3D-Gradient Coil Structures for Breast imaging using Fuzzy Membership Functions. Part II

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Introduction: Quadrupolar gradient coils (QGC) have been extensively used in several MRI and MRS applications [1]. This coil is classically designed using a cos(2θ) current distribution on a cylinder oriented perpendicular to the B_0 field and is characterized by its high efficiency and the capability to produce large regions of uniformity. An ideal current distribution of cos(2θ) around a infinitely long cylindrical generates a perfect gradient uniformity in all points inside. However, for a practical case when the length of the cylinder is comparable with its diameter, the size of the region of uniform gradient is considerably reduced. In applications such as the imaging of breast lesions, where a uniform gradient close to the coil entrance is required, the return current paths also play an essential role to in field generation [1]. The cylindrical geometry together with relatively few degrees of freedom of the standard QGC constrains the design possibilities. In this work, we propose a simple and flexible methodology that allows new QGC solutions which are beyond conventional configurations. The merit of this approach is that it considers the current return paths as being an active and useful part of the gradient coil. Some examples obtained by applying the new methodology are compared with classical QGC to demonstrate the versatility of the proposed approach.

Method: The method is based on the well-known real space discrete coil technique where the conductor pattern is expressed with function of parametric curves whose free parameters are perturbed by an optimization algorithm that minimizes a cost function defined for a variety of design goals. The objective function includes a coil current pattern generator (CPG) and a flexible 3D coil structure generator (CSG) that use fuzzy memberships functions (MFs) as parametric curves to reshape to 3D the coil current pattern [2,3]. The MFs are characterized by generating highly flexible curve shapes that depend on few parameters; which is essential for gradient coil design in discrete real space [3]. The CPG uses an open concept where the coil pattern can be generated by any kinds of existing gradient coil designing methods (e.g., target field and its variations or stream functions [3], etc.). In this work the CPG is based on the cos(2θ) spatial distribution of straight wires. The θ-angular position of the wires is assumed as a degree of freedom.

Using MFs the 3D CSG reshapes the coil and then produces (x,y,z)-coordinates of the coil. Given the (x,y,z) coordinates generated in the CSG module, three components of **dl** (differential of conductor segment length) and **r'** (source position) are calculated, and then magnetic field, force, torque components are calculated, as well as the inductance and the required gradient field. The magnetic field is calculated using The Biot-Savart law. The 3D coil coordinates are obtained by the equations $x = \rho\sin(\theta)$ and $z = \rho\cos(\theta)$ which are modified with the sample shape and design goals [3]. In order

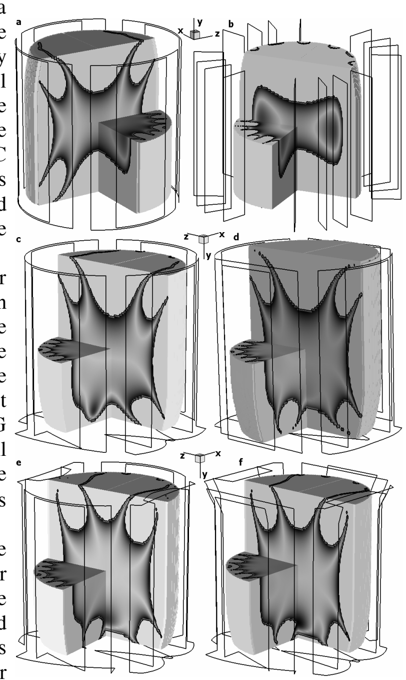
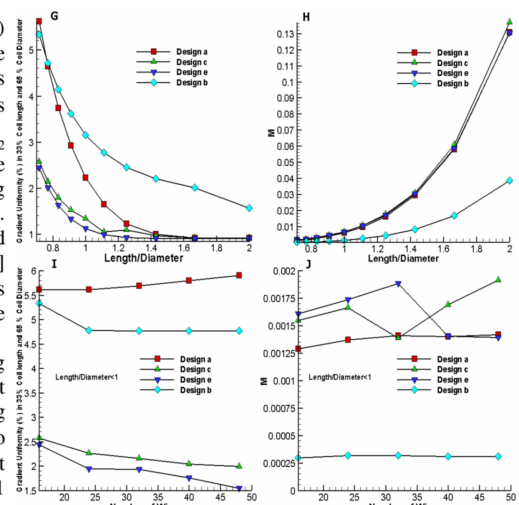


