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Unilateral Magnet for Magnetic Resonance Imaging

ZHANG Yan-li *, *XIE De-xin*, *BAI Bao-dong*, *ZENG Lin-suo*

(School of Biomedical Engineering, Shenyang University of Technology, Shenyang 110023, China)

Abstract：Openness is a desired property for the magnets used in magnetic resonance imaging (MRI). For this reason, the design and making of unilateral magnets, which give a homogenous region, usable for imaging, that is external or "remote" from the magnet, has been attracting more and more attention and research interests. In this paper, the feasibility of designing a unilateral MRI magnet with a slice-shaped imaging region is discussed on the basis of the theory of saddle point. The properties of magnetic fields generated by different types of unilateral magnets were derived by simulation, and the results showed that it is possible to have a unilateral magnet that can generate remote saddle points in the field profile while having sources on one side. The configuration of a unilateral permanent magnet specially designed for MRI is presented, which can produce a slice-shaped imaging region with high field homogeneity. The results obtained in this study will build a strong basis for future scientific researches on fully opened MRI system with homogeneous magnetic field.

Key words：magnetic resonance imaging, unilateral magnets, slice imaging region, saddle point

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Introduction

The solenoid-shaped MRI magnets (superconductive, resistive) as well as iron core C shape permanent magnets are known for imaging the whole body. However, such whole body MRI magnets are not generally well-suited for treatment of the patient with other modalities or for minimally invasive surgical procedures guided by real time MRI because of the limited access of the surgeon to the patient. This limited access results

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Biography：(1975-), female, Shenyang, Ph. D student, School of Biomedical Engineering, Institute of Electrical Engineering. * Corresponding author：Zhang Yan-li, Phone:13940269519, E-mail:zyl_sy@163.com.

from the field producing means surrounding the imaging volume. A recent classification scheme has termed these magnets as bihedral magnets since the magnetic sources are arranged around two sides of the sample or patient[1]. On the other hand, unilateral designs, which have magnetic sources arranged to one side, are truly "open" in that the target region may be considered to be external or "remote" from the magnetic resources. The first application of this type of magnet was used in industry, such as oil well logging[2, 3]. The application field of the unilateral NMR (nuclear magnetic resonance) devices has experienced considerable expansion during the last years[4, 5]. The use of single-sided NMR MOUSE probes[5] makes NMR a truly non-invasive method suitable for surface studies of arbitrarily large objects. Clearly, the primary advantage of unilateral magnets over bihedral magnets is their openness. However, these unilateral NMR magnets is not applicable for biomedical imaging because the magnetic field in the imaging region has great gradient with several tens T/m while the gradient required for body imaging is only several tens mT/m.

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 imaging patients that have claustrophobia. Recently, Pulyer et al. [6, 7] have shown that it is possible for the unilateral MRI system to produce a volumetric region of homogeneous field, which is external or "remote" from the magnetic sources. Moreover, they also proposed a kind of the magnet configuration for MRI system, which comprises a primary coil magnet system and a bias coil magnet system to produce a magnetic field having a substantially remote region with field homogeneity. In order to obtain the sphere shape volumetric region with field homogeneity and field strength, the design of the unilateral MRI system is at the price of low energy efficiency and high cost.

When the permanent material is employed to achieve unilateral magnet, the generation of the volumetric imaging region external from magnet sources is difficult and even impossible because the magnet will be huge and the cost of the magnet is expensive in order to achieve enough field strength and homogeneity. Currently, a novel thought about the unilateral MRI magnet is proposed in[8, 9]. The imaging region of this kind of magnet is a slice region instead of volumetric region and the magnet source is a permanent magnet instead of electromagnet. The magnetic field in the slice region has certain gradient in the direction perpendicular to the main magnetic field and the magnitude of the gradient is as small as that produced by the layer-selection gradient coil. Thus, a set of gradient coils in the layer-selection direction may be saved. The slice selection along the depth direction is obtained by means of the highly constant static magnetic field gradient produced by this magnet geometry. The normal two-dimensional encodes imaging method can be used to obtain the 2-D images. By removing the patient table (shown in Fig. 1), it is possible to make other tissues located in this slice region in order to get the whole body images. Clearly, the primary advantage of the unilateral permanent magnet with a slice imaging region is lower weight and lower cost together with more "openness".

