Design of Open High Magnetic Field MRI Superconducting Magnet With Continuous Current and Genetic Algorithm Method

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***Abstract*—An optimization design method of short-length actively shielded and open structure superconducting MRI magnets is suggested in the paper. Firstly, the section of the solenoid coil is simplified as a current loop with zero section to solve a linear programming problem. The position coordinates in the radius and axial, and current for the loop can be calculated by the linear programming method. Then, the cross-section of the coil is optimized with a genetic algorithm to get appropriate section size. The method of linear programming, especially combining with genetic algorithm, reduces optimizing variables, which makes the design of a magnet feasible. Based on the method, a full open MRI superconducting magnet is designed with maximum radii of 0.8 m and 1.2 m. In the paper, the detailed optimization technologies are presented.**

***Index Terms*—Magnetic resonance imaging (MRI), nonlinear optimization methods, open structure MRI magnet design.**

I. Introduction

THE USE OF Magnetic Resonance Imaging (MRI) for medical diagnosis has grown tremendously over past two decades. Compared to the other imaging modalities, such as X-ray and CT, MRI is non-invasive and has good image quality. The length of the superconducting magnet heretofore typically has been between 1.6 and 2.0 m, often making the MRI system appears confining. The imaging region within the bore has typically been a 40–50 cm in diameter sphere (or diameter sphere volume-DSV) [1]–[3]. In order to improve patient comfort and to accept high field systems, the compact structure and open magnet design have become a new trend. It is desirable to have new and better devices and techniques for biomedical MRI applications such as open magnet MRI systems for imaging while performing surgery or other treatments on patients or for patients that have claustrophobia. In order to obtain the sphere shape volumetric region with reasonable field homogeneity and field strength, the design of the short-length or open MRI system is at the price of low energy efficiency and high cost. A hybrid numerical method for nuclear magnetic resonance magnet design is developed. The method combines the inverse current density approach and genetic optimal nonlinear optimization numerical techniques. An inverse approach by current density method is used to find a suitable current density profile for a specified total magnet length, DSV size and position, and required field strength [3]. The distribution is used as a starting point for the coil block design, and then a nonlinear optimization method is used to refine the configuration of the magnet. The method has been successfully applied to the designs of compact unshielded and active shielded magnets, open magnets and specialized magnets for head imaging. The results highlight that this method is flexible and useful.

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II. Design Method for Superconducting Magnet Used in MRI System

A. Requirements of MRI System

The homogeneous magnetic field in the diameter sphere volume (DSV) with a diameter of 400 mm generated by a coil used in MRI system should be lower than 10 ppm. To protect electronic devices and the safety of peoples, the stray magnetic field generated by the superconducting magnet needs to be as small as possible. Typically, the line of 5 G should be less than 5 m from the center of the magnet.

In order to realize the requirements in an MRI system, multi-solenoid-typed superconducting coils are employed to generate the expected center magnetic field and homogeneity. Shielding of stray magnetic field has three methods, such as, room shield, iron-yoke shield and active shield methods. Generally, it needs a lots of iron for the room shield and iron-yoke shield to reduce 5 G lines, therefore, the method is not popular in the modern MRI system. The active shield method has been employed in the new generation of superconducting magnet for MRI system due to compact configuration, lightweight and very small 5 G line. The active shield method uses reversed currents at the outside of the main coils.

B. Continuous Current Methods

Magnetic field generated by a superconducting solenoid coil with rectangular-shaped cross-section can be calculated by fives parameters, such as position coordinates, size of cross-section, current density etc.. The relationship between magnetic field and position coordinates and size of geometries is nonlinear. If the superconducting solenoids are assumed to be a series of idea current loops located at the center of squares, the magnetic field

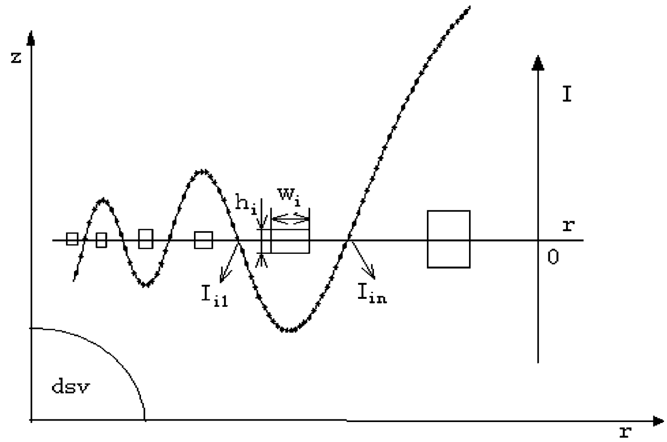


Fig. 1. Schematic of initial dimensions of the coils bulks.

