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Development of human head size magnetic particle imaging system

K. Nomura[a,b,*] · M. Washino[a] · T. Matsuda[a] · T. Nakagawa[b] · S. Seino[b] · T. Kiwa[c] · M. Kanemaru[a]

[a]Mitsubishi Electric Corp., 8-1-1 Tsukaguchi-Honmachi, Amagasaki, Hyogo 661-8661, Japan
[b]Osaka University, 2-1 Yamadaoka, Suita, Osaka 565-0871, Japan
[c]Okayama University, 1-1-1 Tsushimanaka, Okayama Kita-ku, Okayama 700-0082, Japan
[*]Corresponding author, email: Nomura.Kota@cw.MitsubishiElectric.co.jp

Abstract

A magnetic particle imaging device deploys an alternating magnetic field generated by a coil to induce magnetic signals from magnetic particles injected into the body. It then uses these signals to produce a three-dimensional image. A higher frequency alternating magnetic field enhances the sensitivity of signal detection. Frequencies of approximately 25 kHz are used in commercialized, compact devices designed for testing on small animals, such as mice. One of the factors that hitherto hindered the practical application of such systems in the treatment of humans was the extremely large size of the power supply unit needed to drive a much larger coil. Mitsubishi Electric has developed a process that minimizes extraneous signals (noise) that hinder the detection of magnetic signals by leveraging its extensive electromagnetic technology know-how acquired through the development of various devices and fine-tuning the configuration of the coils that generate alternating magnetic fields and those dedicated to signal detection. As a result, we have successfully developed a magnetic particle imaging device that can sensitively image magnetic particles in an area equivalent to the size of the human brain.

I. Introduction

Magnetic particle imaging (MPI) is an attractive imaging modality that detects disease by injecting magnetic nanoparticles (MNPs) into the body as tracers and acquiring their alternating magnetization signals [1]. Recently, MPI systems for human have been designed and developed in various institutes [2-4]. One major difficulty in increasing the size of MPI systems is that the power supply for the coils that form the magnetic field has a large capacity. In particular, the AC magnetic field for exciting signals from magnetic particles is generally about 25 kHz and tends to increase the power supply capacity. We seek to scale MPI for human use by expanding our previous MPI systems. We have been developing a low-frequency MPI system to enable signal acquisition at 1 kHz or less [5]. This paper details a prototype result of our developed MPI system for imaging human-head-sized samples. Air-core coil systems have been developed that can generate and scan field-free line (FFL).

II. Material and methods

The MPI equipment consists of a gradient magnetic field source that generates a static magnetic field region with an FFL, and an excitation coil that applies AC magnetic fields. When MNPs exist on an FFL, their magnetization fluctuates due to the applied AC magnetic field. As a result, the receive coil can detect the magnetic signal as a voltage. When MNPs exist in a non-zero statistical mag-

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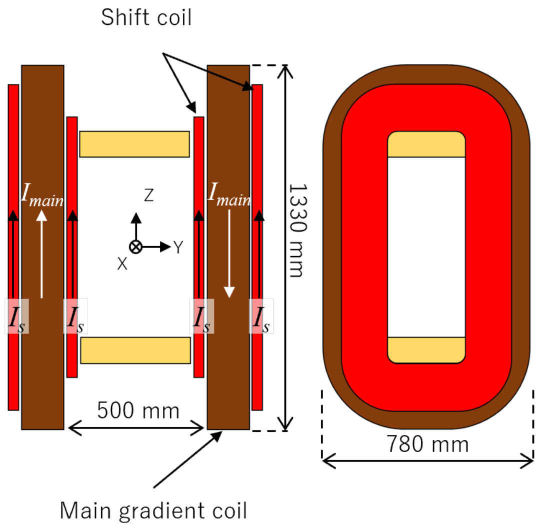

Figure 1: Illustration of electromagnet coil assembly for scanning FFL: Electromagnet consists of four pairs of racetrack coils. Main gradient coil forms a base FFL by applying a constant current. FFL position can be scanned by changing electric current of two pairs of shift coil.

