TECHNICAL NOTE

Improved SNR of Magnetic Resonance Microimaging using a Cooled Resonance Circuit at 0.3T

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Because it is noninvasive, magnetic resonance microimaging (MRMI) can be used for 3-dimensional measurement of living tissues for cell engineering. Thermal noise in the resonance circuit of the radiofrequency (RF) system of the MRMI cannot become ignored as the signal diminishes in accordance with decreasing sample size, and cooling the RF coil of the receiver circuit can effectively reduce thermal noise. We used a low temperature normal conductor circuit to reduce noise and confirmed improved signal-to-noise ratio for a conventional microimaging system at low B_0 field (0.3T) with low cost.

Keywords: *cooled, MRMI, SNR*

Introduction

Magnetic resonance microimaging (MRMI) can be used for 3-dimensional measurement of living tissues for cell engineering because it is noninvasive. Measurement of small samples at high spatial resolution requires a small field of view (FOV) and, so, a small radiofrequency (RF) coil. Because the signal decreases as the sample size decreases, the thermal noise in the resonance circuit of the RF system cannot become ignored. Cooling the RF coil of the receiver circuit effectively reduces thermal noise.

Two methods have been proposed to reduce the thermal noise of RF circuits; use of a superconductor circuit (including a high temperature superconductor [HTS]) and use of a low temperature normal conductor (LTNC) circuit. Cooling RF circuits to reduce thermal noise of a nuclear magnetic resonance (NMR) equipment is not a new idea.¹ The first nuclear magnetic resonance spectroscopy with an LTNC NMR probe cooled by liquid helium was demonstrated in 1979.² Use of a 20-cm square copper surface coil cooled by liquid nitrogen for whole body imaging at low B_0 field (0.15T) achieved an improvement factor of 1.3 signal-to-noise ratio (SNR) for *in vivo* images of the human lumbar

spine with 128×128 image matrix.³ Use of a conventional surface coil cooled by Joule-Thomson micro-refrigerator demonstrated an SNR improvement factor of 2.7.⁴ Use of a superconducting RF system may be expected to show an even higher SNR because the coil resistance is near zero, and such systems have become available for high resolution NMR. These systems can achieve higher sensitivity, and their application in MRMI is reported.⁵ Blank and associates used surface coil resonators with an HTS and measured Q factor (quality factor) exceeding 50,000.^{6,7} Using an HTS coil of 2-cm diameter cooled to 30 K for *in vivo* imaging of a rat brain, Miller and colleagues achieved a nearly 4-fold gain in SNR at 2T compared to that achieved using a warm copper coil.⁸ However, processing superconducting materials to form a circuit of complex design that maintains superconductivity for small volumes is expensive and technically difficult. In this study, we investigated the cost-effectiveness of using an LTNC for a conventional microimaging system at a low B_0 field (0.3T). We obtained relatively high resolution using a conventional system. We believe that this method will be useful for MRMI.

Theory

The noise generated in the receiver circuit is ther-

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mal noise. The Nyquist formula⁹ states that the mean square of the thermal noise voltage fluctuations in a conductor is proportional to the product of its absolute temperature T and its resistance R :

$$\langle V \rangle^2 = 4k_B TR \Delta f, \quad (1)$$

where k_B is Boltzmann's constant and Δf is the frequency bandwidth of the measurement. When the receiver circuit is a series resonance circuit, the Q of the resonance circuit is:

$$Q = \omega L / R, \quad (2)$$

where $\omega = 2\pi f$, f is the resonance frequency, L is the inductance of the coil, and R is the resistance of the resonance circuit. Then the SNR of the resonance circuit is:

$$SNR \propto \sqrt{\frac{Q}{T}}. \quad (3)$$

Therefore, the SNR is larger at higher Q and lower T . From Eq. 2, Q is higher at lower R .

Methods

The circuit of the RF transmitter and receiver (TR) system consisted of a coil and capacitors. The RF coil was designed as a 22-turn solenoid coil of 21-mm diameter using an enameled copper wire (2 mm ϕ). We made a phantom using a glass tube of 10-mm diameter surrounded by an inner vacuum insulated Dewar cylinder made of fiber-reinforced plastics (FRP) for heat insulation. We used a 33-pF fixed and a 40-pF variable capacitor for the resonance circuit and a 22-pF fixed and a 40-pF variable capacitor for the matching circuit (Fig. 1). We cooled the resonance circuit with coil and capacitors using liquid nitrogen (77 K) within an outer

vacuum insulated Dewar box made of FRP for heat insulation. The outer Dewar was surrounded by electromagnetic shields. We connected the resonance circuit with coil and capacitors to a network analyzer, tuned it to 50 Ω at a resonance frequency of 12.91 MHz, and measured the Q factor (Fig. 2). We measured the Q factor at room temperature (300 K) in the same way. The minor diameter of the phantom was 9 mm, and the volume of CuSO₄ solution in the coil was 2163 μ L. We obtained projection images of 25 mmol/L CuSO₄ solution at 77 K and 300 K using a conventional spin echo sequence. The MR imaging magnet was 0.3T. The thickness was 8 cm (60 kg, homogeneity 30 ppm @40 mm bore, Hitachi Metals Company, Japan). The MR imaging system consisted of an RF amplifier and 3-axis gradient coils. The bandwidths of the RF amplifier were 5 to 100 MHz; currents of the gradient coils, ± 5 A; and maximum gradient magnetic fields, 2.9×10^{-2} T/m. The slew rate was 9.7×10^{-2} T/m (ms) (MRTechnology, Tsukuba, Japan). Imaging parameters were: resonance frequency, 12.91 MHz; repetition time (TR), 100 ms; echo time (TE), 13 ms; number of excitations (NEX), one; image matrix, 128 $\times$ 256 pixels (one pixel was 0.47 mm $\times$ 0.17 mm [± 0.01 mm]); and bandwidth (BW), 10 kHz. Slice thickness was 9.4 mm and the first flip angle was 90°. SNR was calculated as the average of the signal area divided by the standard deviation of the background noise.

