A biplanar shielded gradient coil design for hyperpolarized gas lung imaging in the standing position

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Introduction: This study presents the design of an optimized shielded biplanar gradient coil set to be used in an open permanent low field (0.2T) magnet, which will be used for lung MRI of protons and inhaled hyperpolarized (HP) gas (He-3) with subjects in the standing position. Many formulations have been proposed for the design of shielded gradients over recent years [1, 2]. Our design was based on a modified version of the coil set proposed by Tomasi et al [3].

Methods: Simulation of the gradient field using the Biot-Savart law for both longitudinal and transverse gradient coils was performed with Matlab. The design used four planes perpendicular to the z axis where the primary and shielding planes were placed at z = ±a and z = ±b (b>a) respectively for an efficient shielding effect. The centre of the planes was placed at *±180mm*. The gradient optimization used a simplex search method [4] to get the radii (R_p,R_s) for the longitudinal coil (Gz) and the positions (X_p,X_s,Y_p,Y_s) of the straight wires for the transverse coils (Gx, Gy). A half-size prototype of the Gz coil was built (using *0.1mm* diameter copper wire) and MR phase imaging was used for testing the gradient field against the simulations. The full size coil was built using *2mm* diameter copper magnet wire (Comax, UK) wound on two circular *600mm* diameter fibreboard formers with edge slots cut at the appropriate locations to hold the Gx and Gy coils. Wires were fixed in place using fibre glass tape and silicon based adhesive. Fields were measured using a Hall probe Gaussmeter.

Longitudinal gradient coil: The Gz coil used 34 circular wires (with radius R_p, $p = 1$ *to 34*) for the primary plane and 58 circular wires (with radius R_s, $s = 1$ *to 58*) for the shielding (see Table 1). The wire distribution in both primary planes was identical to maximize the field uniformity. The currents in each plane were opposite; the shielding coil current was a quarter of the primary coil current and in opposite direction to the respective primary coil current in each plane [3].

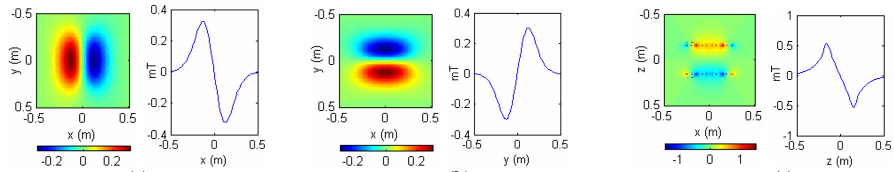
Transverse gradient coil: The Gx and Gy coils used similar designs of *84* wires of various length at $\pm X_p$ and $\pm Y_p$ *(p=1 to 42)* respectively on each primary plane, and another *84* wires of various length at $\pm X_s$ and $\pm Y_s$ *(p=1 to 42)* for the shielding planes (see Table 1). The primary coil currents flow in the same direction for all wires: along y for the Gx coil, and along x for the Gy coil, respectively. The shielding coil is used as a return path for the primary coil current, hence the current flows in the opposite direction to the primary coil current.

Results:

Simulations: Table 1 records the optimized radii of the Gz coil and the positions of the Gx and Gy coils. Fig.1 shows the simulated fields for each of the coils driven with *7A* current and the gradient profile along the middle of each of the respective planes. The field drops to zero outside the shielding plane. The b/a ratio was *1.338*, *1.264*, and *1.195* for the Gx, Gy, and Gz coils respectively which varied because of the coil thickness and the number of overlapping turns required. The simulations gave efficiencies of *0.46, 0.43* and *0.51 mT/m/A* for the Gx, Gy and Gz coils respectively. The theoretical values of inductance were *6.31, 5.55, 0.54mH* and the resistance were *0.46, 0.45, 1.19Ω* for the Gx, Gy, Gz coils respectively.

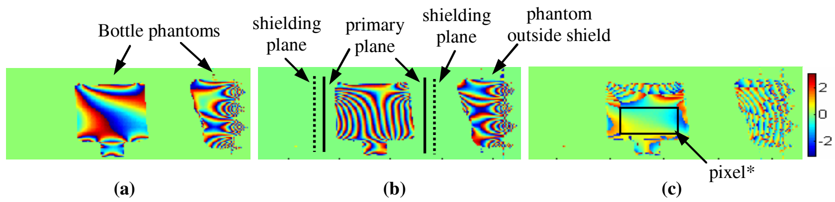
Table 1: Optimized radii and positions for the biplanar gradient coil.

