Design of Rotated Surface Coil Array for Multiple-Subject Imaging at 400 MHz

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ABSTRACT Multiple-subject magnetic resonance imaging (MRI) has been used in phenotyping studies to reduce the total experimental time by simultaneously capturing images of multiple animals using single or multiple radio frequency (RF) coils. However, conventional methods must consider using decoupling circuits to minimize the mutual inductance coupling between RF coils. These decoupling circuits unnecessarily increase the field of view (FOV) and image acquisition times. In this study, we propose a rotated surface (RS) coil that can generate a uniformly transmitted magnetic field (|B1|+) and minimize the distance between the coils when used for multiple-animal imaging without decoupling circuits. The RS coil was designed using electromagnetic (EM) simulation software, and the rotation angle, which creates a uniform |B1|+-component, was derived by mathematical calculations. We compared the |B1|+-component uniformity produced by RS coils with different rotation angles and a 4-leg high-pass filter birdcage coil (HBC) in multiple-coil arrays at the same location using EM simulations. The RS coil displayed greater than 94% and 84% uniformity when used as a single-channel and multiple-coil array in the central axial plane, respectively. Following the EM simulations, a multiple-mouse RS coil array was manufactured, and a bench test was performed to ensure its operation at 400 MHz and for coupling analysis. T1-FLASH images of the four phantoms and mice were acquired using a multiple RS coil array in a 9.4-T pre-clinical MR system. The signal-to-noise ratio (S/N) and uniformity were measured using the phantom images.

INDEX TERMS Rotated surface (RS) coil, rotation angle, magnetic resonance imaging (MRI), multiple subject, phenotyping, radio frequency (RF) coil, uniformity

I. INTRODUCTION

Animals are widely used for research and development in medicine, dentistry, and pharmacy. Each year, a large number of small animals (mice and rats) are used in biomedical studies worldwide, such as those reported in laboratories in the US and the Republic of South Korea [1, 2]. This is because the basic biology and chemistry of mice are similar to those of humans. In addition, numerous studies have revealed the genome sequences of mice and humans. The results of these genome-sequence studies have enabled researchers to develop knockout mice [3, 4]. Therefore, many studies have been conducted using knockout mice to analyze the phenotyping of anatomy, physiology, behavior, and function [5-10]. Phenotypic analysis using knockout mice continues to be conducted, and according to the International Mouse Phenotyping Consortium (IMPC) Data
Release 15.0 Notes, 7824 phenotypic genes have been studied [11].

MRI is a non-invasive imaging modality that uses non-ionizing radiation and is widely used in research and clinical fields because it can provide images with excellent soft tissue contrast and high spatial resolution. It also has the advantage of being able to image various anatomical, metabolic, and functional characteristics. Phenotyping studies using MRI require a large number of mice and high-quality imaging. Phenotypic analysis includes comparative studies of wild-type and knockout mice. Identifying the genes corresponding to the origin of a disease is difficult to accomplish using phenotyping studies, which require a large sample size and even larger numbers of mice when the phenotype is ambiguous. Because imaging each mouse individually is overly time consuming, researchers have investigated the use of radiofrequency (RF) coils for multiple-subject MRI to reduce the total experimental time by capturing images of multiple subjects simultaneously [12-16]. Birdcage coils have generally been used in multiple-subject MRI, with the most important requirement that each coil operates individually by minimizing the mutual inductance coupling between each coil in multiple radio frequency (RF) coil arrays. Therefore, multiple-birdcage coil arrays must use an RF shield or decoupling capacitors to minimize mutual inductance coupling [13-16]. Owing to the decoupling circuitry, the volume of each coil increases, and the distance between the coils increases, which unnecessarily increases the field of view (FOV). Additionally, phenotyping studies using mice that require images with high spatial resolution are more frequently performed on ultra-high-field preclinical MRI. This is because the gradient performance of preclinical MRI is higher, and the signal-to-noise ratio (S/N) also increases with the increase in the main magnetic field (|B₀|) strength. However, pre-clinical MRI has a much smaller bore size compared to clinical MRI; the stronger the |B₀|, the smaller the bore size of pre-clinical MRI. This complicates the placement of many bulky birdcage coils with decoupling circuits.

