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Gradient-based rotational drift and frequency encoding for high-resolution magnetic particle imaging

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Abstract

Magnetic particles exhibit nonlinear rotational drift in rotating magnetic fields, which is dependent on the particle properties and external fields. The rotational drift frequency difference from the particles on an FFL in a gradient rotational excitation field may generate a free-induction decay signal and can be used as a phase encoding method for simultaneous high-resolution and large field-of-view MPI. We propose the use of gradient-based rotational drift to excite magnetic nanoparticles on an FFL. The different rotational drift frequencies from different FFL locations induce free induction decay and refocused echo signals. Magnetic particle concentrations on the FFL can be solved by using Fourier-transform-based reconstruction. The simulation was performed using in-house developed software based on the Interactive Data Language. IDL. The 2D Shepp-Logan phantom and the digital vessel phantom were used for an FFL-based raster scan and image reconstruction simulation. The correlation coefficient between the original and reconstructed images was used for image quality assessment. The dephasing signals and echos form a k-space for image reconstruction. Reconstructed images exhibited an increasing resolution from left to right with an increasing rotational drift frequency slope under a spatially linear excitation field. In the reconstructed brain TOF-MRA image with multi-angle gradient-based rotational excitation from filtered back-projection, the main branches and small vessels are visible with a high peak SNR. In conclusion, we propose the use of gradient-based rotational drift to excite magnetic nanoparticles for high-resolution MPI.

I. Introduction

Magnetic particles exhibit nonlinear rotational drift in rotating magnetic fields [1], which is dependent on the particle properties and external fields [2, 3]. The rotational drift frequency difference from the particles on an FFL in a gradient rotational excitation field may generate a free-induction decay signal and can be used as a phase encoding method [4] for simultaneous high-resolution and large field-of-view MPI.

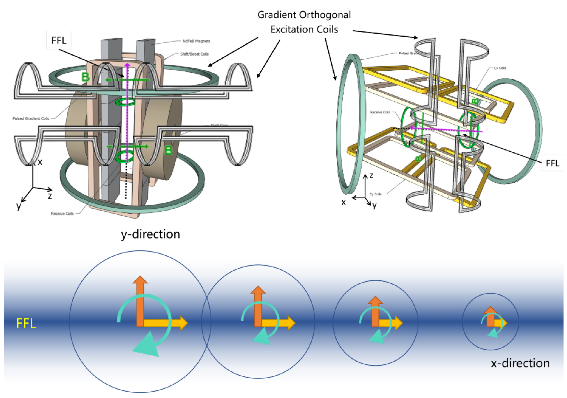

Figure 1: Top: two MPI scanners with gradient rotational excitation coils. Bottom: circular rotating excitation magnetic field along the FFL.

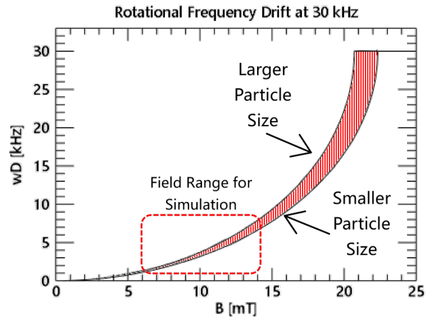

Figure 2: The relationship between rotating frequency drift of magnetic nanoparticles and external field amplitude (30 kHz excitation). The frequency drift of magnetic nanoparticles exhibits a distribution due to particle size variance. An excitation field with a small amplitude and high frequency can be used to minimize the drift distribution.

II. Theory

II.I. Rotational Excitation on FFL

Gradient coils can be used to generate a selection field with a field free line (FFL) in the middle. We assume that the FFL is along the **x** direction and the particle concentration along the FFL is **c**(**x**).

The circular rotating excitation magnetic field can be generated by two sets of gradient coils from MRI (Fig. 1). Both sets of gradient coils provide linearly varying magnetic fields along the FFL with the same gradient field, G_x. The orthogonal excitation fields from the two gradient coils are along the x and y direction respectively and have a 90º phase shift:

$$\mathbf{B}(\mathbf{x},t)=(G_x\cdot x)\cdot\left[cos(\omega_0 t)\hat{\mathbf{i}}+cos(\omega_0 t+90^\circ)\hat{\mathbf{j}}\right], \quad (1)$$

where ω_0 represents the rotating frequency of the external magnetic field.

