Fabrication of an MRI superconducting magnet with an off-center homogeneous field zone for imaging

M. Sekino[1], A. Miyazoe[1], H. Ohsaki[1], T. Hisatsune[1], O. Ozaki[2], T. Kiyoshi[3], and H. Wada[3]
[1]The University of Tokyo, Kashiwa, Japan, [2]Kobe Steel Ltd., Kobe, Japan, [3]National Institute for Materials Science, Tsukuba, Japan

Introduction: Early diagnosis of dementia is difficult, and medication for it becomes less effective at the point where it can clearly be detected. The Benton visual retention test (BVRT) is a widely accepted memory test where suspected patients conduct some drawing tasks [1]. Diagnosis after the test is based on the work they have done, but not including direct observations of their brain activities. Thus, diagnosis of dementia will have a much more rigid, scientific base if hippocampal activities can precisely be observed during BVRT using fMRI. In order to combine fMRI with BVRT, some modification is needed. Fig. 1 illustrates a patient carrying out the BVRT in a specially designed MRI machine. This machine has a homogeneous magnetic field zone for imaging off the center of the superconducting coil, so that the patient can have a field of vision to carry out drawing on a desk. The superconducting magnet has an asymmetric coil structure for realizing this configuration. The feasibility of asymmetric coil structures was discussed in previous papers [2], but an MRI magnet with such coil structures has never been fabricated, as far as we know. The purpose of this study is to develop a prototype novel MRI magnet with an off-center distribution of homogeneous magnetic field zone for phantom and animal imaging. The magnet generates a vertical magnetic field of 0.77 T, with a designed field homogeneity of 5 ppm within a 35-mm diameter spherical volume (DSV). The magnet represents a scale model for human fMRI, revealing that this asymmetric coil configuration can provide a sufficiently homogeneous field for imaging.

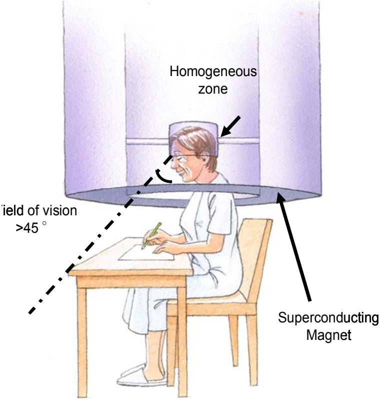

Fig. 1: Illustration of Benton visual retention test using fMRI.

Methods of Design: The distribution of a magnetic field generated by an axisymmetric coil system may be expanded using zonal spherical harmonics $B = \sum a_n r^n P_n(\cos\theta)$, where a_n is the coefficient of the n order term and P_n the Legendre function of the n order. When a current flows through a circular coil coaxial with z, the magnetic field along the z-axis is given by [3]

$$B = \frac{\mu_0 I}{2f} \sum_n \left(\frac{z}{f}\right)^n (n+1)\left[P_n(\cos\alpha) - \cos\alpha P_{n+1}(\cos\alpha)\right] \tag{1}$$

where I is the coil current, f is the distance of a point Q located on the coil conductor from the origin, and α is the angle between the z-axis and the line formed between the origin and Q. Equation (1) gives the contribution of a single circular current to the odd and the even terms in the zonal spherical harmonics expansion. In the case of an MRI magnet where the generation of a homogenous field is required, the coil configuration design is determined so that the summation of the coefficients of all order terms of the component coils satisfies $\sum_k a_n^k = 0$, where a_n^k is the coefficient of the n order component of the field generated by the k-th coil. Conventional coil design has symmetry with respect to the magnet center where the homogeneous field zone occurs. The summation of a_n^k for odd terms is zero for this coil design so, the coil configuration depends only on even terms. On the contrary, if the homogeneous field zone is not in the magnet center, symmetry is lost and odd terms must be taken into account. Equation (1) gives the fundamental relationship between the coil structure and the coefficients of spherical zonal harmonics. We combined linear and nonlinear programming methods in the calculation of the coil configuration, considering the coefficients of the 1st to 7th terms, so as to make $\sum_k a_n^k$ as small as possible.

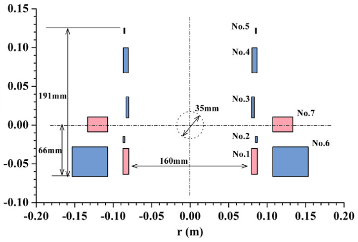

Fig. 2: Coil configuration of a prototype magnet. Dotted circle indicates the homogeneous field zone.

