmatrix}_{1 \times MN}$$  \hspace{1cm} (1.11b)

$$\tilde{M} = \begin{bmatrix} \tilde{M}(\vec{r}_1) \\ \tilde{M}(\vec{r}_2) \\ \vdots \\ \tilde{M}(\vec{r}_{MN}) \end{bmatrix}_{MN \times 1}$$  \hspace{1cm} (1.11c)

$$\tilde{s} = E\tilde{M}$$  \hspace{1cm} (1.11d)

where $\tilde{s}_i(t_j)$ is the sampled signal at time $t_j$, $E_{ij}$ is a row of encoding matrices ($E$) corresponding to the applied encoding fields at $t_j$, and $\tilde{M}$ is a matrix of discretized magnetizations called voxels. Therefore, the reconstruction technique is equivalent to estimating matrix $\tilde{M}$ using a system of linear equations established in Eq. 1.11d. A direct approach involves calculating the inverse of the encoding matrix (decoding matrix $F$) and estimating $\tilde{M}$ using Eq. 1.12a[8].

$$\tilde{M} = F\tilde{s}$$  \hspace{1cm} (1.12a)

$$F = E^{-1}$$  \hspace{1cm} (1.12b)

This is generally a time-consuming process on the order of $N(O^3)$ that requires a lot of memory. If the condition number of the encoding matrix is large, the direct inversion will amplify the noise. An iterative solution has been used to solve these systems of linear equations. A more detailed explanation of how to solve these equations by directly calculating the decoding matrix ($F$) will be presented in Chapter 4.

1.4 Image-guided interventions

Minimally invasive procedures, specifically catheter-based procedures, such as angioplasty, stent placements, and nephrostomy, play an increasingly important role in med-
ical interventions. Their minimally invasive nature reduces patient recovery time [1]. However, these interventions are often not only complex but also require lengthy procedures due to the fact that the interventional devices have to be manoeuvred in small places such as the vasculature and the ureter. Consequently, tracking and visualizing these devices will lead to a significant improvement in the outcome of the interventions [1, 12].

In fact, the use of imaging modalities has already significantly improved the outcome of these interventions because these imaging methods enable the visualization and tracking of the devices with respect to the anatomy. X-ray is currently the modality of choice for many image-guided interventions. However, X-rays also have some disadvantages. For instance, X-ray imaging cannot offer volumetric imaging or soft-tissue contrast, and it involves ionizing radiation. As a result of these disadvantages, CT and MRI have recently been proven to be exquisite imaging modalities because they provide volumetric images with soft-tissue contrast, thus minimizing intra-procedure complications by visualizing the device with respect to the anatomy in three dimensions [1, 12].

1.4.1 The role of MRI in image-guided interventions

X-ray and CT imaging during interventions often require the injection of iodinated contrast agents to visualize the device. In addition, lengthy procedures expose both the interventionist and the patient to elevated radiation doses. In contrast with these modalities, MRI has no harmful radiation and often requires no or limited use of contrast agents [12]. The current limitation of MRI in comparison to X-rays and CT, however, is its lower spatial and temporal resolution.

Tracking interventional tools in MR-guided procedures has been done using various methods, which can be divided into active and passive techniques [13].
CHAPTER 1. BACKGROUND

1.4.2 Active tracking

In active tracking, a micro-coil is placed at the distal end of the interventional tool to record signals in its vicinity, thus providing positional information [14, 15]. Multiple micro-coils are used to simultaneously measure the position and orientation of the tool [14]. Some active tracking techniques use the entire guide wire as an antenna to receive MR signals. This results in high-intensity regions around the guide wire [16].

Active tracking techniques provide superior temporal resolution, which can be beneficial when tracking in fast-moving organs such as coronary arteries [12]. Automatic slice-repositioning is also possible with active tracking techniques [17]. However, these techniques are associated with both hardware complexities and an increased risk of unwanted RF heating from the presence of long conductive wires in the MRI scanner during the power transmission [18].

1.4.3 Passive tracking

In passive tracking techniques, materials with different magnetic properties are used to generate contrast between the interventional tools and the surrounding tissue in order to visualize and track the tools [19, 20]. Unlike active tracking devices, passive tracking devices have lower hardware complexity and fewer safety issues regarding unwanted RF heating. Passive tracking techniques can be divided into two main groups: positive and negative contrast, based on the contrast effect with respect to the surrounding tissue.

Negative contrast often involves a material with a different volume susceptibility ($\Delta \chi$) that will cause signal losses in its vicinity. Bakker et al. [21] used paramagnetic dysprosium-oxide ($Dy_2O_3$) rings mounted on a catheter to visualize the catheter by generating susceptibility artifacts along its length.

For positive contrast, the effect of the magnetic material on and in the catheter has been used to generate higher signal intensities than the surrounding tissues. T1-shortening contrast agents were used to reduce the T1 of surrounding tissues, which enabled the locating of interventional tools based on high signal intensity in T1-weighted
images [1, 12]. Seppenwolde et al. [22] used additional de-phasing gradients to suppress background signals and to generate positive contrast between signals in the vicinity of susceptibility-shifted materials. Dharmakumar et al. [23] have used the steady-state profile of the signal for low-flip-angle (FA) excitations to generate positive contrast for off-resonance signals. Felfoul et al. [24] used off-resonance excitation to acquire spin-echo projections of magnetizations that are only affected by the ferromagnetic ball mounted at the tip of an untethered device.

Unless positive contrast is used in passive tracking, the generated effect (i.e. signal loss) of the interventional device can be precluded by other field inhomogeneity effects, such as air-tissue interfaces [25]. Furthermore, in volume-projection images, the effect of the device may be diluted because of its small spatial extent in comparison to the size of the voxel along the projection axis.

1.5 Endoscopic imaging in interventional MRI

In therapeutic interventions such as stent placement, kidney stone removal, and therapeutic-agents delivery, endoscopic imaging can often be as important as the tracking of the interventional devices. Moreover, the characterization of pathology and pre- and intra-procedural planning can benefit greatly from endoscopic imaging [26].

New forms of endoscopic imaging, such as intravascular ultrasound (IVUS) and optical coherence tomography (OCT), have extended the clinical usefulness of conventional endoscopy [27–29]. In IVUS, the backscattered reflection of ultrasonic waves can be used to generate endoscopic images. The spatial resolution of IVUS imaging is between 19 and 36 \( \mu m \) depending on its frequency (20 to 40 MHz) [26]. Similarly, OCT uses backscattered reflections of infrared light to generate images. The temporal and spatial resolution (10 to 15 \( \mu m \)) of OCT is superior to other endoscopic imaging modalities [26].

Compared with other modalities such as ultrasound and optical endoscopes, MRI offers superior soft-tissue contrast and deeper tissue penetration [30, 31]. The main configuration for endoscopic MRI involves using a receiver micro-coil at the tip of the
catheter inside an external field [26]. Usually, the placement of the micro-coil is such that it can acquire side-view\(^2\) images [32]. An orthogonal configuration of two micro-coils has been used to acquire forward-view endoscopic images [33]. A standalone endoscopic MRI probe has also been developed, which uses a permanent magnet to induce the magnetization and a transmitter/receiver RF coil at the tip of a catheter to excite and record the MR signals [34].

The acquisition of high-resolution endoscopic MR images requires spatial encoding fields that are sufficiently large in order to localize signal sources within a reasonable acquisition time (e.g., 5 ms), while reducing the risk of physiological effects and the loss of signal-to-noise ratio\(^3\) (SNR) [35]. Another drawback of MR endoscopic imaging is the fact that the encoding fields are stationary with respect to the tissue. Therefore, any small movement during the endoscopic imaging will cause blurring in the images [26].

1.6 Percutaneous nephrostomy (PCN)

Percutaneous nephrostomy (PCN) refers to the placement of a catheter or an interventional device in the upper urinary tract. The main indication for PCN is relief of urinary obstructions. Other indications for PCN involve access to the upper urinary tract for delivering chemotherapy, treatment of complications associated with renal transplants, and nephrolithotomy to remove kidney stones. The main complications of PCN include hemorrhage during and after the procedure, sepsis, and catheter dislodging\(^4\).

Urinary tract obstruction may be caused by deposition of calcification or cysts in the tract. The obstruction may result in hydronephrosis, which is swelling of the kidney as a result of urine build-up [36]. The prevalence of urinary obstruction is estimated to be 3% in all individuals, and 12% of the population can be affected by this.

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\(^2\)Side-view imaging refers to an imaging technique that acquires images from a plane perpendicular to the long axis of the imaging device.

\(^3\)Signal-to-noise ratio is defined as the ratio of signal amplitude to the standard deviation of the noise.

\(^4\)Involuntarily displacement of the catheter from its original placement
of patients with urinary tract obstruction who were referred for PCN intervention was approximately 1.3 million.  

1.6.1 Anatomy of the human kidney

The kidney is the main organ in the human collecting system. The main physiological function of the renal system is to remove waste products from the blood. The kidney also plays a role in the regulation of blood pressure by maintaining the optimal level of salt and water, and in the regulation of electrolytes in the body.

Figure 1.6 shows the different parts of a kidney. The renal parenchyma includes the renal cortex and medulla. The main renal artery supplies the blood to the renal cortex, and this is where the initial filtering happens. The waste products are first drained into the renal pelvis and then down into the ureter towards the bladder.

Figure 1.6: The schematic shows the different parts of the human kidney.

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5 According to American Urological Association
1.6.2 PCN methods

PCN is normally performed under local anaesthesia with patients usually placed in a prone-oblique position. A needle is used to puncture the skin and is advanced through the cortex into the renal pelvis. Once there, the drainage catheter is looped inside the upper pelvis.

There are two main techniques in PCN interventions [36]. In the one-step (or Trocar) technique, the drainage catheter is placed over the needle. Thus, with a single pass, the catheter can be looped inside the renal pelvis. The main disadvantage of the Trocar method is that the bulky device will create a sizeable puncture site, which increases the risk of hemorrhage, especially in cases where the interventionist requires multiple attempts to advance the needle into the renal pelvis.

Another technique is the Seldinger technique. The needle is placed into the renal pelvis. A 0.038 guide wire is then advanced through the needle into the ureter. The needle is retracted and a dilator is advanced over the guide wire. The drainage catheter is inserted over the guide wire and looped inside the pelvis. Finally, the guide wire is retracted. Figure 1.7 illustrates how the Seldinger technique is performed.

![Figure 1.7: The Seldinger technique. In this technique, the needle punctures through the skin towards the upper calyx. Once access is achieved, the drainage catheter is inserted and looped inside the calyx.](image-url)
1.6.3 Image-guided PCN and the role of MRI

Imaging techniques have played a significant role in improving the outcome of PCN interventions. The success rate of PCNs is dependent on the technique being used and the degree of hydronephrosis. The success rate in patients with an obstructed and dilated urinary tract is 98-99% [37]. This success rate, however, is reduced to 85% in patients with a non-dilated tract. Dilation of the urinary tract is necessary for successful placement of the catheter and guide wire either in the ureter or into the bladder (the necessary step for successful placement of the drainage catheter into the renal pelvis reducing the risk of catheter dislodging) [38].

The process of choosing a particular imaging modality is dependent on the clinical scenario in terms of the operating procedure (the Seldinger or the Trocar method) and whether the collection system is dilated or not [37]. A combination of sonography and fluoroscopy is the most frequently used imaging technique for guiding PCN. The initial puncture is done under sonography, and the subsequent guide-wire placement, catheter insertion, and looping are done under fluoroscopy. CT has also been used alone or in combination with fluoroscopy or sonography in PCN procedures [4]. These modalities often require the injection of contrast agents, which can be harmful, especially in cases where patients have urinary tract obstructions [4].

In recent years, preliminary studies have shown that MRI can be used in PCNs, particularly in cases with non-dilated collection systems, obese patients (where finding an acoustic window for safe needle advancement can be cumbersome due to the presence of fat or lack of proper kidney visualization), ectopic patients (i.e. the kidneys not in the usual anatomical position), pregnant women, and young patients. It is estimated that 6% of the patients (who were referred for a PCN intervention) may be eligible for MRI-guided nephrostomy [37]. Multi-planar imaging in MRI enables interventionalists to visually track the needle, guide wires, and catheters during the procedure to minimize the risk of inadvertently severing major arteries in the vicinity of the kidney [4, 37].

In PCN procedures, accurate tracking of the devices can greatly impact the outcome of the procedures [2, 4, 37, 38]. In the Trocar method, reliable tracking can prevent
multiple large puncture sites. In the Seldinger method, accurate tracking of the guide wire (in Figure 1.7b) can reduce the risk of catheter misplacement and thereby the risk of dislodgement.

1.7 Outline of the Thesis

The goal of this thesis is to develop novel techniques, using the effects of magnetic material in MRI, for tracking and encoding in MR-guided interventional procedures. The following is a brief chapter outline.

In Chapter 2, my hypothesis is that off-resonance excitation can provide positive contrast in three-axes projections due to the presence of off-resonance sources around a magnetic marker. This chapter demonstrates that library-based projections based on off-resonance excitation can be used to simultaneously measure the position and orientation of the device.

The purpose of Chapter 3 is to show that modulating the susceptibility effect of a marker can be used in subtraction techniques to visualize/locate the device. A short-pulsed laser with a specific wavelength has been used to demagnetize Ni thin films in magneto-optical research. In this chapter, it is hypothesized that the volume susceptibility of the Ni particles, attached to the tip of a fibre optic cable, can be changed by emitting laser photons onto the Ni particles. This new method can easily be implemented in existing catheters with relatively limited hardware complexities.

In Chapter 4, a new endoscopic modality is developed: magnetic markers are used to produce spatially varying magnetic-field perturbations. With the known geometry of the markers, magnetic-field perturbations can be predicted using simulation and be used as necessary encoding fields for signal localization. Field perturbations from susceptibility markers can thus be used to generate the spatial encoding fields. These markers can generate high, local gradients, enabling high-resolution endoscopic imaging. This is the first time this process has been done.
As the final chapter, Chapter 5 provides a summary and discussion of the significance of the research developed in this thesis and its applications in MR-guided nephrostomy. Potential directions for future investigations are also addressed.
References


Chapter 2

Position And Orientation Measurement Of Susceptibility Markers Using Spectrally Selective Spin-echo Projections

2.1 Introduction

Passive tracking often involves negative contrast with surrounding tissues [1, 2], visualized as a dark region. Methods that manipulate the contrast between interventional tools and the surrounding tissue have been developed [3–5], giving a hyperintense signal surrounding devices that would otherwise be dark. One large advantage of the resulting “positive contrast” is that the projection-mode imaging available with active devices becomes possible with susceptibility effects.

