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Theoretical design of gradient coils with minimum power dissipation: Accounting for the discretization of current density into coil windings

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ABSTRACT

Gradient coil windings are typically constructed from either variable width copper tracks or fixed width wires. Excessive power dissipation within these windings during gradient coil operation limits the maximum drive current or duty cycle of the coil. It is common to design gradient coils in terms of a continuous minimum power current density and to perform a discretization to obtain the locations of the coil tracks or wires. However, the existence of finite gaps between these conductors and a maximum conductor width leads to an underestimation of coil resistance when calculated using the continuous current density. Put equivalently, the actual current density within the tracks or wires is higher than that used in the optimization and this departure results in suboptimal coil designs. In this work, a mapping to an effective current density is proposed to account for these effects and provide the correct contribution to the power dissipation. This enables the design of gradient coils that are genuinely optimal in terms of power minimization, post-discretization. The method was applied to the theoretical design of a variety of small *x*- and *z*-gradient coils for use in small animal imaging and coils for human head imaging. Computer-driven comparisons were made between coils designed with and without the current density mapping, in terms of simulated power dissipation. For coils to be built using variable width tracks, the method provides slight reductions in power dissipation in most cases and substantial gains only in cases where the minimum separation between track centre-lines is less than twice the gap size. However, for coils to be built using fixed width wires, very considerable reductions in dissipated power are consistently attainable (up to 60%) when compared to standard approaches of coil optimization.

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1. Introduction

In magnetic resonance imaging (MRI), the gradient coils are responsible for providing the spatial encoding to the MR signal. Their function is to induce strong magnetic fields that vary linearly in three orthogonal spatial directions, which can be switched independently at a rate that is sufficiently fast for the MR pulse sequence at hand. Satisfying these criteria is necessary for achieving the goal of high resolution images and short scan times. It is therefore common to assess gradient coil performance in terms of gradient homogeneity, coil efficiency and inductance [1,2].

However, in reality a great many additional factors must be considered and there is no single coil design, or indeed design technique, that is appropriate for all applications. Such considerations include the minimization of dissipated power [3], the minimization and compensation of eddy currents [4,5], the minimization of torque [6], the avoidance of peripheral nerve stimulation [7,8], the reduction of acoustic noise [9,10], the existence of a minimum separation between adjacent coil conductors [11,12], and the minimization of peak operational temperature [13]. The gradient coil design problem therefore becomes one of optimizing the trade-off between various competing criteria [14] and finding an appropriate metric of optimality for any given application.

For the case of small gradient coils, such as gradient inserts or those used in small animal imaging or MR microscopy, which inherently possess sufficiently low inductance, the primary design criterion is typically the minimization of dissipated power. Power dissipation results in a rise in the average temperature of the system and this limits either the maximum current and therefore the maximum gradient strength, or the maximum duty cycle and therefore the imaging speed [15]. The present paper deals with the design of minimum power coils.

It is common to cast the gradient coil design problem in terms of a distributed surface current density over a surface representing

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the coil former. This approach is both efficient and convenient, since many design parameters can be expressed as linear or quadratic functions of the current density. For example, the magnetic induction vector can be calculated using the Biot-Savart law, whereas power dissipation can be described by the square of the current density integrated over the coil surface. Therefore, a standard technique is to minimize this measure of power dissipation with respect to the current density vector whilst maintaining an acceptable level of gradient field homogeneity [2]. Note that for the purpose of convenience we use the term "standard" throughout the manuscript to refer to this technique, however it is important to stress that there exist several variants to this method for obtaining coils with low power dissipation.

Once a current density solution has been obtained, this must be discretized in some appropriate fashion to obtain the positions of the gradient coil windings, whilst maintaining a satisfactory induced field. Typically, this can be achieved by contouring the associated stream function [16,17]. Coil fabrication then proceeds usually in one of two ways; either a number of copper sheets are cut into a series of coil tracks of variable width, using water-jet cutting, milling or chemical etching, for example, or copper wires of a fixed thickness are machine or hand wound onto the coil former. Coil tracks of variable width have lower resistance and provide a closer approximation to the distributed current density, and these are typically used for building *x*- and *y*-gradient coils. A minimum gap between tracks is required to ensure electrical isolation and also represents a limitation of the cheaper manufacturing methods. A maximum track width is prescribed to ensure that the current flows in the correct places, since electric current tends to take the path of least resistance and therefore will deviate from its ideal position in wide, curved tracks [18]. In contrast, wires of fixed thickness provide a simpler manufacturing method and are typically used for building *z*-gradients or coils built on complex surfaces [19,20]. Furthermore, wire-wound coils are often used in MR-microscopy [21] or in other cases to allow hollow wires with internal water cooling [22].

