Discrete Design of an Insert Gradient Coil for Head Imaging in High Field MRI

S. Shvartsman[1], G. DeMeester[1], M. Morich[1]
[1]Philips Medical Systems, Cleveland, Ohio, United States

Introduction

In this work we present designs for a transverse head insert gradient coil using a discrete coil design method. The targets for the gradient are: 60mT/m gradient strength, 600mT/m/ms slew rate, and 24cm Field-of-View (FoV). The gradient is sized to fit within a whole body gradient coil. In designing such a gradient coil we use the method of z-intercepts. The concept of this approach is that all characteristics of a transverse gradient can be expressed through the set of z-intercepts where the coil current patterns intersect the cardinal axis φ=0, for an X-gradient. Unshielded symmetric design with shoulder cutouts and shielded asymmetric design are both investigated with the new method.

Method

Small diameter gradients for head imaging take advantage of the fact that the stored magnetic energy scales as radius to the fifth power. This means higher efficiency, less inductance to provide high slew rate, lower Lorentz forces, and a smaller stiffer structure for potentially reduced acoustic noise. In addition there is less volume of a patient in the high d**B**/dt field to avoid peripheral nerve stimulation. These performance advantages motivate use of small diameter gradients at high fields, e.g., ≥3T, for proton MRI head studies. Corresponding references could be found in the recent paper [1]. There are many challenges in designing a high performance insert gradient. One key challenge is to have an acceptable head-neck access, due to the patient shoulders restrictions, while maintaining good field quality characteristics. The approaches to overcome this problem include using a symmetric unshielded insert gradient with shoulder cutouts [2], or an asymmetric shielded gradient [3,4]. We propose a new method of transverse gradient coil design using a discrete current density function. It is shown that all characteristics of a transverse gradient, such as coil inductance, spatial behavior of the magnetic filed, and coil resistance can be expressed in terms of a set of z-intercepts $\left\{Z_n^{(in)}, Z_n^{(f)}\right\}$. These are determined as points where the coil current centroids intersect the cardinal axis at φ=0 for an X-gradient or at φ=π/2 for a Y-gradient. The method does not require some of the essential steps of [5] such as finding continuous current densities and discretizing. The z-intercepts positions, and thus the minimum conductor width, are varied. During this variation the stored magnetic energy/inductance of the coil, field quality characteristics within the FoV, and coil's resistance are monitored until desired targets are reached. Uniformity of the residual eddy current effect (RECE) over the FoV is calculated *a posteriori*. Forces and torques are also calculated from the discrete current pattern and knowledge of the Bo field.

Results

We have applied the discrete design method to both unshielded symmetric and shielded asymmetric gradients. Shoulder cutouts are used in the unshielded symmetric case. For compatibility with whole body gradients, the internal diameter of the insert gradient is equal to 360mm. The outer diameter of the insert gradient depends on the topology: 430mm for an unshielded symmetric gradient and 560mm for the shielded asymmetric case. The performance targets in both cases are: 60mT/m gradient strength, slew rate ≥ 600mT/m/ms, good field quality characteristics over 24cm FoV, 175mm patient access (clear axial distance from the imaging center to the patient shoulders). For purposes of establishing the number of coil turns, the target current and voltage is set to 600A and 600V.

In the special case of shoulder cutouts, the difficult axis is the gradient whose cardinal axis is horizontal/left-right. The design method begins with a "pre-discretized" coil that carries current **I** and is determined by the set of z-intercepts. The design method in this special case consists of the following steps: 1) Find the set of z-intercepts that provide the biggest possible (preferably positive) gradient strength non-uniformity over the FoV (gradient falloff in z-direction). The gradient strength non-linearity is preferably negative; 2) Remove several turns near the coil's eye to allow space for creation of the shoulder cutouts. This step reduces the non-uniformity and the coil sensitivity, while increasing, in a positive sense, the gradient strength non-linearity. 3) Reshape the current pattern at the access end of the coil to construct the shoulder cutout. The latter step further decreases the non-uniformity and coils sensitivity, while again increasing, in a positive sense, the non-linearity. This procedure continues until the targets are achieved. Results for this special case are provided in Table 1. Each quadrant of the gradient consists of 6 turns carrying current of I=585.5A, producing G=60mT/m. The current patterns are shown in Figure 1. The inductance of the coil is 73.9μH. The resistance of the coil is 20.1mΩ. Within the 24cm FoV the gradient strength non-uniformity and non-linearity are ~-42% and ~3.2%, respectively. Gradient strength non-uniformity and non-linearity plots are shown in Figure 2. The residual eddy current effect over the 24cm FoV is –1.55% with 0.5% range. For comparison, the properties of a shielded asymmetric gradient are also provided in Table 1. When considering application in a large bore 7T system we find in the asymmetric case there are no theoretical problems of torque and force balancing due to the highly homogeneous magnet field. Likewise, in the symmetric case we find force balancing is not an issue. The performance characteristics of the two coils are comparable.

Conclusion and Discussion

We have designed insert gradient coils with a discrete design method for application at ≥3T. The gradient dimensions are compatible with insertion into whole body gradients. We have considered two topologies: unshielded symmetric insert gradient with shoulder cutouts and a shielded asymmetric insert gradient. In both cases the target performance of 60mT/m gradient strength and 600mT/m/ms slew rate can be achieved using 600A and 600V at the gradient. In this analysis cable and filter losses are ignored. Comparisons of the characteristics of the coil for both topologies are listed in Table 1 below.

References

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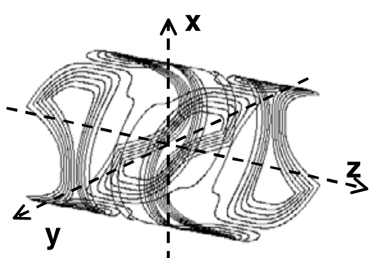

Fig. 1

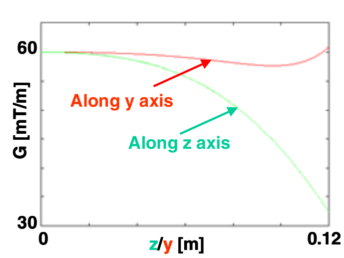

Fig. 2

Table 1

Property	Unshielded symmetric	Shielded asymmetric
Patient access [mm]	175	168
Sensitivity [μT/m/A]	103	112.6
Inductance [μH]	73.9	98.0
Resistance [mΩ]	20.1	20.2
SR [mT/m/ms] at 60mT/m and 600V	819	677
|**B**| [mT] G=60mT/m at ρ = 0.1m	7.95	7.87
Non-linearity over 24cm FoV	+3.2%	+3.7%
Non-uniformity over 24cm FoV	-42%	-42%
RECE range	0.5%	0.25%