II.II. Rotational Frequency Drift

Magnetic nanoparticles in a rotating magnetic field may have a rotating drift in the rotating direction of the external field when the external field [5] is below a critical magnetic field strength $B_c = 2\frac{k_B T}{m}\tau_B\omega_0$. τ_B denotes the Brownian relaxation time of magnetic nanoparticles in water and m is the magnetic moment of the particle.

The rotational frequency drift at each FFL location, $\omega_D(x)$, can be expressed as:

$$\omega_D(x)=\omega_0\left(1-\sqrt{1-\frac{B(x)^2}{B_c^2}}\right), \quad (2)$$

where $B(x)$=$G_x\cdot x$. The magnetic nanoparticles suspended in water show a wide distribution in particle size, relaxation time, and rotational mobility, which may result in a wide distribution in the frequency drift (Fig. 2) and affect the efficiency of gradient-based phase or frequency encoding. This frequency distribution can be minimized by choosing magnetic nanoparticles with a high B_c and less variance in particle size and mobility. We choose a gradient-based excitation field with a small amplitude and high frequency in order to minimize the drift distribution.

The linearly varying rotating field strength on the FFL can induce the nonlinear rotating frequency of the magnetic nanoparticles on the FFL. The magnetization at each FFL location is:

$$\mathbf{M}(x,t)=c(x)m\cdot e^{-i\omega_D t}, \quad (3)$$

The inhomogeneous frequency drift at different FFL locations generates an increasing phase angle $\omega_D t$ with an increase in time, which can be used for one-dimensional frequency-encoding and Fourier-transform-based image reconstruction.

II.III. Dephasing Signal

The dephasing signal is generated by the magnetic nanoparticles on the FFL with different rotating frequencies:

$$\mathbf{S}(t)=-\mu_0\int_{FFL}\frac{d\mathbf{M}(x,t)}{dt}s(x)dx, \quad (4)$$

We can get the dephasing signal as the inverse Fourier transform of the particle concentration **c**(**x**):

$$\mathbf{S}(t)=\mu_0 m\int_{FFL}c(x)s(x)\omega_D(x)e^{-i\omega_D(x)t}dx, \quad (5)$$

where μ_0 denotes the permeability of free space and **s**(**x**) the sensitivity of receive coil. Although the relaxation time is typically very short (e.g., the Brownian relaxation

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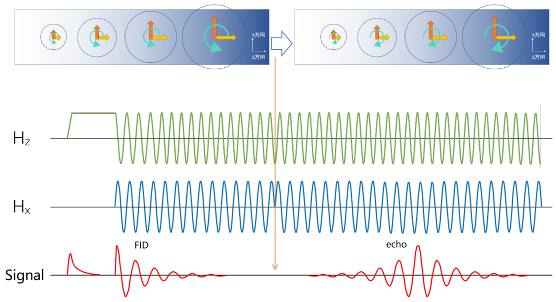

Figure 3: Free induction decay signal and echo can be generated under gradient-based rotational excitation by reversing the rotation direction of the excitation field.

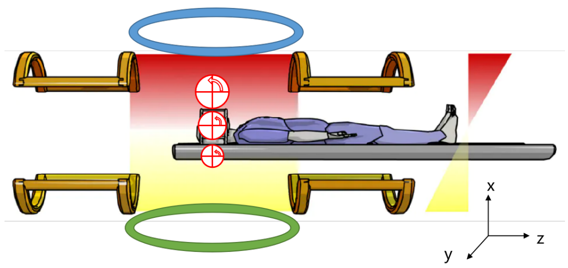

Figure 4: Multi-angle projection images are reconstructed using a circular polarized gradient field along different directions.

time τ_B in water is about ~340 µs for 100 nm and 10 µs for 10 nm hydrodynamic diameter), the relaxation is integrated into the rotational frequency drift and contributes to the dephasing signal as the accumulated rotational angle. If the signal decays too fast to be collected, the individual magnetization could be reformed into an echo by reversing the rotation direction of the excitation field (Fig. 3).