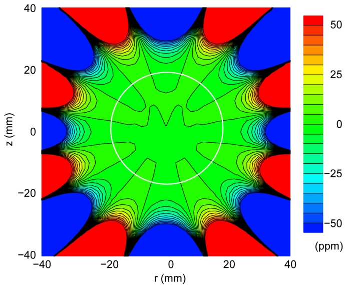

Fig. 3: Magnetic field distribution in the homogeneous field zone.

Coil Structure and Magnetic Field Distribution: The magnet is operated at 4.2 K in the persistent-current mode. There is a homogeneous field zone 29.4 mm below the magnet center. The field in the zone is 0.77 T, while the maximum field in the magnet exceeds 3 T. The magnet consists of 7 coils, as shown in Fig. 2. The direction of current in coil No. 1 and No. 7 is opposite to the direction in the rest of the coils. This configuration enabled us to have a sufficiently large homogeneous field zone in an off-center location. Fig. 3 shows the field distribution in and around the homogeneous field zone. Some fluctuation in the magnetic field due to the 8th order term remains. However, the peak to peak fluctuation is only 1.25 ppm, which is well below the designed homogeneity of 5 ppm. The coil configuration in Fig. 2 indicates that the magnet consists of 5 inner coils and 2 outer coils. There are two bobbins, one for the outer coils and the other for the inner coils. When a magnetic field of 0.77 T is generated in the homogeneity zone, an upward vertical force of ca. 3×10^3 N is exerted on the inner bobbin. This reaction causes a downward vertical force of ca. 3×10^3 N against the outer bobbin. Therefore, two plates are attached to the top and the bottom of the bobbins, in order to confine the relative displacement of the bobbins.

Fig. 4: Fabricated superconducting coils.

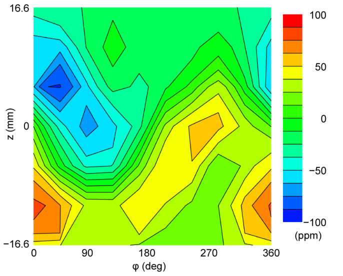

Fig. 5: Projection of magnetic field distribution measured on the surface of a 35-mm sphere after shimming.

Fabrication of Magnet: Fig. 4 shows the model magnet fabricated according to the design described above. The dark belts in the photo are the outer coils shown in Fig. 2. Precise winding is required for a homogeneous field of MRI quality. Irregular winding of a single turn of superconducting wire can cause several hundreds ppm of field inhomogeneity. In order to achieve the field homogeneity designed, the thickness of the insulator sheets between the bobbin and the winding are precisely adjusted. Superconducting joints between coils as well as coils and a persistent current switch in the present magnet do not contain a path for normal conducting material. This assures the high quality of field stability required in the persistent mode operation of the present magnet. The 8 appendices at the top of the magnet in Fig. 4 contain superconducting joints. The superconducting coils were cooled by 98 liters of liquid helium. The room-temperature bore is 89 mm in diameter. The center of the homogeneous field zone is 211 mm above the base-end surface of the cryostat.

Measurement of Magnetic Field Distribution and Shimming: The magnet was maintained in the persistent-current mode with an operating current of 84 A. The magnetic field was measured at 63 points arrayed on the surface of a 35-mm sphere using an NMR magnetometer (METROLAB Instruments, SA). The peak-to-peak inhomogeneity was approximately 1000 ppm, which was significantly larger than the designed value of 5 ppm. This deterioration resulted from various factors, including the limitations of accuracy in machining and fabrication. Similar deteriorations occur also in conventional symmetric MRI magnets. We carried out passive shimming with iron pieces to improve the inhomogeneity. Fig. 5 shows the projection of magnetic field distribution after shimming, exhibiting peak-to-peak inhomogeneity of 165 ppm. We plan to install gradient coils, a radiofrequency coil, and an imaging spectrometer to obtain MRI of small animals.

References: [1] Kawas CH et al., *Neurology* 2003;60:1089-1093. [2] Zhao H et al., *J Magn Reson* 1999;141:340-346. [3] Roméo F et al., *Magn Reson Med* 1984;1:44-65.