In this study, we propose a rotated surface (RS) coil that exhibits a highly uniform transmitted magnetic field (|B₁|) in the axial plane. Uniformity was achieved by rotating the conductors on both sides of a conventional surface coil clockwise along the z-axis. The Biot–Savart equation was used to determine the rotation angle (θ) that produces the most uniform |B₁|, and the results were confirmed using an electromagnetic (EM) simulation software (Sim4Life, ZMT, Zurich). The |B₁| strength and uniformity of the RS coil were compared for each angle of the RS and the 4-leg high-pass filter birdcage coil (HBC). For application to multiple-subject MRI, the angle between adjacent RS coils was kept at 90° spatially, to reduce mutual inductance coupling by designing each coil such that it produces its |B₁| component orthogonally. T₁-FLASH images of four phantoms and mice were acquired using a multiple-subject RS coil array in a 9.4-Tesla (T) preclinical MR system.

II. Materials and Methods

A. Design of the Rotated Surface Coil

The design of the RS coil consists of a surface coil shaped into a cylindrical saddle coil (Ci) with diameter (D₀) and length (L₀) of 30 mm and 70 mm, respectively. Fig. 1 illustrates how the rotation of the coil was performed for the evolution of rotation angles from 0° to 80°. The conductor line with the input source is kept static, while the conductor lines along the z-axis are rotated by an angle (θ), and the face of the conductor is kept orthogonal to the inside cylinder; to achieve this, the conductor is wrapped around cylinder Ci. The distance between the wrapped conductor lines is described by the rotation angle and radius (Rₛ) of Ci as follows:

\[ W_{rs} = \pi R_{rs} \cot(\theta) \]  (1)

FIGURE 1. The RS coil concept illustrating the geometry evolution for rotation angles up to 80° in steps of 5°.

Conventional surface coils have the advantages of higher radiofrequency magnetic field (|B₁|) strength and sensitivity as they are closer to the coil, but they also have the disadvantage of low |B₁| uniformity. We considered two factors to overcome the non-uniform |B₁| generated by the surface coil. The first is to ensure a sufficient penetration depth, and the second is to set the magnitude of the magnetic flux density at two points of reference. Point |dB₁| is defined as the center of the distance separating the conductor lines along the y-axis, and point |dB₂| is the distance between adjacent conductors along the z-axis, the locations of which are illustrated in Fig. 2a with a blue circle (|dB₁|) and an arrow (|dB₂|), respectively. Fig. 2a gives a
detailed description of the geometry of the proposed coil [17, 18].

The penetration depth of a surface coil is determined by the radius of the circle and half width of the rectangle [19]. Similar to the case of the proposed coil, the penetration depth depended on the angle and $W_r$. Therefore, for the stabilized coil diameter, $W_r$ should be at least 30 mm to achieve a penetration depth of at least 15 mm, which is the radius of the RS coil ($R_n$). Setting the condition $W_r \geq 2xR_n$ and solving (Eq. 1) for $\cot(\theta)$, it was found that a range of rotating angle was $2/\pi$ or less, for which the rotation angle must be $57.5^\circ$ or less. Using the Biot–Savart equation, we calculated the rotation angle at which $|dB_1|$ was equal to $|dB_2|$ in free space. The permeability of free space ($\mu_0$) was set to $4\pi \times 10^{-7}$ H/m, and the direct current (I) flowing through the conductor was set to 1 A. The differential length (dl) is a point in the conductor, and the distance (r) is equal to $R_n$. Under the condition that $|dB_1|$ and $|dB_2|$ should be equal, $|dB_1|$ and $|dB_2|$ can be used to obtain the following relationship:

$$|dB_1| = |dB_2| \cdot \frac{k}{R_{rs}} = \frac{k}{(\pi R_{rs} \cot \theta/2)^2} \Rightarrow \cot \theta = \frac{2}{\pi}$$  \hspace{1cm} (2)

The $k$ in Eq (2) is defined as $\mu_0 dl/4\pi$. Solving Eq (2), the rotation angle was found to be $57.5^\circ$. Based on Eq (2), we found that a minimum rotation angle of $57.5^\circ$ can maintain a sufficient penetration depth and ensure that $|dB_1|$ and $|dB_2|$ are equal. Because the RS coil generates the $B$-component in the $xy$-plane, it can be aligned in the $z$-axis direction. Fig. 2b shows an example of the magnetic flux density inside the proposed coil.