In this chapter, we developed a pulse sequence that measures device position with three off-resonance spin echoes and matching these with a library of simulated projections. The method was tested using a 3-mm diameter (9F) controllable susceptibility device [6]. Figure 2.1 shows the schematic of the tracking device. It comprises of three concentric layers of Titanium-Graphite-Titanium. The device was designed such that

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CHAPTER 2. POSITION AND ORIENTATION MEASUREMENT

when the internal layers of graphite and titanium are aligned (OFF state), the magnetic field perturbation is minimized. When the graphite part is retracted (ON state), the balance between the field perturbations no longer exists, thus generating a substantial susceptibility artifact. The tracking device was encapsulated in biocompatible heat-shrink tubing for the in vivo study as shown in Figure 2.2a. The effect of the device in OFF and ON states are shown in Figure 2.2b and c where the device was placed in right carotid artery of a pig. Since the susceptibility effect of the device was orientationally dependent (as shown later in this chapter) in ON state, simultaneous calculation of position and orientation was also demonstrated using the library of projections.

The device was only used in its ON state to generate field perturbations. The reason for using the retractable device is two fold: the extent of retraction can directly affect the abundance of signal generated by the RF pulses; once the device is at the desired location, it can be turned OFF to minimize the susceptibility artifact, and allow imaging in the vicinity of the device.
2.2 Methods

All experiments used a MR750 3.0 T MR system (GE Healthcare, Waukesha, WI, USA) using a custom-built 18 cm × 24 cm surface transceiver RF coil. The transceiver coil was built using a single loop of copper printed on a printed circuit board (PCB) and it was matched and tuned to proton resonance frequency in 3T. The coil was then connected to a transmit/receive (T/R) switch which was connected to the magnet table interface. Once the T/R switch was connected to the magnet table, the built-in body coil was disabled automatically and the signal excitation and reception were performed by the transceiver coil. The tracking technique was evaluated in both phantom and in vivo studies.

2.2.1 Pulse Sequence

There is a spatial distribution of off-resonance precessional frequencies in the vicinity of the device in the ON state. Spectrally-selective RF pulses were used to excite specific isofrequency contours around the device by offsetting the carrier frequency of the excitation pulse ($\Delta f_0$). A train of 180° refocusing RF pulses with the same carrier frequency as the RF excitation pulse was used to form three spin echoes, which were encoded along...
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Figure 2.3: (a) Tracking pulse sequence. A train of 3 refocusing pulses was used to form echoes at $TE_1$, $TE_2$, and $TE_3$ for different projections. In this diagram, $G_x$, $G_y$, and $G_z$ represent the gradients along logical axes which do not necessarily correspond to physical axes. (b) The combined spectral profile of the 90°-180° pulses for different echoes using a Bloch equation simulation.

Each physical axis, thus providing three projections in one repetition time (TR) (see Figure 2.3a). The same RF pulses described in [7] are used here. Simulations using the spectral profile of the RF pulses were performed to predict the refocused bandwidth. The combined spectral profile of the 90°-180° pulses had a bandwidth ($BW_{rf}$) of 400 Hz (see Figure 2.3b).

The performance of the sequence at different parameter combinations was tested with the RF excitation and refocusing pulses in a modified volume-projected 2D GRE sequence so that the refocusing pattern could be imaged. $BW_{rf}$ was varied from 210 to 570 Hz and $\Delta f_0$ varied from 300 to 900 Hz with TR/TE = 500/20 ms, FOV = 6 cm, readout bandwidth = 15.63 kHz, and 128 x 128 matrix.

To test the impact of the parameters above on the tracking accuracy, a set of 30 projections were acquired and averaged for each $BW_{rf}$ and $\Delta f_0$, using the triple spin-echo sequence. Then, sets of projections were acquired in separate scans by moving the device by a known amount (keeping the angle of the device relative to the field constant). These scans were repeated for different $BW_{rf}$ from 210 to 570 Hz and $\Delta f_0$ from 300 to 900 Hz. The error between actual displacement and measured value was used to find the optimal acquisition parameters. The other acquisition parameters were:
TR/TE₁/TE₂/TE₃ = 500/16/32/48 msec, 512 samples, readout bandwidth = 64 kHz, FOV = 48 cm, FA = 90°.

2.2.2 Position Measurement Technique

In order to simultaneously measure the location and orientation of the tracking device, a library of projections was built from simulated projections where the device was placed at the center of the field of view (FOV) and at various angles with respect to main field. These simulations were performed using a method described in the phantom study section.

Cross-correlation (CC) is the gold standard for pattern recognition in 1D signals. Since the effect of the device has a unique shape at each angle with respect to the main field, CC was chosen as the primary method to find the best-matched element in the library. In rapid tracking, a set of projections was acquired, and a CC for each element in the library (see Eq.3.1) was computed. The highest CC coefficient indicates the best-matched library element, which gives the physical location of the device with respect to the isocenter of the magnet as well as the angle of the device with respect to main field as shown in Eq.2.2.

\[
CC_i(n) = \sum_{-N}^{N} \frac{f^*[m] \times g_i[m + n]}{\max(f) \times \max(g_i)}
\]

\[
\max\{\max\{CC_i\}\} \rightarrow g_b(n)
\]

In Eq.3.1, N is the number of samples in a projection, \(f^*[m]\) is the complex conjugate of a new projection, and \(g_i[m + n]\) is \(i\)-th element of the library. In Eq.2.2, \(g_b(n)\) is the best-matched element in the library. The position and orientation measurement technique was implemented using MATLAB (MathWorks, Inc., Natick, MA).

2.2.3 Phantom Study

A phantom study was conducted to evaluate the accuracy of the technique. A tube was embedded in polyacrylic acid gel. A 9F catheter with the device attached to its tip was
placed inside the tube, within the bore of the scanner. The ground-truth position of the device was measured by two MR-compatible stereo cameras mounted on the bore above the scanner isocenter using alignment lights on the scanner. These cameras are capable of measuring the physical position of fiducial markers placed on the catheter with accuracy and precision of $0.04 \pm 0.01$ mm in the Z-axis [8]. The cameras were calibrated using a checkerboard with known position for the Z- and X-axis and the Y-axis was estimated using the dispersion matrix as explained in Rotenberg et al [8]. The coordinates by the cameras were converted to the scanner coordinates with origins at the isocenter using the cameras’ internal software. The tube was aligned diagonally in Y-Z plane at X=0 so that the stereo cameras could track the fiducial markers. The device was then moved through the tube, along Z- and Y-axes and 20 separate measurements were made at each position. Dynamic projection measurements with aforementioned acquisition parameters were acquired while moving the device continuously along the tube.

Simulated projections were used to build the library of elements. Field perturbations for different orientations of the device with respect to the main field were calculated. The field perturbations depend on the physical shape, and volume susceptibilities ($\chi$) of the device. Eq. 2.3 can be used to calculate the field perturbations generated by the magnetic materials of the device.

$$\Delta f(r) = \frac{\gamma}{2\pi} \delta B_d(r) \ast D_{\chi}(r)$$  \hspace{1cm} (2.3)$$

where $\gamma$ is the proton gyromagnetic ratio, $\delta B_d$ is the magnetic field perturbation of a small sphere at $B_0$ as given in Eq. 2.4, and $D_{\chi}$ is the spatial distribution of markers with different volume susceptibilities that describes not only the location and structure of the marker distribution at macroscopic scale, but also the internal structure of the marker [9]. The variable $\Delta f$, which is the convolution of the spatial distribution of magnetic materials with the dipole fields of an infinitesimally small sphere, represents the three-dimensional magnetic field perturbation of the main magnetic field in Hz generated by the magnetic
where $\phi$ is the azimuthal angle in cylindrical coordinates. The angle of the device with respect to the main magnetic field was varied from $0^\circ$ to $180^\circ$ and $0^\circ$ to $360^\circ$ steps of $2^\circ$ around the X- and Y-axes (90 × 180 elements) respectively. A Bloch equation simulation was then performed on the field perturbations to simulate the projections based on the spectral profiles of the RF pulses.

The accuracy of the orientation measurement was also evaluated in the phantom study. With respect to the main field, the angle of the device was varied from $-45^\circ$ to $45^\circ$ every $2^\circ$ about the Y-axis in the coronal plane. Ten 3-axis projections were acquired at each angle step using aforementioned acquisition parameters. Projections were fed to the library to find the best matched element. The angle corresponding to the best matched element was then identified as the orientation of the device with respect to the main field.

### 2.2.4 In vivo Study

The tracking technique was tested in vivo on a healthy 25 kg female Yorkshire pig. The pig was sedated with ketamine (15 mg/kg) and atropine (2 mL, 0.04 mg/kg) IM and intubated under a protocol approved by the institutional animal care and use committee. A 9F catheter with the device attached to its tip was introduced into the carotid artery. The surface transceiver coil was used to excite and record the signals. The catheter was moved to different locations and the position of the device was measured using the proposed technique. Projections were acquired in the order of Z-, X-, and Y-axis using following parameters: single shot, TR/TE = 500/15ms, 512 samples, bandwidth = 64 kHz, FOV = 48 cm, and FA = $90^\circ$. Respiratory-gated sagittal images were acquired at each location to visually inspect the catheter inside the artery (FGRE, TR/TE = 5.7/2.5 ms, FA = 30, 128 x 128) and for comparison with the acquired projections.
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2.3 Results

2.3.1 Tracking Sequence

Bloch equation simulations were performed to predict the excited and refocused magnetization by the RF pulses in the vicinity of the tracking device. Figure 2.4 shows sagittal volume projection images of the excited regions when the device is at different angles with respect to the main field and with the graphite part retracted to varying degrees. The $BW_{rf}$ and $\Delta f_0$ were set to 400 and 500 Hz respectively in the simulation. The bright regions around the device are the excited magnetizations. The volume of the excited region increases with a higher degree of retraction. However, retraction by, for instance, 5 mm will result in a 1.5 cm rigid part of the catheter; thus limiting the application of the device in torturous organs. In the studies in this chapter, 3 mm retraction resulted in enough SNR for tracking and measuring the orientation of the device, while minimizing for the rigid length of the catheter (1.2 cm) facilitating use in clinical applications such as percutaneous nephrostomy.

Different orientations of the device will generate a unique spatial distribution of the magnetic field perturbation thus producing a unique signal region. For example, in Figure 2.4a, the projection along the Z-axis produced two distinct peaks whereas in Figure 2.4c, the projection only contained one wide peak. These unique features were used to search in the library elements to determine the orientation of the device.

A set of 2D images was acquired to characterize the performance of the RF pulses, varying the $BW_{rf}$ and $\Delta f_0$ in small increments. As can be seen in Figure 2.5, frequency offsets closer to on-resonance signals and higher pulse bandwidths both result in the excitation of more background signal, thus providing less positive contrast in the vicinity of the device (top left corner). As the $\Delta f_0$ increases, the excited iso-frequency contour gets closer to the tracking device. There is good qualitative agreement between simulations and experimental results, as shown in Figure 2.4a and Figure 2.5. Figure 2.5 also suggests the optimal range for $\Delta f_0$ to be between 133 to 285 Hz for $BW_{rf}$ equal to 500 Hz.
Figure 2.4: Simulations of regions excited and refocused by the RF pulses where the device is (a) parallel, (b) at 45°, and (c) perpendicular to the main magnetic field. The $BW_{rf}$ and $\Delta f_0$ of the RF excitation pulses were set to 400 Hz and 500 Hz respectively. The black areas are the graphite part of the device and the white areas are titanium. The bright regions around the device are the magnitude of the excited and refocused magnetization.
Figure 2.5: A set of 2D images which was acquired by incorporating the RF excitation and refocusing pulses in a modified volume-projected GRE sequence. The $BW_{rf}$ of the excitation pulses was varied from 210 to 570 Hz, and the $\Delta f_0$ of the RF pulses was varied from 300 to 900 Hz. Other acquisition parameters were follows: TR/TE = 500/20 ms, FOV = 6 cm, bandwidth = 15.63 kHz, NEX = 6, and matrix size = 128 x 128. The scale in this graph shows the signal magnitude in a grayscale image in order to compare the excited regions. All of the images were scaled according to the maximum signal intensity in the top left image. The asterisks in these images refer to the corresponding regions marked similarly in Figure 2.4a.
Samples of three-axis projections acquired by the tracking sequence are shown in Figure 2.6 where the device is parallel (a) and perpendicular (c) to the main field. Corresponding signal regions are marked in Figure 2.4a, Figure 2.5, and Figure 2.6a. Simulated projections with no noise are also shown in Figure 2.6b,d, which are used in the library. As shown in this figure, only signal sources around the tracking device were excited and refocused by the RF pulses. The figure also shows that the orientation of the device affects the shape of the projections. Here, the two separate peaks are on the Z-axis when the device is parallel to the main field, but are on the X-axis when the device is perpendicular to the main field. Some unwanted off-resonance signal, likely due to air adjacent to the phantom is visible (arrowheads).

Figure 2.7 shows a closeup of the simulated projections as a function of the angle with respect to the main field. Simulations were performed in which the device was rotated from 0° to 180° in 10° increments within the coronal (Figure 2.7a) and sagittal planes (Figure 2.7b) about the Y- and X-axis respectively.

In order to evaluate the error in position measurements as a function of device orientation, a series of simulations were performed in which noisy projections for random orientations with random positions were simulated. The standard deviation of the noise was chosen such that the average SNR of the noisy projections was 13.8 to match the SNR of the in vivo measurements. The mean absolute error was calculated at each orientation to build a map. Figure 2.8a,b,c show the mean error maps for each principal axis as a function of rotation of the device with respect to the X- and Y-axis. Figure 2.8d shows the mean absolute error in orientation measurement. The maps exhibit an apparent random distribution of the error suggesting that there was no bias in the accuracy with respect to any particular orientation.

### 2.3.2 Phantom Study

In order to evaluate the accuracy of the tracking technique and sequence, a phantom study was performed. The $BW_{r f}$ and $\Delta f_0$ were set to 400 and 500 Hz respectively, based on optimal acquisition parameters. A set of rapid projections were then acquired in the
Figure 2.6: Real and simulated samples of projections where the device is parallel (a,b) and perpendicular (c,d) to the main field. Some unwanted off-resonance signal, likely due to air adjacent to the phantom is visible (arrowheads).
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Figure 2.7: Closeup plots of simulated projections in which the device was rotated from $0^\circ$ to $180^\circ$ in steps of $10^\circ$ in the (a) coronal and (b) sagittal planes.