Results

We obtained MR images at 77 K and room temperature (300 K) on an X-Z plane (Fig. 2). To evaluate the actual pixel size, we obtained images of a cross phantom (9.0 mm $\times$ 21.8 mm) of plastic material at room temperature (27°C [300 K]). One voxel was 0.47 mm $\times$ 0.17 mm $\times$ 9.4 mm (± 0.01 mm) (Fig. 3). Measurements were repeated 5 times. At 300 K, the Q factor was 160 (standard deviation [SD], 8.4) and the SNR was 21.99 (SD, 1.2), respectively. And at 77 K, the Q factor was 640 (SD, 10.6) and the SNR was 36.01 (SD, 1.5), respectively (Fig. 4). The SNR improved 1.64 fold using the LTNC ($P < 0.05$).

Discussion and Conclusions

The spatial resolution of MR images depends on the SNR . The image area is smaller with larger SNR . As the strength of the static magnetic field increases, the MR signal also increases, whereas T_2 or T_2^* decreases. This trade-off of SNR is not a significant problem for whole body imaging with

Fig. 1. The circuit of the radiofrequency (RF) system consisted of matting and resonance circuits. The matting circuit comprised a 22-pF fixed and a 40-pF variable capacitor, and the resonance circuit, a 33-pF fixed and a 40-pF variable capacitor.

sufficient spins to produce an MR signal, but it is a critical consideration in microimaging. SNR can be improved by cooling the coil, a cost-effective option that can be added to any type of MR unit without modifying the main system.

In this study, we observed a significant reduction in the thermal noise of MR images using liquid nitrogen cooling at sub-millimeter spatial resolution. The SNR of an MR image is in proportion to

$\sqrt{Q/T}$, so we estimated a 4-fold improvement in SNR; however, we only achieved improvement of 1.64 times. This discrepancy may result from decreased conductivity of the sealing material caused by frost consolidation. Frost decreases the Q factor of the resonance circuit and decreases the SNR. The SNR at 77 K was 36.01 at 0.3T and 0.47×0.17 mm spatial resolution in this study. In conclusion, we believe that LTNC will become a practical and cost-effective method to improve images from conventional microimaging systems.

Fig. 2. Probe design for the magnetic resonance microimaging (MRMI) system developed in this study. We dipped the radiofrequency (RF) coil and resonance circuit in liquid nitrogen. The Dewar structure consisted of inner cylinder and outer box. We placed the outer Dewar box between the permanent magnets for B_0 and the gradient coil pairs. The outer Dewar filled it with liquid nitrogen, and it was surrounded by electromagnetic shields. We inserted the samples in the inner Dewar cylinder.

References

1. Hoult DI, Richards RE. The signal-to-noise ratio of the nuclear magnetic resonance experiment. *J Magn Reson* 1976; 24:71–85.
2. Styles P, Soffe NF, Scott CA, et al. A high-resolution NMR probe in which the coil and preamplifier are cooled with liquid helium. *J Magn Reson* 1984;

Fig. 3. Evaluation of the spatial resolution of this system. The left image is a photo of the phantom. The tube includes a cross-shaped plastic phantom 21.8 mm long and 9.0 mm in diameter. The center image an original image obtained by the magnetic resonance microimaging (MRMI) system. The spatial resolution is 0.47 mm/pixel (X) and 0.17 mm/pixel (Z). The right image is a corrected image with aspect ratio of 1:1 in the X and Z directions.

Fig. 4. Projection images of the tubes including 25 mmol/L CuSO₄ solution and their profiles (red lines). The left image was obtained at room temperature (300 K), and its signal-to-noise ratio (SNR) was 21.99. The right image was obtained at low temperature using liquid nitrogen (77 K), and the SNR was 36.01.

- 60:397–404.
3. Hall AS, Barnard B, McArthur P, Gilderdale DJ, Young IR, Bydder GM. Investigation of a whole-body receiver coil operating at liquid nitrogen temperatures. *Magn Reson Med* 1988; 7:230–235.
 4. Wright AC, Song HK, Elliott DM, Wehrli FW. Use of a Joule-Thomson micro-refrigerator to cool a radio-frequency coil for magnetic resonance microscopy. *Rev Sci Instrum* 2005; 76:014301–014301-7.
 5. Darrasse L, Ginefri JC. Perspectives with cryogenic RF probes in biomedical MRI. *Biochimie* 2003; 85: 915–937.
 6. Black RD, Roemer PB, Mogro-Campero A, Turner LG, Rohling KW. High temperature superconducting resonator for use in nuclear magnetic resonance microscopy. *Appl Phys Lett* 1993; 62:771–773.
 7. Black RD, Early TA, Roemer PB, et al. A high-temperature superconducting receiver for nuclear magnetic resonance microscopy. *Science* 1993; 259:793–795.
 8. Miller JR, Hurlston SE, Ma QY, et al. Performance of a high-temperature superconducting probe for *in vivo* microscopy at 2.0T. *Magn Reson Med* 1999; 41:72–79.
 9. Nyquist H. Thermal agitation of electric charge in conductors. *Phys Rev* 1928; 32:110–113.
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