It is a totally new problem about the unilateral magnet design to build a slice region with field homogeneity outside the used magnet. Although the bihedral magnets benefit from a well-developed analytic design theory, which has led to high field and high homogeneous designs, this development about the unilateral magnet with slice region is in its infancy. Because the imaging region is "remote" from the magnet structure, it is hard to make the field strength and homogeneity in the imaging region to meet the requirement. Furthermore, the location of slice imaging region is also not so easy to find as the bilateral MRI magnet, in which the imaging region is located in the center of the magnet structure owing to its closeness and symmetry. In this paper, the properties of magnetic field for different unilateral magnet structures are first discussed. Then, a novel unilateral MRI permanent magnet is presented, which can generate a slice imaging region with field homogeneity outside the magnet sources. The successful design of the unilateral MRI magnet build a foundation for the scientific research of open MRI device.

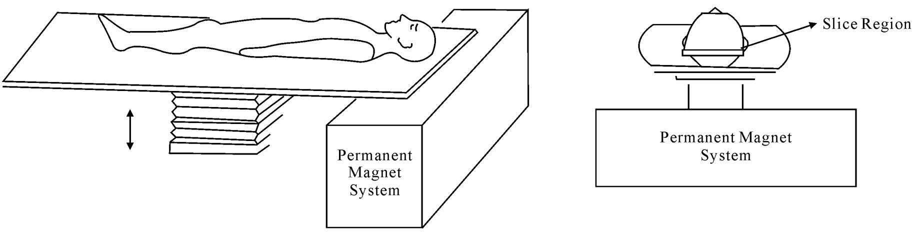

Fig. 1　The unilateral MRI system with a slice imaging region

1　The Discussion of Unilateral Magnet

1.1　The Saddle Point

For the conventional MRI magnet, the imaging region with high homogeneity is located inside the magnet structure. Why can the bilateral magnet build so homogeneous magnetic field? Fig. 2 shows the four-pole shape MRI magnet model and, the magnetic field distribution in the imaging region is shown in Fig. 3.

In Fig. 3, there exits a point at which the magnetic field is maximum along the y axis and is minimum along the x axis. In terms of mathematics this point is called saddle point. Likewise, for the C shape permanent MRI magnet and the superconductivity MRI magnet, there also exists the saddle point in the imaging region. Thus, for the bilateral MRI system, the theoretical feasibility of producing a homogeneous target magnetic field depends upon the existence of a saddle point[1]. The saddle point provides a core of homogeneity and a homogeneous region must begin around the saddle point, at which there is $\nabla B_i = 0 (i = x, y, z)$. For a suitable design of the MRI magnet configuration, there must exist a saddle point in the magnetic field. Thus, it can be concluded that for the unilateral MRI magnet, if the magnet structure can build a saddle point outside the magnet, the magnet structure is feasible and is worth being developed further. The properties of magnetic field for several kinds of unilateral magnets will be discussed in the following section.

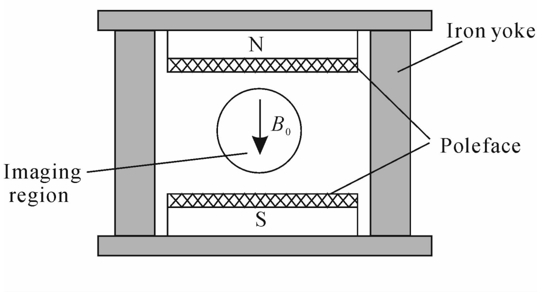

Fig. 2 Four-pole shape MRI magnet model

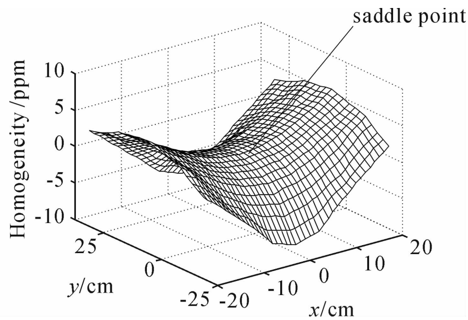

Fig. 3 The distribution of magnetic field in the imaging region

1.2 Unilateral Coil Structure

For a circular loop of radius a carrying current i, the law of Biot-Savart gives the radial and axial components of the magnetic field for any point with the coordinates z and r (with the origin at the center of the loop) as the elliptic integrals

