Effects of End-Ring/Shield Configuration on Homogeneity and Signal-to-Noise Ratio in a Birdcage-Type Coil Loaded With a Human Head

Wanzhan Liu,¹ Christopher M. Collins,¹ Pamela J. Delp,¹ and Michael B. Smith^{1,2*}

We modeled four different end-ring/shield configurations of a birdcage coil to examine their effects on field homogeneity and signal-to-noise ratio (SNR) at 64 MHz and 125 MHz. The configurations are defined as: 1) conventional: a conventional cylindrical shield; 2) surrounding shield: a shield with annular extensions to closely shield the end rings; 3) solid connection: a shield with annular extensions connected to the rungs; and 4) thin wire connection: a shield with thin wires connected to the rungs. At both frequencies, the coil with conventional end-ring/shield configuration produces the most homogeneous RF magnetic (B_1) field when the coil is empty, but produces the least homogeneous B_1 field when the coil is loaded with a human head. The surrounding shield configuration results in the most homogeneous B_1 and highest SNR in the coil loaded with the human head at both frequencies, followed closely by the solid connection configuration. Magn Reson Med 51:217-221, 2004. © 2003 Wiley-Liss, Inc. Key words: end-ring; shield; homogeneity; SNR; birdcage

In the design of radiofrequency (RF) coils for MRI systems. the signal-to-noise ratio (SNR) and the homogeneity of the RF magnetic (B_1) field, which affects the uniformity of the image, are two of the most important considerations. They are affected by many factors, including the current pattern in the coil and the interaction between the fields and the sample. In the absence of wavelength effects, a perfectly homogeneous B_1 field can be generated within an infinitely long cylinder that has surface currents parallel to the cylinder's axis and proportional to the sine of the azimuthal angle. However, the cylinder cannot be infinitely long in practice, and there must be some current return path. The currents can return through two end rings, as in a conventional birdcage coil (1); through the shield (2-4), as in a TEM coil (2); or through thin wire loops (5). The objective of this study was to estimate the effects of these different current patterns on the B_1 field distribution and SNR in a birdcage coil.

We modeled four birdcage coils that are identical except for the end-ring/shield configurations, which produce different available current return paths (Fig. 1). The first is a birdcage coil with the conventional cylindrical shield (conventional). The currents in the rungs can only return

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through the coil's two end rings. The second is a birdcage coil that has a shield with conductive annular extensions that closely shield the end rings (surrounding shield). As in the conventional coil, the currents can only return through the two end rings, but since the end rings are closely shielded, significant eddy currents may be induced in the shield that could affect the RF field. The third coil is a birdcage coil with a shield that has conductive annular extensions connected to the rungs (solid connection). The currents can return through either the shield or the solid annular extensions in a way similar to the currents returning through the end rings in a conventional birdcage coil. The fourth coil is a birdcage coil with a shield that has individual thin wire connections between the shield and the coil's rungs (thin wire connection). The currents in this coil can only return in a path through the shield. We examined the B_1 field distribution and SNR in these four coils at 64 MHz and 125 MHz (1.5 T and 2.93 T, the systems in our lab) using numerical methods.

MATERIALS AND METHODS

The coils were identical (27-cm coil diameter, 34-cm shield diameter, and 22-cm length) 16-rung, low-pass birdcage coils, except for the end-ring/shield configurations. Each coil was loaded with an anatomically accurate human head model (6) that consisted of 18 tissues with corresponding mass density, water content by percent mass, and electrical permittivity and conductivity at either frequency (Fig. 2). A part of the shoulder (about 4 cm thick) was included in the head model to avoid a sharp transition of fields at the end of the neck (7–9). Sixteen voltage sources, which had the same amplitude and phases equal to the azimuthal angle, were used to perform the function of capacitors in a quadrature coil at ideal mode one resonance. This method of simulation was previously proven to be accurate up to 128 MHz for a birdcage coil of this size (10). The finite difference time domain (FDTD) method (11) was used to find the steady-state RF electric field (E) and B_1 field by solving the full wave Maxwell's equations. The spatial resolution of the problem region was 5 mm in all three directions. All FDTD calculations were set up and solved with the aid of commercially available software ("xfdtd"; Remcom, State College, PA).

The magnitude of the transverse circularly polarized component of B_1 that rotates in the same direction of the nuclear spin precession, B_1^+ , was calculated on the central axial plane as (6):

$$B_1^+ = |(\hat{B}_x + i\hat{B}_y) \div 2|$$
 [1a]

¹Department of Radiology, Pennsylvania State University College of Medicine, Hershey, Pennsylvania.

²Department of Cellular and Molecular Physiology, Pennsylvania State University College of Medicine, Hershey, Pennsylvania.