Figure 2.8: (a,b,c) Simulated maps that visualize the mean absolute error in position measurement for projection SNR = 13.8 as a function of rotation for a certain SNR with respect to X-axis ($\theta_x$ in degree) and Y-axis ($\theta_y$ in degree). (d) shows the mean absolute error in orientation measurement as a function of the device orientation. The apparent random distribution of the errors in these figures suggest that the accuracy is independent of the device orientation.
order of Z, Y, and X as the device was moved along the tube. These rapid measurements were then fed into the tracking algorithm to find the best match in the library. The average time for the algorithm to find the best match for each position was 0.22 s (with a standard deviation of 0.056 s) using a computer with an AMD 3.6 GHz processor. The algorithm always returned a single lookup element for any position of the device if the tube was kept at a constant angle to the main magnetic field during device navigation. Figures 2.9a,b show good agreement between the calculated position using the tracking technique and the actual position determined by the stereo cameras. The averages of absolute errors in the Z-, Y-, and X-axes are 0.37, 0.76, and 0.85 mm respectively. Standard deviations of the measured positions along Z-, Y-, and X-axes were 0.25, 0.68, and 1.1 mm respectively. The SNR for second and third echoes drops due to T2 decay of the signal by factors of 1.5 and 1.3 respectively compared to the first echo, resulting in higher errors and standard deviations in those axes measured by these echoes.

The accuracy of the orientation measurement was evaluated by placing the device along known angles. Figure 2.9d shows a good agreement between the actual and measured angles. The mean absolute error and standard deviation are 1.5 and 1.1 degrees respectively.

An additional phantom study was performed with the same acquisition parameters in which the device was moved in all three axes. The projections were acquired in the order of Z, Y and X. Localizer images were acquired to find the best imaging plane that showed the tube and the device. High resolution oblique GRE images (resolution = 0.029 cm, 512 x 512) were then acquired in order to measure the ground truth device position. The images were imported to a viewer software (Osirix), which provides X, Y and Z of any location in the images. The software uses the perpendicular vector to an imaging plane, location of the centre of the plane, and FOV to estimate X, Y, and Z position of any point in the imaging plane. The location of the centre of the artifact was measured using the software as the ground truth device position. Figure 2.10 shows the results for each axis. The average absolute errors with respect to Z-, Y-, and X-axes were 0.45, 0.76, and
Figure 2.9: Calculated positions of the device using the proposed tracking technique versus measured positions using stereo cameras along (a) the Z-axis, (b) the Y-axis, and (c) the X-axis. The average errors in the Z-, Y-, and X-axes are 0.37, 0.76, and 0.85 mm respectively. Standard deviations of the measured positions along Z-, Y-, and X-axes were 0.25, 0.68, and 1.1 mm respectively. The lower errors and standard deviations in the Z-axis are due to the fact that the SNR of the projection in this axis is higher than in the Y-axis. (d) Calculated angles of the device using proposed technique versus the actual angles in the coronal plane.
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0.98 mm, respectively. Standard deviations of the measured positions along Z-, Y-, and X-axes were 0.34, 0.75, and 1.05 mm, respectively.

2.3.3 In vivo Study

Figure 2.11a,b show a series of the images obtained as the device was moved inside the carotid artery. Figure 2.11c,d show the corresponding projections for Figure 2.11a,b respectively, where the effect of the device has been marked by arrows. The standard deviation in Z-, X- and Y-axis measurements were 0.48, 0.87, and 1.2 mm respectively. The average SNR of acquired projections was 13.8 in the in vivo study. The measured orientation of the device for Figure 2.11a was 0° and 0° rotation about the X- and Y-axis, respectively. The measured orientation of the device for Figure 2.11b was 14° and 0° rotation about the X- and Y-axis, respectively. The standard deviation of the orientation measurements was ±1.7°.

Figure 2.12 shows a map of maximum of CCs between in vivo projections and all library elements that were simulated for rotation about the X-axis (θ rotation). Since the device was only rotated about the X-axis, the maps of maximum of CCs were only shown as a function of one rotational axis and all three physical axes. The best matched library element was shown by an arrow in each panel. The FWHM of the peaks was 4° and 1.2 mm in θ and the X-, Y-, and Z-axis respectively. Figure 2.12a and 2.12b visualize the CC maps for the two positions in Figure 2.11. As can be seen in Figure 2.11a,b, there were non-device related peaks in the Z-axis projections which resulted from excitation of the signal sources in air/tissue interface close to trachea. As a result, a secondary peak was appeared in the CC maps shown by black arrowhead. This peak was less than 50% of the peak for the best matched element.

It should also be noted that in CC maps for the Y-axis, a secondary peak with the same value of the best matched element appeared. The projection of the device in the Y-axis does not change for ±Δθ rotation. This leads to a secondary peak at ±Δθ in CC maps. However, the other two projections (X and Z) successfully found a unique library element. Therefore, there was no degeneracy issue in the solution space.
Figure 2.10: Calculated positions of the device using the proposed tracking technique versus measured positions using localizer images along (a) the Z-axis, (b) the Y-axis, and (c) the X-axis. The average errors in the Z-, Y-, and X-axes are 0.45, 0.76, and 0.98 mm respectively. Standard deviations of the measured positions along the Z-, Y-, and X-axes were 0.34, 0.75, and 1.05 mm, respectively.
Figure 2.11: (a, b) Cross-sectional slices with the carotid artery in-plane and the tracking device within the artery. (c, d) 3-axis projections acquired at each position. The standard deviation in Z-, X- and Y-axis measurements were 0.48, 0.87, and 1.2 mm respectively. The measured orientation of the device for a was 0° and 0° rotation about the X- and Y-axis, respectively. The measured orientation of the device for b was 14° and 0° rotation about the X- and Y-axis, respectively. The variation in orientation measurements was ±1.7°.
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Figure 2.12: Maps of the maximum of CCs between in vivo projections and the library elements as a function of $\theta_x$ rotation and all physical axes. **a** shows the CC maps for the first in vivo location and **b** shows the CC maps for the second in vivo location. The arrows show the best matched element for each location. The black arrowhead shows the secondary peak as a result of signal sources excited by the RF pulses at air/tissue interfaces closes to trachea as shown in the Z-projections in Figure 2.11.
2.4 Discussion

In this chapter, a new tracking technique based on off-resonance excitation/refocusing and pattern recognition was validated in phantom and in vivo studies. Background signal was suppressed using the combined spectral profile of the 90 - 180 pulses. Furthermore, due to the presence of field perturbations with high spatial frequencies around the excited regions, $T_2^*$ decay will dramatically decrease the echo amplitude in 3-axis measurements in a single TR with gradient echoes.

Stereo cameras were used to accurately track the position of the device. These cameras recorded tracking markers that were placed at specific locations on the catheter. The cameras were not able to resolve the markings on the catheter when they were inside the gel phantom. Therefore, using a tube with more curvature inside the gel phantom would render the cameras useless. Hybrid imaging techniques such as XMR could also be used to determine the location of the device in a more realistic phantom. It should be noted that any systematic error due to the cameras will propagate into the accuracy measurements. However, the apparent random distribution of the error in Figure 2.9 suggests that there was no apparent systematic error.

During tracking, $90^\circ$ excitation was used to excite the off-resonance signal in order to achieve the highest SNR for signals in the vicinity of the device. If the device was stationary, the successive RF pulses would saturate the signal sources, thus preventing continuous measurement of the position. Therefore, a few seconds will be required to allow the magnetization to return to equilibrium (e.g. 2-3s for normal tissue). However, if the device was moving, new spins would experience the off-resonance field perturbations of the device which were not affected by the previous RF pulses thus producing signal sources in projections. If continuous measurement is required, movement as much as the width of peaks in the projections (at most 3mm) would place new magnetizations in the sensitive region around the device to be excited. A potential problem related to this effect is partially saturated signal around the device, which would change the shape of the excited region such that it no longer matches any of the library elements. This
could be addressed in the tracking algorithm by only updating the position shown to the user once a match has been found, and until then just showing the previous position. In the experiments performed here, the order of the spin-echo projections was set to Z, Y, and X. The SNR dropped incrementally from the first to the third echo, resulting in variable accuracy of position measurements. This ordering can be changed automatically from one TR to the next to improve the accuracy of position measurements in each axis. Magnetization transfer (MT) can be an issue due to the use of off-resonance excitation. However, the effect of MT is negligible in blood or unbound water molecules such as urine [11].

A surface transceiver coil was made for this procedure to reduce the risk of unwanted guidewire heating due to induced current during RF transmission [12], and for greater background signal suppression due to the limited spatial sensitivity of the coil. In the animal study, the SNR (peak of the projections / standard deviation of the noise) of acquired projections was approximately 13.8 using the surface transceiver coil when the device was 5 to 6 cm from the center of the coil. This SNR resulted from excitation of a region with a volume of 47.12 mm$^3$ in the vicinity of the tracking device. Simulations have shown that projections with SNRs less than 4 do not provide a reliable position measurement (results not shown). In the phantom studies, the SNR dropped by a factor of 2 when the device was 15 cm from center of the coil. Therefore, it is expected that the SNR of in vivo projections at this depth (15 cm) will still be greater than 4.

The surface transceiver coil has a spatial sensitivity profile that can affect the shape of the acquired projection. The effect of the coil can be seen in Figure 2.6a in which the amplitudes of the two marked regions (marked with asterisks) did not match those of the simulation in Figure 2.6b. The coil sensitivity profile was ignored in the simulated projections in the library and it seemed to have negligible effect on the accuracy of the tracking technique in the phantom and in vivo studies. However, in future implementations of the technique, the coil sensitivity profile could be acquired prior to the tracking sequence and applied to the stored simulated projections. Fiducial markers can be used to find the location of the coil with respect to the scanner physical axes using rapid pro-
jections. Using this information and previously acquired coil sensitivity projections (or a Biot-Savart simulation), each library element could be multiplied by the coil sensitivity prior to CC calculation for a potential improvement in accuracy.

In general, a more global search for the location of the device is desirable. However, a confined search region will reduce the chance of discontinuities in position measurements of the interventional tools as well as the computing time to find the position. The search area in the library can be confined to the previous measured position ± the length of the device. This limited the distance that the device could be moved between measurements to about 2 cm (twice the length of the device), but this was considered to be a practical limitation for dynamic tracking. Also, using the previously measured orientation, a confined search region in the library (e.g. ±10° in rotation) can be explored to increase the speed of the measurements.

A limited number of simulated library elements was used for the proof of concept. However, a more complete simulation could be performed offline and stored in the memory. There is good qualitative agreement between the real data and simulated projections, as shown in Figure 2.6. It should be noted that the resolution of technique for measuring the angle of the device with respect to the main field is determined by the number of library elements. The finer the angular steps used to generate the library elements, the better the resolution of the measured angle in the tracking technique. The angles adjacent to the best matched element could be used as lower and upper bounds of the angle of the device with respect to main field. Therefore, the search area in the library could be confined to this region and finer simulations could be used to find a more accurate orientation measurement. It is expected that there is a lower limit to the angle steps for which the technique is capable of distinguishing the difference between adjacent elements. This is dependent on the SNR, voxel size, and size of the excited region. For very small angle steps, the projection changes might not be detected which will lead to degeneracy in solution space. However, in practical applications such as MR-guided nephrostomy, it might be practically impossible for the interventionalist to change the orientation of the device with finer steps than 2°.
It should also be noted that rotation of the device solely along its long axis will not generate new projections due to the symmetry of the device. This is a limitation of the specific tracking device used in this study. However, the particular orientation of the device with respect to its long axis may not be essential for navigating the catheter during all MR-guided procedures. The interaction between the error in translational measurements and different orientations was considered here. However, the interaction between the errors in orientations with different positions was not investigated. Ideally, the location of the artifact does not change the accuracy if there is no source altering the shape of projections. CC of two signals with $X_0$ displacement between them is only a shifted version of the CC of the same two signals with $X_1$ displacement. However, there are some effects, seen in Figure 2.6, such as signal variations due to the coil sensitivity profile that change the shape of the projections and reduce the robustness of the match. To mitigate this, the surface coil could simply be moved as the device is moved, similar to movement of the X-ray detector during the X-ray guided interventions. Other solution involves using an array of coils over the area of operation.

Cross-correlation was used to find the best match between acquired projections and library elements. Using CC, the best matched element in the library was rapidly found within approximately 0.22s, so that it can be used in real-time applications. However, other methods, such as mutual information and wavelet transform, can also be used to find the best-matched element. In future studies, we will examine different matching methods and compare them in terms of accuracy of position measurement and convergence time.

Blood flow can have an effect on the measured location of the tracking device because signal sources that have been excited can move away from the device and be refocused at a different location (RF pulses are spatially non-selective). This will give rise to errors in position measurement of the device, especially in later echoes. This issue can be minimized by cardiac gating the acquisitions. In addition, 2-axis measurements can be made in a single TR instead of 3-axis measurements to minimize the blood flow artifact.
Additional background signal sources were excited in projections acquired during the in vivo study. These signal sources were related to other off-resonance sources such as air tissue interfaces. This is particularly visible close to trachea. The CC successfully found the peaks related to the effect of the device without any pre-processing as shown in Figure 2.12. However, subtraction of two consecutive projections can minimize the stationary background, thus reducing the artifacts in the projections.

The catheter tip used in this study, was a 9F (3 mm) prototype of the tracking device. This catheter could be used in applications such as MR-guided nephrostomy, where the catheter could be advanced through a 10-12F (3.33 - 4.00 mm) sheath. Since the upper urinary tract is not as tortuous as arteries, the catheter can be used to guide a guidewire into the renal pelvis and down to the ureter. Materials with larger volume susceptibility can be used to create orientationally dependent field perturbations in smaller size devices for applications that require small and flexible catheters.

### 2.5 Chapter Conclusion

A new device tracking method using three off-resonance spin echoes and matching these with a library of simulated projections was developed and tested. It was shown that simultaneous measurements of the position and orientation are possible with this tracking device since the susceptibility artifacts are orientationally dependent. The average errors vs. ground truth were 0.37, 0.76 and 0.85 mm in the Z-, Y- and X-axes, respectively. The standard deviation of repeated measurements along Z-, Y- and X-axes were 0.25, 0.68 and 1.1 mm respectively. The accuracy of the proposed technique is comparable to that reported in the literature ranging from sub-millimetre to few millimetres [4, 5, 13, 14]. In future work, toggling the order of projections to reduce error and automatic slice repositioning will be investigated.
References


Chapter 3

Modulation of Magnetic Susceptibility Markers with Laser-induced Demagnetization of Nickel Nanoparticles

3.1 Introduction

Magnetic susceptibility effects (i.e. local signal loss, phase changes and “pile-up artifacts”) provide a straightforward means for visualizing and tracking interventional devices such as needles and catheters.[1–6]. The volume susceptibility of ferromagnetic materials such as Ni (600 ppm [7]) is large enough that even a small concentration of these materials will create substantial signal loss. However, visualizing signal loss due to ferromagnetic particles typically requires prior knowledge of the imaging slice containing the device, making tracking difficult when the device moves out of slice. Also, signal loss could be masked by other off-resonance sources, such as air-tissue interfaces. Dominguez-Viqueira et al [8] demonstrated that by mechanically modulating the susceptibility effect of markers every other phase-encode line, the effect of the device shifted by precisely half of field of view (FOV) in the phase-encode direction, enabling the location of the susceptibility marker to be calculated from projection images. It is, however,

1This chapter is adapted from a manuscript that will be submitted to Physics in Medicine and Biology journal.
challenging to fabricate this mechanically-actuated device at a small scale. Accordingly, another mechanism for modulating susceptibility effects has been investigated in this chapter.