in the space can be only determined by the position coordinates and current. In the design, the radii of the main and shield coils can be decided in advance, the n current loops generates the magnetic field at m target points can be calculated [3]

$$AI = B \tag{1}$$

where $\boldsymbol{B_m}$ is the matrix with m row, $\boldsymbol{A_{mn}}$ is the field at the $m^{\rm th}$ target point due to unit current in the $n^{\rm th}$ feasible coil, $\boldsymbol{I}$ is the current matrix with n rows.

The requirement of superconducting wires for coils used in the fabrication is a very important part in a design. It can dictate the total cost of a superconducting magnet system. The inequality constraint for a minimum superconducting wire design for high homogeneous magnetic field can be transferred to following optimization problems [5], [6]:

$$\text{Objective function: } \min \quad f = \sum_{i=1}^{n} 2\pi r_i |i_i| \tag{2}$$

Subject to:

$$\begin{aligned} &AI \le B_0 + \varepsilon B_0 \\ &-AI \le -B_0 + \varepsilon B_0 \\ &|C_r I| \le B_{r,shield} \\ &|C_z I| \le B_{z,shield} \end{aligned} \tag{3}$$

Where: $\boldsymbol{\varepsilon}$ in the region is $1 \sim 10$ ppm, $\mathrm{C_s}$ is a matrix of $l \times n$, $\mathrm{C_s}$ represents the $n^{\rm th}$ current loop generating the magnetic field at the shield location due to a unit current. To avoid the nonlinear constrain conditions, the constrain conditions of r and z components are employed. If magnetic field component of $\boldsymbol{B_r}$ and $\boldsymbol{B_z}$ are 3 G and 4.5 G, respectively, it is adequate to constrain the stray field to 5 G.

The optimization problem can be treated as a linear problem. Based on (3), the problem can be calculated with arguments of current I. In the paper, Matlab is used to solve the problem with several thousands of candidate current loops. The position coordinates $(r_{\rm i}, z_{\rm i})$ for each current loop is randomly generated. As an example for an open structure MRI magnet, a possible current distribution based on the methods is plotted in Fig. 1.

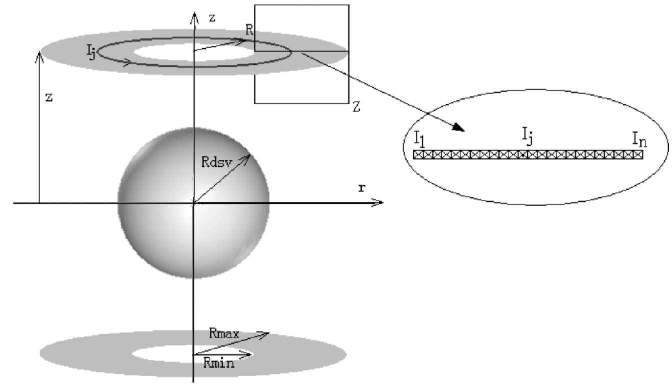


Fig. 2. Discretizing the opened superconducting coils as current loops.

C. Nonlinear Genetic Algorithm Optimal Methods

The current loop has to be partitioned into a number of current bulks according to linear optimization solution as shown in Fig. 1, and each of them has a rectangular cross sections. The number of current bulks is determined by number of distribution in the solution of (3) and dimensions of the bulks are defined by [2], [3]

$$w_i h_i J = \int_{l_i} J(\xi) d\xi \tag{4}$$

where l_i is the length of the jth section of current J, w_j is the width of the bulk, h_j is the height of the bulk and J is the current carried by the unit cross-section. Here, we define the operating current density as about 1.0×10^8 A/m^2. The height of h_j is defined as an optimization argument. If the height of $\boldsymbol{h_j}$ is determined, the cross-section and w_j of each coil can be determined.

