Radio Frequency Probe for Improvement of Signal to Noise Ratio in Magnetic Resonance Image with Inductively Coupled Wireless Coil

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ABSTRACT

An inductively coupled wireless coil for a bidirectional radio frequency (RF) probe coil has been designed to improve the signal to noise ratio (SNR) in the magnetic resonance image (MRI). A birdcage type of a primary coil and a Helmholtz type of a wireless secondary coil was manufactured. The coils were applied to a phantom study with a 3 T MRI system. SNR was calculated in the magnetic resonance images. The experimental results show that the designed coils are effective to increase SNR and to decrease the mean RF power. The low RF power decreases a specific absorption rate to a subject at high frequency of resonance.

Keywords: Magnetic Resonance Image, Radio Frequency Probe, Signal to Noise Ratio, Inductively Coupling, Wireless Coil and Radio Frequency Power.

1. INTRODUCTION

A magnetic resonance image (MRI) system creates an image with signals from an excited subject at high frequency. Its detector is called a radio frequency (RF) probe, because the frequency belongs to the range of radio frequency. The RF probe governs the quality of the image [1]. There are two kinds of RF coils. One is a single coil, and the other is a cross coil [2]. The former coil works for both excitation and reception. In the latter coil, excitation and reception are separated into two coils respectively. In the clinical cross coil, a built-in whole body coil is used for excitation, so that a whole subject has to be excited even in the case of local measurement [3]. The specific absorption rate (SAR) tends to increase in the cross coil method.

A wireless RF coil works with inductive coupling, and has been applied in MRI systems [4]. Hout et al. proposed its clinical application to extend the area of an image for the magnetic resonance [5]. In the system, the intensity of the magnetic field was rather low with the resonance frequency of 8 MHz, so that SAR was also low and limitation of SAR had not to be considered. Graf et al. developed a wireless probe inductively coupled with volume type of RF coil of a 1.5 T MRI system for the head [6]. They applied the probe to small animals, and showed that MRI can be taken with high signal to noise ratio (SNR). In their study, transmitting RF power and SAR were not discussed. A phased array method has been widely used to improve detection efficiency of a radio frequency coil in the MRI system. In the method, SAR tends to increase, because the whole subject is excited even in the case of taking a local image. In MRI with a strong magnetic field, SAR would easily exceed the safety standard value, because SAR is proportional to square intensity of the magnetostatic field.

In the present study, an inductively coupled wireless coil for a bidirectional RF probe coil for a 3 T MRI system has been designed to improve SNR and SAR in MRI.

2. METHODS

RF probe

Signa3.0T (GE Healthcare) was used for the MRI system (Fig. 1). The RF probe consists of two coils: a primary coil and a secondary coil (Fig. 2). The bidirectional primary coil is a birdcage type, which has sixteen legs of 8 mm width each (Figs. 3 & 4). It has cylindrical shape with 200 mm diameter and 290 mm height. The primary coil generates a linear drive type of magnetic field. The secondary coil is Helmholtz type with the dimension of 120 mm diameter and 120 mm height (Figs. 5 & 6). The secondary coil is inserted into the primary coil, which has an electric feeding point for transmission and reception of signals. For the MRI system of 3.0 T, the resonance frequency is adjusted to 127.8 MHz with capacitance in the electric circuit. The excitation efficiency increases, when the resonance frequency of the secondary coil is same as that of the primary coil. The frequency was measured with the radio frequency network analyzer (Agilent Technologies, 8712ET).

Fig. 1: MRI system.

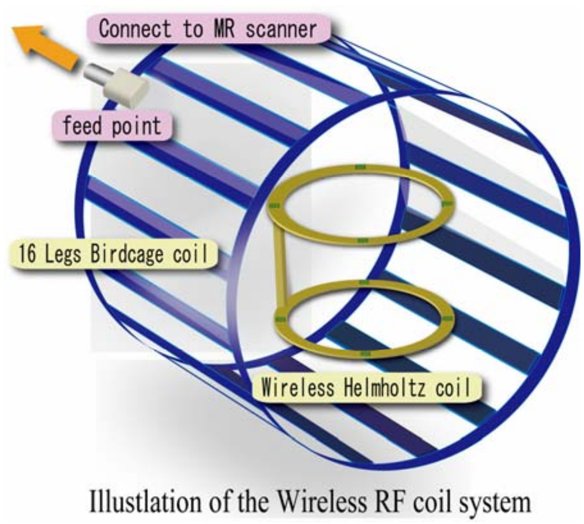


Fig. 2: Two coils system.

