Minimum-Inductance MRI Gradient Coil Design with Arbitrarily-Selected Shape

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자기공명 영상촬영을 위한 임의로 선택된 모양의 최소인덕턴스경사자계코일의 설계

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ABSTRACT

This paper proposes a new inductance minimization scheme for a gradient system of arbitrarily selected shape. Although it is important to minimize the gradient coil inductance to reduce the current switching time, such minimization has been possible only for cylindrical or parallel biplanar coils. By using small current loops on arbitrarily selected surface as optimization elements, the inductance of the whole circuit can be minimized using the loop' s self- and mutual-inductances. Wire positions can be easily derived from the loop current distribution. Preliminary studies for the design of x-directional surface gradient coil show the utility of the proposed gradient coil design scheme.

INTRODUCTION

The conventional cylindrical gradient systems are useful for most MR imaging applications and their design methods have been developed very well [1-4]. However, for specialized imaging applications such as gradient-recalled-echo, echo-planar imaging, or localized high-resolution imaging, surface [5] or insertable coils [6] are preferred to generate stronger gradient field with faster switching time.

Although the minimum inductance design combined with target field approach provides good performance for cylindrical or biplanar insertable coils, the method can not be used for other shapes.

In this paper, we are proposing a new minimum-inductance gradient coil design scheme applicable to any selected shape. Semi-spherical or even helmet-shaped gradient coil can be designed by using the method. The proposed method uses small square-shaped current loops as optimization elements and the total inductance can be minimized by considering their individual and mutual inductance.

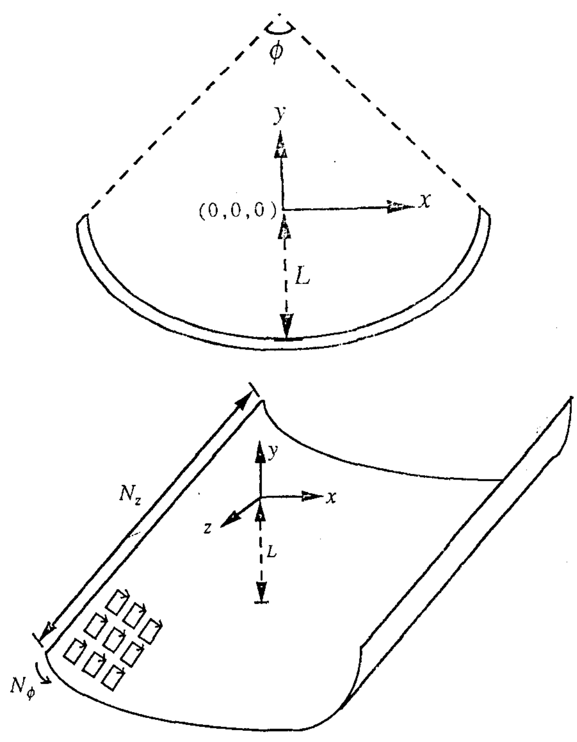

Fig. 1. Current loop elements on a semi-cylindrical surface. L is the distance between the imaging center and the coil surface.

METHOD

A semi-cylindrical-shape gradient coil shown in Fig.1 is designed as follows. If a total of $N = N_\varnothing \cdot N_z$ loop elements are assumed, then a current vector $\mathbf{i}$ is defined as $\mathbf{i} = [i_1\ i_2 \cdots i_N]^T$, where i_k is the current of the k-th loop. The total energy stored in the coil, can then be written as

$$e = \frac{1}{2}[\mathbf{i}^T\ \mathbf{W}\ \ \mathbf{i}],$$

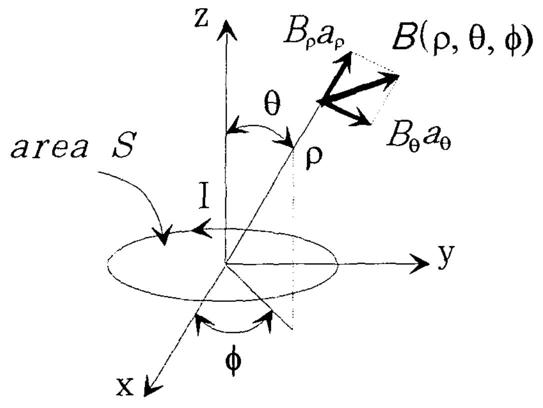

Fig. 2. A small loop current of an area S with current I producing the magnetic field **B** at (ρ,θ,ϕ).

where $\mathbf{W} = \begin{bmatrix} L_{11} & L_{12} & \cdots \\ L_{21} & L_{22} & \cdots \\ \vdots & \vdots & \end{bmatrix}$, L_{ii} is the self inductance of the i-th loop element, and L_{ij} is the mutual inductance between the i-th and j-th elements. Then the loop current distribution can be optimized by minimizing e while producing the field distribution we want to generate (optimization constraint).

The constraint is calculated as follows. Let us assume that the field intensity at (ρ,θ,ϕ) from a small loop current at the origin is to be calculated in the spherical coordinates as in Fig.2.