After discretization, the dephasing signal becomes:

$$\mathbf{S}(t_j) = \mu_0 m \sum_{i=1}^{N} c(x_i) s(x_i) \omega_D(x_i) e^{-i\omega_D(x_i)t_j} \Delta x. \quad (6)$$

The concentration vector of magnetic particles on the FFL can be reconstructed by using Fourier transform:

$$c(x_i) = \frac{1}{\mu_0 m s(x_i) \omega_D(x_i)} \sum_{j=1}^{N} \mathbf{S}(t_j) e^{i\omega_D(x_i)t_j} \Delta t. \quad (7)$$

The pixel size can be calculated from the acquisition time and rotational frequency drift slope:

$$\Delta x_i = \frac{1}{T_{acq} \Delta\omega_D(x_i)}, \quad (8)$$

which indicates a different resolution depending on the pixel location.

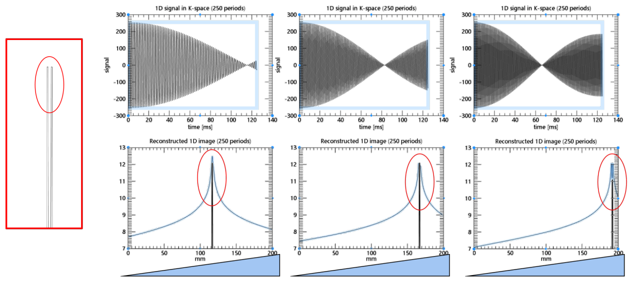

Figure 5: Two points with a width of 0.5mm and a distance of 0.5mm can be separated at a higher excitation field amplitude.

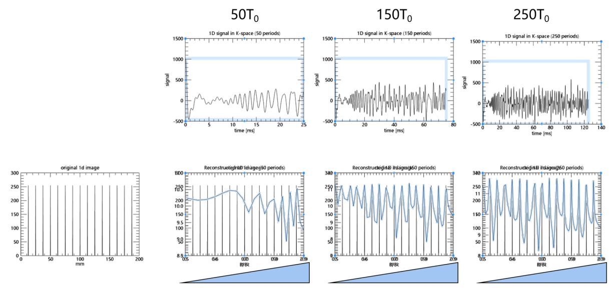

Figure 6: Reconstructed points by using 50, 150, and 250 periods.

III. Materials and Methods

The simulation was performed using in-house developed software based on the Interactive Data Language IDL. We assume that the critical magnetic field strength B_c is 20 mT. The excitation field gradient was set to 0.4 T/m. The FOV was 20 cm with the rotating gradient field ranging from 6 to 14 mT along the FOV (Dashed area in Fig. 2). The excitation frequency was set as 30 kHz. The sampling time interval was 1 to 6 µs. The FFL was translated 399 times resulting in a 2D image composed of 400 1D FFL images. The size of the phantom was 400 × 400 and the pixel size was 0.5 mm.

A circular polarization excitation field with linearly increasing field amplitude along the x direction can be generated by combining two-directional gradient coils. Magnetic nanoparticles in different layers along the x direction can be rotated at different frequencies (Fig. 4). The dephasing signal was used to generate the projection image along the x direction through Fourier transform. By rotating the excitation coils around the patient, multi-angle projection images were generated for the reconstruction of cross-sectional images using the filtered back-projection algorithm in this simulation.

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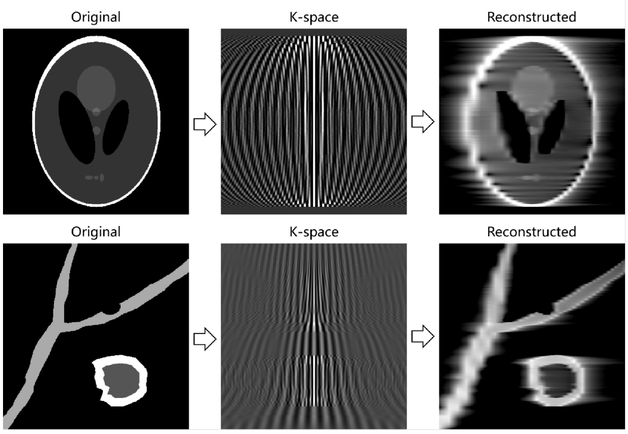

Figure 7: Horizontal FFLs are translated through the phantoms (Left), Signals from horizontal FFLs formed k-space (Middle) for Fourier-transform-based image reconstruction (Right).