^{*}Correspondence to: Michael B. Smith, Center for NMR Research, NMR/MRI Building, Department of Radiology H066, Pennsylvania State University College of Medicine, 500 University Drive, Hershey, PA 17033. E-mail: mbsmith@psu.edu

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FIG. 1. Half of the birdcage coil with a) a cylindrical shield (conventional), b) a shield with annular extensions that closely shield the end rings (surrounding shield), c) a shield with annular extensions connecting the rungs to the shield (solid connection), and d) thin wires connecting the rungs to the shield (thin wire connection).

where the circumflex ([^]) indicates the complex value, and *i* is $\sqrt{-1}$. To calculate B_1^- , the magnitude of the other component of B_1 that rotates in the opposite direction of B_1^+ , a second field calculation was performed with the phases of the voltage sources opposite those in the field calculation for B_1^+ . B_1^- was then calculated as (6):

$$B_1^- = |(\hat{B}_x - i\hat{B}_y)^* \div 2|$$
 [1b]

where the asterisk indicates the complex conjugate.

The SNR was calculated on the central axial plane in the coil, as indicated by the dashed lines in Fig. 2, using the formula (6):

$$SNR \propto f^2 \frac{\sum_{N} |W_n \sin(V | \hat{B}_{1n}^+ | \gamma \tau) | (\hat{B}_{1n}^-)^* \|}{\sqrt{P_{abs}}}$$
[2]

where f is the Larmor frequency, the summation is performed over all N voxels in the plane, W_n is the water content of the *n*th voxel, V is a normalization factor to maximize the amplitude of the total signal contributing to a reconstructed gradient-echo image with a 3-ms 90° rectangular RF pulse, \hat{B}_{1n}^+ and \hat{B}_{1n}^- are \hat{B}_1^+ and \hat{B}_1^- of the *n*th



FIG. 2. Axial (left), sagittal (middle), and coronal (right) planes in the computer model of the conventional birdcage coil loaded with the human head. The dashed lines indicate the location of the axial plane where the homogeneity of B_1^+ and SNR were calculated.



FIG. 3. Normalized B_1^+ on the central axial plane in the unloaded coil with the four different end-ring/shield configurations defined in Fig. 1. B_1^+ at the center is normalized to one.

voxel, respectively, γ is the gyromagnetic ratio of ¹H, τ is the duration of the RF pulse, and P_{abs} , the total absorbed power in the entire model, is calculated as (6):

$$P_{abs} = \frac{1}{2} \sum_{N} \left(\sigma_{xn} E_{xn}^2 + \sigma_{yn} E_{yn}^2 + \sigma_{zn} E_{zn}^2 \right) \Delta_x \Delta_y \Delta_z \qquad [3]$$

where the summation is performed over all N voxels of the entire model; σ_{xn} , σ_{yn} , and σ_{zn} are the conductivity of the material of the *n*th voxel; E_{xn} , E_{yn} , and E_{zn} are the electric field magnitude of the *n*th voxel; and Δ_x , Δ_y , and Δ_z are the dimensions of a Yee cell in the x, y, and z directions, respectively. Equation [3] is a numerical formula of the integral of the power absorbed in an anisotropic conductive medium (12). Considering the size of the coil-sample system and the field strengths used here, we assumed that sample noise dominated, and ignored the thermal noise and radiation loss of the coil (13). This method of calculating SNR has been shown to be in good agreement with experiments comparing SNRs in the human head at different field strengths (14).

The homogeneity in the head is defined as the percentage of the area on the central axial plane inside the head that has the magnitude of B_1^+ within 10% deviation from that at the center of the plane $(||B_1^+(\mathbf{r})| - |B_1^+(\text{center})||/|B_1^+(\text{center})| < 0.1)$. The homogeneity inside 90% of the coil radius on the central axial plane in the empty coil was also calculated to compare with the results of a previous study (15).

RESULTS

The normalized B_1^+ distributions on the central axial plane within a 90% radius of the unloaded coils for the four end-ring/shield configurations at both 64 MHz and 125 MHz are given in Fig. 3. The maps and contour plots of the normalized B_1^+ on the central axial plane inside the head, and the normalized $|B_1|$ on the whole central coronal plane at both frequencies are given in Figs. 4 and 5, respectively. B_1^+ at the center of the coil is equal to one.

The normalization factor, *V*, which is proportional to the input voltage required to produce a maximized amplitude of signal contributing to a reconstructed gradient-echo im-



FIG. 4. Normalized B_1^+ on the central axial plane in the head with the four different end-ring/shield configurations defined in Fig. 1. B_1^+ at the center is normalized to one.

age with a 3-ms 90° rectangular RF pulse (6), is given in Table 1 along with absorbed power (V^2P_{abs}), SNR, and homogeneity of the unloaded and loaded coil for the four different end-ring/shield configurations at 64 MHz and 125 MHz. The SNRs are all normalized to the SNR of the conventional birdcage at 64 MHz.

DISCUSSION

B_1^+ Homogeneity

In the unloaded case, the B_1^+ distribution in the conventional birdcage coil is visibly more homogeneous than that in the other three coils at both frequencies. This is consistent with the results of a previous study (15). However, when the conventional birdcage coil is loaded with the head, it produces the least homogeneous field. In the conventional birdcage coil, from the center to the edge of the head, B_1^+ drops 30% at 64 MHz and 50% at 125 MHz. In the other three coils, B_1^+ drops by only about 10% at 64 MHz and about 30% at 125 MHz.

