Development of a high T_c superconducting bulk magnet with a homogeneous magnetic field using a finite element method and a single-layer shim coil

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TARGET AUDIENCE: MR engineers who are interested in the development of MRI systems using a high-temperature superconductor.

PURPOSE: A magnet using a high critical temperature bulk superconductor is the promising magnet for NMR/MRI[1,2]. The magnet can produce a strong magnetic field without liquid cryogen. However, the homogeneity of the magnet is insufficient (30-100 ppm in peak-to-peak(pp)) because of inhomogeneous crystals, and microcracks[3, 4]. In this study, the magnet structure was optimized using the finite element method (FEM). In addition, a single-layer shim coil[5,6], which can be installed in the magnet with small room temperature bore (bore diameter = 23 mm), was developed. As the results, the homogeneity of the magnet was improved to 6.9 ppm (pp) in the cylindrical volume (ϕ 6.2 × 9.2 mm).

METHODS: The bulk magnet was comprised of vertically stacked (120 mm height) six annular bulk superconductors (60 mm OD) made of c-axis oriented single-domain $EuBa_2Cu_3O_y$ crystals (T_c = 93 K) as shown in Fig. 1 (a). The bulk magnet was magnetized at 4.74 T using the field cooling method[1] as described below. At first, the bulk magnet was cooled down to 50K using a pulse tube refrigerator in a constant external magnetic field (4.74 T) produced by a conventional superconducting widebore NMR magnet (JMTC-300/89, JASTEC, Kobe, Japan). After that, the external field was reduced from 4.74 T to 0.0 T. And then the bulk magnet trapped the external magnetic field with the persistent current mode. In this study, the inner diameter (D_m) and the height (H_m) of middle four layers of the superconductors were determined by using the FEM solver (PHOTO-Series, PHOTON Co., Ltd.) as shown in Fig.2 (a).

To compensate the residual inhomogeneity of the energized bulk magnet, a single-layer shim coil[5,6] was developed using the Target-field[7] approach and the finite-difference method, because the RT bore diameter of the magnet is limited (= 23 mm). The shim coil was wound on a plastic tube with the diameter of 19.0 mm using a 0.2 mm diameter polyurethane coated Cu wire. The shim coil was fixed inside the gradient coil. The magnetic field was measured using a conventional phase-shift method with three-dimensional (3D) spin echo (SE) sequences (TR/TE = 70/10 ms, resolution = (200 μm)^3, matrix size = 64^3, and phase-shift time = 0.0 and 0.3 ms) and the 10-mm NMR tube filled with $CuSO_4$ doped water. The magnetic field distribution in the central ϕ 6.2 × 9.2 mm region of the magnet was evaluated.

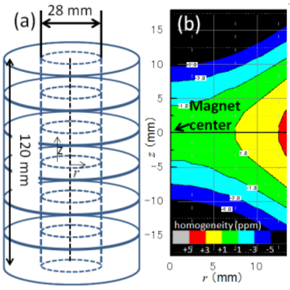

Fig. 1 (a) The structure and (b) the simulation result for the conventional magnet.

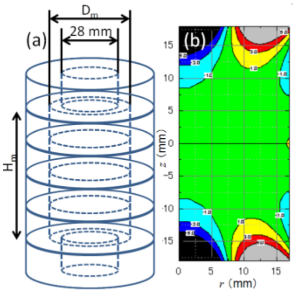

Fig. 2 (a) The structure and (b) the simulation result for the proposed magnet.

To demonstrate the performance of our system, the NMR tube phantom and a mouse embryo were acquired. A 3D gradient echo (GE, TR/TE=60/8 ms, matrix size = 128 × 128 × 32, FOV = (12.8 mm)^2) and a 3DSE (TR/TE=200/20 ms, matrix size = 256×128×96, resolution = (50 μm)^3, NEX = 8) sequences were used for the phantom and the embryo, respectively.

