Quantifying changes in geometric distortion across a gradient coil replacement

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Introduction: Gradient coils may be replaced on MRI systems as part of hardware upgrades or as a result of hardware failure. These events could introduce a step change in longitudinal quantitative neuro-imaging studies. Gradient winding errors cause even nominally identical gradient coils to differ in their field. Additionally the precise alignment of the gradient coil in the magnet and the active shielding coil can alter the gradient profile or eddy current effects, and hence geometric distortion. These considerations are increasingly important for short-bore and head-only gradient coils where the linearity is reduced. It is currently possible to detect pathological volumetric changes as small as 0.25% [1] with MRI and for such applications, maintaining consistent volumetric measurement across a gradient coil replacement is of crucial importance. To quantify the effect of a gradient coil replacement, geometric distortion was measured using a detailed 3D phantom before and after the event. The resulting distortion maps were compared, identifying regions where the change is negligible, and the potential impact on volumetric measures.

Method: The gradient coils were replaced on a GE Signa 1.5T system after spike artifacts appeared in DTI sequences. The original 10-year old coils were replaced by a new set manufactured to the same design. A 3D phantom with high spatial definition was imaged 4 times on separate occasions before and after the replacement. The MR images (FSPGR; FOV=24×24×25.2cm; Matrix=256×256×252; bandwidth=968Hz/pixel; no distortion correction) were non-linearly registered to a CT reference image (assumed artifact free). A set of points around the magnetic field isocenter where the geometric distortion is low were used to estimate the global rigid-body transformation mapping the MR image to the reference image. The rigid-body component was removed from the non-linear transformation leaving a vector map of geometric distortion [2]. The distortion maps from the pre-replacement time points were averaged ($\underline{\mathbf{V}}_{pre,mean}$) and subtracted from the post-replacement average ($\underline{\mathbf{V}}_{post,mean}$), giving a vector field representing change in geometric distortion ($\underline{\mathbf{V}}_{diff}$). An earlier reproducibility study estimated the precision of the method to be 0.3mm over 95% of the phantom volume. Based on this, a lower threshold of 0.3mm was applied to the magnitude of the vector field ($|\underline{\mathbf{V}}_{diff}|$) to highlight areas of change. The standard error of the difference (SE($\underline{\mathbf{V}}_{diff}$)) was calculated using the expression $SE_{diff}(p) = \sqrt{\{\sigma_{pre}(p)^2 + \sigma_{post}(p)^2\}/n}$ where n is the group size (4), and $\sigma_{pre}(\boldsymbol{p})$ and $\sigma_{post}(\boldsymbol{p})$ represent the standard deviation at each point ($\boldsymbol{p}$) for the pre- and post- groups respectively. The standard error of the magnitude of the difference (SE($|\underline{\mathbf{V}}_{diff}|$)) is the quadrature sum of the standard error for the x, y and z directions. The 95% confidence interval (2×SE($|\underline{\mathbf{V}}_{diff}|$)) was calculated to highlight regions where the difference measurement is insensitive to small changes, also with a lower threshold of 0.3mm applied.

A T_1 weighted IR-FSPGR image was acquired with the same scan parameters and without geometric distortion correction on a healthy volunteer (male, age 29). This image was segmented into grey matter (GM), white matter (WM) and CSF in SPM5. To investigate whether the difference between pre- and post-replacement geometric distortion alters the measured brain volumes, these segments were warped into un-distorted space using each of the geometric distortion maps calculated from the phantom data. The volumes of the un-distorted GM, WM and CSF segments were then estimated. The pre- and post-replacement volume measurements were compared with a two-sample T-test.

Results: Figure 1 shows that a typical brain volume lies mostly in an area with less than 0.3mm change in geometric distortion, however there is some overlap (Figure 2a,b). The geometric distortion change exceeds 0.3mm mainly at the superior and inferior extents of the field of view. Regions where the 95% confidence interval of the measurement is larger than 0.3mm have little overlap with the brain volume (Figure 1,2).
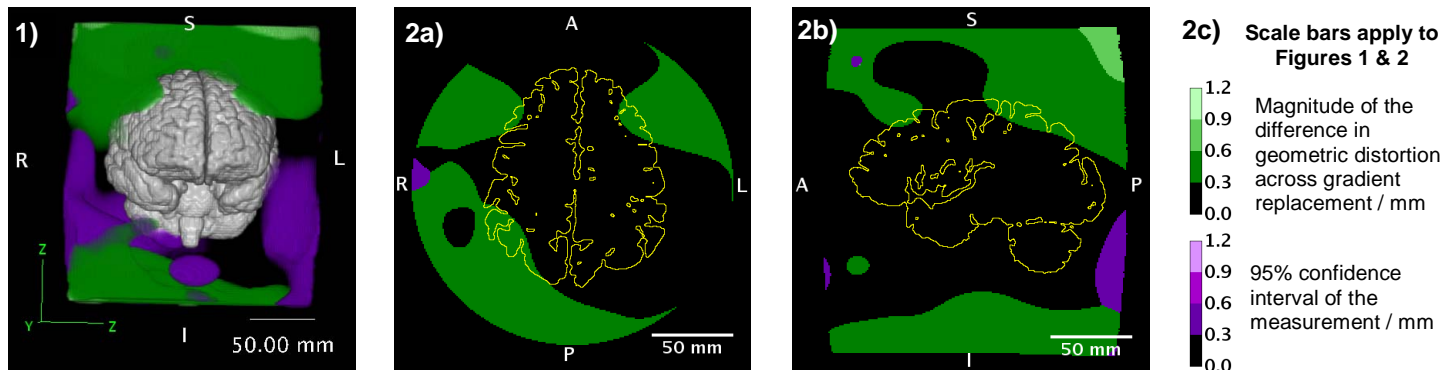

Figure 1. 3D rendering showing areas of change in geometric distortion (green) and areas of insensitivity in the measurement (purple), in relation to a typical brain volume.

Figure 2. Axial (a) and sagittal (b) views showing areas where the absolute change in geometric distortion is 0.3mm or greater (green) and overlaps a typical brain volume (outlined in yellow). Areas where the 95% confidence interval is 0.3mm or greater are shown in purple. c) Legend to the colour scale used in Figures 1 and 2.

When the pre-replacement maps were used for geometric distortion correction, the brain segment volumes differed by 0.03% (WM), 0.13% (GM), 0.36% (CSF) and 0.09% (total volume) compared to using the post-replacement geometric distortion maps. This suggests any volume change introduced by the gradient replacement is negligible for these measures where the coefficient of variance is 1.2–0.5% [3]. The volume change of the brain segments did not reach statistical significance at a p-value threshold of 0.05.

Conclusions: Change in geometric distortion across a gradient coil upgrade can be estimated with a 3D phantom and non-linear registration. The change was less than 0.3mm over 98.5% of a typical brain imaging volume and the impact on SPM-based measurement of grey- and white- matter segment volumes was negligible. Further work is required to confirm this result using the Jacobian of the vector field ($\underline{\mathbf{V}}_{diff}$) to investigate localised volume change and also to assess the impact on more sensitive volumetric techniques such as BBSI [1].

References: [1] Fox, N. C. et al., (1997), *J Magn Reson Imaging* **7**(6), 1069–1075. [2] Jackson, J. S. et al., (2008), *in* 'Proc. 14th Annual Meeting, British Chapter ISMRM, Newcastle, UK', pp. 23. [3] Chard, D. T. et al., (2002), *J Magn Reson Imaging* **15**(3), 259-267.