A Novel Approach in the Network Analysis of Eddy Current Induced by Planar z Gradient Coil

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Introduction

An exact characterization of eddy currents generated by pulsive gradient fields has always been a prime concern in the research field to design optimized gradient coils for different types of MRI systems. In the recent time a very efficient and less time consuming method named as Network method has been showing better promises in electromagnetic analysis [1]. But no work has yet been found in case of planar gradient coil of permanent magnet (PM) MRI systems using this method. In this study a novel approach is implemented in the network analysis of eddy currents transient response in the local shielding box of a compact PM-MRI system for planar z-gradient (Gz) coil by applying scalar potential form of Biot Savart's law. Determination of mutual inductance between gradient coil and any other conducting structure is a crucial issue in the electromagnetic analysis of MRI systems. Following the scalar potential form of Biot Savart's law, we calculated the gradient of the solid angle (Ω) subtended by each turn of the planar coil at any point in the Cu shielding box to find the total flux linkage per unit current between all the turns of z coil and the conductor in interest to calculate mutual inductances [2].

Materials and Methods

Simulation of eddy currents transient response was performed on the radio frequency (Rf) shielding box [3] with outer boundary consisting of Cu plates of thickness 0.5 mm and with dimensions 18 cm×18 cm×9 cm, which is implemented in a 0.3T PM-MRI system of bone-age assessment of children [4]. A 32-turn circular planar z gradient coil with 1 mm wire diameter, 10 cm coil gap (Figure 1(a)) and 15.7 cm maximum diameter coil has been implemented in the system as a Gz coil [3]. Following the network method, the shielding box is divided into contiguous slices with thickness 0.5 mm, which is thinner than the skin depth ($\delta = \sqrt{\frac{2}{\mu\omega\sigma}}$) of the corresponding signal frequency (ω). We have assumed a uniform current flow in each slice and resistance equal to its dc value, $\mathrm{R} = \frac{\rho l}{wt}$, where l is the length, w is the width, and t is the thickness of each slice. It is also assumed that no resistive coupling exists between the slices but they are inductively coupled. The Gz coil and the slices can be represented as a network of *R-L* circuits consisting of the mutual inductances between Gz coil and each slice, and also the self and mutual inductances of the slices itself. The network equation becomes, $\mathrm{M}_{ii} \times \frac{d\mathrm{I}}{dt} + \mathrm{R} \times \mathrm{I} = -\mathrm{M}_{si} \times \frac{di_0}{dt}$, where, I is the unknown eddy current matrix with dimension equal to the number of slices taken.

We have calculated flux density (B) in each slice due to the current flowing in each turn of Gz coil using magnetic scalar potential, $\mathrm{B} \propto \nabla\Omega$, by which total flux linking that particular slice is calculated and, mutual inductance matrix is, $\mathrm{M}_{si} = \frac{flux\ linking\ a\ slice\ due\ to\ current\ in\ Gz\ coil}{current\ in\ Gz\ coil}$, where, Ω is the solid angle subtended by each current loop at any slice, *s* denotes the coil as source, and *i* denotes the slice no. [2]. Solid angle calculation was performed in 3D space. The self and mutual inductances (inductance matrixM_{ii}) of the slices are calculated using the following formulas respectively, $self\ inductance = \frac{\mu_0 l}{2}\left[Log\frac{2l}{w} + 0.5\right]$ and, $mutual\ inductance = \frac{\mu_0 l}{2}\left[Log\left(\frac{l}{d} + \sqrt{1+\frac{l^2}{d^2}}\right) - \sqrt{1+\frac{l^2}{d^2}} + \frac{d}{l}\right]$, where d is the distance between any two slices, and i denotes the slice no. [5]. The gradient coil is driven by a 1 ms trapezoidal pulse (i_0) with 8.125 Ampere maximum current. The ramp-up and ramp-down times of the pulse are 100 μs each and flat-top is 800 μs.

Results and Discussion

Figure 1(b) shows eddy currents as a function of time for different slices, and Figure 1(c) shows secondary magnetic field generated by eddy currents at three different positions in the region of interest (ROI) along z axis. The secondary magnetic field opposes the original gradient field. Figure 1(d) shows the time dependence of the z gradient field 2 cm from the center of the z axis. The total calculation time was only approximately 2 minutes. The formulation of inductive coupling between the source and the conducting structures by applying the Biot Savart's law directly is very efficient process. Our analysis on planar gradient coil is also very unique and much less time consuming. This approach of calculating mutual inductance can be implemented for any shape of planner gradient coil to calculate inductances at any point in the conducting structures of the MRI systems, and hence it will ease the eddy current analysis to a greater level.

References

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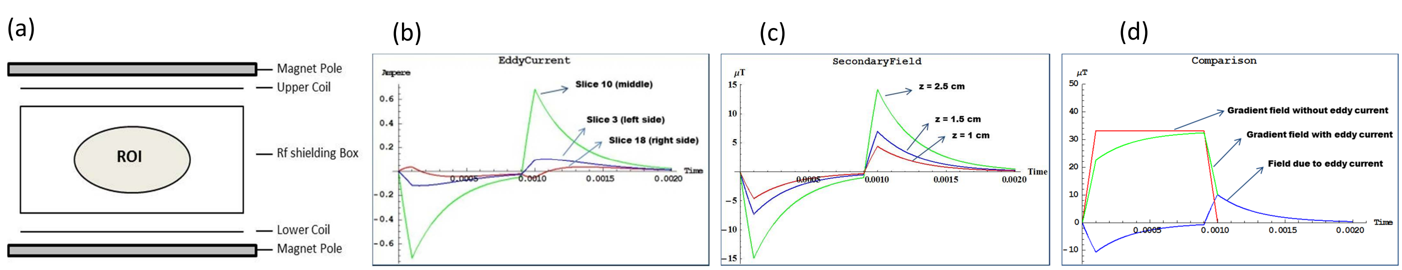

Figure 1 (a) Schematic representation of the PM-MRI system; (b) Transient response of eddy currents in the shielding box; (c) Magnetic field generated by eddy current at different positions along the z axis in the ROI; and (d) Time dependence of the gradient field.