Development of a temperature variable MRI system using a 1.0 Tesla yokeless permanent magnet

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INTRODUCTION Temperature is one of the essential physical or physiological parameters. In a conventional MRI system, if one wants to measure a specimen at a specified temperature, it should be stored in a temperature regulated cell (mostly) in the RF coil. In this situation, the filling factor of the RF coil is not optimized and the sample size is limited by the cell size. We have overcome these problems [1, 2] by installing the magnet, the gradient set, and the RF coil in a temperature-controlled room. However, the controllable temperature range was narrow (-5 to +10 °C) and the evaluation of the inhomogeneity of magnetic field was insufficient. In this study, we have installed a 1.0 T yokeless permanent magnet in a thermostatic bath and measured the magnetic field inhomogeneity in detail to realize MR imaging over a wide temperature range of -5 to +45 °C.

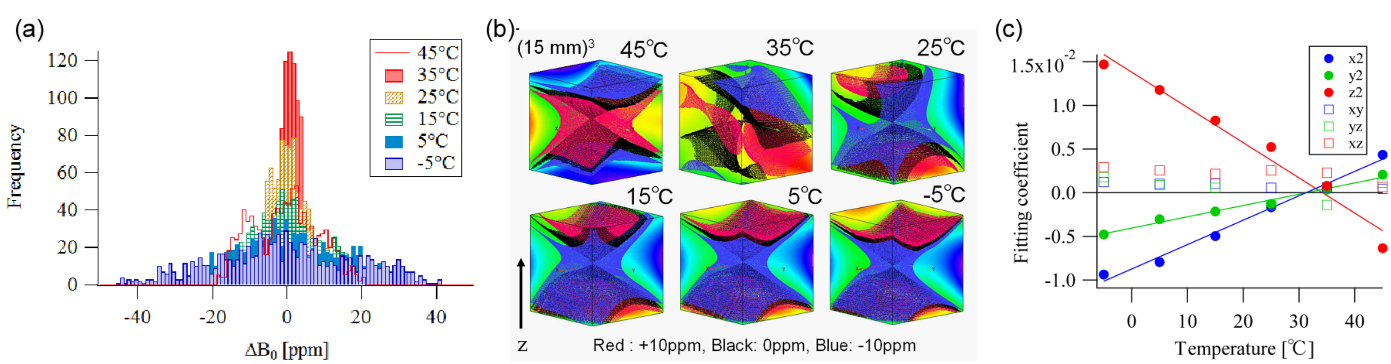
MATERIALS AND METHODS The MRI system consists of a small yokeless permanent magnet (field strength = 1.04 T at +25 °C, gap width = 40 mm, homogeneity about 20 ppm in 20 mm dsv, weight = 85 kg), a planar gradient coil set, a solenoid RF coil (diameter = 30 mm, length = 35 mm), and an MRI console. The permanent magnet was installed in a variable temperature thermostatic bath (temperature range = -15 to +50 °C, inner size = 620 mm (W) × 340 mm (D) × 1154 mm (H); FMU-263I, Fukushima Industries, Osaka, Japan).

The magnetic field distribution of the magnet was measured using a 3D lattice phantom when the temperature of the magnet was at -5, 5, 15, 25, 35, and 45 °C. The phantom consists of 11 acrylic discs (diameter = 23.9 mm, thickness = 3.0 mm) with square trenches (width = 1.0 mm, depth = 1.0 mm, interval = 3.0 mm) stacked in a cylindrical container (ID = 24.0 mm, OD = 26.2 mm, length = 62 mm) filled with baby oil. The 3D images of the phantom were measured using 3D SE sequences (matrix = 256^3, voxel size = $(125\ \mu m)^3$) with positive and negative readout gradients. The spatial coordinates of the vertex points of the square lattice in the images were detected using a GUI program. The magnetic field strengths at the vertex points were calculated from positional shifts of the vertices along the readout direction. As a demonstration, a plant specimen was imaged using a 2D SE sequence (TR=1000 ms, TE=16, 48 ms, 2 mm slice thickness, 256^2 matrix, $(100\ \mu m)^2$ pixel) at -5 and +5 °C.

RESULTS AND DISCUSSION Figure 1(a) shows temperature dependence of the histogram of the magnetic field in the central cubic area ($(15\ mm)^3$) for different magnet temperatures. The magnetic field is almost homogenous at +35 °C (5 ppm (RMS)), but it becomes inhomogenous as the temperature varies, and reaches its maximum at -5 °C (18 ppm (RMS)). Figure 1(b) shows spatial distributions of the magnetic field in the same cubic area, which were approximated using a polynomial in the Cartesian coordinate. This figure reveals that the magnetic field inhomogeneity has a specific spatial symmetry nearly independent of temperature. Indeed, the temperature dependence of the second-order polynomial coefficients (Fig. 1(c)) shows that the inhomogeneity variation is mainly caused by z^2 and x^2 terms, which is valid for different temperatures. More importantly, the magnitude of z^2, y^2, and x^2 terms has a linear dependence on temperature. This scaling behavior is reasonable because the spatial inhomogeneity of the magnetic field is mainly determined by the geometric layout of the magnetic circuit, and is changed only in magnitude as the temperature varies. The simple scaling behavior assures that the magnet field inhomogeneity can be easily corrected over a wide temperature range, for example, by using a single channel shim coil that is properly designed to compensate the inhomogeneity at a certain temperature. As an application of the temperature variable MRI system, we acquired 2D images of an okra at +5 and -5 °C (Figs. 1(d)-1(f)). These images clearly show that free water visualized in the T_2 weighted images (TE=48 ms) at +5 °C has no signal at -5 °C because the water freezes below the freezing point of water. In contrast, the signal from structured water is still visible even at -5 °C.

CONCLUSION We have developed the temperature variable MRI system using the small yokeless permanent magnet and evaluated the magnetic field inhomogeneity. The inhomogeneity has the specific spatial feature over the wide temperature range, which would be properly compensated or corrected with shim coils. The system demonstrated here has potential applications in fluid flow imaging at low temperatures and *in situ* plant imaging at various temperatures, where the accurate temperature control is a crucial design parameter.

REFERENCES: [1] S. Adachi et al. Proc. ISMRM, 2009, p3096. [2] S. Adachi et al. Rev. Sci. Instrum **80**, 054701 (2009).



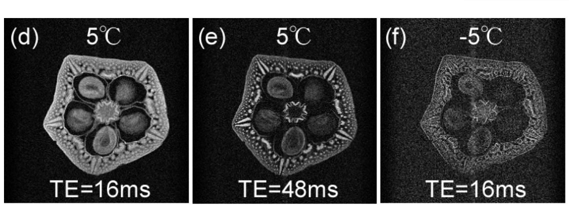


Fig. 1 (a)-(c) Temperature dependence of magnetic field inhomogeneity. (a) Histogram. (b) Magnetic field distribution in the central cubic area. (c) Coefficients of second order terms. **(d)-(f) MR images of okra at 5 and -5 °C. (TR=1000ms)** (d) TE =16 ms, +5 °C. (e) TE = 48 ms, +5 °C. (f) TE =16 ms, -5 °C.