Ultra-Efficient Shielded Dome Gradient Coils

M. Poole[1], and R. Bowtell[1]

[1]Sir Peter Mansfield Magnetic Resonance Centre, University of Nottingham, Nottingham, Nottinghamshire, United Kingdom

Introduction

Insert gradient coils built specifically for head imaging offer access to large field gradients which are useful for many applications, including diffusion-weighted imaging and high-resolution anatomical imaging. It has recently been shown that dome-shaped head gradient coils offer particularly large gains in coil performance because all their windings lie close to the imaging region [1]. Previously described dome-shaped coils were wound on a coil surface formed from a hemisphere [1] or hemisphere with a short cylindrical extension [2], and designed using a conventional target-field approach. Here, we show that further significant improvements in coil performance can be achieved by using a dome surface with an elliptical, inner cross section (Fig.1). A boundary element method after Pissanetzky [3] and Lemdiasov and Ludwig [4] was used to design shielded gradient coils with high efficiency and low inductance on this surface of low symmetry. A prototype coil has been constructed and tested at 3T in field mapping experiments

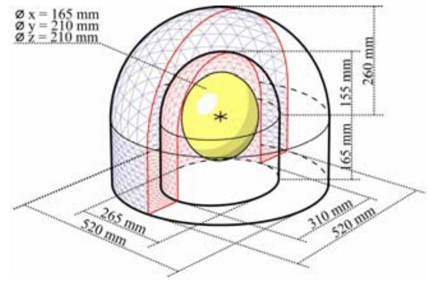

Figure 1. Geometry of the inner (red triangles) and outer (blue triangles) current carrying surfaces. The ROU is also shown (yellow ovoid).

Methods

The boundary element method [3,4] allows gradient coils to be designed with arbitrary geometry. It works by discretising the surface current density into a weighted set of divergence-free basis-functions. The inductance, resistance, and torque of the coil, as well as the magnetic flux density at any point can be derived in terms of these basis functions, allowing a functional reflecting the coil characteristics to be optimised. The inner surface of the dome coils (shown in red in Fig.1), is formed from a 310 mm diameter hemisphere with a 165 mm cylindrical extension by squashing the x-dimension to yield an elliptical cross section with a ratio of major/minor axis lengths of 1.17. This geometry was chosen to provide a 5 cm gap between the subject's head and inner coil surface, so as to accommodate a dome RF coil. The outer surface (blue) was formed from a 520 mm diameter hemisphere and 165 mm cylindrical extension. Coil windings on the inner surface were allowed to spread onto the flat annulus at the inferior end of the coil. A surface mesh of 4224 triangular elements was created in 3D Studio MAX® (Autodesk® Inc., San Rafael, CA, USA) and imported into Matlab® (Mathworks® Inc., Natick, CA, USA) for calculation of the optimal shielded coil designs that generate a field with 5% deviation from linearity over the region of uniformity (Fig. 1). A 3D-contouring algorithm was written to generate the wire-paths of the gradient coils. A Biot-Savart calculation was performed on these wire-paths to obtain the magnetic flux density distributions, and FastHenry© [5] was used to calculate the coil inductances and resistances. The torque per unit field strength experienced by the coils when carrying unit current was also calculated. A half-scale prototype *x*-gradient coil was wound on a rapid-prototyped former using 0.9mm diameter copper wire and tested via field mapping using a double, gradient echo sequence on a Philips Achieva 3T scanner.

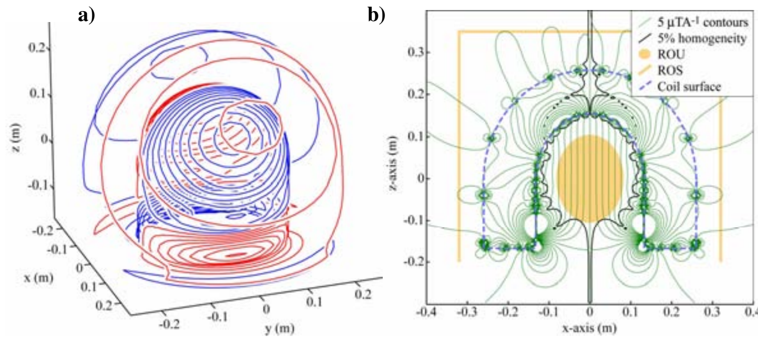

*Figure 2. **a)** wire paths of the x-gradient coil (red wires indicate reversed current flow), and **b)** the associated field plot.*