The magnetization of certain metals such as Ni can be manipulated using laser light [9–11]. Laser-induced demagnetization of Nickel (Ni) particles has been a research topic for read-write processes in computer technology. There are two main effects on the Ni particles once they are placed in an external magnetic field and are excited by a femtosecond laser burst. The first phenomena is the optical effect in which the light interact with the electron system by increasing the electron system temperature ($T_e$) on a picosecond scale [9]. The second effect is the thermal effect: if the temperature of a ferromagnetic material is increased to the Curie temperature ($T_C$), the ordered magnetic domains will become disordered, thus reducing the volume susceptibility of the material. The absorbed light can also increase the bulk temperature of the Ni particles above $T_C$ (358 °C [7, 11]). Ultrafast demagnetization has been measured for Ni particles using the magneto-optical Kerr effect (MOKE) with a 50-femtosecond near-infrared (NIR) light (800 nm wavelength), and 2.5 to 5 mJ/cm$^2$ output power fluences [9].

In this chapter, we used the laser-induced demagnetization phenomena to modulate susceptibility effects in Magnetic Resonance (MR) images by delivering laser power to Ni particles using a fiber optic. If the laser was off, the baseline susceptibility effect of the Ni particles was observed. However, with the laser on, the Ni particles were demagnetized resulting in reduced susceptibility artifacts. This effect can be used for locating devices by subtracting the two images acquired when the laser was off and on. Preliminary experiments showed that voxels in proximity to the Ni particles followed the trend of laser output power. This suggests that Ni particles were demagnetized as a result of incident laser light.
3.2 Methods

3.2.1 Experimental Setup

A continuous-wave 808 nm fiber-coupled laser diode source (S1FC808, Thorlabs Inc., Newton, New Jersey, USA) with the maximum achievable power of 20.9 mW was used to demagnetize Ni particles. A 5 μm diameter optical fiber with 125 μm cladding in a protective kevlar layer was coupled to the laser machine using a FC/PC connector. The maximum output power fluence at the tip of the fiber was 127 W/cm². The distal end of the fiber optic cable was stripped to expose the cladding layer. Ni nano-powder particles with an average size smaller than 100 nm (Sigma-Aldrich Co., #577995-5G, St Louis, MO, USA) were glued to the tip of the stripped fiber using superglue (Loctite 401, Westlake, Ohio, USA) at a sufficient density (10% by weight using 10 g of Ni nanoparticles mixed with 100 g of superglue) to block the light from emitting out of a 5 μm inner core. Figure 3.1a shows a microscopic image of the tip of the fiber optic (cladding) and the Ni particles. The tip of the fiber was immersed into a 50 ml container filled with water, shown in Figure 3.1b.
All of the experiments were performed in a 1.5 T wide-bore scanner (Optima MR450w, GE Healthcare, Waukesha, WI, USA). A 5-inch receive-only surface coil was used to acquire MR images. A multiphase balanced SSFP (bSSFP) sequence with the following parameters was used to acquire \( N = 36 \) images: matrix size = \( 128 \times 128 \), flip angle = 40, bandwidth = 31.3 kHz, FOV = 13 cm, slice thickness = 5 mm, TR/TE = 5.6/2.6 ms, 5 second delay between images. Figure 3.1b shows one of the bSSFP images with susceptibility artifacts of Ni particles. The laser output power was changed with a trend shown in Figure 3.2a. For comparison, the same imaging protocol was repeated when the laser was off throughout the acquisition of all 36 images.

A complex signal from each voxel through all \( N = 36 \) images was correlated to the laser output power trend. Eq. 3.1 was used to calculate the cross covariance and an image was built based on the absolute values of \( CC(i, j) \) for each voxel at zero lag between the signal of the voxel and the laser output power trend. \( S^*_i,j(n+0) \) is the real signal trend of voxel \( i, j \) at zero lag and \( L_{i,j}(n) \) is the laser output power.

\[
CC_{i,j}(0) = \left| \sum_{n=1}^{N} S^*_i,j(n+0)L_{i,j}(n) \right| \tag{3.1}
\]

Furthermore, in another experiment, the changes in the signal of voxels affected by the Ni particles as a function of time were also considered to investigate the transient effect and time constant of the changes in susceptibility artifact. A train of pulses with 6s full power and 15s off, was used to excite the laser source with maximum achievable power (20.9 mW). Rapid images were acquired using multiphase bSSFP sequence with the following parameters: TR/TE = 3.4/2.6 ms, bandwidth = 64 kHz, slice thickness = 5 mm, FOV = 20 cm, wait time between images = 50 ms and matrix size = \( 64 \times 64 \). The signal curve was sampled every 250 ms in this experiment by processing these images.

In order to further validate that the susceptibility of the Ni particles changed due to interaction with the light, the fiber tip was placed inside a 4mm NMR tube and the relative volume susceptibility was measured with the laser off and on using an Evan’s balance [12].

\(^{2}\)The 5 sec delay was the startup time for the laser diode control circuit.
Figure 3.2: (a) The normalized magnitude of the signal of a voxel (shown by arrow in Figure 3.1b) as a function of time. It also shows the laser output power for each image. (b) The plot shows the correlation between normalized magnitude of the signal of the voxel and the laser power for each image. For a voxel whose signal is positively correlated with a laser power trend, decreasing the susceptibility value will result in increasing the signal amplitude in the voxel.
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Bloch equation simulations incorporating the time-varying rotation matrix during slice-selective excitation and the evolution of the off-resonance magnetization during spatial encoding were performed to estimate the percentage change in average volume susceptibility of the Ni tip due to the laser with 20.9 mW power. Two GRE images with the laser off and on were acquired using the following parameters: flip angle = 90, matrix size = 256 × 256, bandwidth = 15.63 kHz, TR/TE = 1/0.0092 s, FOV = 5 cm, and slice thickness = 5 mm. In the simulation, a fixed geometry of the tip with uniform distribution of Ni particles was used to calculate field perturbations due to the tip. Signal was simulated under the influence of these field perturbations and an image was reconstructed by taking the inverse Fourier transform. The volume susceptibility of the Ni tip was varied systematically to minimize, by least squares, the difference between the simulated and the acquired images for both the off and on states. These two fitted values for the volume susceptibility were then used to calculate the percentage change of the volume susceptibility of the Ni tip from off to on state.

3.3 Results

Figure 3.2a shows the normalized magnitude of the signal of a voxel (shown by arrow in Figure 3.1b) that is positively correlated with the laser output power. Figure 3.2b shows the positive correlation between the magnitude of the signal of the voxel and the laser power. This is consistent with a change in the signal amplitude resulting from changes in susceptibility of the Ni particles interacting with the light. As a result, the susceptibility artifact changes, and the magnitude of the voxel signal increases or decreases. However, Figure 3.2b also suggests that there is an apparent non-linear relationship between the signal changes and the laser power.

The changes in the signal amplitude were in the order of 15% of the bulk signal in the images. The experimentally observed 15% change in the magnitude of averaged signal within the susceptibility artifact between the off and on state agreed with a Bloch simulation corresponding to approximately 14% changes in the bulk volume susceptibil-
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Figure 3.3: Simulation (top row) versus acquired images (bottom row). Bloch equation simulation revealed approximately 14% changes in the bulk volume susceptibility of the Ni tip that gave the observed change in the signal magnitude between off and on state. This result is in good agreement with Evan’s balance experiment. This was further validated with the Evan’s balance experiment, which showed approximately a 16% decrease in the volume susceptibility of the ensemble of the Ni particles at the tip of the fiber cable when the laser was activated. Figure 3.3 shows the simulated and acquired images. The right column shows the percentage changes in signal magnitude for different locations in the imaging plane.

Figure 3.4a and b show the cross covariance maps for NEX equal to 10 and 1 respectively. The highly correlated voxels were located where there was susceptibility artifact from the Ni particles. Figure 3.4c and d illustrate the cross covariance maps for NEX equal to 10 and 1 respectively, when the laser was off for all acquired images. In Figure 3.4c and d, cross covariance of the complex signal of each voxel with the laser output power trend, shown in Figure 3.2a, was calculated. The results in Figure 3.4c and d suggest that the signals of the highly correlated voxels, shown in Figure 3.4a and b, were not changing if the laser was not used during the image acquisition.
Figure 3.4: Cross covariance maps of voxels with the laser power trend. The top row shows the experiments results for when the laser was used to demagnetize the Ni particles with (a) NEX = 10 and (b) NEX =1. The bottom row shows the experimental results from when the laser was not used but the signal trend of all voxels was correlated with the trend of the laser output power shown in Figure 3.2 for (c) NEX = 10 and (d) NEX = 1. The bottom row suggests that the effect of the demagnetization shown in the top row was the direct effect of the laser on the Ni particles.
Figure 3.5: (a,d) Images that were taken in which the laser was off with NEX = 10 and 1 respectively. (b,e) Images were taken with the output power of the laser set to 20.9 mW for NEX = 10 and 1 respectively. (c,f) show the magnitude of subtraction visual changes in susceptibility artifact due to the effect of the laser on the Ni particles. This is the mode that could be used for tracking using projection images.

In order to visualize the changes in susceptibility artifact of the Ni particles, an image with the output power of 20.9 mW was subtracted from another image when the laser was off. Figure 3.5a and d show the images for when the laser was on and had a NEX equal to 10 and 1 respectively. Similarly, Figure 3.5b and e show the images for when the laser was off. The subtraction images, Figure 3.5c and f, show differences in the signal intensity when the laser was on vs off. Figure 3.5f is the mode that could be used for tracking using rapid projections, which will be discussed in detail in section 5.3.2.

Figure 3.6 show the changes of the signal of a voxel as a function of time. The solid line is the fitted exponential curve to the measured data and the dashed line is the laser power as a function of time. The estimated time constants were estimated to be 5 ± 0.3 and 7 ± 0.2 s for a rise and fall of signal amplitude resulting from the demagnetization process.

The delay between the input voltage to the laser source and the actual photons coming out of the fiber tip, was measured by taking a movie of the input voltage on the
scope and the laser spot on a fluoroscopic card. By analyzing the movie frame by frame, the delay was estimated to be less than 100 ms which was negligible compared to the time constant of the demagnetization process.

3.4 Discussion

In this work, the phenomenon of laser-induced demagnetization of Ni particles was investigated. Experimental results suggested that light interact with Ni particles to cause a demagnetization process, thus changing the mean volume susceptibility of the particles. Signals of the voxels around the Ni particles were highly correlated with the laser output power trend, suggesting that the susceptibility artifacts were changing due to the effect of the laser. Additional experiments have been performed (data not shown) to investigate whether the observed effect was merely due to the Ni particles. A tip was fabricated with graphite particles to absorb the light in a similar fashion. No changes were observed in phase/magnitude of the signals of the voxels close to the tip. This suggests that the observed changes in experiments with the Ni particles were entirely driven by the effect of the laser on the Ni particles.

The exact nature of the demagnetization process is still a source of controversy in ultrafast demagnetization research [9]. The time-scale of changes of the susceptibility artifact in the experiment was estimated to be in the order of a few seconds. Based on the specification of the laser source and the time-scale, the nature of the demagnetization process is likely to be thermal. This means that the temperature of the Ni particles increased, thus reducing the bulk volume susceptibility. The tip of the fiber was not warm to the touch under any of the experimental conditions, suggesting that the ensemble of Ni particles at the tip was not heated to the extent that is detectable by finger tip. The tip was made as a mixture of Ni particles and superglue at 10% concentration. For 14% changes in the bulk susceptibility of the tip (assuming a uniform distribution of the particles in the mixture and a spherical geometry for the tip), at most 14% of the Ni particles had to be demagnetized by reaching to the Curie temperature (358 °C). This seems to be
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Figure 3.6: Changes in the magnitude of the signal as a function of time. A train of pulses with 6s full power and 15s off, was used to excite the laser source with maximum achievable power (20.9 mW). The solid line is the fitted exponential curve with a time constant of $5 \pm 0.3$ and $7 \pm 0.2$ s for rise and decay, respectively. The dashed line is the laser power.

...a high temperature, since only a small portion of the tip is being affected by the laser. A careful temperature measurement will be needed in future studies, which could be done using an infrared camera.

It should be noted that the signal rise in Figure 3.6 was not due to a steady state approach of the off-resonance signal. This is because: 1) there was a catalyzing $\frac{\alpha}{2}$ ($\alpha =$ flip angle) block before the imaging sequence for each timepoint, and; 2) the duration of the signal rise was $6 \pm 0.2$ s which exactly matched the pulse duration for the experiment in Figure 3.6.

The time constant was measured by keeping the laser on for 6s and turning it off for 15s. The signal changes in Figure 3.6, was increased by more than 50% of the baseline. In the subtraction experiment, the signal changed by 15% and the SNR was good enough to visualize the tip of the device. Therefore, there can be tradeoff between the SNR of signal changes and the temporal resolution. Figure 3.6 shows that a 15% change in signal can be achieved in approximately 2s. In some applications, such as MR-guided
nephrostomy, a temporal resolution of 2s was enough to visualize the interventional device. Currently we are exploring ways to decrease this time constant.

In the experimental setup, Ni particles were glued to the cladding of the fiber optic which was 125 \( \mu m \) diameter. However, light emanated from the inner core which was 5 \( \mu m \) diameter. Therefore, only a small fraction of the Ni particles (estimated to be less than 10\% of total volume, shown in Figure 3.1a) were exposed to light. Despite this, the SNR of the signal change was 36 in Figure 3.5c; this is equivalent to a 15\% change in the bulk signal. In future studies, it can be expected that with a better coupling between the Ni particles and the laser output, a more substantial change in susceptibility values of the particles could be achieved. Such efficient coupling could be accomplished by embedding Ni particles in the fiber inner core during the manufacturing process.