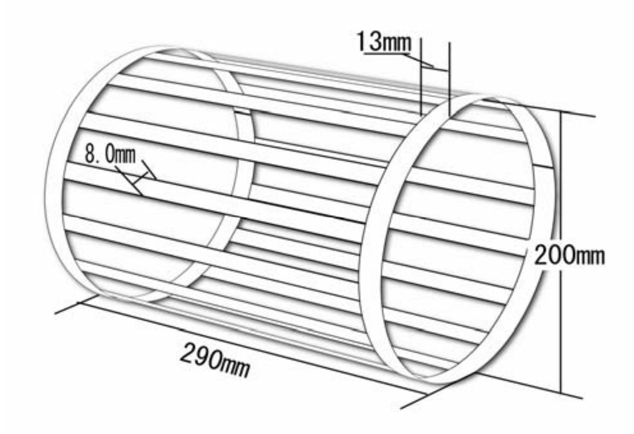


Fig. 3: Dimension of primary coil.

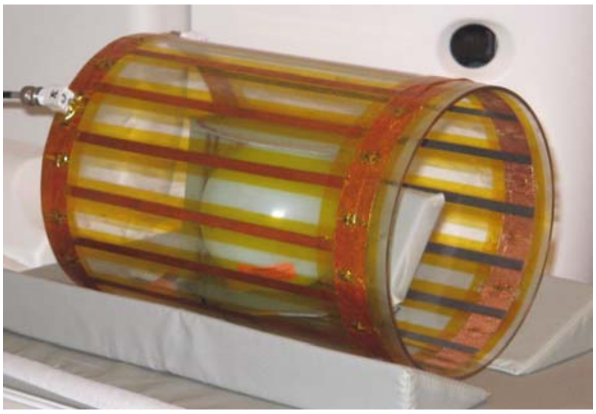


Fig. 4: Primary coil with phantom ball.

Signal to Noise Ratio

The SNR in inductively coupled coils is a function of a mutual inductance (m), a resonance frequency (f), and a resistance of a primary coil (Rp) (Fig. 7). The SNR increases, when the product of the mutual inductance and the resonance frequency is larger than the resistance of the primary coil (Rp<<mf).

The methodology to calculate SNR from I (the mean intensity of the signal in the region of interest) and B (the standard deviation of the mean intensity in the background) in equation (1) is based on the National Electric Manufactures Association (NEMA) standards.

$$SNR = I / B \qquad (1)$$

The standard deviation in the background is calculated on the image composed from subtraction signal of two images with the same parameter.

SAR

A specific absorption rate into a subject (SAR, W/kg) is calculated with the following equations.

$$SAR=0.5\ Pabs/d \qquad (2)$$

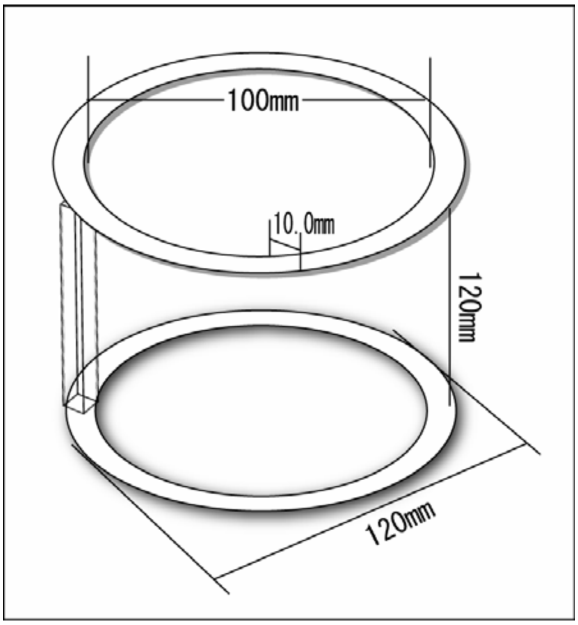


Fig. 5: Dimension of secondary coil.

