Investigation of Magnetic Interference Induced via Gradient Field Coils for Ultra-Low-Field MRI Systems

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Abstract. We are developing a compact ultra-low-field MRI system that is composed of a SQUID gradiometer and a coil set that generates magnetic fields for capturing MR images. The magnetic interference induced from a power amplifier potentially disturbs MRI measurements. We investigated the path of the interference by experimental measurements and calculation of the magnetic field generated by the coil set. We found that the magnetic field generated from a particular gradient coil affected the SQUID gradiometer and that the level of the interference was strongly dependent on the shape of the gradient coils. When the coils’ shapes are designed, minimizing the noise introduced from the power amplifier is crucial, in addition to consideration of the homogeneities of the magnetic field.

1. Introduction

Ultra-low-field nuclear magnetic resonance (NMR) and magnetic resonance imaging (MRI) have been widely studied in the past 10 years [1-2]. Our group is now developing a compact ultra-low-field MRI system that can be integrated with a magnetoencephalography (MEG) system for small animals based on SQUID techniques [3-4]. This integration will greatly benefit the development of new medicines and pathological studies because it enables concurrent capturing of the biomagnetic signals and anatomical images of small animal.

The ultra-low-field MRI system was equipped with a set of coils to generate magnetic fields oriented in the direction of the y-axis. The coil set was composed of coils for so-called the measurement field (Bm) and coils to produce gradient fields along the x-, y-, and z-axes (Gx, Gy, and Gz, respectively). The electric noise induced from a power amplifier via such coils sometimes affects the magnetic sensors and eventually disturbs MRI measurements even if the power amplifiers have been carefully designed to have low electric noise. We found that the magnetic noise induced from the Gy coil was significantly greater than the noise induced from the other coils.

Therefore, we investigated the magnetic interference induced via the coils through the experimental measurement and numerical calculation of the magnetic field distribution.
2. Configuration of the System

Figure 1 shows a block diagram of our ultra-low field MRI system. The coil set was fabricated to be desktop-sized for installation inside a magnetically shielded box.

The four pairs of shielded planar coils for the measurement field and the 3D gradient fields were optimized through a target field method and sufficient homogeneity for ultra-low-field MRI measurements was obtained [3]. In this paper, the noise induced via the polarization coil ($B_p$ coil) was not considered because the $B_p$ coil was disconnected from the power amplifier using a mechanical relay during the acquisition of the MR signal.

A SQUID sensor with a first-order axial gradiometer pick up coil was employed to detect the MR signal. The pickup coil was perpendicular to the $z$-axis in Figure 1, and its diameter was 15 mm. The baseline of the gradiometer was set along the $z$-axis and its length was 50 mm. The noise level of the SQUID gradiometer was $5.3 \, \text{fT/Hz}^{1/2}$.

3. Magnetic Field Measurements

Experimental magnetic field measurements were performed in order to examine the path of the magnetic noise, which we discuss in this study.

First, we measured the free-induction decay (FID) signal from 100 ml of tap water; this signal is shown in Figure 2. The intensity of the measurement field was 33 $\mu$T; the corresponding Larmor frequency of protons was 1.4 kHz. The FID signal was clearly observed when only the $G_x$ and $G_z$ coils were connected to the current driver unit, whereas the noise level of the measured waveform considerably increased to 520 $\text{fT/Hz}^{1/2}$ and the FID signal could not be seen when the $G_y$ coil was connected to the power amplifier. This result suggests that the $G_y$ coil was the main noise introduction path from the power amplifiers.

Second, a sinusoidal electric current was individually applied to each coil from the power amplifier which was connected to a function generator. The frequency and the amplitude of the electric current were 5 Hz and 10 mA, respectively. A fluxgate magnetometer, instead of the SQUID gradiometer, was employed to measure the magnetic signal, in order to clarify magnetic field distribution.

Figure 3 shows the results of the measurements. The horizontal axis shows the position of the fluxgate magnetometer along the $z$-axis. Here, $z = 0$ was defined to be equal to the center of the planar coils. The measured range corresponded to the position of the SQUID gradiometer during MRI measurement. The vertical axis shows the $z$-component of the measured magnetic signals ($B_z$). The intensity of the measured signal generated by the $G_y$ coil was 25 to 400 times larger than those generated by the other coils. This means that the SQUID sensor’s sensitivity to the magnetic field induced from the $G_y$ coil was 25 to 400 times greater than its sensitivity to the magnetic fields induced from the other coils.
4. Analysis of Magnetic Field Distribution

In order to investigate why the Gy coil induced the most magnetic noise, the distributions of the magnetic fields generated by each coil were calculated around the field-of-view (FOV) area and the SQUID sensor. The coils were assumed to be composed of small segments, and the magnetic field generated by each segment was calculated using the Biot-Savart equation. Thus, the magnetic fields generated by the coils were sums over the calculated magnetic fields corresponding to the coil segments.

The directions of the calculated magnetic fields are indicated by the arrows in Figure 4. The arrows have a fixed length in this figure and hence do not represent the field magnitudes. The circles drawn with dashed lines indicate the FOV area; its diameter was 40 mm. The position of the pickup coil of the SQUID gradiometer for the MRI measurements is also shown in the Figure 4 diagrams.

Figure 2. The FID signal from 100 ml of tap water in the measurement field of 33 μT. The noise level of the measured waveform increased when the Gy coil was connected to the power amplifier.

Figure 3. Intensity of Bz generated from the coil set.

Figure 4. Directions of the calculated magnetic field corresponding to each coil: (a) Bm coil, (b) Gx coil, (c) Gy coil, and (d) Gz coil. The length of each arrow is fixed and do not represent the intensity of the magnetic field.
Only the magnetic field generated by the G_y coil pointed toward the z direction around the SQUID gradiometer. Therefore, the intensity of the magnetic field generated by the G_y coil in z direction was much larger than that generated by the other coils.

Figure 5 shows the calculated intensity of \( B_z \) generated by the Gy coil along the z-axis. The \( B_z \) generated from the G_y coil had a gradient parallel to the baseline of the gradiometer. This calculation agreed well with the result of the magnetic measurement shown in Figure 3. The gradient of \( B_z \) functioned to deteriorate the gradiometric cancellation effect of the gradiometer.

5. Discussion
It is possible to deduce from the section 3 and 4 that the level of magnetic interference to the SQUID sensor strongly depends on the shape of the coils. In our system, large magnetic noise induction resulted from the shape of the G_y coil, because the magnetic field generated by the G_y coil was dominated over \( B_z \) and had a gradient along the z-axis.

Previously, only the direction parallel to the measurement field has been considered in designing coil patterns. In addition to the consideration of the homogeneities of the magnetic field, minimizing the noise introduced from the power amplifier to SQUID sensors is also crucial.

With regard to the SQUID sensor design, the second-order axial gradiometer (\( d^2B_z/dz^2 \)) or the planar gradiometer (\( dB_z/dx \) of \( dB_z/dy \)) might be useful to reduce the magnetic interference induced via the G_y coil, which we plan to introduce in our ultra-low-field MRI system.

6. Conclusion
The level of magnetic interference strongly depends on the shape of the gradient coils. Reducing the magnetic interference induced via gradient coils requires optimization of coils’ shape or appropriate design of the SQUID gradiometer.

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