The degree of demagnetization is dependent on the ambient temperature and the power of the laser. It has been shown that the higher the power, the more the demagnetization of Ni particles [9]. The maximum achievable power of the laser that has been used for this work was only 20.9 mW. A higher power laser could be used to increase the laser-induced demagnetization effect, but bulk heating of the device tip must obviously be avoided for in vivo applications.

Once a good coupling has been achieved, the technique proposed by Dominguez-Viqueira et al [8], can be used by modulating the volume susceptibility of Ni particles every other TR. Also, a simple subtraction between images acquired with laser on and off can be used to locate the device even in projection images.

Fiber optics could be easily be attached to the existing catheters without the risk of increasing the diameter of the catheters or of reducing their mechanical properties.

In conclusion, the laser-induced demagnetization was used in MRI for the first time. The numerical simulation results were in good agreement with the experimental data. With a 23 mW NIR laser, 14\% change in the bulk volume susceptibility of the Ni tip was achieved which resulted in 15\% change in average signal magnitude. The results showed that the modulation in susceptibility artifact in MRI images has potential to be used to track interventional devices in MR-guided procedures.
References


Chapter 4

Spatial encoding using the nonlinear field perturbations from magnetic materials

4.1 Introduction

New forms of endoscopic imaging, such as intravascular ultrasound (IVUS) and optical coherence tomography (OCT), have extended the clinical usefulness of conventional endoscopy [1–3]. Magnetic resonance imaging can offer superior soft tissue contrast over modalities such as ultrasound, as well as deeper tissue penetration compared to optical endoscopes [4, 5]. However, MRI-based endoscopy is limited in spatial resolution due to insufficient localization of MR signals.

Localization of MR signals is mainly performed using the linear gradient coils mounted on the magnet [6]. The acquisition of high resolution MR images, especially in applications such as intravascular imaging, requires spatial encoding fields that are sufficiently large to localize signal sources within a reasonable signal acquisition time (e.g. 5 ms) while reducing the risk of physiological effects and loss of the signal-to-noise ratio (SNR) [7]. A number of solutions have been proposed to increase the spatial resolution of MR imaging, ranging from improvements in image acquisition methods to

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1This chapter is adapted from a paper accepted for publication in an upcoming issue of Magnetic Resonance in Medicine. It has been modified slightly for clarity. Karimi, H, Dominguez-Viqueira, W, Cunningham, CH (2014). Spatial encoding using the nonlinear field perturbations from magnetic materials. Magn Reson Med, 72, 2:399-408.
advances in imaging hardware. Wiesmann et al. [8] used cardiac and respiratory-gated gradient echo imaging to examine atherosclerosis plaques by achieving an isotropic voxel size of less than 100 \( \mu \)m. Sathyanarayana et al. [7, 9] developed a method for high-resolution intravascular MRI that employs a catheter-based transceiver coil, using the sensitivity profile of the coil for localizations. Blank et al. [10] developed a self-contained intravascular magnetic resonance probe for characterizing vulnerable plaques and for visualizing thin fibrous caps [11].

Spatial resolution can also be increased by using high-performance gradient inserts. However, the strength and switching rate (slew rate) of gradient fields are usually insufficient due to power limitations of the gradient amplifiers. Also, gradient performance is limited by peripheral nerve stimulation [12] and RF-related heating during interventional MRI [13]. In recent years, the use of nonlinear encoding fields, or spatial encoding magnetic fields (SEMs) [14], has attracted much attention because of their ability to reduce peripheral nerve stimulation [14], to image more rapidly [15], and to tailor the field of view (FOV) to suit the target application [16]. These new gradient inserts with tailored image reconstruction schemes are relatively expensive, difficult to install on current clinical magnets, and are still limited to imaging of large organs (e.g., the brain).

The aim of this chapter is to investigate the use of materials with different magnetic susceptibilities and their corresponding magnetic field perturbations as a source of spatial encoding fields. These materials distort the homogeneity of the main magnetic field, imposing predictable spatial distributions that can be used to localize signal sources in an imaged object. A variant of the strong approach in parallel imaging techniques [17] is used here to reconstruct images using signals encoded with the new SEMs. The use of magnetic materials to produce encoding fields may enable the development of high local gradients at low costs with less complexity in terms of additional hardware, and at reduced acoustic noise associated with pulsed gradients.
4.2 Theory

4.2.1 Spatial Encoding Fields

Spatial encoding in MRI is typically performed using linearly varying magnetic fields (gradients) in space. These fields are generated using the gradient coils mounted on the scanner. Signals are spatially encoded in such a way that the Fourier transform can be applied to reconstruct the image. Local variations of the main magnetic field ($B_0$), due to either field inhomogeneities or susceptibility effects, can significantly degrade the image quality. However, these local variations, especially susceptibility induced fields, can be treated as spatial encoding fields, thereby providing a new regime of image acquisition.

In this chapter, we investigate whether the magnetic field perturbations from the apparatus depicted in Figure 4.1 can be used to encode spatial information into received MR signals [20]. Excited spin isochromats precess with frequency offsets that vary based on their positions relative to the magnetic markers. Sampled signals are simply projections of the spins in the object onto isocontours of frequency offsets. Because of the non-linear and non-bijective nature of the SEM, multiple spin isochromats can be projected onto the same isocontour, which leads to ambiguities in their spatial location. However, if the SEM is rotated about its center in the next pulse repetition time (TR), magnetization is encoded by a different SEM than in the previous TR. If the excitation and acquisition process continue until one full revolution of the marker housing is completed, the resulting signal equation is described by:

$$s(t, \theta) = \int \rho(r)e^{-i2\pi \Delta f(r, \theta)t} dr$$  \hspace{1cm} (4.1)

In this equation, $\Delta f(r, \theta)$ is calculated from Eq. 2.3, $\theta$ refers to the angle of the device with respect to its own axis (see Figure 4.2) and $\rho(r)$ is the spin density vector. $T_2$ relaxation is neglected in the scope of this chapter. The term $\exp(-i2\pi \Delta f(r, \theta)t)$ represents the set of encoding functions generated by the SEM. Figure 4.2 shows the encoding functions for different angular positions of the marker housing. The linear relationship between signal and spin densities makes the reconstruction process equivalent to find-
Figure 4.1: (a) Spatial distribution of a susceptibility marker, which is a titanium marker with $\chi_p = 124.8$ ppm and radius of 1.5 mm. The arrow shows the direction of rotation of the housing. (b) Axial view of magnetic field perturbations in Hz representing the precessional frequency shift for $B_0 = 1.5T$ at a displacement 800 µm in axial plane in front of the marker. The field perturbations are reduced as $1/r^3$, where $r$ is the distance from the markers. (c) The coronal view shows the relative position of the marker housing (arrowhead) and the phantom (arrow). (d) Coronal view of the field perturbations showing the marker housing (solid lines) and the axial plane (dotted line). Note that this configuration is merely an example, among many possible configurations, that was selected for this proof-of-concept study.
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Figure 4.2: (a) Angular positions of the magnetic marker (b) Encoding functions for different angular steps at 1 ms, 5 ms and 10 ms from RF excitation, which are calculated for the marker, shown in Figure 4.1a, using Eq. 2.3. Phase of each voxel is changing as a function of time due to off-resonance.

Using a decoding matrix \( F \), which produces a spin density vector \( \tilde{\rho} \) from the encoded signals. Once the decoding matrix is calculated, Eq. 4.2 can be used to reconstruct the spin density vector where \( s \) is a vector containing sampled signals.

\[
\tilde{\rho} = Fs
\]

If the marker housing is rotated every \( \delta \theta \) \( (N_a = 2\pi / \delta \theta) \) and \( N_t \) samples are acquired for each angular position of the housing, for \( N_k \) voxels, the dimensions of the decoding matrix are \( N_k \times (N_aN_t) \). Therefore, \( s \) has dimensions of \( (N_aN_t) \times 1 \) and \( \tilde{\rho} \) has dimensions of \( N_k \times 1 \).
4.2.2 From SRFs to Decoding Matrix

The spatial response function (SRF) has been used in the context of parallel imaging in MRI in order to characterize the performance of encoding fields [21]. The SRF represents the spatial distribution of signal sources that contribute to the reconstructed intensity of a voxel. In the ideal case of linear SEMs, the SRF is a sinc function, indicating that most of the signal originates from spins located in the voxel (the main lobe of the sinc function). However, when nonlinear and non-bijective SEMs are used, contamination from signals originating from spins located outside of each voxel often occurs [22]. The more localized the SRF is for the voxel being reconstructed, the less the signal contamination from distant voxels will be. SRFs are computed using a linear combination of encoding functions as shown in Eq. 4.3 for voxel $\kappa$ [17].

$$SRF(\rho_\kappa) = \sum_{i,j} F_{\kappa,(i,j)} e^{-i2\pi f(r,\theta_i) t_j}$$  \hspace{1cm} (4.3)

$F_{\kappa,(i,j)}$ is the decoding matrix element for voxel $\kappa$ at angle step $\theta_i$ and sampling time point $t_j$. Figure 4.3 shows an in-plane localized SRF (c) and an in-plane SRF that is not as well localized (d). As shown in Figure 4.3, nonlinear SEMs resulted in signal contamination from voxels symmetrically located across the center of the marker housing.

Unlike the strong approach in parallel imaging techniques in which the decoding matrix entries for a voxel can be found via matrix inversion [21], here these coefficients are found through an iterative optimization process that searches for the best estimate of a 3D boxcar SRF at the location of the voxel. The boxcar SRF represents the perfect SRF in which there are no signal sources, in-plane or out-of-plane, except those located within the voxel being reconstructed. The least-square error between the computed SRF and the desired SRF at each iteration was minimized. Eq. 4.4 shows the cost function used to estimate the decoding matrix entries for the voxel being reconstructed ($\rho_\kappa$).

$$F_\kappa = \arg \min \left\{ \iint_{V} |SRF_c(\rho_\kappa, r) - SRF_d(\rho_\kappa, r)|^2 \, dr \right\}$$  \hspace{1cm} (4.4)

where $SRF_c$ is the computed SRF at each iteration and $SRF_d$ is the desired SRF. Note that with this optimization technique, $SRF_d$, which is the voxel function, can have any arbitrary shape. In other words, voxel functions do not necessarily reside on a rectangular
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grid. The result of the minimization process shown in Eq. 4.4, is the $\kappa$-th row of the decoding matrix. Once the decoding matrix entries for all voxels were computed, Eq. 4.2 was used to reconstruct the spin density distribution matrix.

4.2.3 Encoded Voxels

Spatial encoding with non-bijective SEMs will result in ambiguities in the reconstruction due to the indistinguishable mixture of signals originating from different voxels. In parallel imaging methodologies using nonlinear encoding fields, such as PatLoc [14] or O-space imaging [15], a consideration of the sensitivities of multiple receiver coils is crucial in the reconstruction of all voxels in the FOV. This additional information reduces the ambiguities of the encoding functions [14]. However, some voxels in the FOV can be distinguishably encoded and reconstructed reliably without using the additional spatial information from multiple receiver coils.

In the technique proposed here, not all of the locations in the FOV experience a variation of SEMs as the marker housing is rotated. Figure 4.3a shows an example of two voxels, one that lies in the path of the marker (Voxel 1) as it is being rotated and one that does not lie in the path of the marker (Voxel 2). Figure 4.3b shows the field perturbation variation as a function of the angle of the marker housing for the two voxels. Voxel 1 undergoes larger local field variations than voxel 2, which results in a more distinguishable spatial encoding for different angular steps. This is analogous to using different gradient strengths in linear spatial encoding.

The local k-space concept [23] can also be used to provide further evidence about the encoding capability of the SEMs at the locations of voxel 1 and voxel 2. Figure 4.3e shows that the extent of local k-space at the location of voxel 1 is much larger than at voxel 2 (Figure 4.3f). This corresponds to a higher resolution for voxel 1 compared to voxel 2. Readout lines in Figure 4.3e represent local k-space trajectories for different angular steps. Undersampling in local k-space causes the signal contamination as visible in Figure 4.3d. However, the SRF shown in Figure 4.3c shows that this problem seems to be controllable for the technique described here.
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Figure 4.3: (a) Reconstruction voxels (e.g., voxel 1) lie on an annulus, which corresponds to the span of one radius of the marker, as the housing was rotated. Voxel 2 is an example of voxels that are located outside this span. (b) Field perturbation for $B_0=1.5$ T at the locations of voxel 1 and voxel 2 as a function of the angle of rotation ($\theta$). Voxel are located 800 $\mu$m from the front of the marker housing. (c) The well-localized SRF for voxel 1, which shows little signal contamination. (d) Poorly localized SRF for voxel 2, which shows increased signal contamination from distant voxels. Note the difference in colour scales. The two-dimensional local $k$-space coverages for voxel 1 (e) and voxel 2 (f) show more coverage at the location of voxel 1 than at the location of voxel 2, suggesting a more localized SRF for voxel 1 than voxel 2.
Although this technique is not capable of reconstructing all of the voxels in the FOV using the marker configuration in Figure 4.1a, it is possible to obtain sufficient spatial-encoding information if the path traversed by the markers (the reconstruction annulus) were well-encoded (i.e. the SRF is localized). Hence, changing the positions of the markers in the marker housing can result in the displacement of the reconstruction annulus, thus achieving a larger in-plane FOV.

The field perturbations weaken as a function of $1/r^3$ ($r =$ radial distance from the marker) in the out-of-plane direction. Therefore, the local gradients produced by these field perturbations will be reduced as a function of distance. This limits the encoding capabilities of the field perturbation needed to achieve well-localized SRFs in a reasonable readout window for voxels located far from the marker.

4.3 Methods

4.3.1 Simulation Study

Simulation studies were performed to assess the feasibility of the new encoding technique. In order to reconstruct a large two-dimensional region in the image, 10 different marker housings were examined. A magnetic marker was placed in a different position in each housing, ranging from 7.88 to 15.82 mm (in steps of 0.794 mm) from the center of the marker housing (see Figure 4.4b). Each marker housing contained one paramagnetic marker (titanium) with a volume susceptibility ($\Delta \chi$) of 124.8 ppm and a diameter of 3.175 mm. A numerical phantom, shown in Figure 4.4a, consisting of $256 \times 256$ spin isochromats, located at a distance of 800 $\mu$m in front of the marker housing, was used to simulate signals at angular steps of $2^\circ$ and $8^\circ$. Assuming a non-selective excitation, signals were generated for readout windows of 1.6 and 3.2 ms starting 300 $\mu$s after the excitation, with a sampling bandwidth of 2.5 kHz. Random Gaussian noise was added to the complex signal values (standard deviation of the noise was calculated using SNR of the experimental result). Simulated field perturbations for each marker housing were
used to generate encoding functions. Figure 4.2 shows the simulated encoding functions. These encoding functions were then fed into the SRF optimization process to estimate the decoding matrix entries. One well-encoded annulus was reconstructed from the simulation of each marker housing. The final image was formed by piecing together the 10 different reconstructed annuli.