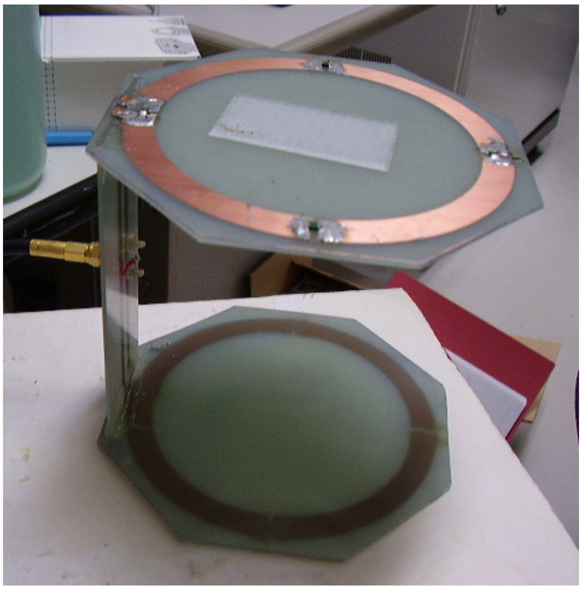


Fig. 6: Secondary coil.

$$Pall=Pabs+Prad \quad (3)$$

Where Pall is all of radio frequency power (W/m^3), Pabs is absorbed power into a subject (W/m^3), Prad is radiated power (W/m^3), and d is density of the tissue (kg/m^3). SAR can be approximately calculated with Pall in the present study, because Prad is very small compared with Pabs in the frequency range around 127.8 MHz (Eq. 4).

$$SAR=0.5\ Pall/d \quad (4)$$

The mean radio frequency power were measured to estimate SAR in the present study.

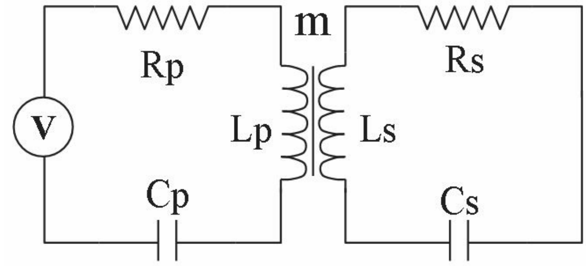
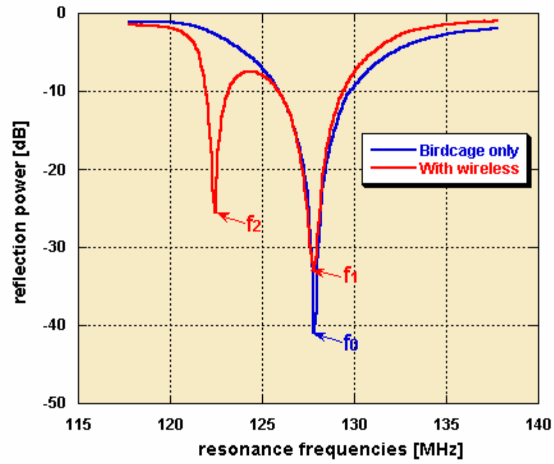


Fig. 7: Inductively coupled coils.



Resonance frequency [MHz]	f0	f1	f2	Q_{load}
Birdcage coil	127.77	–	–	639
With wireless	–	127.84	122.43	319.5

Fig. 8: Relationship between reflection power and frequency.

Phantom Study

A cupper sulfate aqueous solution packed in a plastic ball of 100 mm diameter was used for the phantom measurement (Fig, 4). To check the contrast of magnetic resonance image, an orange of 120 mm diameter was also used for a phantom.

While T1 and T2 weighted magnetic resonance images were taken at the phantom with and without the wireless coil, the transmitting gain and the mean radio frequency power (W) were measured. The maximum transmitting gain is 200, which generates the maximum transmitting radio frequency power of 8 kW.

SNR was calculated for the phantom study of the cupper sulfate aqueous solution in the ball.

3. RESULTS

Fig. 8 illustrates relationship between reflection power and

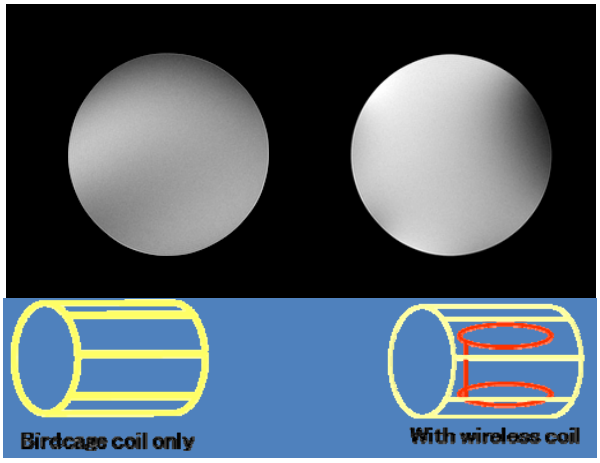


Fig. 9: T1 weighted images for cupper sulfate aqueous solution packed in plastic ball: without wireless coil (left), and with wireless coil (right).