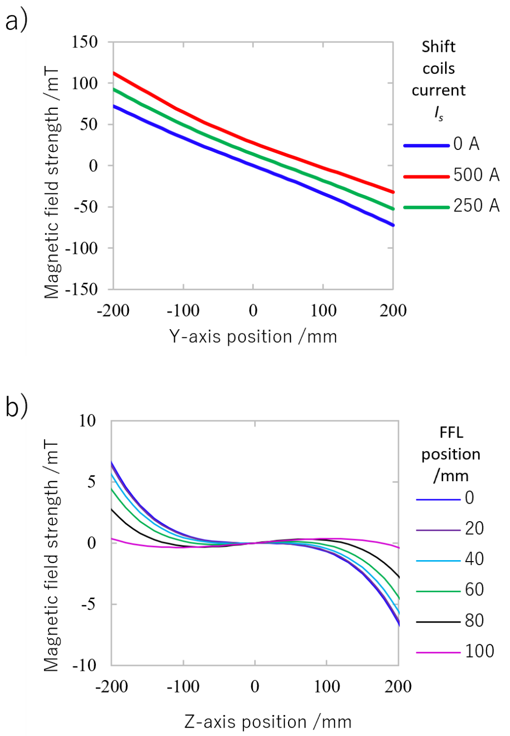

Figure 2: a) Gradient coils were designed to form a gradient magnetic field of 0.37 T/m at coil's center by applying a direct current of 500 A. b) FFL position is scanned in a 95-mm positive direction. The FFL within 95-mm scan area is +/- 100 mm long in z direction.

netic field, MNP magnetization might become saturated, and no magnetic signal will be detected. An FFL-based MPI offers greater benefit of sensitivity from 2D projection encoding than a field free point since the sensing area is broadened. To take a 2D image, the FFL must be moved across the field-of-view (FOV). FFLs and their shifts can be combined in the same electromagnet system, but their generation is achieved using four pairs of coils. Fig. 1 shows a model of the designed coil assembly. The main gradient coil consists of a pair of racetrack coils. The main gradient coils are applied with a constant current to form an FFL, which can be shifted by applying a variable current to a shift coil that consists of two sets. The shift coil and the main gradient coil are wound by a hollow conductor, which has square cross sections of 12 x 12 mm^2 and 13.5 x 13.5 mm^2 and includes a hollow channel 8-mm diameter to allow water cooling. The main gradient coil, which consists of 143 turn racetrack coils, is designed to generate an inclined magnetic field of 0.37 T/m with a DC current of 500 A. The shift coil enables the FFL to be scanned by arranging two pairs of coils that sandwich the main gradient coil. The outer and inner coils consist of 12 and 40 turn racetrack coils. The FFL position can be scanned by changing the current from -500 to 500 A in the shift coil. Fig. 2 shows the gradient magnetic field distribution when the shift coil's current amount is changed from 0 to 500 A, and the magnetic field distribution is in the no-field region at each FFL position. We confirmed that the designed coil system has a FOV of about φ 180 mm. Fig. 3 shows a human head as a developed MPI device and a CAD diagram of the drive coil and the receive coil. The receiving coil inner diameter and bore diameter are 300 mm. The drive coil is a pair of pancake coils having an indirect water-cooling structure with a copper plate. The copper plate is designed to be in two parts with the coil circumferential direction to reduce the generation of the eddy currents. An AC current of 500 Hz can be loaded up to 140 Ap-p using a conductor consisting of 2 x 6 mm^2 conductors. The distance between the two coils (wound in 28 turns) can be variable, and the center field strength can be excited by a 20 mTp-p in the Helmholtz configuration.

