WRAPPED EDGE GRADIENT COIL FOR MRI-PET ANIMAL IMAGING

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Introduction: The medical imaging methods of Computed Tomography (CT), Magnetic Resonance Imaging (MRI) and Positron Emission Tomography (PET) are well established, each with their own advantages and disadvantages. There appears to be an ever increasing need to combine these modes of imaging, with the view of providing earlier diagnosis of disease, improved treatment planning and better patient care. Moreover, the concepts of MRI-PET and CT-PET are well established, but in the case of MRI-PET many challenges persist. Nevertheless, our work is aimed at MRI-PET scanners and primarily on the improvement of magnets and gradients [1]. Here we propose an open transverse gradient coil arrangement, which is suitable for small animal imaging. The z-gradient coil is known to be simpler to design; therefore main challenges are concerned with the design of x- and y-gradient coil windings, whereby the x-gradient coil is the same as a y-gradient when a 90^o rotation is applied.

In the past we developed a wrapped edge gradient coil for compact clinical MRI scanners [2]. Recent work by Poole *et al* on novel split gradient coils useable in MRI-PET scanners showed that these scanners can be made compact [3]. Our work is aimed at the design of compact superconducting magnets and gradients coils, with the hope of reducing the overall scanner size and associated room needed to perform experiments.

Method: The magnetic field is defined through pure harmonic coefficients to appropriately represent the gradient magnetic field. The full spherical harmonic representation in spherical coordinates is used to define the required *z*-component variation in the magnetic field [4]. Divergent free current is assumed, which means that $\nabla\cdot(\nabla\times\mathbf{B})=0$ is satisfied. As a result, a stream function $\mathbf{S}$ from which wire paths are obtained is calculated using $\nabla\cdot\mathbf{S}=0$ meaning that $\mathbf{S}=\nabla\times\mathbf{B}$. Assuming that the conductor paths exist on cylindrical surfaces, the following explicit stream function relationship is used to define current densities [2]:

$$S_\rho=\frac{1}{\rho}\frac{\partial B_\theta}{\partial z}-\frac{\partial B_z}{\partial\theta},\quad S_\theta=\frac{\partial B_\rho}{\partial z}-\frac{\partial B_z}{\partial\rho},\quad S_z=\frac{1}{\rho}\left(\frac{\partial}{\partial\rho}(\rho B_\theta)-\frac{\partial B_\rho}{\partial\theta}\right).$$

Above, the component of $\mathbf{S}$ normal to the cylinder surface is S_r, and is the component of interest from which current paths are defined through contouring. The Cartesian wire path coordinates ($\mathbf{C}_{ij}$) at the extremities of the design ($z_0=\pm 0.4m$) have been mapped to the transverse plane in the following way:

$$\mathbf{C}_{ij}=\left(l_{ij}x_{ij},l_{ij}y_{ij},z_0\right),\quad l_{ij}=1+\frac{|z_{ij}|-|z_0|}{2\sqrt{x_{ij}^2+y_{ij}^2}}.$$

Results and Discussion: Fig. 1(a) illustrates the x-gradient design useable in an animal scanner [3]. At the edge of the gradient coil the wire paths have been wrapped into the transverse plane to have the ability to retain compactness, and at the same time increase the linearity of the gradient magnetic field. In Fig. 1(b) the magnetic field produced by the gradient coil of Fig. 1(a) is depicted. In Fig. 1(c) the error in the gradient magnetic field is shown. In Fig. 1(c) the contour lines start at the 5% error level, followed by 10%, 15% and so on. The highlighted circles in Figs. 1(b) and 1(c) are of diameter 10cm, 15cm and 20cm to illustrate that the size of the DSV is rather large, for a gradient coil which has an inner diameter of 30cm and the separation between the primary and shield windings is 5cm. The gradient field produced by this design is 0.2mT/m/A. For illustration purposes only every second winding path is shown in Fig. 1(a), otherwise the design would be hard to comprehend. Our presentation will show details of this gradient coil, along with the y- and z-gradient coil layouts and specifications. The separation between the quadrants of the gradient coil is 15cm, which is in accordance with a 1T animal scanner recently published [3].

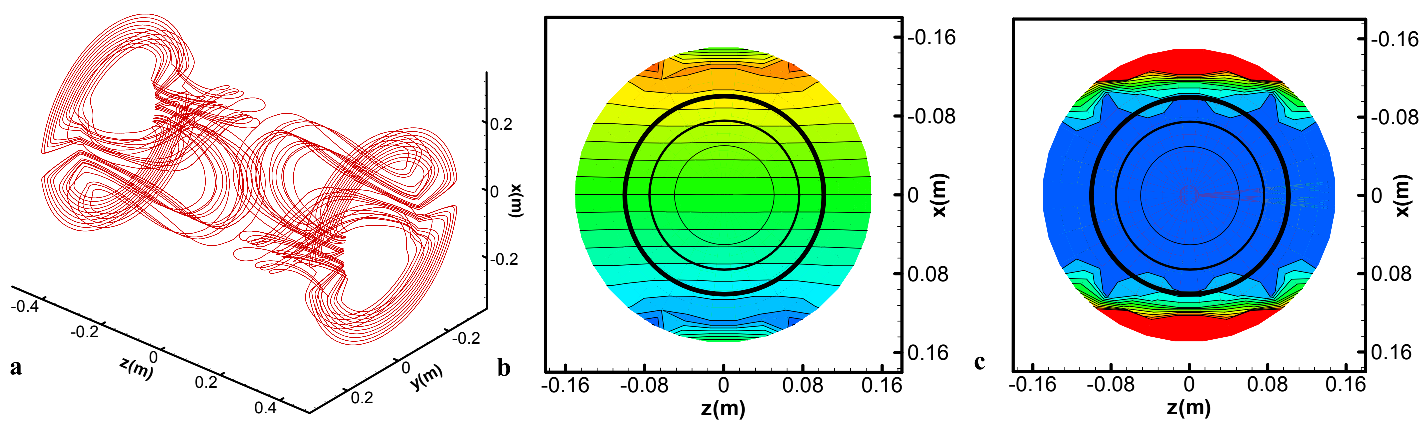

Figure 1 Illustrated is the result obtained after optimisation of spherical harmonic coefficients, which are used to establish the winding pattern. The winding pattern is then used to find the magnetic field, gradient field and error in the gradient field given a fixed current. Shown are (a) the actual winding pattern (only every second contour is shown for visual clarity), (b) the z-component of the magnetic field produced by the transverse gradient coil winding and (c) the error in the gradient magnetic field with 5% to 100% error levels in increments of 5%. The circles in (b) and (c) have diameters of 10cm, 15cm and 20cm to help establish the region in which acceptable images can be attained.

References

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