Here, assuming that $\rho^2 >> a^2$, the magnetic field $\boldsymbol{B} = B_\rho \boldsymbol{a}_\rho + B_\theta \boldsymbol{a}_\theta$ at (ρ,θ,ϕ) (see Fig. 2) is calculated as $B_\rho = \dfrac{2\mu_0 IS}{4\pi\rho^3}\cos\theta$ and $B_\theta = \dfrac{\mu_0 IS}{4\pi\rho^3}\sin\theta$. From this equation, we can find the z-directional field $B_z(x, y, z)$ from a loop current at any selected position. The $x^l y^m z^n$ gradient component at (0, 0, 0) can then be expressed as

$$G_{l,m,n} = \left[\frac{\partial^l}{\partial x^l} \frac{\partial^m}{\partial y^m} \frac{\partial^n}{\partial z^n} \left[B_z(x,y,z) \right] \right]_{x=y=z=0} .$$

The constraint for the optimization is formulated by specifying the fields from loop current elements and their derivatives in a matrix form as $\mathbf{G} \cdot \mathbf{i} = \mathbf{1}$, or

$$\begin{bmatrix} 1 \\ 0 \\ 0 \\ \vdots \\ \vdots \end{bmatrix} = \begin{bmatrix} \mathbf{G} \;\; (\; N_L \;\; by \;\; N \;) \\ \\ \\ \\ \end{bmatrix} \begin{bmatrix} i_1 \\ i_2 \\ i_3 \\ \vdots \\ \vdots \\ \vdots \\ i_N \end{bmatrix},$$

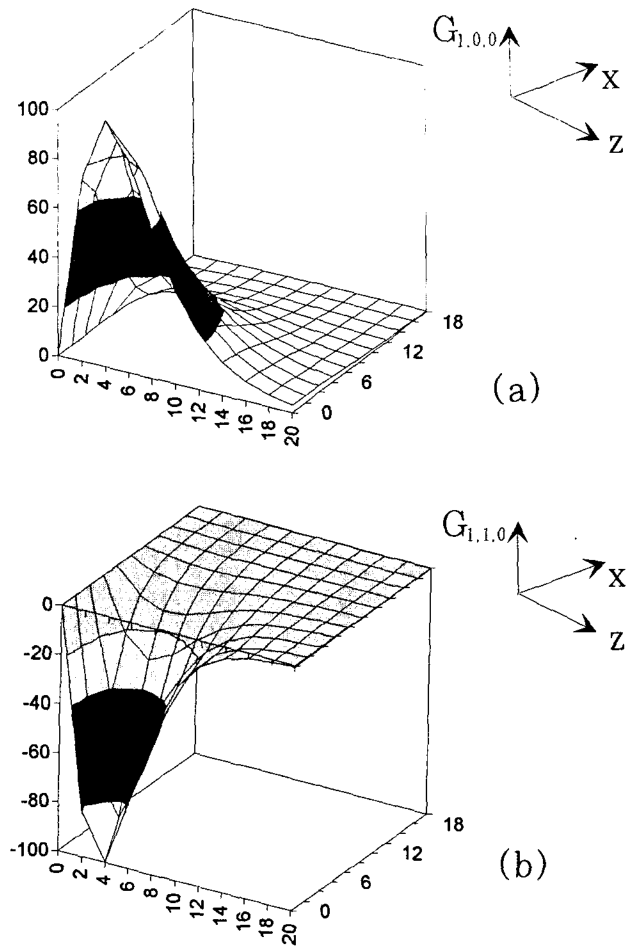

Fig. 3. $G_{1,0,0}$,(a), and $G_{1,1,0}$,(b), for $0 \le x, z \le 20cm$.

where **G** is the matrix form of $G_{l,m,n}$ for the corresponding current-loop elements, i_k and N_L is the number of gradient components we want control. The first row of **G** corresponds to the gradient component we want, such as (1,0,0) for x-gradient. The other rows correspond to the gradient components we want to remove, such as (1, 1, 0) or (3, 0, 0) component for xy or x^3 gradients, respectively.

Using the above constraint, the **i** with the minimum energy is calculated by using Lagrange multipliers as $\mathbf{i} = \mathbf{W}^{-1}\mathbf{G}^{T}\left[\mathbf{G}\mathbf{W}^{-1}\mathbf{G}^{T}\right]^{-1}\mathbf{1}$ and the wire distribution derived from **i** shows the minimum inductance design for a given gradient pattern.

RESULTS AND DISCUSSION

The above scheme has been used to design an x-directional SGC with loop-current elements distributed on a plane predefined as $-20 \le x \le 20cm$, and $-20 \le z \le 20cm$, and $y = -10cm$. The stored energy is minimized to produce a given x component while removing xy component (vertical change of x-gradient intensity).

Figures 3a and 3b shows $G_{1,0,0}$ and $G_{1,1,0}$ for $0 \leq x,\ z \leq 20cm$. Using the distribution, **G** can be generated. L_{ij} 's are generated by experimental measurements and simulation. An x- gradient coil has been designed and other-directional gradient coils are being designed. Preliminary results show good agreement to the theoretically predicted behavior. Eventually a surface RF coil will be inserted between the subject and the gradient coil set for NMR signal reception.

REFERENCES

1. F. Romeo, D.I. Hoult, "Magnetic field profiling: Analysis and correcting coil design." *Magn. Reson. Med.*, **1**, 44-65, 1984.
2. J.W. Carlson, "An optimized, highly homogeneous shielded gradient coil design." *Proc. SMRM VII*, 28,1988.
3. R. Turner, "Comparison of minimum inductance and minimum power gradient coil design strategies." *Proc. SMRM XI*, 4031, 1992.
4. P. Mansfield, B. Chapman, "Multishied active magnetic screenig of coil structure in NMR." *J. Magn. Reson.*, **72**, 211-233, 1987.
5. Z.H. Cho, J.H. Yi, "A novel type of surface gradient coil." *J. Magn. Reson.*, **94**, 471, 1991.
6. M.A. Martens, L.S. Petropoulos, R.W. Brown, *et al.*, "Insertable biplanar coils for magnetic resonance imaging." *Rev. Sci. Instrum*, **62-11**, 2639, 1991.