The end-ring currents contribute to this change in homogeneity upon addition of the load in the conventional birdcage coil compared to the other coils. In the conventional birdcage coil, where the currents in the rungs can only return through the two end rings, the end-ring currents induce additional transverse field at the coil center (16). This additional transverse field can compensate for the low field strength at the center of the unloaded coil. In the loaded coil, however, the B_1^+ at the center tends to be higher than that at peripheral areas because of wavelength effects, so the further contribution to B_1^+ at the center by the end-ring currents actually decreases B_1^+ homogeneity in the head. In the coil with the surrounding shield, the currents in the rungs also return only through the end rings; however, the eddy currents in the shield induced by the end-ring currents are much larger than those in the conventional birdcage coil because of the close shielding, and partially cancel out the effects of the end-ring currents on the B_1^+ field homogeneity. In the coil with the solid connection, the end-ring currents are limited because much of the current in the rungs returns through the shield. However, since there are no end rings in the coil with the thin wire connection, no end-ring currents can exist.

The quantified homogeneity listed in Table 1 is consistent with the graphical results. In the unloaded case, the B_1^+ field homogeneity does not change significantly with the increase of the frequency from 64 MHz to 125 MHz because the wavelength (~4.7 m at 64 MHz, and 2.4 m at 125 MHz in free space) is much larger than the coil dimensions at both frequencies. Within the



FIG. 5. Normalized $|B_1|$ on the central coronal plane in the coil with four different end-ring/shield configurations and loaded with the human head model at 64 MHz and 125 MHz. $|B_1|$ at the center is normalized to one. The black line inside the coil indicates the position of the head.

Table 1

Normalization Factor, V, Total Absorbed Power (V ² P _{abs}), SNR, Homogeneity in the Empty Coil, and Homogeneity in the Coil Loaded
With a Human Head for the Four Different End Ring-Shield Configurations at 64 MHz and 125 MHz*

f	Configuration	$V^2 P_{abs}$			Homogeneity	Homogeneity
		V	Watts	SNR	in empty coil	in loaded coil
64 MHz	Conventional	57.05	1.51	1.00	92.48	78.51
	Surrounding shield	69.30	1.35	1.06	48.04	97.83
	Solid connection	58.10	1.48	1.01	56.53	96.79
	Thin Wire connection	84.30	1.56	0.98	44.81	96.79
125 MHz	Conventional	97.50	6.51	1.74	94.47	18.80
	Surrounding shield	122.90	5.97	1.85	64.54	20.80
	Solid connection	94.70	6.43	1.78	65.50	20.28
	Thin wire connection	156.20	7.37	1.66	51.32	20.02

*Homogeneity is defined as the percentage of the area on the center axial plane having B⁺₁ within 10% deviation of that at center of the coil.

head, the wavelength is much shorter (~0.49 m at 64 MHz, and 0.29 m at 125 MHz, assuming the head is filled with a material that has a dielectric constant equal to the average of that of white matter and gray matter) such that it is comparable to the dimensions of the coil. These wavelength effects account for the significantly degraded B_1^+ field homogeneity in the head at 125 MHz compared to that at 64 MHz.

SNR

The difference between the highest and lowest SNRs for different end-ring/shield configurations at a given frequency is within 11% of the maximum SNR. At both frequencies, the coil with the surrounding shield results in the lowest $V^2 P_{abs}$ and the highest SNR. This can be better understood by examining the RF field distribution in the coil-sample system. Figure 5 shows the normalized $|B_1|$ maps and the contour plots on the center coronal plane of the coils. Compared with the other configurations, the coil with the surrounding shield best contains the B_1 flux within the volume of the coil. This leads to less RF power coupling to the neck and the shoulders, and thus better SNR in the central part of the coil. A coil with a surrounding shield has not yet been used for imaging. Of the RF coils currently used in MRI systems, the end-ring/shield configuration of the TEM coil is most similar to the coil with a solid connection, which has the second highest SNR of the four coils.

For all four end-ring/shield configurations, a greaterthan-quadratic increase of V^2P_{abs} leads to a less-than-linear increase of SNR (from 64 MHz to 125 MHz) because of wavelength effects. This phenomenon was also observed in an experiment with a surface coil on a human subject's chest (17), and in a computer simulation of an unshielded birdcage coil loaded with a human head (6).

Limitations of the Methods

The performance of the coils may not be comprehensively represented in this study, since homogeneity and SNR were only calculated on the central axial plane. The center axial plane was chosen in order to be consistent with previous publications (6,7,15) and birdcage coil theory (1), which is based on the use of infinitely long cylinders. In the SNR calculation, sample noise dominance was assumed, and for simplicity the coil noise and radiation loss were not considered. This is consistent with the definition of intrinsic SNR (ISNR) (13).

CONCLUSIONS

The birdcage coil with a conventional cylindrical shield has the best B_1 homogeneity on the center axial plane in the unloaded coil at 64 MHz and 125 MHz. However, when the coil is loaded with a human head, this configuration has the lowest homogeneity at both frequencies. The coil with the surrounding shield has the least energy lost in the sample and the highest SNR at both frequencies. The difference in SNR between the different end-ring/shield configurations is within 11%.

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