In the nonlinear optimization, the genetic algorithm is employed to calculate the parameters for the coil configuration. Genetic algorithm were developed in the mid 1970s by Holland [7]. The genetic algorithms have been given considerable attention lately and have been applied to problems in electromagnetic field computation. Due to the coding into binary strings, genetic algorithms are able to treat mixed continuous and integer design variables and are therefore the choice for magnet design optimization. The genetic optimization method is a global optimization method and can treat large-scale, multi-value and discrete optimization problems with very fast calculation speeds. The nonlinear optimization problem is defined as:

Objective function

$$f_{\min} = \lambda_1 |B_0' - B_0| + \lambda_2 h_{\max} + \lambda_3 \left| B'_{stary_{\max}} - B_{stray} \right| \tag{5}$$

here:

$$\begin{aligned} \Delta_i &= \left| \frac{B_i - B_0'}{B_0'} \right| \times 10^6 \\ \Delta_{\max} &= \max\{\Delta_1, \Delta_2, \ldots, \Delta_n\} \\ B'_{stray_{\max}} &= \max\left\{B'_{stray_1}, B'_{stray_2}, \ldots, B'_{stray_m}\right\} \end{aligned} \tag{6}$$

where B_0' is the center magnetic field of the superconducting magnet, B_0 is the expected magnetic field located at the center of the magnet, $B_{\rm i}$ is the magnetic field at the boundary of DSV, Δ is the inhomogeneity of magnetic field in DSV, B'_{stray_i} is the

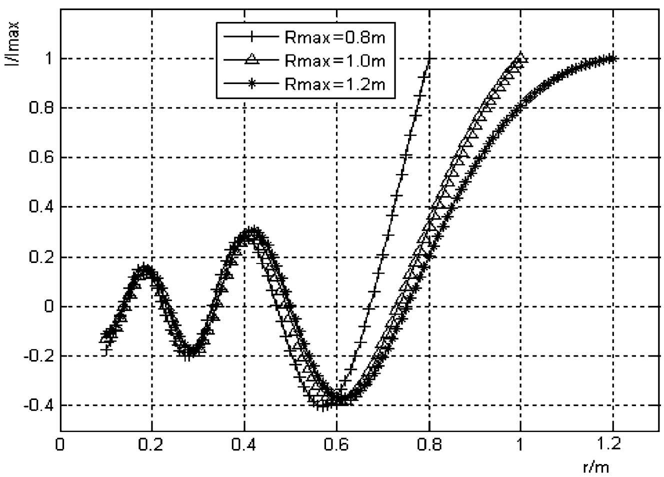


Fig. 3. Normal current distribution for various R_{max}.