### B. EM Simulation Set-up

The geometry of the RS coil derived from mathematical calculations was modeled using the EM simulation software. The $[B_z]^*$ distribution of the RS coil was compared for each rotation angle and the 4-HBC. All conductors of the coil were specified as perfect electric conductors (PECs), and the width of all conductors was set to 5 mm. The input sources of all coils were normalized to an EM input power of 1 W at a center frequency of 400 MHz and bandwidth of 800 MHz for accurate comparison. The cylindrical phantoms were set to be 24 mm in diameter and 60 mm in length, and they were placed in the center of each coil. The electrical properties of the phantom were set as follows: fat ($\sigma = 0.0806$ S/m, $\varepsilon_r = 11.6227$) and brain ($\sigma = 1.03038$ S/m, $\varepsilon_r = 55.9962$), at 400 MHz. The $[B_z]^*$ obtained through the EM simulation were analyzed by calculating the following three values using MATLAB (MathWorks, Natick, MA).

$$\bar{P} = N^{-1} \sum_{i=1}^{N} x_i$$ \hspace{1cm} (3)

$$S = \sqrt{(N-1)^{-1} \sum_{i=1}^{N} |P_i - \bar{P}|^2}$$ \hspace{1cm} (4)

$$NAAD = 100 \times (1 - (N \cdot \bar{I})^{-1} \sum_{i=1}^{N} |P_i - \bar{P}|)$$ \hspace{1cm} (5)

$P_i$ is the $[B_z]^*$ intensity of each pixel, and $N$ is the number of pixels. The $[B_z]^*$ uniformity was evaluated by calculating the standard deviation ($S$) and normalized absolute average deviation (NAAD) [20]. We tested two arrangements consisting of four RS coils; the 1×2×1 arrangement consisted of one coil at the top, two coils in the middle, and one at the bottom. The 1×4 configuration consisted of all coils placed in the same row. The mutual inductance coupling between the coils was measured so that it could be used for multiple-subject MRI. To ensure that the $[B_z]^*$-component generated between adjacent coils is orthogonal, the difference in the spatial angle between adjacent coils was maintained at 90°. In the 1×2×1 arrangement, the center-to-center spacings of the adjacent coils were set to 34 mm and 48.08 mm, respectively. In the 1×4 arrangement, the center-to-center spacing of adjacent coils was set to 34 mm.

FIGURE 2. The RS coil (a) geometry with the respective design variables, (b) the B-field
C. Manufacturing of Multiple-Subject Coil Array

A multiple-subject RS coil array is manufactured based on the results of the EM simulation. The resonance frequency and characteristic impedance of each RS coil were tuned to 400 MHz and matched at 50 Ω using non-magnetic fixed capacitors with \( C = 3 \) pF and \( C_M = 470 \) pF (Dalian Dalicap Technology, Liaoning, China) and variable capacitors with \( C_V = 8-30 \) pF (EW Electronics, Chatsworth, CA, USA). In the bench test, the performance of the multiple-subject RS coil array was measured using a vector network analyzer (N9913A, Keysight, Santa Rosa, CA). The tuning and matching conditions of each coil and mutual inductance coupling between the coils were measured. For measurements that were conducted using a vector network analyzer, all unmeasured coils were connected to a 50 Ω termination. We manufactured a 4-way Wilkinson power divider that evenly distributes the transmit (Tx) output power of the MR system to each coil [21, 22]. Four T/Rx switches were manufactured [22].

The transmitting/receiving operation of the coil was controlled using a coaxial cable with an electrical length of \( \lambda/4 \) (RG-316, 90 mm) and a PIN diode (MACOM, Lowell, MA). Coaxial cables of the same length were used to connect each coil and additional circuits, to maintain the same phase. The conditions of the 4-way Wilkinson power divider and each T/Rx switch were measured using a vector network analyzer.

