

Design of High Performance Gradient Coil for 3T Head Specialty Scanner

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Introduction: As compared to whole-body gradient coils, a head gradient coil can have superior neuroimaging capabilities due to increased gradient amplitude and switching speeds [1-2]. For small field-of-view (FOV) gradient coils, the peripheral nerve stimulation (PNS) threshold is also known to be higher. A specialty MRI scanner for head imaging can deliver the performance of a head gradient with optimal design integration of the resonator module and magnet and lower total cost of ownership than a gradient insert added to a whole-body scanner. However, this type of development requires a holistic system design approach and more careful consideration of anthropometric data while meeting stringent requirements on gradient linearity, uniformity, eddy-currents, leakage fields and ultimately imaging performance. We report here on the design of a head gradient prototype to be integrated in a specialty 3T head only magnet for high performance imaging. It delivers a force and torque balanced shielded design for asymmetric X and Y axes and symmetric Z axis with an inner diameter of 42 cm. This coil will serve as a benchmark for model validation and assessment of PNS, higher order eddy current effects, acoustics, dielectric and thermal performance.

Methods: An object-oriented design code based on the stream function method [3] was used to target the specifications of a shielded head gradient coil (Fig. 1a). In addition to targeting gradient strength, linearity and uniformity, the design code can constrain individual harmonic orders of the gradient field and eddy current field (generated at magnet's radiation shield radius, Fig. 1b) and limit the maximum allowable leakage field, net force and torque and minimum turn spacing. The dimensioning study evaluated the impact of the distance between the center of the field of view and the point of contact between the X , Y , or Z coil with the shoulders of the patient, defined as z_{patient} (Fig. 1c). The upper limit used for optimization of transverse axes z_{patient} was 16 cm, which corresponds to the 8th percentile of the shoulder to tragon distance (mean \pm stdev = 18.7cm \pm 1.9cm) in an anthropometric study of a male population [4]. The selected design configuration was evaluated through harmonic analysis of the nonlinear gradient field and simulated brain images based on static gradient field maps with and without gradient warping correction.

Results and Discussion: The design proposed here delivers peak specifications of 88mT/m and 630T/m/s and a linearity of 16.9% over a 26 cm diameter spherical volume (DSV) based on a commercial gradient driver (GE, MR750). An asymmetric configuration was selected for X and Y based on anthropometric data to provide a large inner diameter of 42 cm and a stepped configuration on the shoulder to maximize the insertion length. An optimum design point was found for the transverse axes z_{patient} within the 16 cm optimization range. This allowed the solutions obtained to meet stringent performance targets, while keeping patient access a foremost priority. Having the ability to image down to the C2-C3 junction ensures that the base of the brain can be imaged in a distortion-free field. Our design optimization algorithm resulted in a z_{patient} for the Z axis exceeding the 16 cm requirement for patient access. However, this did not adversely affect patient access due to the implementation of a stepped design (Fig. 1c). This allows access for 95% of the population based on anthropometric data from a male and female combined population [5]. Figure 2 shows that the results from gradient warping with and without the even order harmonics were quite similar, which reflects the design constraints imposed to suppress the even order harmonics (Fig.3). Residual misalignments were observed in both corrected images of Fig.2 due to effects from the higher-order ($>5^{\text{th}}$) harmonics.

Conclusion: We report here on the progress towards an actively shielded head gradient coil designed for improved patient access and optimal performance. This design matches the gradient driver cost functions and a dedicated head-only magnet design. The algorithms developed to design this gradient allowed design trade-offs that generated an optimal gradient coil dedicated for imaging the brain. Modeling the coil based on the design parameters yielded simulated data confirming distortion-free imaging of the whole brain (achieving the desired gradient linearity over the targeted FOV). Experimental verification of the theoretical performance data is currently underway. Initial experiments will be performed with a whole

body MR scanner at 3T to determine and validate the model predictions for image quality, eddy current effects, thermal performance, acoustics and PNS.

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References: [1]Turner, MRM, 11:903-920 (1993), [2]Chronik et al., MRM 44:955-963 (2000), [3] Schenck et al. US Patent 4,646,024 (1987), [4] Churchill et al., technical report, MRL-TR-77-2 (1978), [5] Gordon et al., technical Report NATICK/TR-89/044 (1988)

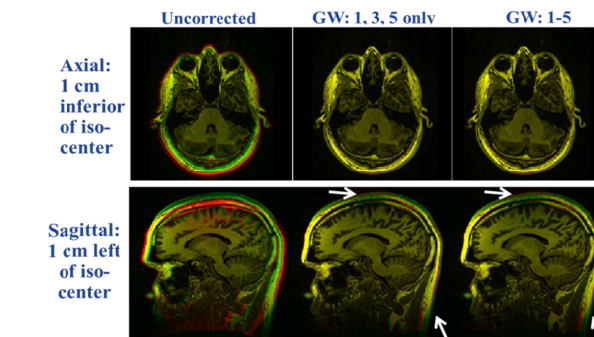


Figure 2: Simulated brain images. The red channel images in the left, middle, right columns show uncorrected (static field-simulated), odd-order gradient distortion (gradwarp, GW) corrected, and both odd- and even-order GW-corrected images, respectively. The green channel in each image shows 3D-GW corrected source image acquired at 3T with a whole-body MRI system (GE, MR750). Residual misalignments are seen in the extreme locations in the superior-inferior directions.

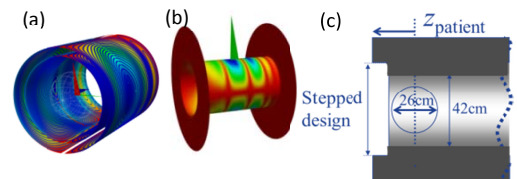


Figure 1: a) Asymmetric transverse gradient pattern. b) Eddy current prediction from optimization design tool. c) Layout of step design in asymmetric gradient.

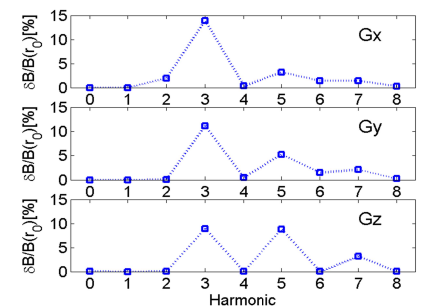


Figure 3: Harmonic analysis of the head gradient field map. The plots show the maximum field strengths within the 26 cm DSV contributed by the harmonics of order $n \leq 8$, expressed as the percentage of the applied gradient field at $r_0 = 13$ cm. The $n = 1$ harmonic is zero by definition.