Fig.1. Different *Gz* QGCs designed with the present method. a, b: standard QGC. Ratio for the coil diameter between the primary and the secondary 1.4.The contour defines the 5% of gradient uniformity. G,H,I,J: Coil performance versus coil parameters.

to follow the breast anatomical shape the original coil's radius ρ is reduced in the range y=[L/2…L) using the expression $\rho_N = zmf(y, L/2, L\cdot a_{M+1})\cdot\rho$. *zmf* is a Z-shape curve MF [2]; *M* is the number of wire elements and a_{M+1} is a degree of freedom. The *x* and *z*-coordinates can be written as $x = \rho_N \sin(\theta - \alpha(y))$ and $z = \rho_N \cos(\theta - \alpha(y))$. With $\alpha(y) = smf(y, y_1\cdot a_{M+2}, y_2\cdot a_{M+3})$ the straight wires can be twisted in order to increase the space of possible solutions. *zmf* is a S-shape curve MF [2], y_1, y_2 defines the '*y*- range' domain where the coil is twisted, a_{M+2} and a_{M+3} are degrees of freedom. The wire poles can be in series connected using current arcs between adjacent quadrants, or extending horizontally the wires along the *z* direction [1] (this is easy to apply for one of the two QGC *Gx* or *Gz*). Another connection method is using a secondary coil parallel to the primary at a larger radius (shielded design). In this work the arc connection at *y~L* (coil's bottom) is modified through a bell-shape MF [2,3] in order to produce a useful gradient contribution to efficiency η and gradient uniformity in DSV. This transformation could be achieved at the coil entrance taking into account that enough room must be provided for patient comfort.



Results and Discussions: The design method was applied to design *Gx/Gz* unshielded QGC using different combinations of MFs in the 3D-CSG. Fig1. a and b, shows the standard QGC for breast imaging and there is no MFs in the CSG. Fig1. c and e illustrates two of the several coils designed using MFs in the CSG. In all these configurations the coil radius ρ was set to 11.7 cm and the axial length to 23.4 cm in order to assure a ratio of $L/(2\rho)$ =1. The DSV size and shape was defined in such a way that it follows the shape and sizes of the sample [1]; 16 wires were used to represent the discrete coil. Fig.1 demonstrates the large uniformity region produced by the new designs (c,d,e,f) and the current return paths impact over the uniformity along the *y* direction. The new design (c) and its variant (d) shows a superior extension of the (5%) uniformity along the *y* direction due to the reshaping of the current return paths at the coil bottom, however, the same effect is desired at the coil top. Fig 1d shows a design where the sample accessibility to the coil DSV has been scarified to extend the uniformity towards the coil top. In the design f, the coil entrance has been opened for a better access. The inductance (L) and η of the design showed in Fig.1 (f) is 5.7 μH and 0.2 mT/mA, respectively. The torque is 3×10^{-3} Nm/AT. The figure of merit ($M=\eta^2/L$) is 7.1×10^{-3} $T^2A^{-2}m^{-2}H^{-1}$; which is 1.2 and 7 times larger than those produced by designs a and b, respectively. The distance from the coil top to the 5% contours along the *y*-axis is 6.8 cm in the design f and 7.8 cm and 8 cm in the designs a and b, respectively. Fig.1 G shows that, for Length/Diameter (*L/D*) ratios smaller than one, the designs c and e have the best gradient uniformity for a fixed DSV size. In this study the ratio between the primary and secondary coil diameter was set to 2 (design b). Fig.1 G shows approximately the same coil performance as the designs a, c, e. Fig. I indicates the changes of the gradient uniformity for a fixed region size when the number of wires increases for *L/D*<1. In the standard design a, the gradient uniformity tends to decrease due to the influence of the arcs. In design b) the gradient uniformity keeps approximately a constant value for a certain number of wires. However, the two new designs can potentially increase the gradient uniformity with larger number of wires. For the condition *L/D*<1, the figure of merit tends to increases slightly and to keep a constant value. For designs c and e, however, there is not any clear dependency due the high flexibility of the MFs.

Conclusions: A flexible and simple methodology for the design of unconventional 3D-*Gz/Gx* gradient coils with superior performance and gradient uniformity over conventional quadrupolar structures has been presented. The appropriate 3D-distribution of the current return paths can considerably increase the coil performance and enlarge the gradient uniformity region to fit well with the requirements of breast imaging.

References.

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Acknowledgments

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