D. MRI Experimental Set-up

MR images were obtained using the T1-FLASH sequence (TR/TE, 200/4 ms; FA, 30°; FOV, 80 × 80 mm²; matrix, 512 × 512; resolution, 156 × 156 μm²; slice thickness, 0.5 mm) in a 9.4-T preclinical MR system (Bruker, Ettlingen, Germany). The axial plane of the cylindrical phantom (0.044 g MnCl₂ + 0.0667 g NaCl per 1000 g distilled water) [23] images was acquired, and the S/N and NAAD of each phantom region were measured in the single-channel receiving state. The S/N was measured using the NEMA MS 1-4 method [24], and the uniformity was measured using the same method used in the simulation. Four mice (C57BL/6 (wild-type), 26 g) were anesthetized by injection and imaged. Animal research protocols were approved by the Sungkyunkwan University Institutional Animal Care and Use Committee (IACUC).

III. Results

The \( |B_1|^+ \) distribution in the single-channel RS coil is illustrated in Figs. 3 and 4 for fat and brain phantoms, respectively. We set the region of interest (ROI) to 75% of each phantom area. The results of the \( |B_1|^+ \) analysis of ROI are presented in Tables 1 and 2 for the fat and brain phantoms, respectively. The highest \( \bar{P} \) values for the fat and brain phantoms were for the RS coils at 57.5° and 65°, respectively. In terms of uniformity, the NAAD metric indicated that the 55° angle provided the best performance. In the case of SD, the RS coil with 80° had the lowest value; however, it had a
poor $|\mathbf{B}_1^+|$ intensity. Therefore, it cannot be said that it generated an optimal $|\mathbf{B}_1^+|$. Additionally, the $|\mathbf{B}_1^+|$ intensity 1-D profile crossing the two points, as marked in Figs. 3a and 4a, was compared with those of the RS coils with $0^\circ$, $55^\circ$, and $57.5^\circ$ and the reference 4-HBC, as illustrated in Fig. 5. The NAAD measurements are summarized in Table 3. In the case of $P_1$-$P_2$, $P_3$-$P_4$, and $P_9$-$P_{10}$, the RS coil with $55^\circ$ displayed the highest NAAD value, and the 4-HBC had a better performance for $P_5$-$P_6$, $P_7$-$P_8$, and $P_{11}$-$P_{12}$. Based on the previous results, the RS coil of $55^\circ$ was expanded to a 4-channel array to apply multiple-subject MRI. The RS coils were configured in $1\times2\times1$ and $1\times4$ arrangements, and the scattering parameters (S-parameters) and $|\mathbf{B}_1^+|$ distributions had the same arrangements as the 4-HBC. Both multiple-RS coil arrangements showed a reflection coefficient ($S_{ll}$) below $-65$ dB and a transmission coefficient ($S_{ij}$) of $-30$ dB; however, multiple 4-HBC arrangements had severe inductance coupling between each coil. The $|\mathbf{B}_1^+|$ for these arrangements is illustrated in Fig. 6, and the statistical analysis is summarized in Table 4.

### TABLE I

| $\bar{P}$ [μT] | 0° | 5° | 10° | 15° | 20° | 25° | 30° | 35° | 40° |
|----------------|----|----|-----|-----|-----|-----|-----|-----|-----|
| 24.2454        | 23.9955 | 23.8655 | 23.6918 | 23.8623 | 24.2639 | 24.6624 | 25.5524 |
| 0.34574        | 0.42473 | 0.90268 | 1.2242 | 1.2648 | 0.83734 | 0.65234 | 0.84203 | 0.54493 |
| 88.7916        | 89.5847 | 89.9523 | 90.2892 | 90.8797 | 91.2603 | 91.9791 | 92.7881 |

### TABLE II

| $\bar{P}$ [μT] | 0° | 5° | 10° | 15° | 20° | 25° | 30° | 35° | 40° |
|----------------|----|----|-----|-----|-----|-----|-----|-----|-----|
| 26.2536        | 30.352 | 29.8832 | 31.3433 | 27.7599 | 28.2986 | 18.7108 | 30.2926 | 0.29737 |
| 0.86161        | 0.5554 | 0.18463 | 0.51809 | 0.64891 | 0.93972 | 0.94908 | 1.3954 | 0.00166 |
| 93.4131        | 93.9454 | 94.4755 | 93.7779 | 93.0718 | 88.9481 | 83.0981 | 70.2067 | 91.8378 |