$$\begin{cases} B_z = \dfrac{\mu_0 ia}{4\pi}\displaystyle\int_0^{2\pi}\dfrac{(a - r\cos\theta)\,\mathrm{d}\theta}{(a^2 + z^2 + r^2 - 2ar\cos\theta)^{3/2}} \\ B_r = \dfrac{\mu_0 ia}{4\pi}\displaystyle\int_0^{2\pi}\dfrac{z\cos\theta\,\mathrm{d}\theta}{(a^2 + z^2 + r^2 - 2ar\cos\theta)^{3/2}} \end{cases} \tag{1}$$

where B_z is the component of the field parallel to the z axis, and is the radial component. If a coil of radius r_b is nested coaxially within a larger coil of radius r_a and the currents flow in opposite senses (Fig. 4(a)), the magnetic field created is the sum of the fields generated by the two coils respectively. Let $r_a = 10$ and $r_b = 5$, the magnetic field at the scope of $-1 \leqslant r \leqslant +1$, $0 \leqslant z \leqslant 15$ at R-Z plane will be considered. Because the radial component of the magnetic field is intrinsically very small at the above scope, only the axial component is considered. With proper selection of the currents in the two coils, the axial magnetic field produced by the two coils at z axis is plotted in Fig. 4(b). The relative field strength refers to the ratio of the axial magnetic field strength at any point to that at the center of each loop. The middle curve represents the sum of the individual field generated by the inner and outer coils. It can be seen that there exists a maximum of the magnetic field at about $z = 5.73$ in Fig. 4(b). The magnetic field at the scope of $z = 5.73$ and $-1 \leqslant r \leqslant +1$ is depicted in Fig. 4(c). It shows that there exists a minimum of the magnetic field B_z at $r = 0$. Thus, the opposed coil design, in which the diameters or ampere-turns of the two coils are different, can provide a remote saddle point at $z=5.73$, $r=0$.

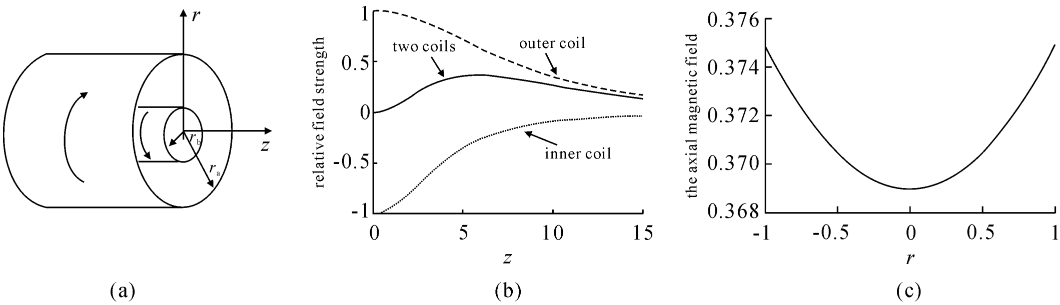

Fig. 4　Opposed coils (a), the magnetic field B_z at the z axis (b), the magnetic field B_z at $z=5.73$, $-1\leqslant r\leqslant+1$ (c).

Reference [1] discussed other unilateral coil structures. The illustration of the coil designs with qualitative field profiles and the equivalent dipole schematic of each magnet are depicted in Fig. 5. All the coil magnets in this figure can provide remote saddle points.