Since the spatial distribution and magnitude of the estimated SRFs are dependent on the spatial location and shape of the desired SRF in Eq. 4.4, intensity corrections have to be made after reconstruction to avoid any systematic changes in intensity of the voxels. This has been done by dividing the intensity of the reconstructed voxel by the integral of the computed SRF (using Eq. 4.4 once the decoding matrix entries for the voxel have been estimated) for that voxel.

4.3.2 Phantom Experiment

An experimental setup, shown in Figure 4.5, was constructed for further validation of the new encoding technique at $B_0 = 1.5 \text{ T}$. Samples of titanium were measured with an Evans balance and were found to have volume susceptibilities ($\Delta \chi$) of $128 \pm 0.23 \text{ ppm}$. Small rods made from these materials were implanted into three marker housings, with dimensions and arrangements as described in the simulation study. A non-magnetic rotary station with $2^\circ$ accuracy (CRM1L, Thorlabs Inc., Newton, NJ) was used to accurately measure the marker housing angle at each signal acquisition step. The rotary station was attached to one side of the setup to allow manual rotation of the housing outside the MR scanner bore. A uniform phantom, consisting of a 3.5-mm thick cylinder filled with CuSO$_4$ doped water (70 mM, $T_2 = 20 \text{ ms}$) was used to measure the magnetic field perturbations arising from the markers. The field-map measurement was obtained using a three-dimensional spin-echo chemical shift imaging with hard pulse excitation and refocusing pulses ($300 \mu \text{s}$ for the excitation and $400 \mu \text{s}$ for the refocusing pulse) for multiple echo times. Linear regression was applied to the phase evolutions as a function of time for each voxel in a three-dimensional matrix to estimate the spatial distribution of fre-
Figure 4.4: (a) shows the numerical phantom of size $256 \times 256$ used for signal simulations. The phantom represents an occluded artery with different structures (different magnetization densities) and a thrombus region in the center (brightest region). (b) depicts the simulation scheme which shows the placement of marker for each annulus. Each full revolution of the housing will reconstruct the corresponding annulus. The arrow shows the direction of the marker translation after each full revolution of the housing. The annuli shown here correspond to the first and last reconstructed annuli. (c-f) show the reconstructed images from simulated signals encoded with the 10 marker placements, thus reconstructing 10 annuli in each image. The upper and lower row show reconstructed images encoded every $8^\circ$ and $2^\circ$ respectively. Different readout acquisition times are shown in the left and right columns (c,e and d,f). Reconstructed image (f) correspond well with the numerical phantom (a).
frequency offsets \((32 \times 32 \times 7\) phase encoding steps with \(TE = 15\) ms, \(TR = 80\) ms, \(FA = 90^\circ\), 512 samples, \(BW = 19.23\) kHz and \(55\) mm\(\times 55\) mm\(\times 6\) mm FOV).

Experimentally measured field perturbations were used in an optimization process to further validate the numerical model described in Eq. 2.3. The least-square error between the experimental and simulated field perturbations was minimized to estimate the volume susceptibility values of the marker as well as its physical location. The estimated volume susceptibility for the titanium marker was \(124.8 \pm 1.5\) ppm. Simulated field perturbations were ultimately used to reconstruct the image using signals from the phantom experiment.

A 2-mm deep annulus phantom with a triangle (edge = 1.5 mm), circle (diameter = 1.5 mm), and square (edge = 1.8 mm) pillars was constructed and filled with 14 mM CuSO\(_4\) doped water \((T_2 = 100\) ms), and placed 800 \(\mu\)m away from the marker housing on the other side of the experimental setup. The phantom was placed at the isocenter of the scanner parallel to \(B_0\). A 5-inch single channel receive-only surface coil was used to record the signals at each angular step (see Figure 4.5b). Longitudinal magnetization was excited with a 300 \(\mu\)s hard pulse \((FA = 90^\circ)\) as shown in Figure 4.5c. During the excitation and acquisition of the signals, all of the gradient coils were turned off except for crushers applied at the end of each TR along all three axes in order to provide an audible reference for manually rotating the marker housing. A TR of 5 s was selected to allow manual rotation of the housing in \(4^\circ\) angular steps \((N_a = 90)\). Signal acquisition was started 300 \(\mu\)s after excitation \((N_t = 2048\) samples, \(BW = 19.23\) kHz). The whole experiment was repeated for three marker housings in which the titanium marker was displaced by one radius of the marker (1.5 mm), thus reconstructing three annuli.

### 4.4 Results

Figure 4.4 shows the reconstructed image from the simulation. Different magnetization densities were successfully localized for the voxels that lie on the path of the markers as the housing was rotated. Voxel functions were chosen to be sectors on the 10 recon-
Figure 4.5: This figure shows the experimental setup. (a) Top view of the rotary station used for measuring angular steps. (b) Top view of phantom and the marker housing. A 5-inch receive-only surface coil is located in the sagittal plane on top of the phantom. (c) Pulse sequence diagram showing a 300 µs hard pulse excitation followed by the signal readout at each angle step.
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struction annuli. Figure 4.4 suggests that longer readout times will increase radial and circumferential resolutions by increasing the extent of local k-space. Also, finer angular positions will increase the density of the sampled local k-space which will decrease artifacts during reconstruction. Although a large pool of spins in the middle of the phantom (mimicking signals from thrombus) was excited by the non-selective pulse, these spins were not affected by field perturbations from the marker. In the reconstruction procedure, these unencoded (on-resonance) signals did not significantly contaminate the signals that originated from the encoded voxels in the outer annuli because the SRFs for the encoded voxels were well localized.

Figure 4.6 shows the simulated and experimentally measured magnetic field perturbations for different planes with respect to the marker housing. The first plane was located 800 µm away from the marker housing and the remaining planes were spatially separated approximately by 430 µm in the axial direction. The highest in-plane local gradient in front of the marker was 4.66 kHz/cm, which is visible in Figure 4.6 (a) and (e). The measured field perturbations were used to calibrate the simulation to obtain field perturbation maps with arbitrary resolution.

Figure 4.7a shows the first 300 values of the decoding matrix coefficients for voxel 1 in Figure 4.3. Figure 4.7b shows the magnitude of encoded signals for three angular steps for one of the marker housings. Variation of the signals for different angular positions of the device suggests that spatial information was encoded in both the magnitude and the phase of the signals.

Figure 4.8 shows the gradient recalled echo (GRE) image of the phantom (a), a masked version of the GRE image showing only the middle reconstructed annulus (b), and the reconstructed image (c). The radial resolution was 0.79 mm and the circumferential resolution for the inner, middle, and outer annuli were 0.64, 0.74, and 0.84 mm respectively. Regions of lower intensity in the middle reconstructed annulus correspond well to the void regions of the shaped pillars as shown in the GRE image, demonstrating that the new spatial encoding technique was capable of encoding different regions in the
Figure 4.6: (a-d) Simulated field perturbations and (e-h) measured field perturbations generated from the modified spin-echo chemical shift imaging (CSI) pulse sequence, which shows the spatial variation of frequency shifts from the Larmor frequency of a proton at 1.5 T for different planes separated by 430 μm. The measured field perturbations agree well with the simulated ones. The maximum local gradient produced by the marker is 4.66 kHz/cm in images (a) and (e) images. Linear gradient coils can produce gradients up to 21.3 kHz/cm (maximum gradient strength = 50 mT/m).
Figure 4.7: (a) The first 300 elements of the decoding matrix coefficients for voxel 1 in Figure 4.3 ($N_a \times N_t = 90 \times 2048$). The region marked by $t_1$ shows the first time point for all 90 angular steps. Peaks 1 and 2 depict the increase in the magnitude of the coefficients when the paramagnetic marker was passing in front of the voxel at different time points. (b) Magnitude of the signals for 3 different angular positions of the marker housing, corresponding to middle annulus of the reconstructed image. As shown in the plot, the magnitudes of the signals have changed from one angle to another, suggesting that the spatial information was encoded in the signal. Although only the signal magnitudes have been shown here, the phase of the signals also contains spatial information, according to Eq. 4.1, which is visible in Figure 4.2.

phantom. The first and third annuli correspond to uniform regions of the phantom with no embedded structures.

4.5 Discussion

A new spatial encoding technique using field perturbations emanating from a marker with a different volume susceptibility than the surrounding environment has been proposed here. The spatially variant fields were rotated in increments and signals were acquired at each angular step. A titanium marker was used in three different housings, producing multiple reconstructed annuli. The experimental study presented here was performed with non-selective RF excitation. Even though signal sources from the entire annular phantom were excited, the encoding technique was able to distinguish between signals originating from within and from outside the effective reconstruction region.
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Figure 4.8: (a) GRE image of the phantom encoded by linear gradients (SNR=53.8). (b) Masked GRE image corresponding to the middle reconstruction annulus. (c) Reconstructed image from signals that are spatially encoded by the markers (SNR=7.4). The radial resolution was 0.79 mm and the circumferential resolution for the inner, middle, and outer annuli were 0.64, 0.74, and 0.84 mm respectively.

Spectrally selective pulses can be used to limit the excited magnetization to voxels within the proximity of the markers, thus reducing signal contamination from distant voxels. This excitation technique is expected to improve the accuracy of the reconstruction. It should be noted that by incorporating spectrally selective excitation, the encoding technique becomes similar to sensitive-point imaging techniques [24]. However, the fundamental difference between these two techniques stems from the fact that in sensitive-point imaging, the region from which the signal emanates is considered as one voxel, but in the new technique, the signal region is further encoded by the phase profile of the magnetic field perturbations to achieve smaller voxels.

Based on the simulations and phantom study [20], the one-marker configuration was a reasonable choice to be considered as an encoding device for the proof-of-concept experimental study. It should be appreciated that the marker combinations, their magnetic properties, and shapes presented in this study are just examples of the many possible configurations of susceptibility markers that can be used to generate encoding fields. Future studies will involve a thorough investigation of the effects of the design of encoding fields on the performance of image acquisition and reconstruction, as assessed by metrics such as the SNR, out-of-plane encoding capability, and image contrast.

Intensity variations in the reconstructed image (Figure 4.8c) have two primary sources in the new encoding technique. First is the variation in magnitude of the SRF
for each voxel. Unlike Fourier imaging, SRFs are spatially variant, thus leading to systematic variation of intensities throughout the FOV. The second source is the random noise in MR signals. Our analysis showed that the SNR in the reconstructed image is about 7.4. This low SNR will lead to noticeable changes in the intensity of the voxels in the reconstructed image as can be seen in Figure 4.8c.

Further investigations are required to increase the effective reconstruction region with the objective of reconstructing the full FOV. In nonlinear encoding schemes presented in [14, 15], multi-channel receiver coils were used to provide additional spatial information, which reduced the intrinsic ambiguities in the encoding fields. Parallel receiving can also be used with the encoding technique with susceptibility markers to obtain additional information that can be included in the optimization process to estimate the decoding matrix entries. Another approach to increasing the effective reconstruction region has already been used in this study. If the marker is moved after each full revolution, the reconstructed annulus moves with the marker, thus covering the full two-dimensional FOV. Therefore, an optimal choice for the number of markers, their configuration, and their magnetic properties, can be achieved for a given reconstruction region. It should be noted that using susceptibility markers will provide a great degree of freedom in the generation of the spatial encoding fields required for a target application. In section 5.3.4, alternative marker configurations and their field perturbations will be discussed, which may be better suited for clinical applications.

The encoding performance of susceptibility field perturbations must be further investigated. Layton et al. [25] proposed a new performance metric that analyzes the pixel covariance of an arbitrary encoding field using frame theory. Because the decoding matrix described in this chapter was built by minimizing the signal contamination from distant voxels, a new performance metric based on voxel-wise contamination variance must be developed. This new performance metric would provide the spatial distribution of contamination variance, thus providing a good approximation for the locations of well-encoded voxels in the FOV.
In this chapter, a marker housing 3 cm in diameter has been used for the proof-of-concept study. Miniaturizing the device for attaching onto a catheter poses a number of challenges that need to be overcome in future studies. Electric discharge machining can be used to fabricate markers with diameters of 300-400 $\mu$m [26]. Similar practical techniques that have been utilized in the fabrication of OCT or IVUS catheters could be used to build and assemble the marker housing [3]. In order to compensate for SNR loss in intravascular imaging, micro-coils have been implemented at the tip of interventional catheters [27]. An orthogonal-solenoid coil, presented by Anderson et al. [28], provides a good SNR distribution for forward-looking endoscopy by moving signal nulls away from the front of the catheter. To further improve the SNR, a Carr-Purcell-Meiboom-Gill sequence can be incorporated to average the signal over an echo train similar to the method of Perlo et al. [29].

Miniaturization of the markers will result in the generation of the field perturbations (e.g., $\pm$ 2000 Hz) over much smaller distances, thus increasing the local gradients and achievable spatial resolution. For example, for a titanium marker 300 $\mu$m in diameter, the maximum local gradients would be approximately 46.6 kHz/cm, which is at least 10 times larger than the local gradients produced by a 3-mm titanium marker. We expect to achieve higher in-plane spatial resolutions with the same readout time as was used in the experimental study. However, the effective spatial extent of the local gradients in the out-of-plane direction would also be decreased. As a result, the depth of penetration of the imaging device will be decreased. Achieving localized SRFs for voxels further away from the marker housing requires a longer readout time, which may render the imaging approach impractical for some clinical applications. A 300 $\mu$m marker is expected to provide a depth of penetration of as much as 250-300 $\mu$m.

Similar to Fourier encoding, the amount of the signal from a voxel is proportional to the volume of the SRF for that voxel. However, unlike Fourier encoding for which the coefficients of the decoding matrix have equal magnitude of 1, for the proposed encoding technique, the coefficients of the decoding matrix do not have a constant magnitude, as shown in Figure 4.7a. The implication of this is that a white Gaussian noise will not re-
main white and will be spatially dependent. Therefore, a more complex noise behaviour analysis is needed in future studies to be able to find SNR performance of the proposed technique.

Although the magnetic marker was used to generate forward-view images in this study, the marker can also be placed on the side of a catheter to produce side-view images similar to OCT images. As described earlier, the depth of penetration of a 300 µm marker is close to that achieved in OCT imaging [5].

