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Skin and proximity effects in the conductors of split gradient coils for a hybrid Linac-MRI scanner

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ABSTRACT

In magnetic resonance imaging (MRI), rapidly changing gradient fields are applied to encode the magnetic resonance signal with spatial position; however eddy currents are induced in the surrounding conducting structures depending on the geometry of the conductor and the excitation waveform. These alternating fields change the spatial profile of the current density within the coil track with the applied frequencies of the input waveform and by their proximity to other conductors. In this paper, the impact of the conductor width and the excited frequency over the parameters that characterise the performance of split transverse and longitudinal gradient coils are studied. Thirty *x*-gradient coils were designed using a "free-surface" coil design method and the track width was varied from 1 mm to 30 mm with an increment value of 1 mm; a frequency sweep analysis in the range of 100 Hz to 10 kHz was performed using the multi-layer integral method (MIM) and parameters such as power loss produced by the coil and generated in the cryostat, inductance, coil efficiency (field strength/operating current), magnetic field profile produced by the coil and the eddy currents were studied. An experimental validation of the theoretical model was performed on an example coil. Coils with filamentary conductor segments were also studied to compare the simulated parameters with those produced by coils with a finite track. There was found to be a significant difference between the parameters calculated using filamentary coils and those obtained when the coil is simulated using finite size tracks. A wider track width produces coil with superior efficiency and low resistance; however, due to the skin effect, the power loss increases faster in wider tracks than in those generated in coils with narrow tracks. It was demonstrated that rapidly changing current paths must be avoided in order to mitigate the power loss and the spatial asymmetry in the current density profile. The decision of using narrow or wider tracks in split coils should be carefully investigated using a temperature analysis which includes skin and proximity effects.

1. Introduction

Gradient coils are one of the essential components in magnetic resonance imaging scanners as they are used to encode the nuclear magnetic resonance signal with spatial position. They also play an important role in determining how fast and accurately an image is obtained [1].

Gradient coils are usually etched or machined using a copper sheet with track widths varying from 6 mm to 30 mm and consist of straight strips, circular loops, arcs and their combinations [2]. The coils are designed to produce a linear variation of the axial component of the magnetic field along the Cartesian coordinates in the DSV [2]. Eddy currents are inevitably induced when gradient coils are quickly switched on and off which causes energy loss, mechanical vibration and Joule power deposition in conductive parts of MRI scanners [3,4]. More importantly, eddy currents produce deleterious effects on the desired gradient field, leading to distortion and artefacts in the images [5]. The use of litz wire is one of the possible solutions used to lessen the severity of the skin effect and consequently lower the effective resistance and resistive losses. However, gradient coils manufactured using litz wires can be an expensive process as grooves typically need to be machined in fibre glass tubes to position the wires and this can be particularly difficult for some transverse coil designs [6].

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To guarantee uniform gradient in the DSV and high quality images, a range of parameters such as minimum inductance, resistance, self-inductance, field linearity, wire density and shielding need to be considered carefully in the gradient coil design process [2]. A variety of different design methods have been developed to effectively constrain these parameters [2,7,8]. Some authors have introduced "free-surface" coil design methods [9–11] based on a pioneering work of Pissanetzky [12]. These methods are able to produce gradient coils on arbitrary surfaces and, at the same time, control coil parameters with linear and quadratic dependence on the unknown current density. The current density is expressed as a function of a set of basis functions which represents the value of the stream function in the coil domain [13]. Equally spaced contours of the stream function define the position of the current paths [13]. Commonly, the Biot–Savart law is used to evaluate the magnetic field produced by the coil assuming discrete and infinitely thin wire segments or tracks of finite dimensions. Usually coil tracks with a uniform current density are used to evaluate the magnetic field and other parameters from the designed coil. However, due to the skin and proximity effects, the current density tends to be non-uniform across the width of the conductor [14] and it is known that these effects increase with the frequency [14]. Not taking into account the skin and proximity effects may lead to inaccurate modelling of the magnetic field profile expected from the gradient coil design. Moreover, it is known that power loss tends to increase with the frequency but the implications of the skin and proximity effect have not been investigated in detail for MRI gradient coils. A mathematical model that explains coupling between tracks caused by edge-effects was developed by Kroot et al. [14,15]. However, in his work, only the current distribution in strips and islands were studied, but the effects caused by a non-uniform current density distribution along the track width on the coil parameters and field harmonics were not considered.