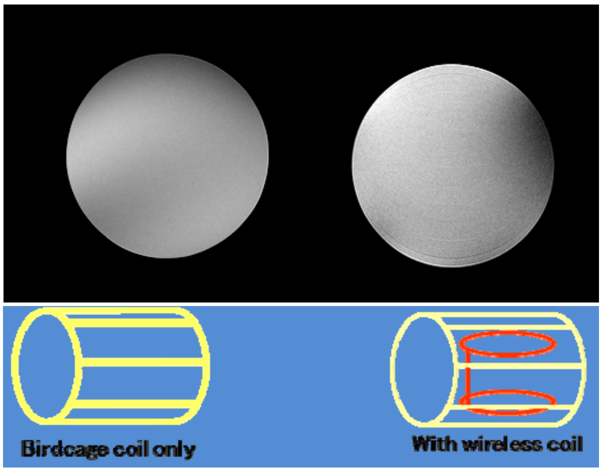


Fig. 10: T2 weighted images for cupper sulfate aqueous solution packed in plastic ball: without wireless coil (left), and with wireless coil (right).

Table 1: Transmit gain and mean RF power at ball.

	Transmit gain		RF power [W]	
	T1	T2	T1	T2
Without wireless coil	120	70	3	2
With wireless coil	20	7	0.05	0.07

frequency in the electric power measurement. The secondary coil induced the secondary resonance frequency of 122.4 MHz (f2), which is 5.4 MHz smaller than the primary resonance frequency of 127.8 MHz (f1). The experimental result shows that the primary resonance frequency does not shift with the secondary wireless coil: f1 is equal to f0.

Table 1 shows the transmitting gain and the mean RF power, while the magnetic resonance images were taken at the cupper

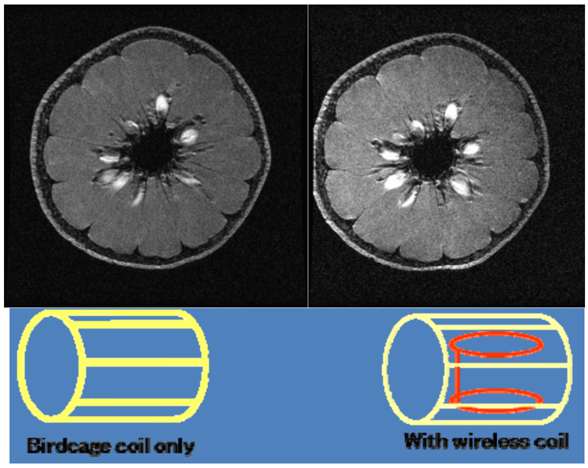


Fig. 11: T1 weighted images for orange: without wireless coil (left), and with wireless coil (right).

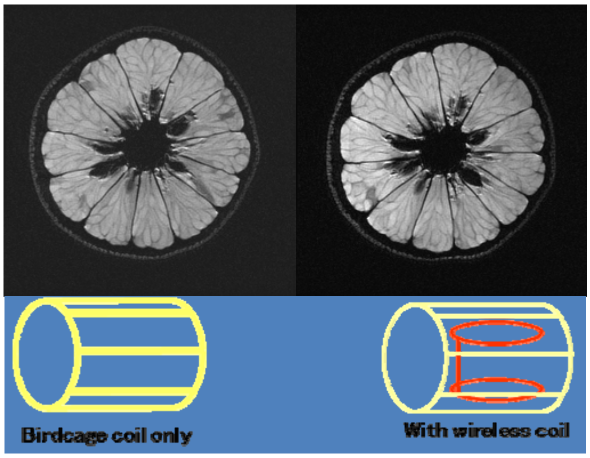


Fig. 12: T2 weighted images for orange: without wireless coil (left), and with wireless coil (right).

sulfate aqueous solution packed in the plastic ball. The results show that both transmitting gain and the mean RF power decrease considerably with the wireless secondary coil at T1 and T2.

Figs. 9 and 10 show the T1 and T2 weighted images for the cupper sulfate aqueous solution packed in the plastic ball, respectively. The left picture shows the image without the wireless coil, and the right picture shows one with the wireless coil in Figs. 9 and 10.