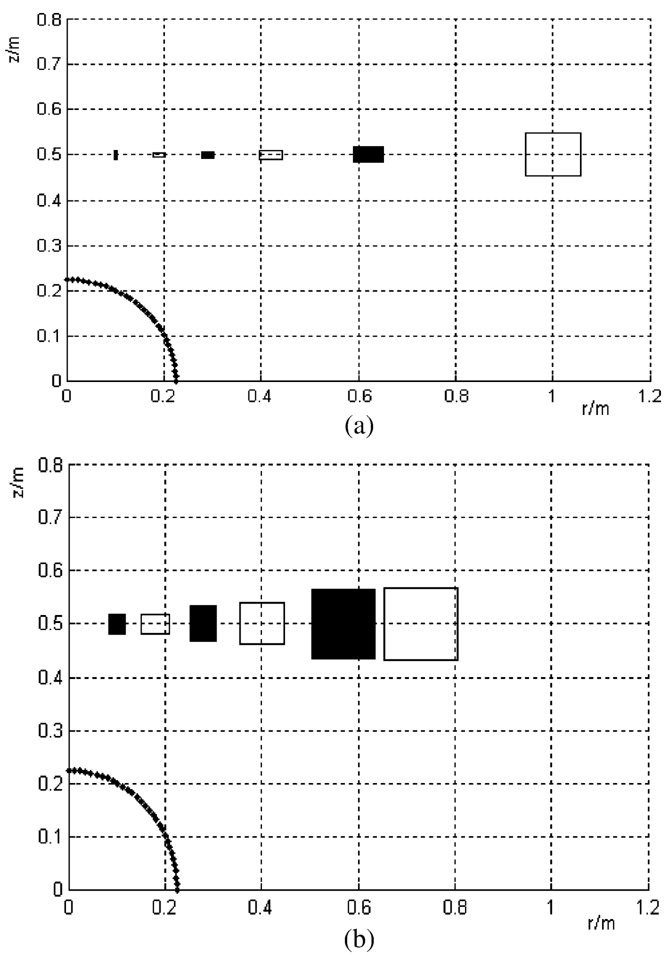


Fig. 4. Configuration of fully open MRI superconducting coils with $R_{\rm amx} = 1.2$ m (a) and $R_{\rm max} = 0.8$ m (b).

stray magnetic field,λ_1, λ_2 and λ_3 are the weighting factors for the each parameters, $B_{\rm stray}$ is equal to 5 G.

The code for the genetic optimization method is employed the Matlab genetic toolbox, the mutation number is 100, the number of generation is equal to 300, and the adaptive crossover is used in the code.

TABLE I
MAIN PARAMETERS FOR $\mathrm{R\,max} = 1.2$ m

	R1/m	R2/m	Z1/m	Z2/m	J
1	0.0977	0.1023	0.4900	0.5100	-
2	0.1770	0.2030	0.4950	0.5050	+
3	0.2776	0.3024	0.4933	0.5067	-
4	0.3959	0.4441	0.4896	0.5104	+
5	0.5902	0.6498	0.4834	0.5166	-
6	0.9434	1.0566	0.4531	0.5469	+

TABLE II
MAIN PARAMETERS FOR $\mathrm{R\,max} = 0.8$ m

	R1/m	R2/m	Z1/m	Z2/m	J
1	0.0837	0.1163	0.4817	0.5183	-
2	0.1501	0.2099	0.4822	0.5178	+
3	0.2536	0.3064	0.4666	0.5334	-
4	0.3541	0.4459	0.4616	0.5384	+
5	0.5058	0.6342	0.4357	0.5643	-
6	0.6551	0.8049	0.4317	0.5683	+

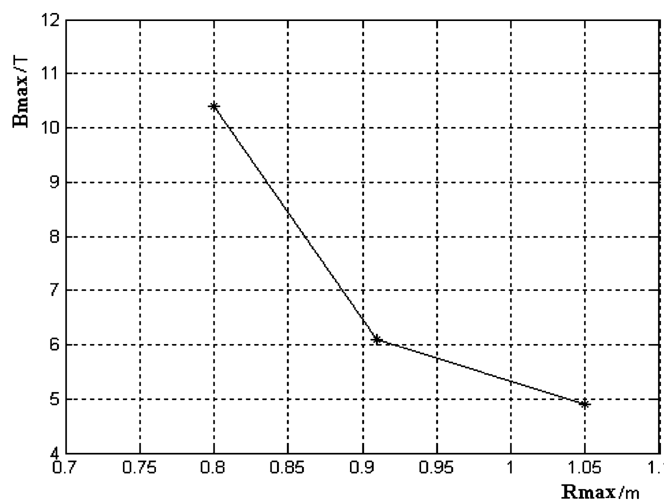


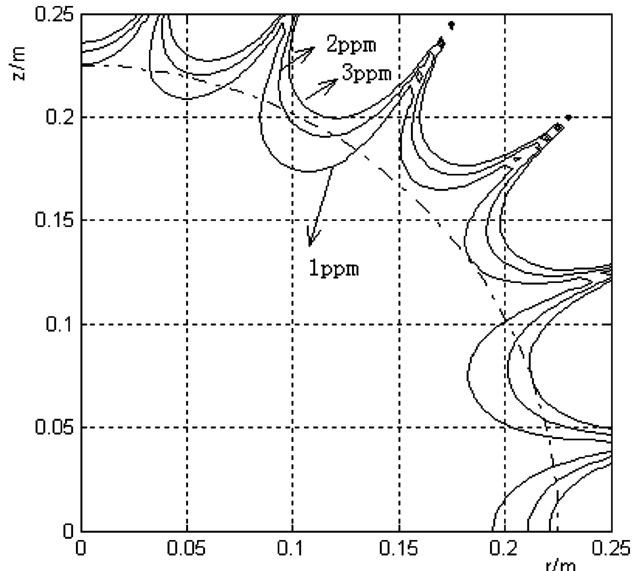
Fig. 5. Peak magnetic field for $\mathrm{R\,max} = 1.2$ m, 1.0 m and $\mathrm{R\,max} = 0.8$ m.