One of the important properties of this encoding method is that it is not sensitive to the initial angular position of the device. For example, if the marker housing was positioned at \( \theta = 0^\circ \) and the phantom was rotated by 45°, the reconstructed image would have been rotated by 45°. This property is similar to IVUS and OCT imaging. However, in the new encoding technique, it is important to accurately measure the angular steps between each signal acquisition, because they are necessary input parameters for the field perturbation simulations. In order to orient the reconstructed image properly, optical sensors can be mounted on the marker housing to measure the initial angular position.

Another important characteristic of these encoding fields is that the orientation of the marker(s) with respect to \( B_0 \) can affect the magnetic field perturbations that were originally used to estimate the decoding matrix entries. This results in a complex artifact which, in its simplest form, is similar to a geometric distortion that may occur when using linear encoding. Therefore, one of the challenges that needs to be addressed in future studies is the measurement of the orientation of the device prior to signal acquisition at each TR. This drawback can be more pronounced in applications such as the imaging of coronary arteries, where the arteries undergo excessive motion and orientational changes. The library-based approach in Chapter 2 can be used to find the orientation of the device if the effect of the encoding device is orientationally dependent. One approach is to use the retractable device in Chapter 2 to produce the orientationally dependent effect.
Although the signal acquisition time was intentionally prolonged to allow manual rotation of the device, non-magnetic step motors could be used to automatically rotate the device with precise angular steps, thus increasing the signal acquisition speed. Manual rotation of the device gives rise to errors in the reconstruction process, such as changes in magnetic field perturbation, which leads to mis-registration of the less intense regions corresponding to the structures in the phantom. Extensive offline pre-processing can be performed to calculate decoding matrix entries for different acquisition parameters and changes in encoding fields due to changes in device orientation with respect to $B_0$ in order to accelerate the reconstruction process to clinically acceptable times ($< 1$ s) [29]. It should be appreciated that once the decoding matrix entries are calculated and stored, the reconstruction process is a simple matrix multiplication as shown in Eq. 4.2 (e.g., taking 0.6 s in our phantom study).

### 4.6 Chapter Conclusion

A new MRI method using field variations generated by magnetic materials as the spatial-encoding fields was described and demonstrated. Through simulation, it was found that only voxels near the path of the moving magnetic markers could be reconstructed, and that symmetry in the marker distribution reduced the reconstruction accuracy. Proof-of-concept is supported by the correspondence of void regions in an image reconstructed from experimental data compared to those in a conventional gradient-echo image.
References


Chapter 5

Conclusions

5.1 Thesis Summary

Magnetic materials are often considered sources of image artifacts in MRI. However, in conjunction with sophisticated excitation and signal-processing techniques, magnetic materials can be used to generate signals for tracking devices during MRI-guided procedures and to acquire endoscopic images. The focus of this thesis is to advance reliable passive tracking and imaging techniques using magnetic materials for application in the emerging field of MR-guided interventions.

Chapters 2 and 3 focused on developing tracking techniques and devices that can be useful for MR-guided PCN. The research in Chapter 4 laid the groundwork for developing an imaging technique using magnetic materials for MR endoscopes and for PCN.

Chapter 2: A new tracking method and device that matched a library of simulated projections with three off-resonance spin echoes were developed and tested. It was shown that simultaneous measuring of the position and orientation of the device is possible using this tracking method since the susceptibility artifacts are orientationally dependent. The average errors vs. the ground truth were 0.37, 0.76, and 0.85 mm in the Z, Y, and X axes, respectively. The standard deviation of repeated measurements along the Z, Y, and X axes were 0.25, 0.68, and 1.1 mm, respectively. This accuracy is shown to be comparable to the accuracy reported in the literature of other MRI tracking meth-
ods, which ranges from a sub-millimetre to a few millimetres [1–4]. However, the device modifications required to enable tracking are much simpler to implement than in other methods.

Chapter 3: In this chapter, the feasibility of using laser light to alter the magnetic susceptibility of Ni particles in order to operate as a tracking device was investigated. A fibre optic cable was used to deliver laser photons to Ni nanoparticles that were glued to the tip of a 125-µm fibre. The phantom study showed that upon being excited by the laser, the bulk volume susceptibility of the Ni tip had changed. This change enabled tracking techniques, such as the subtraction between laser on and off images, that were capable of visualizing and tracking the device. The simplicity and size of the fibre and Ni tip enabled the modification of existing MR-compatible PCN devices, with limited hardware complexities.

Chapter 4: A new MRI method using field variations generated by magnetic materials as the spatial-encoding fields was described and demonstrated. Through simulation, it was found that only the voxels near the path of the moving magnetic markers could be reconstructed, and that symmetry in the marker distribution reduced the reconstruction accuracy. In comparison to the void regions in a conventional gradient-echo image, proof of concept is supported by the correspondence of void regions in an image that had been reconstructed from experimental data. This new endoscopic imaging method paves the way for the development of endoscopes that can be used to image inside the collecting system and to improve the pre-planning of collecting-system interventions, such as nephrolithotomy.

5.2 Significance

Magnetic susceptibility difference is usually considered a source of artifacts in MRI images. The field perturbations that are a result of the differences in volume susceptibility will generate further field perturbations, which will result in intravoxel de-phasing, slice profile deformation, and geometric distortions. There are contrast mechanisms, such
as susceptibility-weighted imaging (SWI), that benefit from the susceptibility artifacts with respect to visualizing veins in the brain. In this thesis, the aim was to convert this source of artifacts in MRI images into a potential source for tracking interventional devices and for generating the spatial encoding fields necessary for MRI images. Field perturbations from magnetic markers were used for spatial encoding for the first time. It was shown that these fields could generate the large magnetic gradients necessary for achieving small voxel sizes.

Furthermore, when used in conjunction with sophisticated excitation pulses, the field perturbations were used to track and estimate the orientation of a catheter. It was also shown for the first time that a laser can be used to modulate the susceptibility of a ferromagnetic material, leading to the innovation of a contrast mechanism for tracking interventional devices.

5.3 Future Directions

With development of the tracking and imaging techniques in this thesis, the flexibility of using magnetic material can be further exploited to develop new devices tailored with specific applications. The targeted application for the techniques developed in this thesis is percutaneous nephrostomy. However, these techniques are not limited to MR-guided PCNs and would be beneficial if adopted for use in other interventions. The following sections will thus outline the future directions for improving the developed techniques, and the directions toward applying the techniques in clinical applications.

5.3.1 Clinical Applications Beyond the Kidney

The intention of developing these techniques was to advance MR-guided PCNs. Recent research shows that MRI can replace CT, and sometimes X-ray, for non-dilated, obese, young, or infant patients.

Catheter-based techniques, such as stent placement in peripheral vessels and coronary arteries, have increasingly been replacing invasive surgeries because they are both
minimally invasive and reduce recovery time. The role of MRI has also been increasing over the past few years. In fact, iMRI suites are being installed in many hospitals. To contribute to these current advancements, one of the many possible applications of the techniques developed in this thesis could be the placement of a stent in peripheral vessels. This application is backed by recent studies [5] that have shown the feasibility of using MRI for image guidance during the placement of a stent for femoral stenosis. Therefore, the tracking techniques can be applied to these procedures, especially since high temporal resolution is not particularly important when compared to the placement of a stent in the coronary artery. In general, these new applications could benefit from a smaller tracking device with a tailored tracking method.

5.3.2 Tracking Miniaturized Devices

The device that was used for the experiments in Chapter 2 was a 9F (∼ 3 mm) catheter. It has three layers of titanium and graphite in a concentric layout. Once all the layers are aligned, the field perturbation of the device is minimized, thus reducing the susceptibility artifacts in the image. However, if the graphite is retracted, substantial susceptibility artifacts will be generated. The application of the device is limited to relatively large vasculatures and organs (such as the ureter). It is therefore necessary to miniaturize the tracking device for interventions in smaller vasculatures including distal peripheral arteries. A new 6F (∼ 2 mm) catheter was developed [6] with two concentric layers of titanium and graphite.

The volume of the excited region using the RF excitation in Chapter 2 was approximately 47.12 mm$^3$ (based on simulation), which resulted in an SNR equal to 13.8. Simulations have shown that projections with SNRs less than 4 do not provide a reliable position measurement (results not shown). Smaller devices, such as the developed 6F (∼ 2 mm) catheter, will produce projections with much smaller SNR levels than the 9F device. The SNR of the projection using this device was less than 4, which made tracking it with the technique in Chapter 2 impractical.
The motivation for this future project is to develop a tracking technique that is capable of tracking smaller devices such as the 6F catheter [6]. The RF pulses used in Chapter 2 had a narrow bandwidth and could excite only positive or negative off-resonance magnetizations. However, a continuous range of frequencies from negative to positive exists around the markers, as shown in Figure 1.1. Any of these magnetizations can be used as a signal source in positive contrast tracking, which increases the SNR of the projections.

An on-resonance, non-selective, and hard pulse (300 µs) will excite a wide range of frequencies. However, substantial background signal sources are also excited by the pulse. Therefore, a careful background-signal suppression method is required in order to effectively visualize the signal around the device only. In Chapter 2, 1D projections were acquired in a single-line trajectory along each axis, from negative spatial frequencies to positive ones in 3D k-space. Traversing back on the same trajectory in the same repetition time, and completing point-wise subtraction from forward data, would effectively suppress any signal sources that are on-resonance (neglecting the $T_2$ decay, which is a fair assumption compared to readout time). Yet, doing this would not suppress any off-resonance sources that cause $T_2^*$ decay.

Figure 5.1 shows a simulated 2D image of the 9F device. The spiral k-space trajectory is traversed forward and backward. Point-wise subtraction was performed on all k-space points, and the final image was reconstructed by re-gridding the k-space onto the Cartesian grid. As shown in this figure, only off-resonance signal sources affected by the device are highlighted, as they gave rise to a positive contrast. Similar to Figure 5.1, 1D projections can be acquired with the same idea. Figure 5.2 shows the pulse sequence. Three-axes projections can be acquired in three repetition times.

If the effect of the device can be changed from one TR to another TR, any off-resonance sources that are not changing between the TRs will stay the same in the subtraction of the first and second echoes in the second TR. Therefore, subtracting the results from the first and second TRs will suppress any off-resonance signals that are not changing between the TRs.
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Figure 5.1: A 2D Bloch equation simulation of the flyback technique, in which the same trajectory traversed forward and backward in the same TR, and point-wise subtraction was performed on the $k$-space points to reconstruct the image.

Figure 5.2: Pulse sequence for flyback technique. The whole process will be done with the device on and off separately.
Figure 5.3: Flowchart for flyback technique showing the final signal for the Z axis for two TRs with the device on and off. The magnitude of complex signals is shown here. The maximum of the final signal will be the location of the device as shown by the arrowhead.

The signal sources around the device precessed with off-resonance frequencies ranging from -3 to 3 kHz. Knowing this, a high-pass filter can be applied to further suppress additional rolling background signals and non-constant off-resonance sources, such as air-tissue interfaces. Figure 5.3 shows the steps that are taken to generate a 1D signal. Its maximum point indicates the location of the device as shown by the arrowhead. This process will be done for all three axes.

A phantom study was performed to validate the technique. The sequence in Figure 5.2 was used with the following parameters: TR/TE = 8.3/2 ms, FA = 5°, 256 samples, bandwidth = 128 kHz, and FOV = 48 cm. All of these experiments were performed using GE MR750 3T and a built-in body coil for transmitting and receiving the signal. A tube
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Figure 5.4: (a) The subtraction of only the first echoes for the acquisition with the device being on and off. (b) The processed projection using the flyback technique with one repetition time. (c) The processed projection using 10 TRs. The device-related peak (marked by an arrow) is clearly visible and detectable in this figure.

was placed in a 40×20×10 cm bucket filled with tissue mimicking gel. The 6F catheter [6] was advanced in the tube. Three-axes projections were acquired for multiple locations of the device in the tube and for the device in both the on and off mode. The data was retrospectively combined to generate the processed projections. The centre of the susceptibility artifact of the device (in off mode) was used as the ground truth for the location of the device.

The importance of the second echo is to suppress background signals and to increase the SNR of the final processed signal. Figure 5.4 visualizes the importance of the second echo in each TR. The first projection in Figure 5.4a shows the subtraction of only the first echoes for the acquisition with the device being on and off (after high-pass filtering). The two peaks resulted from an air-phantom interface at the boundary of the phantom and the device itself (shown by the arrow). The projection in Figure 5.4b shows the processed projection using two echoes in each TR. This resulted in further suppression of the peak from the boundary of the phantom and an amplification of the peak related to the device. The peak-to-noise ratio (PNR) for the Figures 5.4a, b, and c was 6.5, 11, and 31, respectively.

Figure 5.5 shows the 3-axes flyback projections acquired in two separate signal acquisitions: one for when the device was on, one for when it was off. The echoes were retrospectively fed into the processing pipeline. Four TRs were averaged to increase the
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Figure 5.5: The 3-axes flyback projections acquired in two separate acquisitions, one for the device on and one for the device off. The echoes were retrospectively fed into the processing pipeline. Four TRs were averaged to increase the SNR. The maximum point at each projection represents the location of the device on the respective axis.

All of the phantom studies were performed with separate acquisitions with the device in on and off modes. Signal sources around the device reached a steady state in each acquisition. Offline processing of the echoes resulted in the peak with a PNR equal to 12, as shown in Figure 5.6a. A second phantom study was performed, and the device was turned off during the signal acquisition. The resulting changes in field inhomogeneities of the device perturbed the steady state of the signal sources close to the device. This increased the SNR of the subtraction in the flyback technique, leading to a higher PNR peak at the location of the device, as shown in Figure 5.6b. This is the more realistic scenario for the signal acquisition during MR-guided interventions.

Using a built-in body coil for the scanner, the flyback technique showed promising results in tracking smaller devices. Averaging multiple TRs can be used to increase the
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Figure 5.6: (a) The Z-axis projection for offline processing of echoes recorded in separate acquisitions with the device in on and off modes. (b) The Z-axis projection from processing the echoes recorded in a single acquisition in which the device was turned off during the signal acquisition.

SNR of the final projections. Future improvements for the flyback include performing more detailed phantom experiments to assess the accuracy of the tracking technique, and performing an MR-guided PCN using the 6F catheter.

The on-and-off capability of the retractable device was not used during tracking in Chapter 2. However, the flyback technique used this capability to increase the SNR of the projections, thus enabling tracking for even smaller devices such as the 6F catheter. If the effect of the device is orientationally dependent, the library-based approach in Chapter 2 can be combined with the flyback technique to find the orientation of the device in addition to its location.