| $\bar{P}$ [μT] | 0° | 5° | 10° | 15° | 20° | 25° | 30° | 35° | 40° |
|----------------|----|----|-----|-----|-----|-----|-----|-----|-----|
| 9.7935         | 9.7221 | 9.6418 | 9.5287 | 9.4397 | 9.3774 | 9.3779 | 9.3978 | 9.5581 |
| 0.16066        | 0.14883 | 0.2903 | 0.38649 | 0.40321 | 0.27837 | 0.18235 | 0.19754 | 0.18546 |
| 90.1374        | 91.3896 | 91.5551 | 91.6344 | 91.7058 | 91.908 | 92.2567 | 93.0098 | 94.0891 |

| $\bar{P}$ [μT] | 0° | 5° | 10° | 15° | 20° | 25° | 30° | 35° | 40° |
|----------------|----|----|-----|-----|-----|-----|-----|-----|-----|
| 9.7934         | 10.034 | 10.0262 | 10.1061 | 10.06 | 10.2702 | 9.6997 | 8.8185 | 0.47369 |
| 0.28638        | 0.13647 | 0.06398 | 0.14314 | 0.2197 | 0.27694 | 0.47361 | 0.38093 | 0.00567 |
| 95.1304        | 95.6466 | 96.0128 | 95.0899 | 94.2769 | 89.9189 | 83.9831 | 70.6546 | 91.659 |
Although the multiple 4-HBC arrangement exhibits a higher $\bar{P}$ value, in terms of uniformity, the multiple RS coil has a better performance in addition to the high decoupling that was achieved with the RS coils, whereas the 4-HBC has a high coupling between coils.

The $1\times2\times1$ arrangement was selected for manufacturing because of its high decoupling. Multiple-mice RS coil arrays were adjusted such that $S_{ij}$ was less than $-25$ dB, and the characteristic impedance was $50 \pm 5 \Omega$ when the cylindrical phantom was loaded into the coil at 400 MHz. $S_{ij}$ between the coils was measured to be less than $-20$ dB. The parameters are summarized in Table 5. We measured the power from the Tx of a 4-way Wilkinson power divider for each coil. $S_{ij}$ between the Tx and each coil was $-6.02 \pm 0.3$ dB when using the network analyzer. In the T/Rx switch, the 

**TABLE III**

| NAAD MEASUREMENTS OF 1-D PROFILES (FIG. 5.) |
|-------------------------------------------|
| $\theta$ | $55^\circ$ | $57.5^\circ$ | 4-HBC |
| $P_1$-$P_2$ | 80.9103 | 87.2685 | 86.1146 | 76.2727 |
| $P_1$-$P_3$ | 86.1270 | 96.9555 | 93.1191 | 94.1453 |
| $P_2$-$P_3$ | 90.0016 | 89.7455 | 84.1917 | 93.4946 |
| $P_3$-$P_4$ | 85.1971 | 90.9353 | 89.8270 | 91.8902 |
| $P_4$-$P_5$ | 85.7961 | 97.4664 | 94.2645 | 91.8922 |
| $P_5$-$P_6$ | 90.0224 | 91.4699 | 86.5923 | 96.7292 |

**TABLE IV**

| $\bar{P}$, SD, AND NAAD MEASUREMENTS OF FIG. 6. |
|-----------------------------------------------|
| $\bar{P}$ [μT] | 12.9425 | 19.439 | 12.9411 | 13.2686 |
| SD [μT] | 0.99034 | 1.4804 | 0.74212 | 2.261 |
| NAAD [%] | 84.0233 | 76.8982 | 86.1564 | 62.9265 |

**TABLE V**

| S-PARAMETERS OF THE MULTIPLE-RS COIL ARRAY |
|--------------------------------------------|
| $\text{RS}_1$ | $\text{RS}_2$ | $\text{RS}_3$ | $\text{RS}_4$ |
| $\text{RS}_1$ | -25.91 | -23.08 | -21.86 | -22.70 |
| $\text{RS}_2$ | -28.92 | -26.57 | -20.27 |
| $\text{RS}_3$ | -28.82 | -21.57 |
| $\text{RS}_4$ | -27.10 |

**TABLE VI**

| S/N, NAAD MEASUREMENTS OF FIG. 7 (C-II) |
|-----------------------------------------|
| $\text{S/N}$ | 111.32 | 88.91 | 73.88 | 92.48 |
| NAAD [%] | 94.64 | 93.08 | 93.80 | 95.06 |

