Gradient nonlinearity effects of diffusion weighted imaging in a dedicated head-only MRI system

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Target Audience: Physicists and radiologists interested in quantitative diffusion neuroimaging, especially in stroke and oncology.

Purpose: A dedicated, head-only MRI system [1-2] can potentially reduce the cost of neuro-MRI, require a smaller installation footprint, and produce higher gradient performance. The improved gradient amplitude and slew rate are particularly beneficial in diffusion-weighted/tensor-imaging (DWI/DTI), which has seen increased utility in applications such as quantitative stroke and oncology assessments. For a fixed, head-sized field of view (FOV), a head-only specialty gradient coil tends to have a greater gradient field nonlinearity (GN) compared to a conventional whole-body gradient coil optimized for a body-sized FOV. This nonlinearity results in a larger spatial variation of the diffusion-encoding b-value [3-4]. The clinical impact of this increased b-value variation on quantitative neuro-DWI, and the validity of GN correction (GNC) [5] have yet to be evaluated.

Methods: The gradient field maps of a head-only gradient system [2] were used to calculate spatially-varying b-value maps. The field maps were fitted with 5th-order spherical harmonic coefficients [6] for retrospective GNC [5]. DTI brain acquisitions were made on a whole-body 3T system (MR750, GE Healthcare, Waukesha WI USA), and its uncorrected apparent diffusion coefficient (ADC) and fractional anisotropy (FA) results were compared against that from GNC. The GN-corrected images were used to simulate GN-effects from the head-only system, after which a 5th-order GNC was applied. While the isocenter position in the whole-body images did not necessarily coincide with the central position of the imaged volume, it was assumed that these positions were coincident in the head-only simulation. Spatial gradient-warping [6] effects are independent to effects due to GN in diffusion, and were not simulated in this work.

To account for diffusion anisotropy, it was necessary to use DTI. A b-value of 2000 sec/mm^2 with 20 directions were chosen, along with an axial FOV = 24 cm, 128×128, 42×4 mm slices, TR/TE = 6000/100 msec. To characterize GN-effects independently of the chosen b-value, %ADC differences and FA differences from the simulation were compared.

Results: Fig. 1 shows patterns of increased ADC with increased in-plane distance from the isocenter, and reduced ADC with increased axial (S/I) distance from the isocenter. Using ADC difference < ±5% as an arbitrary limit, the closest axial plane with a significant (P = 0.01) ADC difference in the whole-body images was 58 mm from isocenter; for the head-only situation, every image plane produced a significantly different ADC. The differences in FA also increased with respect to isocenter. Similar inferences could be made using an FA difference < ±0.025 as an arbitrary limit. However, the distribution of FA-difference values tended to be centered on zero, and regions of FA under- or over-estimation tended to be clustered. Applying 5th-order GNC restores accuracy in both ADC and FA.

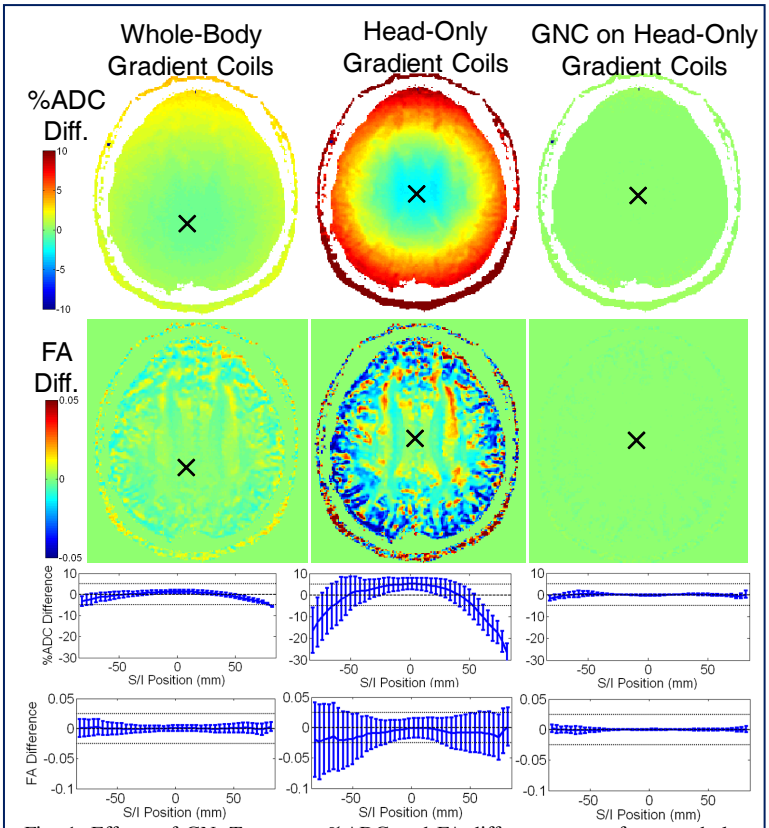

Fig. 1. Effects of GN. Top rows: %ADC and FA difference maps from a whole-body, head-only, and GN-corrected head-only axial diffusion images 30 mm superior to isocenter. The x marks the position of the in-plane isocenter. Bottom rows: corresponding mean and standard error plots of %ADC and FA differences averaged in-plane, as a function of the S/I position. Dashed lines indicating zero, ±5% ADC difference and ±0.025 FA difference are shown.

Discussion and Conclusion: Diffusion imaging using a dedicated, head-only gradient coil without nonlinearity correction could produce significant, spatially-varying b-values that result in inaccurate trace diffusion images, ADC and FA. This could result in erroneous diagnosis and quantitation in several clinical applications of diffusion MRI, such as stroke and oncology. However, these errors are correctable retrospectively with GNC. The high gradient amplitude available with a head-only gradient could also result in increased ADC bias due to the concomitant field effects associated with a dual-spin-echo preparation [7-8], and these effects are generally increased in asymmetric, transverse coil designs [7]. Concomitant field effects were not studied here, but could be mitigated with either prospective modifications to the pulse sequence [8] or a standard, single-spin-echo preparation.

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