Construction of compact MRI magnet with superconducting shield

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Abstract. We developed a new method to design linear field gradient coils wound inside of a superconducting shield, which is attached to the outside of a superconducting solenoid. A compact palm-sized MRI magnet is constructed under these design and is tested to have enough performance.

1. Introduction

Nowadays MRI is commonly employed as a useful diagnosis tool in a hospital. However its usage should not be limited to medical applications. We have pioneered the use of MRI at ultra low temperature environment (ULT-MRI) to study quantum condensed matter.[1] It was achieved with superconducting magnet with linear field gradient coil attached to the outside of an IVC of a dilution refrigerator. This design limits the use of our technology to be used on a specially designed cryostat. Of course one could operate small superconducting magnet inside the IVC. However a possible coupling to a superconducting material, which locates near by, or another magnet as the one for nuclear adiabatic demagnetization cooling limits the possibility again. One way to avoid this problem is to attach superconducting shield outside of a compact magnet. When superconducting magnet has superconducting shield outside, one must count the effect of shielding current and adjust the winding density to produce a desired field profile in total. The design method for the case of solenoid coil in a superconducting shield was developed by Muething \textit{et al.}[2] They divided a superconducting shield into a set of rings and treated shielding currents as a set of circular currents along inner perimeters of a superconducting shield. Each circular current produces radial magnetic field on the inner surface of other rings. By summing up all the radial magnetic field contributed from each rings, they obtained a mutual inductance matrix, which related shielding currents and the radial magnetic fields. They could determine shielding currents so that the shielding current generated radial magnetic fields on the inner surface of the shield canceled radial magnetic fields, which were produced by the solenoid coil in the shield. Then, they could estimate axial magnetic field near the solenoid center, as a sum of the magnetic field produced by the solenoid itself and the shielding currents. By adjusting the size of shim layers on outside of a homogeneous layer of the solenoid, so that the total axial magnetic field near the center is as homogeneous as possible, they determined a design of their superconducting magnet with superconducting shield. This method is commonly used among low temperature physicists to design small superconducting NMR magnet, which can
be installed inside the IVC of their refrigerator. However it is not applicable to gradient coils, which are also under the influence of superconducting shield, because of a symmetry problem. Later, we used a small superconducting solenoid with superconducting shield and improperly designed gradient coils to perform the first MRI experiment below 1 mK[3]. Even though it was successful in imaging the domain structure in U2D2 He, the obtained images were blurred probably because of the nonlinear field gradient. These images were improved significantly by the use of properly designed magnet installed on the specially designed cryostat.[4] To introduce the usability of our ULT-MRI technology to other non-specially designed cryostat, we developed a new method to design gradient coils in a superconducting shield with counting the effect of shielding current.

2. Design Principle

The design principle of Muething et al.[2] assumes the rotational symmetry, which is very adequate in their analysis because all the components, solenoid coils and shield, are symmetric in that sense. However, when one need to introduce linear field gradient in three orthogonal axes, which are necessarily in MRI measurement, the symmetry breaks down. Let’s take the axes so that the solenoid axis is parallel to z and is perpendicular to x and y. For the case of linear field gradient in z axis, \( \frac{\partial B_z}{\partial z} (G_z) \), a split pair of coils with opposite current direction located at the same distance \( z_0 \) from the field center as shown in Fig.1 is necessarily. In this case, one can handle by simply introducing antisymmetric part in negative z axis. Then a condition to have smallest deviation from linear field gradient \( \frac{\partial B_z}{\partial z} \) near the center, \( x=y=z=0 \), was found by adjusting the location of coils \( z_0 \).

However in the case of linear field gradient in x and y axes, namely \( \frac{\partial B_x}{\partial x} (G_x) \) and \( \frac{\partial B_y}{\partial y} (G_y) \), situation is more complicated. One need to introduce two pairs of saddle coils, as shown in Fig.2. In this case more elaborated numerical method like as FEM might be necessarily to handle shielding current properly. However, as will be explained later in detail, simplified approximation can be used for this case since the shielding current must be conserved on the entire surface of the shield.

As shown in Fig.3, we divide a superconducting shield into a number of small panels, each of which is surrounded by two arcs and two line segments. Then we approximate shielding current on the inner surface of superconducting shield as a summation of loop currents on the rim of the each panels. By taking this approximation, a current conservation on the surface of superconductor is automatically fulfilled. Each loop current produces radial magnetic field on the inner surface of other panels. We approximate the field strength with the value at the center of each panel. By summing up all the radial magnetic field contributed from each loop currents, we obtain a mutual inductance matrix, which relates shielding loop currents and the radial magnetic fields. Then we could determine the value of each loop currents, so that they cancel radial magnetic fields, which are produced by the saddle coils. After this evaluation, we can estimate the field distribution near the center of magnet. Our results indicate that 22 (in z axis) \( \times \) 20 (in x-y circumference) panels are enough for this design method with \( \sim 10^{-2} \) accuracy. Again, the shape parameters, \( z_0 \) and \( z_1 \), are adjusted so that field distribution near the center of magnet has the smallest deviation from linear field gradient \( \frac{\partial B_z}{\partial z} \) or \( \frac{\partial B_x}{\partial x} \). The value of angle \( \alpha \) is set to 60 degree during this adjustment, because the estimated linearity was not very sensitive to the angle near 60 degree.

3. Construction of magnet

Following the design consideration, we have constructed a small palm-sized MRI magnet with Nb shield, whose size is 60 mm in diameter and 110 mm in length. Main solenoid coil was designed following the Muething et al.[2] method with improved design adjustment scheme. This magnet can produce magnetic field up to 0.8 T at 4 A without exceeding \( H_{C1} \) of Nb shield, above which
the field distribution is not as homogeneous as designed due to the penetration of flux into a shield. Three gradient coils, namely $\frac{\partial B_z}{\partial x}$, $\frac{\partial B_z}{\partial y}$, and $\frac{\partial B_z}{\partial z}$, was designed as explained above. Each windings were adjusted so that three coils had equivalent field-current-ratio, 0.18 T/m/A. The assembled magnet system was tested at pumped $^4$He bath with liquid $^3$He in cylindrical sample cell whose size was 5 mm in diameter and 5 mm in length, as shown in Fig.4. The observed NMR spectrum gave a field homogeneity of $7 \times 10^{-5}$ with field compensation by three gradient coils of $\frac{\partial B_z}{\partial x}$, $\frac{\partial B_z}{\partial y}$, and $\frac{\partial B_z}{\partial z}$. The achieved field homogeneity is acceptable for our MRI measurement, where the typical size of a sample is in a millimeter scale.

Due to the compact size, this MRI magnet system can be installed into a working space of a standard nuclear adiabatic demagnetization refrigerator or a moderate sized dilution refrigerator.
Figure 4. Schematic view of the assembled magnets and superconducting shield and test sample cell for NMR of liquid $^3$He.

The stable operation of this sort of superconducting magnet in a vacuum space of a refrigerator needs some care not to generate extra heat load into refrigerator. Once we filter out the remaining ripple currents coming from current sources for driving these coils, and after experiencing the excess heat generation due to the flux penetration into superconducting wire during the first time magnetization after cooling down, we can normally ramp up and down the current for each coils without disturbing the refrigerator as far as the sweeping speed is moderate enough not to generate eddy current heating on the metallic pieces near magnet.

References
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