Regardless of which construction method is used, both represent an approximation to the original current density solution. While this approximation is reasonable in terms of calculating the induced magnetic field within the region of interest and the coil inductance, it can lead to gross errors in the calculation of coil resistance. As such, the resulting gradient coil is no longer optimal with respect to power dissipation. For example, it has been observed that a reduction in coil resistance is possible by deviating from the supposed minimum power solution and spreading out some of the more closely packed windings [11,18,13].

There exists another class of gradient coil design methods, referred to here as discrete coil design, that attempt to simulate and adjust the wire positions directly, rather than considering a distributed current density. These methods provide accurate estimates of coil resistance; however, due to the computational difficulty associated with calculating the resistance, inductance and magnetic field, these methods are limited to specific coil geometries and furthermore they often require troublesome stochastic optimization [23–26]. In a sense, the present work bridges the gap between the continuous and discrete approaches; it approximates more accurately the current density and power dissipation when built, while retaining the ability to design coils with arbitrary shape and to use well-behaved optimization techniques.

In this work, a design method is presented that accounts for the method of coil discretization and the effect it has on coil resistance and power dissipation calculations in the low frequency limit. That is, the method produces genuine minimum power coil designs for cases in which the construction method is known prior to the theoretical design stage. Note that the study is concerned entirely with investigating changes in simulated power dissipation as a result of altering the coil winding pattern and does not deal with the mechanical and material engineering aspects of coil fabrication. The key idea behind the technique involves a mapping to an effective current density used in the minimization and this will be described in the following section. In Section 3, a survey of results will be presented for some common gradient coil types. These will be discussed in Section 4 and some concluding remarks will be given in Section 5.

2. Methods

A coil design method can be split into three parts. The first establishes the geometry of the coil surface and parameterizes the surface current density in terms of free variables that define the coil. It also defines how these variables relate to the electromagnetic coil properties. The second step arranges an optimization function that will give the best coil for a given application and then solves this with an appropriate algorithm. Finally, the relation between the free variables, the optimal solution and the coil definition is used to produce the physical coil design and simulate its performance.

2.1. Parameterization and standard minimum power coil design

In all coils we first establish a geometry, or surface Ω, and define the stream function, ψ, of the surface current density, $\boldsymbol{j}_c$, as a weighted sum of basis functions, Ψ_n. That is,

$$\psi(\mathbf{r}') = \sum_{n=1}^{N} \psi_n \Psi_n(\mathbf{r}'), \tag{1}$$

where $\mathbf{r}' \in \Omega$. The current density on the coil surface can be obtained by taking the curl of the stream function:

$$\boldsymbol{j}_c(\mathbf{r}') = \nabla \times [\psi(\mathbf{r}')\hat{\mathbf{n}}(\mathbf{r}')], \tag{2}$$

where $\hat{\mathbf{n}}(\mathbf{r}')$ is the unit surface normal at $\mathbf{r}'$. The vector of stream function coefficients $\boldsymbol{\psi} = [\psi_1, \psi_2, \ldots, \psi_N]$ then defines everything about the coil. The magnetic field, $B_z(\mathbf{r})$, away from the surface (i.e. $\mathbf{r} \notin \Omega$) can be calculated from the current density using the Biot-Savart law, and the power dissipation, P, is proportional to the square of the current density integrated over Ω.

The choice of geometry and basis functions, $\Psi_n(\mathbf{r}')$, follows, but is not limited to, two approaches in this work. The sum-of-sinusoids approach [27,28] was used in the design of symmetric, unshielded, cylindrical gradient coils (see also [29] for pertinent expressions) and the axisymmetric boundary element method [30,31] was used in the case of asymmetric, actively-shielded, head-only gradient coils. Note that cylindrical geometry has been considered in both cases for reasons of convenience in testing the design concept. The same methods could be applied in a straightforward manner to the design of coils with altogether different underlying structure.

Standard minimum power gradient coil design may proceed by minimizing, with respect to the stream function coefficients, a functional of the form:

$$F = \Phi + \lambda P. \tag{3}$$

The term Φ represents the deviation of the induced field, $B_z(\boldsymbol{j}_c)$, from a target field, B_t, at a set of points that span the imaging volume of interest (e.g. the square of the Euclidean or l_2-norm of the field error), and the term P is the power dissipation in the coil. The weighting λ controls the trade-off between field error and power dissipation and must be sufficiently large to obtain a well-conditioned system. Alternatively, constrained optimization may be used to solve the problem.

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