In this paper, we study the impact of the coil track width and frequency on the coil performance taking into account the skin and proximity effects for transverse split gradient coils [3]. Coil performance is characterised by the following coil parameters: resistance, inductance, field harmonic and power loss produced by the coil and the surrounding cryostat, shielding ratio, figure of merit (FoM) and η^2/R (where η is efficiency and R is resistance). The shielding ratio is measured as ratio of the maximum absolute value of the field produced by the eddy current and field produced by the coil ($\max(|B_z^{eddy}|)/\max(|B_z^{coil}|) \cdot 100\%$) in the region of interest (RoI) at a given frequency. The FoM is defined as η^2/L where L is inductance and η is efficiency [2,16]. We use the equivalent magnetisation current method to design split and shielded whole body x-gradient coils, in both scenarios track width changes from 1 mm to 30 mm [11,17]. The multi-layer integral method is extended to analyse the designed coils and predict the influence of the track width and frequency on gradient coil performance [18].

2. Method

2.1. Coil design

In order to study the influence of the track width and frequency variation over the coil parameters we designed 30 whole-body x-split gradient coils using equivalent magnetisation current [11]. The track width was changed linearly from 1 mm to 30 mm, the target gradient strength G_0 was kept at a constant value of 10 mT/m for all designs, the field linearity was fixed to 5% in a DSV of 460 mm and the target FoM was kept constant for all the designs.

We focused our study on the x-folded gradient coil due to its geometrical complexity compared to that of the z-gradient coil. In this paper we neglected the inductive coupling between the x-coil and the surrounding coils. This assumption simplifies the model and significantly reduces the computational burden and the number of variables that may influence the skin and proximity effect due to the coil self-inductive coupling. However, this represents a simplification of the scenario that is present in conventional MRI scanners. Depending on the spatial distribution of current and track width of the surrounding coils, it is possible for inductive coupling to occur between the x-coils and the rest of the gradient and shim coils. This complex interaction will be considered in a detailed analysis in future work. Fig. 1 shows the split coil support, DSV and a simplified shield cryostat used to design the split gradient coils.

The track width was changed from 1 mm to 30 mm with equal steps of 1 mm to produce a total of 30 x-gradient coils. The minimum gap between wires was constrained to 2.8 mm. Each of the modelled coils were simplified to 24 turns (4 turns per quadrant), as both the computing time and memory requirements increases with a factor O (N^2), where N is the number of nodes. A current of 1065 A was used to generate the target field of G_0. High current was required due to the small number of turns used to model the coil. The conductor thickness was uniform for each of the designed coils. We assumed that the coil domain is made of 2 mm thick copper sheet for the 30 coils. Two conductive cryostat cylinders, warm bore and cold shield were used as a simplified cryostat to analyse the eddy currents. The split cryostat structure is shown in Fig. 1b. All the properties of the split transverse coils are listed in Table 1.

2.2. Extending the MIM

The multi-layer integral method (MIM) was used for eddy current simulations and to analyse inductive interaction between the

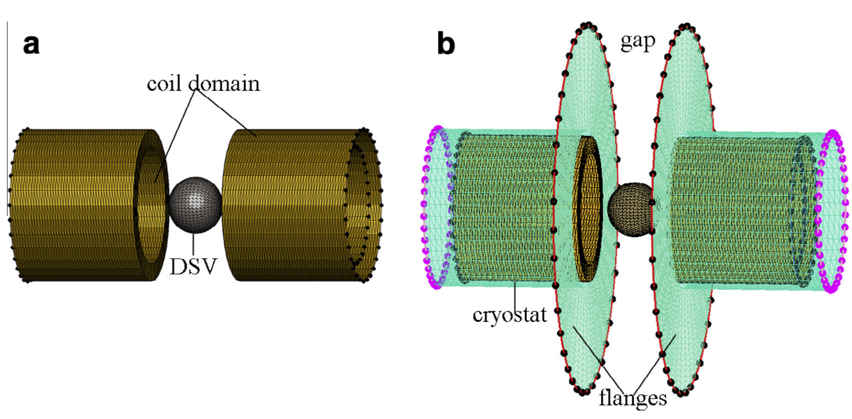

Fig. 1. (a) Split gradient coil geometry support and (b) simplified split cold shield cryostat.