The calculated SNR is shown in Table 2. The results show that SNR increases with the wireless coil both in T1 and in T2 weighted images.

Table 3 shows the transmitting gain and the mean radio frequency (RF) power, while the magnetic resonance images were taken at the orange. The results show that both transmitting gain and the mean RF power decrease considerably with the wireless secondary coil at T1 and T2. Figs. 11 and 12

Table 2: SNR in phantom study.

	T1	T2
Without wireless coil	127	62
With wireless coil	151	75

Table 3: Transmit gain and mean RF power at orange.

	Transmit gain		RF power [W]	
	T1	T2	T1	T2
Without wireless coil	97	81	2.95	1.79
With wireless coil	31	17	0.03	0.5

show the T1 and T2 weighted images for the orange, respectively. The left picture shows the image without the wireless coil, and the right picture shows one with the wireless coil in Figs. 11 and 12.

4. DISCUSSION

There are several ways to improve resolution in MRI: one is to increase the resonance frequency, and another is to increase SNR. Increase the resonance frequency has an advantage to improve resolution of MRI, but has disadvantage to decrease SAR.

The signal to noise ratio (SNR) decreases as the dimension of the RF coil, so that expansion of imaging area decreases resolution of image [7]. Roemer proposed a NMR phased array, which consists of many small probes in array arrangement [8]. Their system employs the cross coil method, so that SAR tends to increase with the whole body excitation. SAR is limited to smaller value at the cross coil than at the single coil in the standard of the International Electrotechnical Commission [9]. In the previous study, transmission power and SAR have not been examined on the inductively coupled RF coils.

Although secondary resonance frequency was induced with the secondary wireless coil, primary resonance frequency was maintained at the same value in the present study. Mutual inductance is very small compared with self-inductance of the primary coil, so that the resonance frequency does not shift. The difference between first and second resonance frequencies is 5 MHz, which is large enough to be distinguished in the MRI system. Simultaneous resonance both at primary and at secondary coils decreases the mean electric power to take MRI, which decreases SAR.

Fitting the secondary wireless coil increases the magnetic filling factor, which improves SNR. Morphological preciseness of the coils would provide a uniform magnetic field, which improves MRI in the present system with the secondary wireless coil. The designed coil is simple and easily applied to a conventional magnetic resonance image system.

5. CONCLUSION

The phantom study shows that the designed inductively coupled wireless coil for a bidirectional radio frequency probe coil improves the signal to noise ratio in the magnetic resonance image and decrease the specific absorption rate into a subject.

6. ACKNOWLEDGMENT

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7. REFERENCES

[1] C. N. Chen, D. I. Hoult, ***Biomedical Magnetic Resonance Technology,*** Adam Hilger, 1989.
[2] J. Jin, ***Electromagnetic Analysis and Design in Magnetic Resonance Imaging***, CRC Press, 1998.
[3] E. B. Boskamp, "Improved Surface Coil Imaging in MR: Decoupling of the Excitation and Receiver Coils", ***Radiology***, Vol. 157, 1985, p. 449.
[4] M. D. Schnall, C. Barlow, V. H. Subramanian and J. S. Leigh, "Wireless Implanted Magnetic Resonance Probes for *in vivo* NMR". ***J Magn Reson***, Vol. 68, 1986, pp. 161-167.
[5] D. I. Hoult and B. Tomanek, "Use of Mutually Inductive Coupling in Probe Design", ***Concepts in Magnetic Resonance (Magnetic Resonance Engineering),*** Vol. 15, No. 4, 2002, pp. 262-285.
[6] H. Graf, P. Martirosian, F. Schick, M. Grieser and M. E. Bellemann, "Inductively Coupled RF Coils for Examination of Small Animals and Objects in Standard Whole-Body MR Scanners", ***Medical Physics***, Vol. 30, No. 6, 2003, pp. 1241-1245.
[7] F. D. Doty, "Probe Design and Construction," ***The Encyclopedia of NMR***, Vol. 6, Wiley, 1996, pp. 3753-3762.
[8] P. B. Roemer, W. A. Edelstein, C. E. Hayes, S. P. Souza and O. M. Mueller, "The NMR phased array", ***Magn Reson Med***. Vol. 16, No. 2, 1990, Nov, pp. 192-225.
[9] IEC 60601-2-33 Particular Requirements for Safety – Magnetic Resonance Equipment for Medical Diagnosis Second Edition 2002-5, International Electrotechnical Commission, Geneve.