하계학술대회
논문86-15-1

2.0 T 핵자기 공명 영상시스템용 고주파 코일의 설계

(Design of RF Coil for 2.0 T NMR Imaging System)

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Since the advent of superconducting magnets for NMR imaging, there has been a trend towards higher magnetic field strength for higher sensitivity or ietter signal-to-noise ratio (SNR). Although high field imaging has a engineering difficulty as well as inherent complexity to high frequency operation such as RF attenuation and phase delay in biological tissues [1, 2], the demand for spectroscopic imaging in proton [3] and lesser degree in imaging other than proton stimulated development of high field NMR systems.

As NMR imaging moves towards higher field strength, radiofrequency (RF) coils associated with the system require new design principles and fabrication techniques, since as frequency goes higher with larger dimension the length of the coil wire approaches to that of the r.f wavelength, RF coils of conventional type becomes exceedingly difficult to realize and becomes inefficient. Currently there are two types of high field RF coils gaining popularity; i.e., distributed phase and slotted tube resonator (STR) coils. Although these coils have been built and their usefulness has been emonstrated by many investigators [4-9], technical information in the published literature are scare, especially regarding the fabrication of RF coils for the fields as high as 2.0 T with the dimensions as large as 50 cm. The purpose of present paper is to report principles, fabrication details and performance of slotted tube resonator whole body RF coils at 85.14 MHz developed as an integral part of the KAIS 2.0 Tesla NMR imaging system [10].

Design of RF coils for NMR imaging experiment is usually based on the following criteria. The self-resonant frequency of the coil should be well above the operating frequency to allow appropriate tuning. Due to this requirement and the coil length limitation, the classical coils such as saddle-shaped coils and cross elliptical coils have their maximum field limitation of approximately 0.7 T in head size and 0.5 T in body size. Further improvement has been made possible by introduction of the distributed phase coil griginally proposed by Hinshaw and Gauss [4] and enable to extend the operating frequency up to above 40 MHz (1.0 T). For frequency higher than 40 MHz, with the original scheme of this distributed phase coil, has been limited by the dimension and wavelength. Q of a coil is another factor of concern. High Q not only provides good SNR in the receiving mode, but also allows low power in the transmitting mode. As it is known, the major impediments to the improvement of the Q of coils with high fields is the loss mechanisms in human body [11-13]. Although magnetically induced loss is unavoidable, it is important to reduce dielectric losses of the human body by minimizing the electric field inside the coil.

Although the ringresonator coil [5] which is seemingly a high frequency extension of the distributed phase coil, and capable of working up to 100 MHz or more the STR coil introduced recently by Alderman and Grant [8] appears to be more suitable for high frequency peration due to low electric field nature. While both coils have the structure with separated regions of distinct magnetic and electric RF fields, it is possible to select only the high electric field part of the structure by shielding so that the electric field can effectively be minimized. It has been demonstrated that, because of this low electric field, the STR coil is superior to that of the ringresonator RF coils even at the low frequency range (40 MHz) [9].

We have developed a STR type of resonators for whole body proton imaging for 2.0 Tesla NMR system by improving the frequency characteristics of the original STR. The structures of the main resonator and auxilliary parts of the STR coil are shown in fig. 1(a). STR coil consists of three parts: a main resonator, an inductive coupler (fig. 2), and an outer shield cage. The main resonator loop generates RF magnetic field. The inductive coupler links the resonator with ohe transmitter/receiver of NMR imaging system and the outer shield prevents the resonator from radiation. In fig. 1(b), equivalent circuits of the improved STR coils.

The main resonator of the STR is formed by two sheet inductors and four capacitors attached at edges of the sheets. Its equivalent circuit is in fig. 1 (b) and the resonant frequency is simply given by

$$f_o = 1 \;/\; 2\pi\sqrt{2LC_t} \qquad (1)$$

where L is the inductance of an inductor sheet and C_t is the capacitance of each capacitor attached.

Present STR body coil has an elliptic shape transaxial geometry so as to increase the filling factor by fitting the coil to human body, the lengths of the major and minor axis of the ellipsoid was 56 cm, 50 cm, respectively and the longitudinal length of the STR was 50 cm. 2mm-thick copper sheets (43 cm x 50 cm) are employed to cover the elliptic surface of the STR coil former. The inductance of each sheet was 0.16 uH, and the capacitance of each capacitor were from 8 to 17 pF.

