A Split Gradient Coil for High Speed Imaging with Application to MRI-RT

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Introduction: It is desirable to have a high speed whole body gradient coil with a central gap between coils of 200mm for applications in MRI guided Radiation Therapy. Multileaved Collimators (MLCs) near the coil gap have conducting and possible eddy current surfaces that indicate avoidance of 3D coil design in this region. Our gapped coil design is different from the approach described in [1]. A similar design for a whole body gradient was analyzed in [2]. Radiation treatment monitoring requires continuous fast imaging for time intervals of 20min, so gapped gradient inefficiency and duty cycle requirements combine to increase cooling requirements.

Method: The geometry for a shielded whole body gradient with a central coil gap and with continuous inner former is shown in Figure 1. This geometry is different from that reported in [1] because there are no coil currents between the primary and shield at the gap. This reduces the eddy currents in the MLCs that can move during the imaging and cause asymmetry. Our design method allows current only on cylindrical surfaces to minimize eddy currents. The ID/OD of the gradient is 800mm/1040mm with an overall length of 2m. The gradient design includes minimization of the force/torque exerted by the magnet that the system is intended to be used with. The relatively thin and uniform inner former in the center helps to maintain alignment and mechanically react the forces and torques while providing uniform radiation attenuation. Figure 2 shows the current patterns of half of the gradient coil and Table 1 lists the electrical characteristics of the coil.

Results: One of the consequences of a coil gap is that the conductors of the X and Y fingerprint coils are compressed between the gap's edge and the eye of the coil. There are also associated sensitivity losses caused both by the gap and by not allowing coil sections on the gap surface. This means that for desired coil sensitivity the power density in this region is higher than with no gap coil designs. An anticipated gradient duty cycle of $G_{RMS}\approx 10$mT/m with $I_{RMS}\approx 370$A increases cooling requirements at the hottest spot of the gradient coil and enhances the need for direct cooling. In order to achieve these requirements with adequate cooling, directly cooled X, Y, and Z coils were used. This changes the heat removal problem from conduction to a problem of heat removal from directly cooled coils. The temperature increase in the conductor circuit can be limited to 32°C when cooling 15kW on any axis. By comparison the conventional X/Y sheet coil cooling method to the Z-gradient direct cooling water channels would result in maximum temperatures near 100°C. The Z coil is rather naturally gapped like a Maxwell pair coil and relatively more efficient than the X/Y coils. It is also relatively easy to directly cool.

High speed imaging applications benefit from good gradient slew rate. Because the conductor tubes have wall thickness ≤ 2mm we expect little skin effect increase in resistance as a function of frequency. A gradient slew rate of about 200mT/m/ms requires a current driver with high voltage capability. This high voltage requires good partial discharge immunity of the gradient coil. The onset of partial discharge of the coil described was above $V_{RMS} \geq 2$kV. This is achieved by design, construction methods, and selection of materials.

Conclusions: We have designed and built a gapped whole body gradient coil for applications in MRI guided Radiation Therapy. High performance requirements of the conventional two surface split coil gradient design is best realized with direct cooling on all three axes. Also because MRI system will be used for continuous motion monitoring using high speed imaging a 600A-2kV gradient driver is used. This adds the construction requirement of good partial discharge immunity. Eddy current asymmetry (due to the presence of the MLCs) was modeled using the Boundary Element software package "Faraday". The simulations show that the eddy current asymmetry over 50cm DSV is about 0.5% with the design described (Figure 3) as compared to over 1.5% reported in [2].

References: [1] M. Poole et al, "Split Gradient Coil for Simultaneous PET-MRI", MRM, Vol. 62, #5, p. 1106-1111, 2009
[2] Sh. Shvartsman et al, "Gradient Coil Induced Eddy Current Computation Using the Boundary Elements Method", ISMRM 2008

Figure 1. Whole Body Gradient Coil Geometry

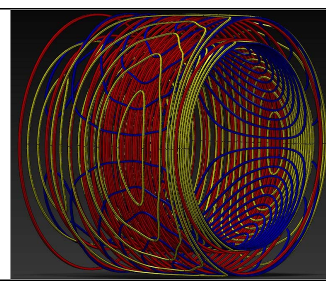

Figure 2. SolidWorks Picture of Half Gradient Coil

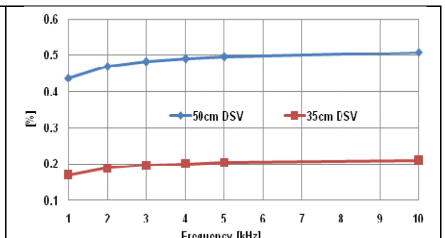

Figure 3. Eddy Current Asymmetry as a Function of Frequency

Table 1. Coil Characteristics

	Inductance [mH]	Resistance [mΩ]	Sensitivity [μT/m/A]
X gradient	0.270	0.11	31.35
Y-gradient	0.270	0.11	30.25
Z gradient	0.380	0.10	44.90