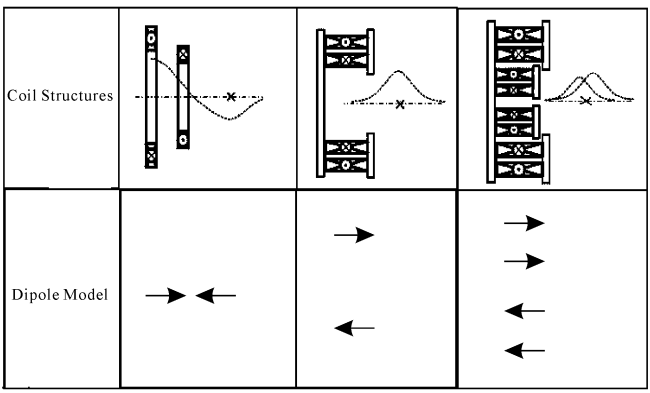

Fig. 5　Unilateral coil structures ("x" represents the saddle point)[1]

1.3　Unilateral Permanent Magnet Blocks

As shown in the last section, there exists a remote saddle point in the magnetic field produced by opposed coils. It is helpful for the unilateral permanent magnet design. Fig. 6 shows the opposed permanent magnet blocks in which the direction of their magnetization is opposite. Similar to the opposed coils, the permanent magnet also provides a remote saddle point at the x axis and the x coordinate of the saddle point is represented as $x_{\max}$. The magnetic field along x axis produced by this magnet can be described by two characteristics: the maximum field strength $B_{y\max}$ (only the y component of the magnetic field need to be considered since the x component vanished on the x axis); and the region of homogeneity around the saddle point, $x_{\max}$, over a distance Δx, in which the homogeneity is about 10^{-3}. The field going through an extremum $B_{y\max}$ at the point, $x_{\max}$, is remote from the permanent magnet configuration.

A survey of various combinations of opposed permanent magnets has been made by arbitrarily letting $h_a = 8$ cm and $w_a = 4$ cm, while varying h_b and w_b. The variation of $x_{\max}$, $B_{y\max}$ and Δx with h_b and w_b is presented for a representative set of cases, shown in Fig. 7 and Table 1.

From Table 1, we can see that $x_{\max}$ and Δx decreases and $B_{y\max}$ increases when the ratio of w_b/w_a scales up with ratio of h_b/h_a remaining the same. It is worth noting, however, that for a given ratio of w_b/w_a, the value of $x_{\max}$, Δx , and $B_{y\max}$ are all decreasing. Therefore, the location of the saddle point is dependent upon the size of the opposed permanent magnet blocks, and thus the remoteness of the field may be controlled by the ratio of length and width of these opposed permanent magnet blocks. This flexibility is important for the design of a homogeneous region, which is "remote" to the magnetic sources.

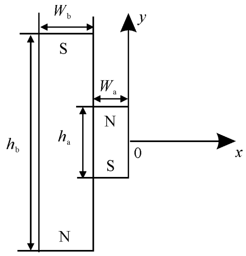

Fig. 6 Opposed permanent magnet blocks

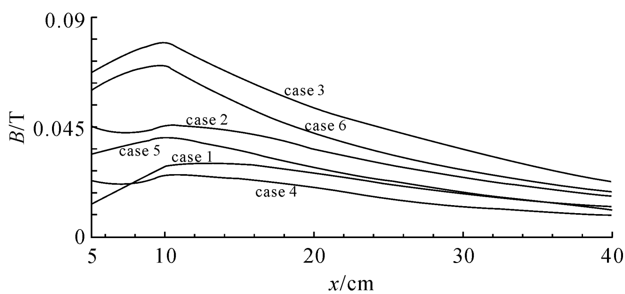

Fig. 7 The value of B_y at the X-axis for two opposed permanent magnets

A set of cases (from case 1 to case 6) correspond to different sizes of two opposed permanent magnet blocks in Table 1.

Table 1 Maximum Field Strength for a Set of Cases ($h_a = 8$ cm, $w_a = 4$ cm)

Case	h_b/cm	w_b/cm	$x_{\max}$/cm	$B_{y\max}$/T	Δx/cm
1	32	6	15	0.030	5
2	32	8	12	0.048	2
3	32	12	10	0.080	0
4	24	6	11	0.026	1
5	24	8	10	0.044	0
6	24	12	10	0.070	0

Based upon the magnet introduced above, it is clear that a suitable unilateral magnet will have the following features: 1) It must generate a remote saddle point, which makes it possible to generate a homogeneous target magnetic field. 2) Permanent materials and current coils may be employed to achieve unilateral magnet. 3) The remoteness of the saddle points is related to the size and structural form of magnet sources.