In addition, present STR coil employed an induvtive coupling instead of the Alderman's scheme where two end guard rings and capacitive coupling [8]. Figure 1(c) shows the equivalent circuit of the STR coil with the guard rings. The resonant frequency of this circuit, f_o' is given as

$$f_o' = \frac{1}{2\pi}\;\frac{1}{[2LC_t\;(1 + C_g/2C_t)]} = \frac{f_o}{(1 + C_g/2C_t)} \qquad (2)$$

where f_o is the resonant frequency of the STR coil without the guard rings and C_g represents the capacitance between the guard ring and the main resonator of the coil. If the width of the gap is 1 cm, the C_g for whole body coil becomes about 67 pF. It is found, therefore, that the guard rings are the main obstacle for the high frequency operation. In present case, the resonant frequency is decreased as much as 58 % as compared with that of the coil without the guard rings.

A 2mm-thick copper plate (16 cm x 50 cm) is used to couple the resonator inductively as shown in fig. 2. Coupling coefficient is adjusted to be small in order to increase Q of the coil. The Q without load was found to be 430 and while in the loaded case Q was 53 at 85.14 MHz. Here the measured Q is given by the ratio of the resonant frequency to the bandwidth in which input VSWR of the coil is below 2.61.

Figure 3 shows a magnetic field map of the STR regarding the shape of the main resonator inside the STR as a part of circular cylinder whose arc angle is 80°.

Figure 4 shows an axial human body image obtained using the STR whole body RF coil. From the imaging results, it is noted that the RF magnetic field homogeneity does not depend on the coil type but on the optimization of each coil geometry.

Although ringresonator is capable of raising the operation frequency up to 130 MHz by reducing the inductance and increasing the empedance of the coaxial cable, the STR coil is considered to have higher frequency charateristics than the ringresonator coil. By decreasing the capacitances, the frequency of the coil is expected to be extendable up to 170 MHz (4.0 T) for the whole body coil.

References

1. C. C. Johnson and A. W. Guy, IEEE Proc. 60, 692 (1972)
2. H. S. Kim, Y. S. Kim, and Z. H. Cho, IEEE Nucl. Sci., 28, 142 (1981)
3. Z. H. Cho, O. Nalcioglu, H. W. Park, J. B. Ra, and S. K. Hilal, Magn. Reson. Med., 2, 253 (1985)
4. W. S. Hinshaw and R. C. Gauss, European Patent No81303522.7 (1981)
5. P. Roschmann, Society of Magn. Reson. Med. 3rd annual meeting, Aug. 13 - 17, 634 (1984)
6. S. Kan, P. Gonord, and C. Duret, Rev. Sci. Instrum. 44, 1725 (1973)
7. H. J. Schneider and P. Dullenkopf, Rev. Sci. Instrum. 48, 68 (1977)
8. D. W. Alderman and D. M. Grant, J. Mag. Reson. 36, 447 (1979)
9. A. LeRoy-Willig, L. Darrasse, J. Taquin, and M. Sauzade, Magn. Reson. Med. 2, 20 (1985)
10. Z. H. Cho et. al., to be published.
11. D. I. Hoult and R. E. Richards, J. of Mag. Reson., 24, 71 (1976)
12. D. G. Gadian and F. N. H. Robinson, J. of Mag. Reson., 34, 449 (1979)
13. D. I. Hoult and P. C. Lauterbur, J. of Mag. Reson., 34, 425 (1979)