III. Results and discussion

The image reconstruction results with the MNPs samples are detailed below. The MNPs used Ferucarbotran (γ-Fe2O3, Meito Sangyo Co., Ltd.), which is an iron oxide particle that is a material used for MRI contrast agents. Fig. 4 shows the measurement results of the DC magnetization characteristics of the MNPs that we used. To evaluate the imaging performance of the developed device, we fabricated a 150 × 140 mm, M-shaped phantom. Fig. 5 shows an M-shaped phantom, formed by sealing MNPs in a silicon tube with a 6-mm inner diameter and

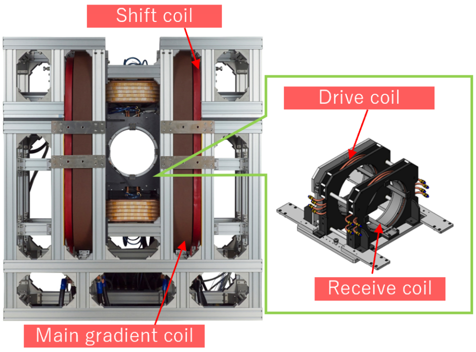

Figure 3: Figure 3: Photographs of appearance of human-head-sized MPI system and CAD diagrams of drive and receive coil

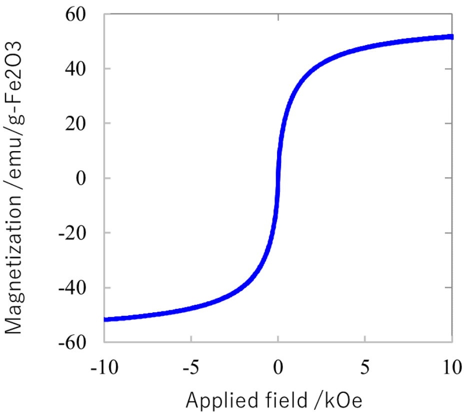

Figure 4: DC magnetization characteristics of Ferucarbotran (γ-Fe2O3, Meito Sangyo Co., Ltd.), which is an MNP used in a phantom

fixing them to a polymer base in an M-shape.

Image reconstruction was done using third-harmonic signal data with an excitation frequency of 500 Hz (corresponding to the magnetic field strength of 20 mTp-p). It was difficult to reconstruct an image of a signal of the fifth harmonic or higher because of the high electromagnetic noise during measurement. Fig. 6(a) shows the result of operating the FFL at ± 500 A with an inclined magnetic field strength of 0.37 T/m and mechanically rotating the sample at a pitch of 3° to obtain the detection signal. Fig. 6(b) presents an image reconstructed from a sinogram.

IV. Conclusions

We developed a human-head-sized MPI system with a 300 mm bore diameter. The developed MPI system imaged at a low frequency below 1 kHz, and the power supply's footprint was reduced to about 2.0 m^2. An M-shaped phantom of about 140 mm^2 was successfully imaged under an alternating magnetic field condition of 500 Hz. In the future, the SNR of the high-order harmonic signal will be improved to increase the image quality, and a scanning method will be optimized to speed up the measurement time.

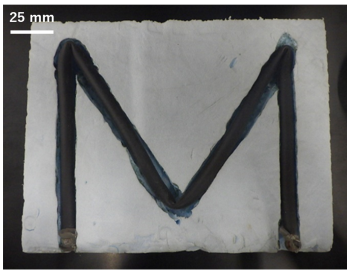

Figure 5: Photograph of phantom encapsulated in a tube with a 6-mm inner diameter

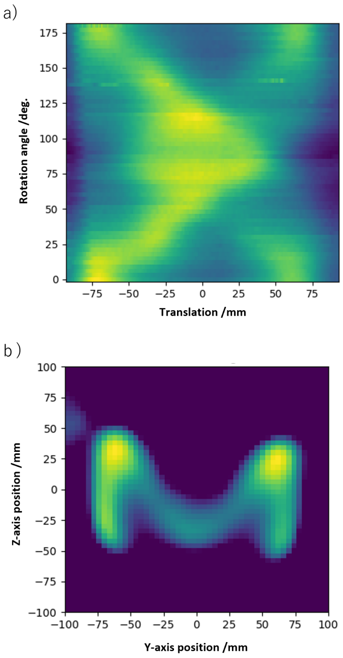

Figure 6: a) Sinogram of third-harmonic signal: Horizontal axis is FFL position, and vertical axis is relative angle between the FFL and the phantom. b) Resulting reconstructed MPI image by sinogram of third-harmonic signal.

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Author's statement

Conflicts of interest: The authors declare no conflicts of interest. Informed consent: Informed consent was obtained from all individuals included in this study.

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