R_P (mm) 2 turns each	R_s (mm) (*N*) indicate *N* turns with same radius.		X_p or Y_p (mm) 3 wires each	X_s or Y_s (mm) 3 wires each
54	34	216 (4)	± 0	± 2
80	54	226 (4)	± 18	± 11
150	64	236 (4)	± 30	± 25
160	74	246 (4)	± 50	± 46
166	84	256 (3)	± 51	± 167
176	94	266 (3)	± 52	± 168
178	104	276	± 52	± 170
182	114		± 53	± 175
198	124		± 64	± 175
204	134		± 66	± 177
206	146 (2)		± 67	± 177
212	156 (3)		± 70	± 177
226	166 (3)		± 88	± 179
230	176 (4)		± 180	± 185
240	186 (4)			
244	196 (5)			
252	206 (4)			



(a) (b) (c)

Fig.1: Simulated fields (*mT*) for (a) Gx coil on x-y plane, (b) Gy coil on x-y plane, (c) Gz coil on x-z plane.



(a) (b) (c)

Fig. 2: Half-size Gz prototype: (a) Phase image acquired without current applied to the Gz coil. (b) Phase image acquired with *28mA* applied to the shielded Gz coil. (c) Residual Gz coil's inhomogeneity. Rectangle shows a target region. (*pixel with largest phase shift due to inhomogeneity.)

Half-sized Gz prototype: Figs.2a and 2b show the phase images (sagittal plane) obtained on a 1.5T MR scanner where a bottle phantom was placed in the centre of the coil and another placed outside one of the shields. When a current (*28mA*) was applied to the Gz coil (Fig.2b), the phase image showed the contours of the constant Bz field from the Gz coil but only a small effect on the phantom outside the shield. The linear Gz gradient was obtained by linear fitting a first order polynomial to the linear phase ramp in Fig.2b (using function *polyfit* and *polyval* in Matlab) and was found to be approximately *4.0mT/m/A* along the coil axis of the half-size Gz prototype. The linear gradient was then taken out to investigate any residual inhomogeneity due to the Gz coil as shown in Fig.2c, which was fairly smooth in the target region. The largest phase shift in this region due to the inhomogeneity was *2.77 radians* at the bottom right corner pixel (see arrow), equivalent to ≈ *0.62μT*. The measured fields of the shielded and unshielded Gz coils along the coil axis are displayed in Fig. 3. The ratio of *shielded:unshielded* field is 0.28 at $z = 15cm$ if both peaks are normalised to the same value, which is a suppression of 72% of the z component of the magnetic field.

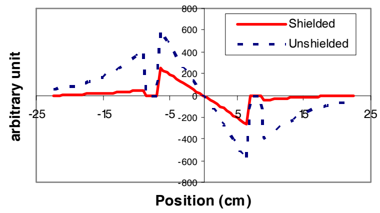
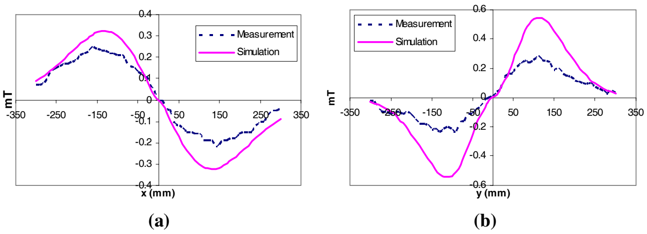


Fig. 3: Measured fields of the half-size Gz coil using an oscilloscope from the signal induced in a search coil when the Gz coil was driven with a low frequency AC current.

Full-size coil: The efficiencies of the full-size shielded biplanar coils, measured from the mean field strength, were *0.39, 0.35 and 0.39 mT/m/A* for the Gx, Gy and Gz coils respectively which are in reasonable agreement with the predicted values. Figs.4a and 4b show the measurement and simulation results for the field strength (in *mT*) of the Gx and Gy coils when driven with a current of *7A*. From these measurements, the region of uniformity (ROU) of the transverse coils is expected to lie within an elliptical volume with major axes of *250* x *250* x *200 mm³*.

Conclusions: An optimized shielded biplanar coil has been designed for a low field MRI system which successfully suppresses the magnetic field outside the coil.



(a) (b)

Fig. 4: Field strength (*mT*) for (a) Gx coil measured along x-direction at *y=-10mm, z=0*. (b) Gy coil measured along y-direction at *x=0, z=90mm*.

References:

[1] Turner et.al. MRI.1993;11:903 [2] Yoda et al. J.Appl.Phys. 1990;67:4349

[3] Tomasi et.al. JMR 1999;140:325 [4] Lagarias et.al. J.Optimization 1998;9:112