2 Proposed Unilateral MRI Permanent Magnet with Slice Imaging Region

By considering the above unilateral magnets, the proposed unilateral MRI permanent magnet with slice imaging region is shown in Fig. 8, which is composed of the permanent material chosen as NdFeB and the iron with the size of $a=80$ cm, $b=22$ cm. When the permanent material is employed to achieve unilateral magnet, the imaging region external from magnet sources should be designed as the slice shape, which is different from unilateral coil magnet, in which the current of the coil can be adjusted to be high enough to provide homogeneous volumetric region. As we know, the conventional imaging region for the whole body image is a sphere with a diameter of 30 cm and during the course of the imaging, the depth of the selected layer is about 1～10 mm. For the conventional permanent MRI magnet, the main magnetic field in the imaging region is about 0.1～0.5 T. Thus, for the unilateral MRI design the requirement of design is that the depth and length of the slice region may be $S_b=4$ mm and $S_a=12$ cm, the main magnetic field strength is about 0.1～0.2 T and the homogeneity of the imaging region is about 10^{-4}.

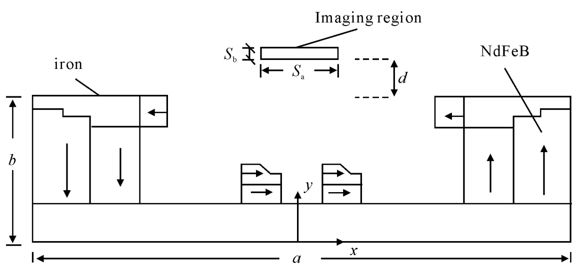

Fig. 8　The unilateral permanent magnet model ($a=80$ cm, $b=22$ cm)

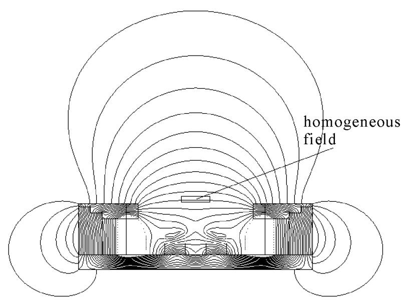

Fig. 9　The distribution of magnetic field

Over the past several decades, the magnetic field produced by permanent magnet has been well solved by numerical computation, such as finite element method (FEM). But this method cannot give the analytical expression of magnetic field, thus it is difficult to find the accurate location of saddle point according to the definition of saddle point, i.e. $\nabla B_i=0(i=x,y,z)$. Based on the theorem of equivalency and separation of variables, the new-style equivalent source method (NESM)[10], a semi-analytical method, can give series expansion in a field region, in which its coefficients are determined by making the errors at boundaries minimum by using the least square method. However, the semi-analytical method will become very onerous for the problems with complex boundaries and multi-material regions. Thus, A mixed NESM-FEM method[11] with higher precision is proposed here to probe the existence possibility of slice region for such a unilateral magnet and analyze the characteristic of the magnetic field distribution. By the NESM-FEM method, the analytical expression of the magnetic field strength can be obtained, and then by finding the $\nabla B_i = 0 (i = x, y, z)$, the location of saddle point can be found.

The magnetic field distribution is shown in Fig. 9. The saddle point in the unilateral magnet is located at $x=0$, $y=24$ cm and the distance from the magnet is $d=2$ cm. The main magnetic field in the slice region is 0.16 T and the unhomogeneity is about 8.56×10^{-3}. Obviously, the homogeneity is not sufficient for MRI imaging, but the magnet can be served as the initial design and by carrying out some optimization the homogeneity can be improved.

For the conventional MRI magnet design, the iron pole is usually added to the magnet structure in order to improve the homogeneity. In this paper on the basis of the MRI magnet shown in Fig. 8, the genetic algorithm is used to optimize the shape of the iron pole and the 3-D model after optimization is shown in Fig. 10[12]. The field homogeneity at different xz planes near the saddle point is shown in Table 2. From Table 2, it can be seen that the homogeneity of the xz plane at about $y=24.0$ cm is the best. The magnetic field strength distribution at $y=24$ cm plane is shown in Fig. 11. Thus, we may propose that the y-direction location of the slice region with field homogeneity is from 23.8 cm to 24.2 cm and the depth is 0.4 cm. The size of the slice region is $S_b = 0.4$ cm, $S_a = 12$ cm, $S_c = 12$ cm, $d = 2.0$ cm, and the homogeneity is about 3.43×10^{-4}. The slice region produced by the proposed unilateral magnet configuration is sufficient for imaging. The normal two-dimensional imaging method can be used to obtain the 2-D images. By removing the patient table, it is possible to make other tissues located in this slice region in order to get the whole body images. Thus, the proposed unilateral MRI magnet is fully open and the magnet design is successful. The experiment test about this magnet and the study of the shimming technique will be considered in another paper.