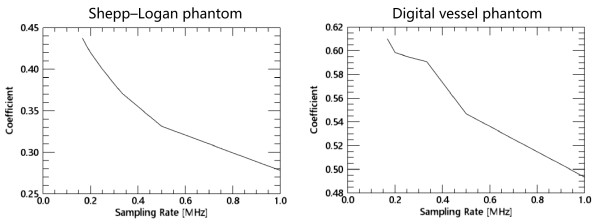

Figure 8: The correlation coefficient between the reconstructed and original image decreased with a higher sampling rate.

IV. Results and Discussions

Fig. 5 shows the reconstructed two points at locations with different excitation field amplitudes. The two points can be separated by using a higher field amplitude due to a greater frequency drift slope.

Fig. 6 shows that the pixel size is inversely proportional to the acquisition time (Eq. 8). The reconstructed points can be separated by using a long acquisition time.

Fig. 7 shows the echo signals that form a k-space for image reconstruction. Reconstructed images exhibited an increasing resolution from left to right due to a higher frequency drift gradient at a greater field amplitude. This may be a non-stationary imaging system with a different resolution, depending on the location in the field of view.

Fig. 8 shows the higher image quality with a low sampling rate, which corresponds to a narrower bandwidth and greater signal-to-noise ratio (SNR).

Fig. 9 shows the reconstructed brain TOF-MRA image with multi-angle gradient-based rotational excitation. The main branches and small vessels are visible in the reconstructed image. The peak SNR (pSNR) was 68.0337dB and the structural similarity (SSIM) was 0.4832.

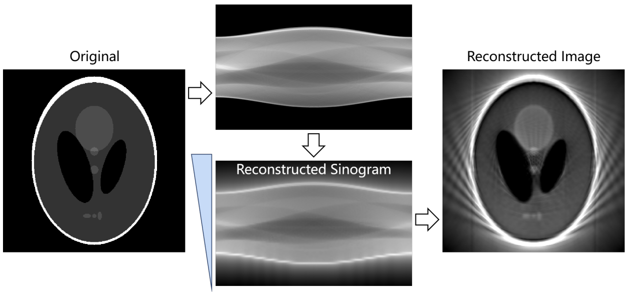

Figure 9: A reconstructed sinogram from gradient-based rotational excitation was used to reconstruct the brain vessel image.

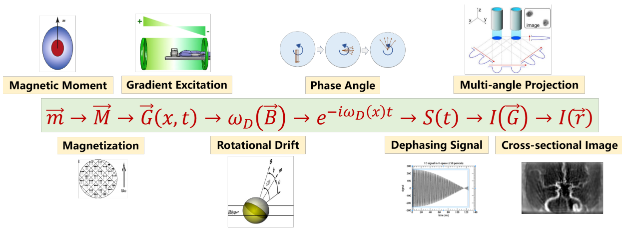

Figure 10: Imaging chain of rotation-drift-based MPI.

V. Conclusions

The imaging chain was summarized in Fig. 10. In this method, we keep rotating the magnetic particles using circularly polarized magnetic field. The phase angle of each magnetic moment keeps increasing and accumulating with different frequency drift, which is not limited by the short Brownian relaxation time. The rotational frequency drift under a gradient-based excitation field plays a key role for spatial encoding in high-resolution MPI.

In this simulation using 30 kHz rotational excitation, 1-10 kHz signals were generated, which may be smaller than the higher harmonics with higher frequencies in "conventional" MPI. Increasing wire layers and turns of the receive coils may improve the sensitivity and SNR.

In conclusion, we propose the use of gradient-based rotational drift to excite magnetic nanoparticles for high-resolution MPI. The different rotational drift frequencies from different FFL locations induce free induction decay and refocused echo signals for 1D image reconstruction. Magnetic nanoparticle concentrations on the FFL can be solved by using Fourier-transform-based reconstruction. Multi-angle gradient-based rotation drifts generate the sinogram for high-resolution 2D image reconstruction.

Author's statement

Conflict of interest: Authors state no conflict of interest. Informed consent: Not applicable. Ethical approval: Not applicable

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