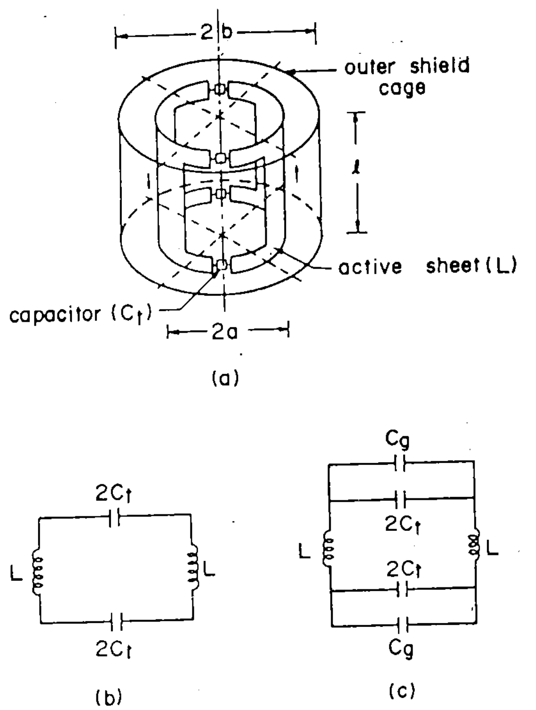

Fig. 1 Configuration of the STR coil and its equivalent circuits.
(a) Four wings are attached at edges of a sheet inductor, variable capacitors are serially connected to each wing.
(b) A sheet inductor surrounded by the outer shield is simply modelled by an inductor. $2C_t$ represents the parallel capacitance of two attached capacitors.
(c) Two end guard rings are modeled by two capacitances C_g's.

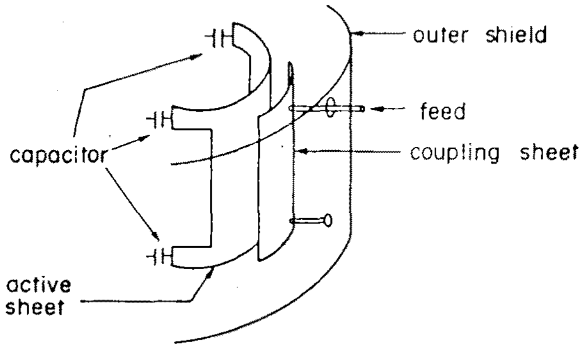

Fig. 2 Illustration of coupling schemes.
A coupling sheet is used for inductive coupling. Coupling coefficient can be changed by adjusting the width and length of the coupling sheet. Impedance matching is accomplished by a matching capacitor connected serially to the feeder

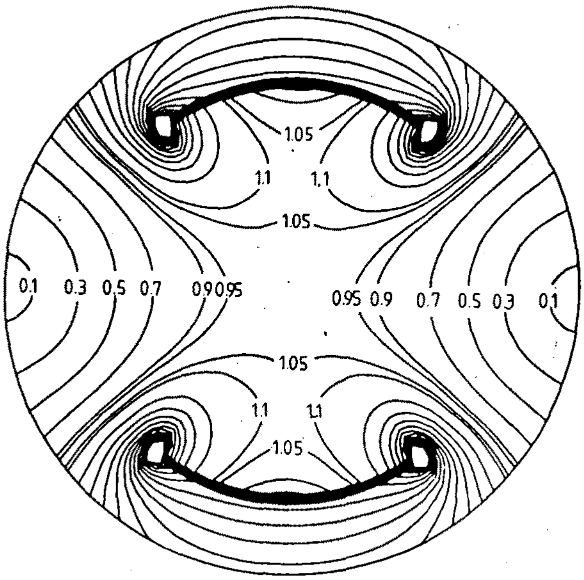

Fig. 3 Equi-level lines of B1(r)/B1(0) for the STR coil when b/a = 1.4 , and the arc angle of an inductor sheet is 80°.

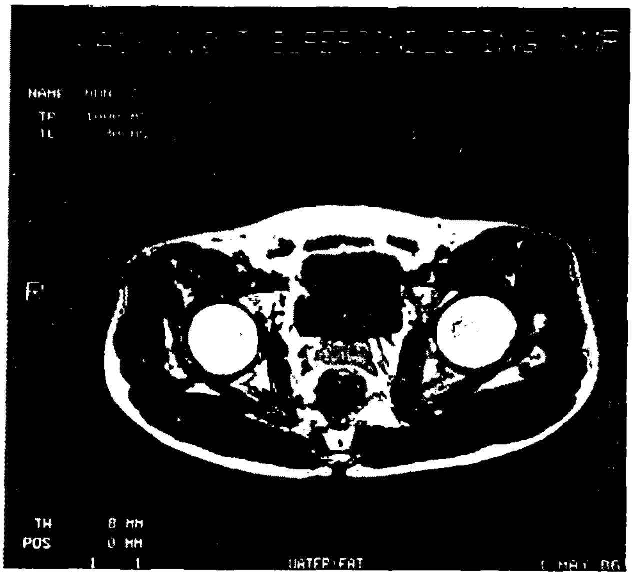

Fig. 4 An axial image of human abdomen obtained with the STR coil. (TR = 1000 msec, TE = 30 msec)