5.3.3 Laser-Induced Demagnetization and Its Future Directions

In Chapter 3, the feasibility of using a laser for the demagnetization of the Ni nanoparticles was assessed. The Ni particles were superglued to the tip of a 5µm fibre. A 23-mW NIR fibre-coupled laser (808 nm) was used to demagnetize the particles. A 15% change in signal amplitude was observed. This was a result of the 14 to 16% changes in the bulk volume susceptibility of the Ni tip. The observed SNR of signal changes was enough to visualize the device in volume-projected images.
It was estimated that less than 10% of the Ni particles at the tip were being affected by photons, and that the rest of them were just creating a non-modulating susceptibility effect. Scattering at the end of the fibre tip was likely limited to the critical angle of the fibre core in front of the inner core. The time scale of the effect was estimated to be between three to four seconds, as based on the imaging and on Evans balance experiments. It has been concluded that it was likely that the thermal effect was more dominant than the optical effect in the demagnetization process.

An improved and optimized Ni tip and setup is required to see an increased demagnetization effect with a shorter time scale. As such, the following are some of the future directions for this part of the project:

**Fusing the Ni particles inside the fibre optic**

Usually, melted silicon dioxide glass is used to make the fibre optic. Different chemical compounds were added during the process of making the soot with the desired thickness. Ni particles can be mixed with the molten glass, so that a fibre core can be made. With this modification, the Ni particles will be directly in contact with the photons in the inner layer of the fibre. It is expected that with this technique, more Ni particles will be affected by the photons, resulting in a more observable demagnetization.

**Coating with a heat-sink material**

Most often, copper has been used to build RF coils in MRI scanners. The volume susceptibility of copper is close to that of water, thus producing minimal susceptibility artifacts. It can also act as a heat sink, transferring any heat generated by the Ni and reducing the time constant of the effect.

The increase in temperature of the particles does not cause heating of the surrounding materials, which would result in proton resonance frequency (PRF) shifts. PRF shifts are a direct result of the temperature increase after thermal therapies. These shifts directly affect the phase images that have been used for MR-thermometry. The temperature rise required to observe phase changes in the experiments in Chapter 3 would have
to be more than $120^\circ\text{C}$. If this was the case, air bubbles would have been visible in the phantom. Furthermore, manual inspection of the fibre tip (holding the tip by hand) did not result in any sensible increase in temperate. And finally, Evans balance experiment did not show any changes as a result of a temperature rise in the tip, as this exclusively measures any changes in susceptibility. For all of these reasons, it was concluded that the changes in appearance of the susceptibility artifacts (refer to Figure 3.5) was a direct result of susceptibility changes of the Ni particles.

However, the underlying mechanism that caused the modulation in the volume susceptibility of the particles was likely a thermal effect. This means that the Ni particles were heated to a high temperature (close to the Curie temperature), thus reducing the volume susceptibility. Therefore, a better mechanism for transferring heat is needed to rapidly cool down the Ni particles after turning the laser off.

**Higher-Powered Laser**

The laser source that was used in the experiments in Chapter 3 had a maximum nominal power of 23 mW. Depending on the coupling of the fibre-cable connector, a maximum power of 20.9 mW was achieved. A higher fluence of the laser could increase the fluence of scattered photons, as this would demagnetize more Ni particles.

It is expected that using a higher-powered laser will achieve a higher fluence of photons. This is expected to have two effects: absorption and scattering. A higher absorption of photons by the Ni particles will increase the temperature of the particles and lead to a higher degree of demagnetization. Scattering will have a higher fluence, resulting in an increase in the volume of the affected region at the tip of the device. In addition, a short pulse of a higher-powered laser could evoke the optical effect, as this mimics the experimental setups for laser-induced demagnetization of thin films of Ni[7]. In deploying these techniques, it is expected that demagnetization higher than a 14% reduction in volume susceptibility can be achieved.
5.3.4 Forward Solutions for Spatial Encoding Using Magnetic Materials

In Chapter 4, the decoding matrix elements were iteratively estimated directly by minimizing the least-square error between the ideal voxel function and the estimated voxel function. The image was then reconstructed by using Eq. 5.1.

\[ \tilde{\rho} = Fs \]  

(5.1)

where \( F \) is the decoding matrix, and \( s \) is a vector of the measured signals. However, a forward solution could also be used to reconstruct the image. Using this technique, the encoding matrix (\( E \)) was built using the calculated field perturbations. In order to validate this approach, the following section discusses the forward solution and the phantom studies.

Methods and Phantom Studies

Two 4-marker configurations of graphite and titanium markers, shown in Figure 5.7a and c, were used to generate the magnetic encoding fields. Figure 5.7b and d show these field perturbations at \( \theta^\circ = 0 \). The non-homogeneous magnetic fields are then used for the spatial encoding of signals. The resulting magnetic field perturbations are moved as we rotate the device by a certain degree (\( \theta^\circ \)). By rotating the device 360\(^\circ\), sufficient information can be acquired to reconstruct a forward-view image of a phantom at the tip of the device.

Eq.5.2 shows the encoding matrix (\( E \)). In this equation, we assume \( M \times N \) discrete 2D-voxels and perform acquisitions for \( K \) angles, recording FIDs for a readout time equal to \( t_L \). Elements of the encoding matrix have been computed using the 2D field perturbations in front of the markers at a 2-mm distance and prescribed acquisition parameters.
Figure 5.7: Two different configurations of markers: (a) Square configuration of markers (b) its field perturbation at 45° rotation (c) Linear configuration of markers (d) its field perturbation at 1.5T
as described.

\[
E = \begin{bmatrix}
  e^{-i\omega_1(x_1,y_1)t_1} & e^{-i\omega_1(x_1,y_2)t_1} & \cdots & e^{-i\omega_1(x_M,y_N)t_1} \\
  e^{-i\omega_1(x_1,y_1)t_2} & e^{-i\omega_1(x_1,y_2)t_2} & \cdots & e^{-i\omega_1(x_M,y_N)t_2} \\
  \vdots & \vdots & \ddots & \vdots \\
  e^{-i\omega_1(x_1,y_1)t_L} & e^{-i\omega_1(x_1,y_2)t_L} & \cdots & e^{-i\omega_1(x_M,y_N)t_L} \\
  e^{-i\omega_2(x_1,y_1)t_1} & e^{-i\omega_2(x_1,y_2)t_1} & \cdots & e^{-i\omega_2(x_M,y_N)t_1} \\
  \vdots & \vdots & \ddots & \vdots \\
  e^{-i\omega_K(x_1,y_1)t_L} & e^{-i\omega_K(x_1,y_2)t_L} & \cdots & e^{-i\omega_K(x_M,y_N)t_L}
\end{bmatrix}_{KL\times MN} (5.2)
\]

Signal values (\(s_{ij}\)) at each time point (\(t_j\)) for each specific angle (\(\theta_i\)) are computed by multiplying one row of the encoding matrix, given in Eq. 5.3b, by the magnetization density vectors \(\tilde{\rho}\) as shown in Eq. 5.3a. Eq. 5.3a is the discretization of the continuous signal equation.

\[
\tilde{s}_{ij} = E_{ij}\tilde{\rho} (5.3a)
\]

\[
E_{ij} = \begin{bmatrix}
  e^{-i\omega_1(x_1,y_1)t_j} & e^{-i\omega_1(x_2,y_2)t_j} & \cdots & e^{-i\omega_1(x_M,y_N)t_j}
\end{bmatrix}_{1\times MN} (5.3b)
\]

Due to the fact that spatial encoding is a linear process, the reconstructed image pixel values, arranged as a vector in \(\tilde{\rho}\), can be computed by solving a set of overdetermined linear equations as shown in Eq. 5.4, or equivalently by computing the decoding matrix (\(F\)).

\[
\tilde{s} = E\tilde{\rho} (5.4)
\]

If the spatial encoding fields were linear, solving Eq. 5.4 would be equivalent to performing a Fourier transform on the signals (conventional 2DFT). The encoding matrix in Eq. 5.2 is usually ill-conditioned, so taking the pseudo-inverse of the encoding matrix to compute the decoding matrix (\(F\)) will result in a poor reconstruction. A cascade regularization technique was used to reduce the ill-posedness of the encoding matrix and to improve the accuracy of the reconstruction process with truncated singular value decomposition (TSVD), followed by Tikhonov regularization [8]. Fig. 5.8 depicts the flowchart of the proposed technique for computing the decoding matrix (\(F\)) using the encoding matrix (\(E\)).
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The reconstruction technique described above is considered a forward-problem solution for the new encoding technique. In order to validate the solution, we performed simulations and phantom experiments at a 1.5T scanner using a prototype device. A numerical phantom study was performed using the simulated phase pattern resulting from paramagnetic and diamagnetic markers with susceptibility differences equal to 200 and -200 ppm, as depicted in Figure 5.7a. Simulated signals were computed using the phase pattern with specific acquisition parameters (TR, TE, BW, and number of samples in the readout direction); a Shepp-Logan numerical phantom as our magnetization density; and Eq.5.4 to compute $\tilde{\rho}$.

The results from the numerical phantom study are presented in Figure 5.9, with Figure 5.9b showing the image reconstructed with Eq.5.4. This simulation also showed that the reconstruction will not be affected significantly if the angle of the device with respect to $B_0$ changes by $\pm 15^\circ$ (results not shown).

The method was further validated by a phantom study that is presented in [8]. In this proof-of-concept study, we used real signal values from a phantom containing holes of different sizes, as shown in Figure 5.10a (GRE, TE=6ms, TR=20ms, FA=30, $128 \times 128$ pixels). A square configuration of markers, shown in Figure 5.7a, was used to encode spatial information into signals (TR=12s, TE=256µs, FA=30°, FOV=4cm, BW=0.5kHz, and 256 samples in the readout direction). Signal values were acquired with TR=12s to allow time to manually rotate the device in $10^\circ$ steps between each TR. The reconstructed image, using a simulated phase pattern, is shown in Figure 5.10b.

In this study, we computed the decoding matrix by taking the pseudo-inverse of the encoding matrix ($F = E^\dagger$). This preliminary result shows that the new technique is capable of encoding spatial information into MR signals. However, there are a few artifacts in the reconstructed image, such as a replication of magnetization densities. This is due to the fact that the inherent symmetry of the resultant field perturbation, with
Figure 5.9: Numerical phantom study and result (a) Numerical phantom (1024 × 1024) (b) Reconstructed phantom using our technique (28 × 28)

respect to its centre from the square configuration of markers, results in ambiguities relating to the encoded spatial information. To address this issue, we proposed the linear configuration of markers shown in Figure 5.7c, in which there is no inherent symmetry.

By incorporating cascade regularization, we performed another phantom study using a home-built device. The data was acquired with TE=300 µs, FA=90°, FOV=3 cm, BW=10 kHz, 2048 samples in the readout direction, and TR=6 s to allow manual rotation of the device. A linear configuration of the markers, shown in Figures 5.7c and 5.7d, has been shown to reduce the symmetry artifact, as visible in Figure 5.10b.

Two phantoms have been built for the new phantom study. The first one is completely filled with CuSO₄-doped water, and the second one is filled with a layer of oil and a layer of water as shown in Figures 5.11a and 5.11b, respectively. The reconstructed images of the phantoms (using the same decoding matrix coefficients) are shown in Figures 5.11c and 5.11d. The regions of different contrast in Figure 5.11d correlate well (qualitatively) with the oil and water regions seen in Figure 5.11b. Furthermore, the replication artifact that we observed in Figure 5.10b has been eliminated, as expected. These phantom studies ignored the intravoxel phase dispersion that occurs due to a non-
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Figure 5.10: (a) GRE image of the phantom (FOV=3cm) (b) Image reconstructed using the new method (28 × 28)

Figure 5.11: (a) GRE image of water-only phantom (b) GRE image of water-and-oil phantom (c) Reconstructed image of water-only phantom 16 × 16 voxels (d) Reconstructed image of water-and-oil phantom 16 × 16 voxels. The resolution in the reconstructed images is 2 mm.
homogenous field in the out-of-plane direction. As such, current studies do not analyze the range of phase-dispersion patterns. For instance, field perturbations due to markers vary dramatically in the out-of-plane direction, which can result in signal loss in the voxel. Based on the simulation, the sensitive region of the device is defined as the distance from the markers to the point in the out-of-plane direction, as this is where the off-resonance frequency due to the device has decreased tenfold. The sensitive region in front of the markers is $1.5R$ ($R$ is the radius of each marker). Beyond $1.5R$, there are no significant off-resonance field perturbations for different angle-steps of the device, which creates the basis for spatial encoding. Intravoxel dephasing could be addressed by considering 3D field perturbations of the markers and by building encoding and decoding matrices for a 3D reconstruction grid ($M \times N \times P$) with a voxel size smaller than $1.5R$ in the out-of-plane direction.

An artifact is visible as a hypointense region at the centre of both 5.11c and 5.11d. This artifact may be the result of $T_2^*$ decay of the FIDs, or of an intravoxel phase dispersion in the out-of-plane direction due to nonuniform field perturbation in this direction. To address this issue, the above technique can be used. A refocusing pulse can be used to reduce signal loss due to $T_2^*$ decay.

**Other Marker Configurations**

The phantom studies in the previous section were performed using four-marker configurations shown in Figure 5.7a and c. The studies in Chapter 4 were also performed using a single marker placed at three different radii to reconstruct three annuli. The main advantage of encoding with magnetic markers is that it provides a great degree of freedom in terms of marker position, size, and orientation.

An example of such a marker configuration is a multi-layer combination. Multiple layers can be made out of different mixtures of graphite and titanium, as these produce linearly varying magnetic-field perturbations. By rotating the device, different orientations of the magnetic field can be achieved. By using the linear field perturbation, a radial line in $k$-space can be acquired at each rotation step of the device. The reconstruction of
an image from the signals encoded with this field is done by simply re-gridding the $k$-space points onto a Cartesian grid and taking the inverse Fourier transform. A micro-coil at the tip of the device must be used to limit the signal-sensitivity region, thus reducing the ambiguities caused by wrapping magnetic fields far away from the device.

5.4 Conclusion

The research in this thesis is aimed at developing techniques that are useful for MR-guided nephrostomy interventions. The robust and accurate tracking of interventional devices can increase the likelihood that these interventions will have successful outcomes. Endoscopic imaging can also be used to assess the blockage or the lesion to improve treatment planning. The most impactful and relevant contribution of this thesis is the encoding technique presented in Chapter 4, as this is the first time that field perturbations from magnetic markers are used as the necessary encoding fields. The work in this chapter enables a framework for spatial encoding using any magnetic marker configuration, tailored to an application such as side-view imaging similar to IVUS and OCT.

Furthermore, the research in Chapter 3 uses a laser for the first time as a source of manipulating the contrast in MRI images. The limited hardware complexities in using a laser with Ni nanoparticles enables simple hardware modifications to existing interventional tools in MRI.
References