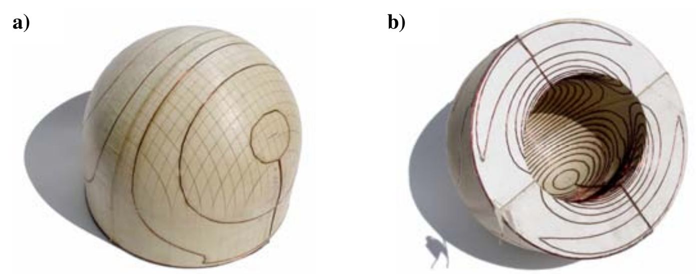

***Figure 3.** Prototype x-gradient coil showing **a)** the outer and **b)** the inner surfaces.*

Results

The wire-paths of the x-gradient coil are shown in Fig. 2 a), while Fig. 2 b) shows a contour plot of the field generated per unit current by this coil. The latter shows that a field deviating from linearity by less than 5 % is generated throughout the ROU and that the field outside the outer coil surface is small. Table 1 describes some calculated properties of the three different gradient coils, indicating that all have high efficiency and low inductance. The minimum wire spacing in all three coils was 3 mm or greater and the torque values were all less than 5.2 $mNm^{-1}A^{-1}T^{-1}$. The prototype coil (Fig. 3) was measured to have L=34μH and R=0.52mΩ in good agreement with the values of L=33μH and R=0.42mΩ calculated using FastHenry. Figure 4 shows that measured and target field maps from the prototype coil are in excellent agreement. The efficiency measured from the field maps (Fig. 4) was 1.36 $mTm^{-1}A^{-1}$ which is equivalent to a value of 0.34 $mTm^{-1}A^{-1}$ for a full size coil.

Gradient	Efficiency, η ($mTm^{-1}A^{-1}$)	Inductance, L (μH)	Resistance, R (mΩ)	Figure of Merit, η^2/L ($T^2m^{-2}A^{-2}H^{-1}$)
x	0.33	63	76	1.70×10^{-3}
y	0.29	56	74	1.47×10^{-3}
z	0.46	112	72	1.86×10^{-3}

***Table 1**. Properties of the gradient coils. Inductance, Resistance and FOM were calculated using FastHenry© [5].*

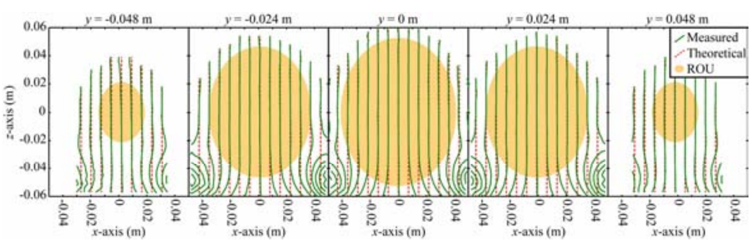

***Figure 4.** Contour maps of the magnetic field generated by the prototype x-gradient coil in different y-planes. The measured (solid green) and fitted target (red) 10μTA^{-1} contour lines are shown, and the central ovals indicate the region of uniformity (ROU).*

Discussion and Conclusion

A torque-balanced and highly efficient, shielded, dome gradient coil set of low inductance has been designed using a boundary element method [3,4]. The figures of merit (η^2/L) of these coils are 3.9, 2.8, and 4.2 times greater than previously described *x*, *y* and *z* asymmetric cylindrical gradient coils [6], but the dome coils have much larger ROU's and are also shielded. The coils described here also have better performance than recently described unshielded, hemispherical dome coils [1,2]. The *x*-gradient coil is capable of generating 100mTm^{-1} with a current of 306A and a stored energy of just 15J. A half-scale prototype has been built and tested so as to validate the theoretical results.

References

[1] D. Green, J. Leggett, and R. Bowtell, *Magn Reson. Med.* **54**, 656-668 (2005) **[2]** J. Leggett, D. Green, and R. Bowtell, *Proc. ISMRM* **14**, 779 (2006) **[3]** S. Pissanetzky, *Meas. Sci. Technol.* **3**, 667-673 (1992). **[4]** R. A. Lemdiasov and R. Ludwig, *Concepts Magn. Reson. B*. **26B**, 67-80 (2005). **[5]** M. Kamon, M. J. Tsuk and J. K. White, *IEEE Trans. On MTT*. **35**, 875-886 (1996). **[6]** D. C. Alsop and T. J. Connick, *Magn. Reson. Med.* **35**, 875-886 (1996)