Design of Gradient and Shim Coils for a Head-Only, Vertical, HTS MRI System

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Introduction The design of a gradient and shim set for a high-temperature superconducting (HTS), vertical, human head-only MRI system [1] is described in this paper. Diffusion and functional MRI studies (for example) benefit from increased gradient strength and slew-rate which is achieved in a head-only system by reducing the diameter of the gradient coils; gradient strength scales with the radius to the power of -2 (a^{-2}) and the slew rate scales with the radius to the power of -3 (a^{-3}). The presence of the shoulders of the subject means that the gradient coil cannot be extended a in the patient direction and so asymmetric cylindrical designs are adopted for head-only gradient and shim coils [2,3]. The manufacturing technique, restricted space and the special type of HTS magnet all impose different priorities and constraints on the design of the gradient and shim coils. These are presented here as well as a description of the approaches taken to address these priorities.

Methods

Geometric constraints: The inner (ID) and outer diameters (OD) of the cylindrical gradient and shim coil set were 350 mm and 494 mm respectively, with axial extents of 180 mm and 410 mm in the patient (downwards) and service (upwards) directions respectively. The region of interest was an oblate spheroid with major and minor diameters of 250 mm and 200 mm respectively.

Manufacturing issues: The primary constraint placed on the coil design from the manufacturing method is the minimum practicable wire spacing. These gradient coils are being constructed by a chemical etching process with 2 layers of copper (each layer 1.5 mm thick) and a central layer of FR4. This approach is convenient in that no return paths are required as connections can be made from one layer to another to ensure the correct current flow in the whole coil. Chemical etching requires a minimum distance between centres of adjacent paths was required to be 22 mm on each layer (11 mm for design purposes). For this reason we use the minimax|j| technique to spread the wires apart in critical areas whilst maintaining field linearity and some power minimisation [4].

Desired performance: The gradient field was required to have an accuracy better that 5% in the ROI. Minimal power was desired, which naturally exhibits low inductance for high slew rates. Coil efficiency, η, was maximised to obtain high gradient strengths.

Protecting the HTS magnet: Active shielding is required to prevent eddy currents that are generated in the magnet cryostat structures from disrupting the imaging process. Moreover, eddy currents in the cryostat vessels are known to cause heating and vibration as well as the secondary eddy-current-induced magnetic field effects. This system will employ a HTS magnet and so the power of the eddy currents in the cryostat were minimised as much as possible; rather than minimising the field produced from the eddy currents, as some approaches do.

Coil design: Coils were designed using an axisymmetric boundary element method [5] and minimised **a)** the sum of squares of the field error, **b)** the power deposited in the cryostat vessel, **c)** the power of the coil and **d)** the maximum current density magnitude [4] whilst **e)** ensuring zero net torque. In the φ direction, cos($m\varphi$) variation was used with m = 1, 3, 5 and 7. Shim coils were designed in the same manner but ensured zero coupling with the gradient coils and are not actively shielded.

Performance simulation: The performance of the coils were simulated with Biot-Savart law, Fasthenry inductance simulation and our new eddy current simulation method [6].

Results Table 1 shows the important performance properties of the designed gradient coils and Figure 1 shows one half of the X gradient primary and shield coils. Inductance and resistance of the coils are naturally low, so prioritising the wire spacing increase the inductance and resistance, but only by a tolerable amount. The efficiency, η, is more than doubled by using minimax|j| minimisation for the same minimum wire spacing (min(P) designed X-gradient coil had N = 7 and η = 35.8 μTm^{-1}A^{-1}).

Coil type	a_P/a_S (mm)	N	η (μTm^{-1}A^{-1})	max(ΔB_z)	min(ws) (mm)	L (μH)	R (mΩ)
Z	182/226	10	160	4.90%	5.3	130	79
X	195/237	14	73.4	5.10%	11.2	67	26
Y	200/242	13	65.5	4.40%	11.2	64	25

Table 1. Gradient coil performance properties; including the primary and shield radii (a_p and a_s), number of contour levels (N), efficiency (η), field error (ΔB_z), wire spacing (ws), inductance (L) and resistance (R).

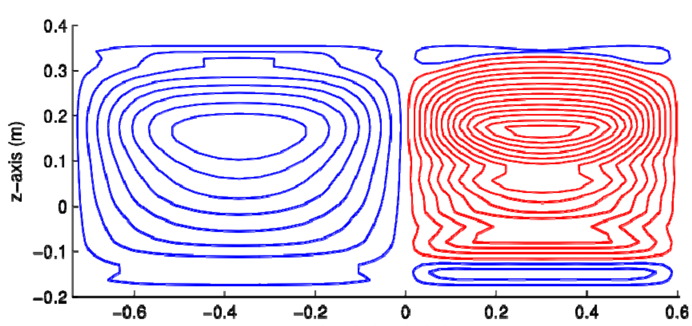

Figure 1. Wire-paths for ½ the X-gradient coil primary (right) and shield (left) coils. Current flows in the opposite sense in red wires with respect to the blue wires. Every odd level wire loop is placed on the top layer, and even on the bottom layer.

Discussion and Conclusions Gradient coils for head only imaging have been designed and built before, e.g. [2,3]. It is known that in such coils there is a difficulty in designing practicable coils because the highly-restricted space at the patient end results in very close and thin wires. So in the present work we use a technique to spread the wires [4] to produce a strong set of head gradient coils. These coils have more than double the efficiency (105% more) for the transverse axes with a 20% increase in the Z gradient when compared to standard min(P) designs. Some areas of the coils have uniform wire spacing due to the minimax|j| design as can be seen in Fig. 1. Also, particular care was taken to reduce the total amount of eddy currents flowing in the cryostat because of the experimental nature of the HTS magnet that these coils will operate in. This is to avoid as much as possible any heating and/or vibration in the magnet cryostat. A 4-fold reduction in total eddy current power was predicted when compared to coils designed to minimise the field generated by the eddy currents in the ROI.

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