III. DESIGN OF SUPERCONDUCTING MAGNET FOR MRI

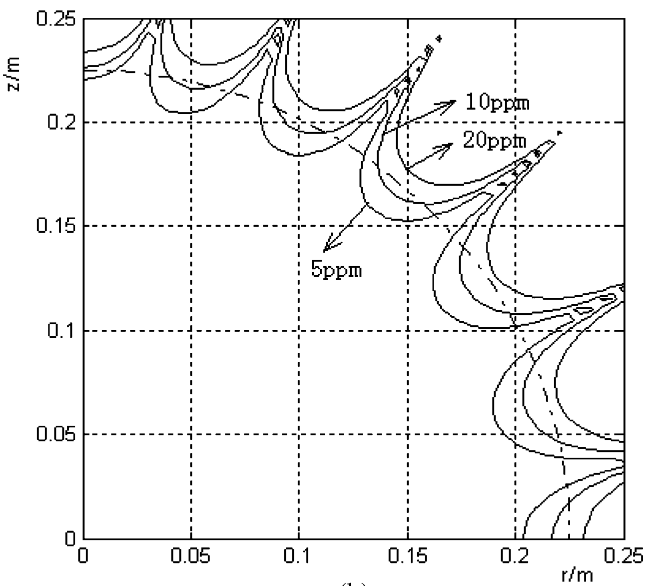
A. Open Superconducting Magnet

The design model as shown in Fig. 2. In the design, the minimum radius of superconducting coil is set to $R_{\rm min} = 0.1$ m,

The maximum radius of coils is defined as $R_{\rm max} = 1.2$ m, 1.0 mm and 0.8 m. As shown in Fig. 2, the region radii for the current loops from $R_{\rm min}$ to $R_{\rm max}$ are discretized into 100 points, the radius of DSV is set to 0.225 m, and the angular coordinate is set as from 0 to π. There are 150 points for the target field points. The distance between the center plane of current to the center of magnet is about $\mathrm{z} = 0.5$ m. The center field of magnet is 1 T. The current distributions based on the linear optimization are plotted in Fig. 3 for various $R_{\rm max}$. From the calculation results, the distribution of current is almost the same for the three scenarios. The number of current bulks is determined by the current distribution. Based on the nonlinear genetic algorithm optimization methods, the cross-sections of superconducting coils are calculated and shown in Figs. 4(a) and (b). The dark blocks stand



(a)



(b)

Fig. 6. Homogeneity for $R_{\max} = 1.2$ m for (a) and 0.8 m for (b).

for reversed currents. With the reduction of radius of superconducting coils, the coil cross-sectional areas are increased. The results show that the maximum outer radius can be a limiting factor. The parameters of each coils are listed in Tables I and II. Further, the peak magnetic field is calculated for maximum radii of 1.2 m and 0.8 m. The peak magnetic field for 1.2 m is much lower than that for $R_{\max} = 0.8$ m. It shows that the very small $R_{\max}$ leads to increase of the peak magnetic field located at the superconducting winding.

The peak magnetic field and homogeneity in DSV region of superconducting coils are calculated as shown in Figs. 5 and 6. The results show that increasing $R_{\max}$ can result in better homogeneity for a superconducting magnet. After using genetic optimization methods, the maximum radius of superconducting coil is about 1.05 m, 0.92 m and 0.80 m. Although the radius of superconducting coils is reduced, it results in increased reverse current. In order to increase the center field, the current and conductor size have to be increased to compensate the negative current generated field. The weight of superconducting wire is increased for the whole magnet.

IV. Conclusions

A method for the design of open magnetic resonance imaging magnets has been presented in this paper. The first step in the design process is to find the source current distribution by using an inverse approach. Nonlinear optimization method are then introduced to fit practical magnet coils to the desired current density. These design methods can be used to design more open superconducting magnets for real MRI applications. The calculations are very effective with very short CPU time required to design a superconducting magnet.

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