MINIMUM STORED ENERGY SPLIT SUPERCONDUCTING MAGNET FOR 3T MRI-PET ANIMAL IMAGING SYSTEM

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Introduction: The different medical imaging modalities 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. For this reason, our work is on MRI-PET scanners and primarily on the improvement of magnets and gradients [1]. Here we propose an open superconducting magnet coil arrangement capable of 3T, which is suitable for small animal MRI-PET imaging systems. The proposed magnet is designed using the minimum stored energy (MSE) approach [2, 3]. The approach provides a globally optimal solution and has been used in designing a wide range of magnets [4], from compact clinical [3], interventional [5], to whole body high field magnet [6].

Method: The MSE method of designing superconducting magnets consists of two independent steps:

(1) To determine the MSE current density map over a predefined magnet domain, where superconducting coils will be laid out, subject to constraints, such as the homogeneity of the FOV and the size of the magnet footprint. The optimal MSE current density map is determined by changing the size of the magnet domain and by adjusting the number of internal and external spherical harmonic coefficients (defining the magnetic field) to be vanished.

(2) To determine the magnet arrangement with the initial coil layout based on the MSE current density map. The coil locations and sizes are refined to enhance the field homogeneity and to decrease the footprint of the magnet, without changing the MSE current density map constraints.

Particulars and greater detail of the MSE method can be found in [2, 3]. Simply, the magnet domain is divided into small elements and their current density distribution referred as the MSE current density map which is solved by minimising the objective function (F) subject to particular constraints:

$$\min_{I_k} F = \frac{1}{2}\sum_{k=0}^{K-1} L_k I_k^2 A_k^2, \quad \text{subject to}: \sum_{k=0}^{K-1}\alpha_{k,1} I_k = \frac{B_0}{2}, \quad \sum_{k=0}^{K-1}\alpha_{k,2n+1} I_k = 0, \quad n = 1 \ldots \frac{N}{2}, \quad \sum_{k=0}^{K-1}\beta_{k,2m+1} I_k = 0, \quad m = 0 \ldots \frac{M}{2}, \quad I_{\min} \le I_k \le I_{\max},$$

where, A_k is the wire cross-sectional area corresponding to element k and L_k is the associated self inductance [7]. The above equation is solved subject to linear constraints on the magnetic field strength, the vanished N-1 internal (a) and M external (b) spherical harmonics to obtain appropriate field linearity within the Field-of-View (FOV). The current densities I_k are bounded to ensure that the conductor superconductivity criterion is not breached. Once the current density map has been obtained through quadratic program (QP) [8], the positive maxima and negative minima extremities of the resulting current density map are used as seed data to configure the individual coils. After the seed coil locations have been defined in this manner, the coils are further optimised using the following cost function [3]:

$$\min_{I, L_k} F^* = \frac{I^2}{2}\sum_{k=0}^{K^*-1} A_k^2 L_k, \quad \text{subject to}: I\sum_{k=0}^{K^*-1}\alpha_{k,1} = \frac{B_0}{2}, \quad I\sum_{k=0}^{K^*-1}\alpha_{k,2n+1} = 0, \quad n = 1 \ldots \frac{N}{2}, \quad I\sum_{k=0}^{K^*-1}\beta_{k,2m+1} = 0, \quad m = 0 \ldots \frac{M}{2}, 0 < I \le I_{\max},$$

where, I is the current density in the coils. To solve this equation, sequential quadratic programming (SQP) [9] was implemented subject to the constraints. Figure 1 provides a particular result for the MRI/PET superconducting magnet coil arrangement. In the figure, the proposed location for the PET camera is at the centre of the magnet. This magnet coil arrangement provides a FOV of 20cm in diameter at 1ppm, a footprint of approximately 6.5m^2 to the 5 Gauss level and a length of 0.8m. We have also computed the magnet stored energy and coil stresses, which will be provided as part of the presentation.

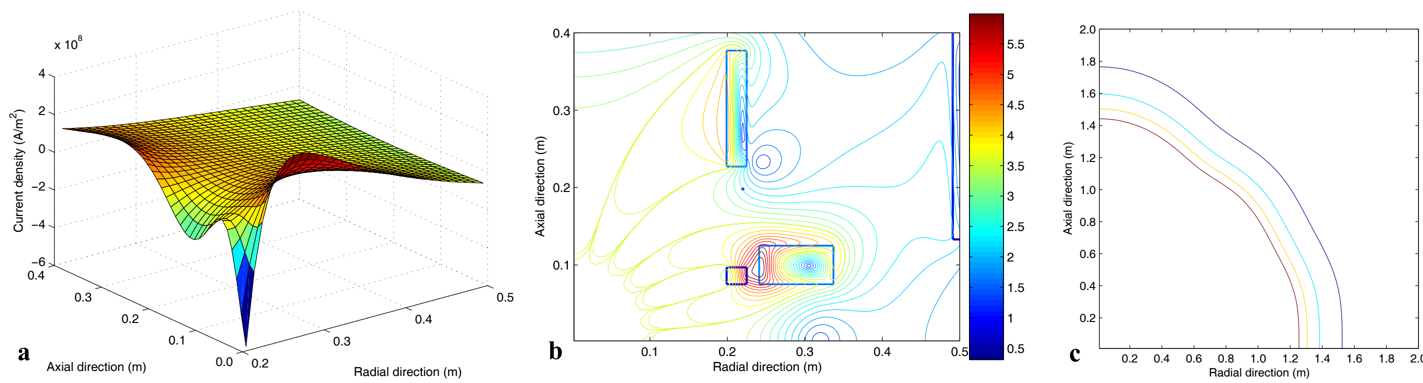

Figure 1 Illustrated is a particular result of the open 3T superconducting magnet designed for combined MRI/PET imaging. Shown are (a) the current density map obtained using Step 1 from this initial seed coils for Step 2 are defined, (b) the final optimised coil arrangement and magnetic field produced, as obtained using Step 2 and (c) the stray field associated with the coil arrangement. The smallest inner and outer coils in (b) are negative current coils. All coils have a transport current of 250A using 1mm by 1mm NbTi superconductor. In (b) the most inner contour corresponds to 1ppm variation, and in (c) 5, 10, 15 and 20 Gauss contours are depicted, outside in. The magnet coil arrangement has an inner diameter of 0.4m and outer diameter of 1m, length of 0.8m and a footprint comparable to current symmetric compact magnet designs. The peak field on any superconductor is less than 5.8T, which is well below the NbTi limit.

References

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