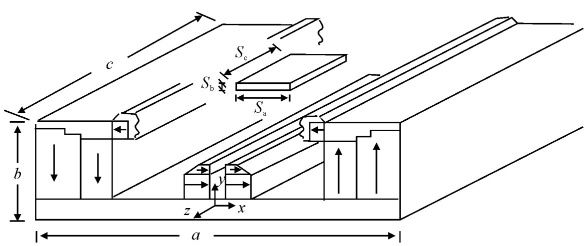

Fig. 10 The 3-D unilateral MRI magnet structure [12] ($a=80$ cm, $b=22$ cm, $c=75$ cm; $S_a=12$ cm, $S_b=0.4$ cm, $S_c=12$ cm)

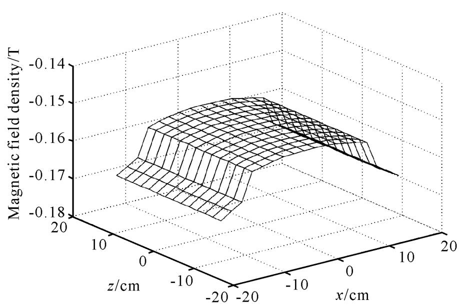

Fig. 11 The magnetic field strength at $y=24$ cm plane

Table 2　The field homogeneity at the different *xz* planes around *y*=24 cm

$-6\text{ cm}\leqslant x\leqslant 6\text{ cm}$，$-6\text{ cm}\leqslant z\leqslant 6\text{ cm}$

y/cm	23.7	23.8	23.9	24.0	24.1	24.2	24.3
field homogeneity(10^{-4})	10.21	4.19	3.07	2.75	2.96	4.18	8.57

3　Conclusion

This paper discusses the properties of magnetic field for different unilateral magnets and a novel unilateral permanent magnet configuration for MRI system，which can produce a slice imaging region with field homogeneity，is presented successfully. Some conclusions about the unilateral magnet design can be concluded as follows：

(1) The theoretical feasibility of producing a homogeneous target magnetic field depends upon the existence of a saddle point. The saddle point provides a core of homogeneity and a homogeneous region must begin around a saddle point，at which there is $\nabla B_i=0(i=x,y,z)$.

(2) A suitable unilateral magnet must generate a remote saddle point.

(3) Permanent materials and current coils may be employed to achieve unilateral magnet.

(4) When the permanent material is employed to achieve the unilateral magnet，the imaging region should be designed as slice shape instead of volumetric region.

The success of such a unilateral permanent magnet with a slice region for MRI builds a strong basis for some scientific researches of fully open MRI system with homogeneous magnetic field.

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单边磁共振成像仪磁体系统研究

张艳丽*，谢德馨，白保东，曾林锁
（沈阳工业大学 生物医学工程系，辽宁 沈阳 110023）

摘　要：单边磁共振成像(MRI)系统由于其良好的开放性引起了越来越多的关注，这种系统的成像样品区位于磁体外侧，其形状为薄片形，其磁体结构设计是一个全新的问题. 本文基于鞍点理论，探索这种单边磁体在其外侧产生均匀样品区的可行性，通过对几种单边磁体结构磁场特性的研究，指出适当设计的永磁磁体能够在磁体外侧产生鞍点. 最后给出一种符合成像要求的单边永磁 MRI 磁体结构. 单边 MRI 磁体结构的可行性分析及磁体结构的成功设计，为开展完全开放式磁共振成像装置的研究打下了坚实的基础.

关键词：磁共振成像；单边磁体；薄片形样品区；鞍点

* 通讯联系人：张艳丽，电话：13940269519，E-mail：zyl_sy@163.com.