RESULTS AND DISCUSSION: As the result of the simulation, D_m and H_m were determined as 36, and 74 mm, respectively. Figs. 1(b) and 2(b) show the calculated magnetic field for the conventional and the proposed magnet structures, respectively. The homogeneous volumes within ±1 ppm (indicated by green color) of the conventional and the proposed structures were ϕ15 × 8.0 mm and ϕ35 × 18.0 mm, respectively. Fig.3 (a) shows the bulk magnet developed in this study. Figs. 3(b) and (c) show the field distribution of the bulk magnet, which was measured using the phase-shift method. As shown in Figs. 3(b) and (c), the homogeneity of the bulk magnet was 42 ppm (pp) in the evaluation volume, although the simulation results imply that the homogeneity of the bulk magnet was improved significantly. The difference between the simulation and the measurement results seemed to be due to microcracks, which was cased by the electromagnetic hoop stress during the magnetization process[3], and the inhomogeneity in the crystals. Theses undesirable factors were neglected in the simulation. Figs. 3 (d) and (e) are the coil winding pattern and the fabricated shim coil, respectively. The current value of 600 mA was applied for the shimming. And Figs. 3 (f) and (g) are the field distribution for the bulk magnet with shim coil. Finally, the homogeneity was improved to 6.9 ppm (pp) using the shim coil.

Fig.3 (h) and (i) are the MR images acquired with the GE sequence without and with the shim coil. There are intensity variations and artifacts due to the off-resonance effect without shimming. On the other hands, these problems were improved by using the shim coil as shown in Fig. 3 (i). And the high resolution imaging of the mouse embryo was achieved using our system as shown in Fig. 3 (j).

CONCLUSION: The bulk magnet with the optimized structure was developed using FEM to achieve a homogeneous magnetic field. In addition to this, the single-layer shim coil was developed and installed. As the result, the field homogeneity of 6.9 ppm (pp) was achieved. The imaging experiments demonstrated the performance of our magnet.

REFERENCES: [1] Nakamura T., et. al., Conc. Magn. Reson. B, 2007; 31(2): 65-70. [2] Ogawa K., et. al., Appl. Phys. Lett., 2011; 98(23): 234101. [3] Diko P, Mater. Sci. Eng.,1998; B53: 149. [4] Tamada D, et. al., Physica C, 2014; 492: 174-177. [5] Terada Y, et. al., Appl. Phys. Exp., 2013; 6(2): 026701. [6] Tamada D, et. al., Proc. ISMRM, 2012; 2578. [7] Liu W, J. Phys. D: Appl. Phys., 2007; 40(15): 4418.

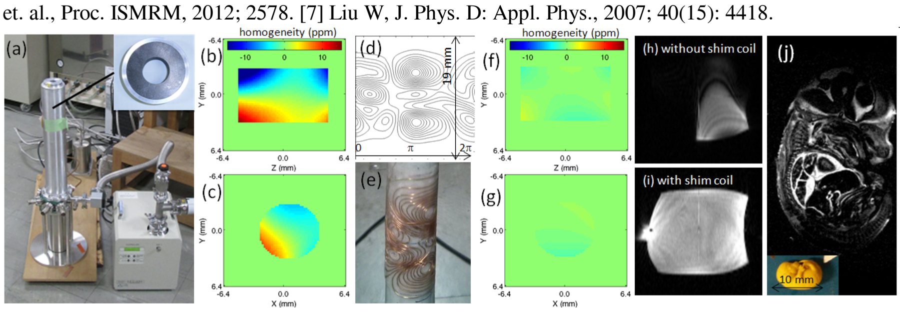

Fig. 3 (a) The developed bulk magnet. (b, c) The cross-sectional field distribution of the magnet without shimming. (d) The winding pattern for the shim coil. (e) The fabricated shim coil. (f, g) The field distribution of the magnet with shimming. (h, i) The MR images acquired (h) without and (i) with shimming. (j) The MR image of mouse embryo.