**Development of a compact MRI system for a low temperature (-5 degree) room**

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## Introduction

Most biomedical MRI applications are performed at room temperature. If the objects are imaged at other temperatures, they would be placed in a temperature regulated cell space in the RF coil. However, this approach has several disadvantages. The first is that the RF coil cannot be optimized for the objects and the SNR of the MR images is sacrificed. The second is that this approach cannot be used to objects sensitive to temperature change. A solution to this problem is to install all of the MRI system except the electronic instruments in a temperature regulated room. In this work, we developed a compact MRI system for a low temperature room to perform several useful applications.

## Hardware system

The compact MRI system consists of a permanent magnet (magnetic field strength = 1.0 T, gap = 60 mm, homogeneous region = 30 mm DSV, weight = 350 kg), a 2nd order shim-coil set, a gradient coil probe, and a compact MRI console. The magnet has a yokeless design and consists of NdFeB material blocks [1]. The magnet, shim coil set, and gradient probe were installed in a low temperature (LT) room (size: 4.8 m(W) × 2.5 m(D) × 2.4 m(H) ) and the MRI console was installed in a usual temperature (~20°C) operation room next to the LT room. The temperature of the LT room can be kept at any temperature from -5 to 10°C. The control lines for transmitter, receiver, and gradient coils were connected through a hole opened in the wall between the LT room and the operation room. Figure 1 shows an overview of the low temperature part of the compact MRI system.

## Experiments

Because the permanent magnet was shimmed using magnetic materials at 25°C in the magnet factory (Hitachi Metals Co., Saga, Japan), we measured the temperature dependence of the homogeneity using a geometrical phantom. The phantom consists of stacked 3 mm-thick plastic disks with grid-shaped trenches (2 mm depth) and a cylindrical plastic container (inner diameter = 23.9 mm, length = 40 mm) filled with a baby oil (Johnson & Johnson, USA). MR images of the phantom were acquired with a 3D driven-equilibrium spin-echo sequence (TR = 200 ms, TE = 8 ms, NEX = 4, image matrix = $256^3$, voxel size = $(100\ \mu m)^3$) at -5, 0, 5, and 10°C with positive and negative readout gradients. During the image acquisition, the 2nd order shim-coil current was off at 0, 5, and 10°C, and on and off at -5°C. For a demonstration of this system, 3D images of a snowpack (depth hoar) immersed in a dodecane were acquired.

## Results and Discussion

Figure 2 shows a 2D cross section selected from a 3D image dataset of the phantom. Because the inplane axes are phase encoding directions, image distortion in this plane is caused by gradient field nonlinearity. Magnetic field inhomogeneity was calculated in the 17 mm × 17 mm × 19 mm rectangular parallelepiped region using the frequency shift (position shift) at the edges of the square trenches. Figure 3 shows the temperature dependence of the maximum and minimum of the magnetic field observed in the region. This graph clearly shows that the inhomogeneity approaches to zero at about 25°C where the magnet shimming was performed. When the 2nd order shim-coil current was turned on at -5°C, the inhomogeneity (difference between the maximum and the minimum) of the magnetic field was improved from 117 to 59 ppm. The 59 ppm value is about ±3 pixel shift in the present pulse sequence. Therefore, the geometric distortion caused by the pixel shift should be corrected for geometrical quantification of the MR images.

Figure 4 shows a 2D cross section selected from a 3D image dataset of a snowpack acquired at -5°C. Other biomedical applications such as visualization of low temperature NMR property of living tissue or animals will be promising.

Fig.1

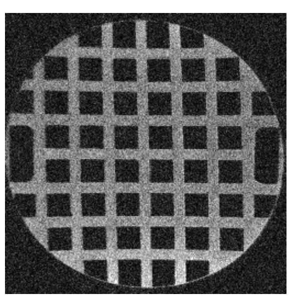

Fig.2

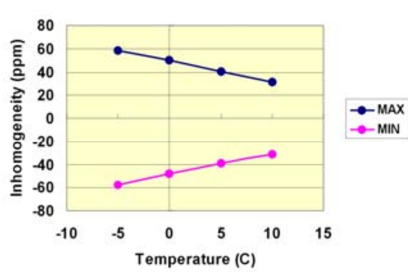


Fig.3

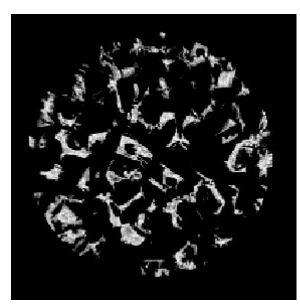

Fig.4

## References

[1] M. Aoki, Compact MRI, Kyoritsu pub. ed. K. Kose, p13-20, 2004.