**FIGURE 5.** | $|B_1|$ intensity 1-D profiles: (a) fat phantom, (b) brain phantom |
DC bias, which determines the transmitting and receiving operations of the coil, is provided by the MR system. When a DC bias is applied, the Tx and coil are connected. Conversely, when a DC bias is not applied, the Rx and coil are connected. The measurements obtained using the network analyzer confirmed that the T/Rx switch operated normally. In Fig. 7a, the constructed coil arrangement is illustrated with the 9.4-T preclinical MR scanner. In Fig. 7b-i and ii, the coil array is unloaded and loaded with four mice, respectively. Each phantom and mouse were carefully mounted on each RS coil, and images were acquired, as illustrated in Fig. 7c and d for the phantom and mice, respectively. S/N and NAAD were measured for each image in Fig. 7c-ii and are summarized in Table 6. The measurements indicate that S/N greater than 73 and NAAD higher than 93% were achieved in each phantom area. These values correspond to the measurements obtained from the EM simulations. Fig. 7d-i and ii show the acquired MR images of multiple-mice brains and bodies in the axial plane. Although the images produced with the proposed coil are of good quality, there are small motion artifacts due to respiration and movement, because the ECG and respiratory monitoring systems for each mouse were not utilized. Moreover, long-term image acquisition was not possible because the anesthesia nozzle and body temperature maintenance systems for each mouse were not installed.

FIGURE 6. Multiple RF coil arrangements: 1×2×1 arrangement (a) RS coil array, (b) 4-HBC array; 1×4 arrangement (c) RS coil array, (d) 4-HBC array
IV. Discussion and Conclusion

In this study, an RS coil that creates a uniform $|B_r|^1$ was derived using a geometrical description of the coil in terms of a rotation angle and by wrapping the volumetric cylinder along the z-axis direction. Furthermore, the boundary condition for the optimal penetration depth was computed based on the Biot–Savart equation, which assumes a DC current of 1 A flowing through the conductor in free space, and the optimal rotation angle was obtained based on the uniformity of the entire volume inside the coil, but the rotation angle could vary depending on variables such as the frequency, current, and electrical properties of the loading subject. In addition, depending on the extent to which the ROI was set inside the coil, the rotation angle changed slightly. Our study confirmed that the highest uniformity was observed for a rotation angle between 54.5° and 57.5°, when 50–99% of the ROI was located inside the coil.

RS coils may not be valid for single-channel MR images; however, when extended to multiple coil arrays for multiple-mice MRI applications, their high decoupling performance can produce a uniform $|B_r|^1$, even when the coils are placed close to each other. By applying decoupling methods, it was possible to create a uniform $|B_r|^1$ in the birdcage coil using multiple coil arrays; however, the uniformity of the $|B_r|^1$ in the multiple RS coil array is expected to be superior in the axial plane. The main reason for using the multiple-mice MRI is to reduce the experimental time by imaging multiple mice simultaneously. In this regard, an RS coil that allows more coils to be placed in the same space is advantageous for reducing the overall experimental time. Imaging the same number of mice reduced the FOV and shortened the image acquisition time. High-resolution images are required for phenotyping. Using smaller voxels lowered the intensity of the signal emitted by each voxel, which reduced the S/N. Therefore, to obtain high spatial resolution images with a high S/N, phenotyping was performed on images acquired using pre-clinical MRI with ultra-high-field MRI. As the strength of the $|B_0|$ increases, the size of the bore in which the coil can be located decreases. Therefore, the RS coil, which enables more coils to be placed in a smaller space, is advantageous for multiple-subject MRI scans.

The solenoid coils can generate the highest $|B_r|^1$ intensity and uniformity depending on the number of turns; however, they must be positioned in the x- or y-axis direction. This makes it difficult to place the coil inside the MRI system, which should be optimized by considering decoupling methods. In future studies, we plan to construct a small animal cradle to ensure that the placement position of the mice, anesthesia conditions, temperature maintenance system, and animal monitoring system (ECG, respiratory) are consistent. In addition, optimization of the imaging sequence is anticipated to enable acquisition of high-quality multiple-subject MR images for phenotyping.

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