A generic method of quantifying geometric distortion using non-linear registration and a 3D phantom

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Introduction: Geometric distortion due to gradient non-uniformity needs to be accounted for in volumetric or interventional MRI. 3D correction of gradient distortion is possible with knowledge of the gradient coil spherical harmonics or with phantom measurements [1]. Both methods have advantages, however only phantom measurements can assess the accuracy of geometric distortion corrections, or monitor changes over time. Existing phantom techniques employ phantom-specific image-processing to measure distortion. Here we demonstrate a generic method using off-the-shelf non-linear registration tools [2], allowing geometric distortion measurement with any sufficiently detailed 3D phantom.

An MR image is first non-linearly registered to a reference image (eg. a CT scan, considered artefact free). The rigid-body transform that maps the MR image to the reference image is estimated from points around the magnet isocenter, where there should be little or no distortion. This also makes the method insensitive to the precise location of the phantom. Finally, a distortion map in MRI space is calculated using this first-step rigid-body transform and the non-linear registration. This geometric distortion measurement is easily extended to geometric distortion correction.

Methods: The phantom contained a set of 18 plates each with an array of circular holes [3] (Figure 1), however the method is not specific to this design. MP-RAGE images with and without gradient non-uniformity correction were acquired on a Siemens 1.5T Espree short-bore interventional scanner (FOV=24×24×25.6cm; matrix=256×256×256). These were non-linearly registered to a reference CT image using vtkCISG software [2] with optimised parameters, giving a non-linear transformation (**N**:mri→ref). It is also necessary to know the rigid-body transformation (**R**:phantom→ref). Unless a stereotactic system is available to determine this accurately, **R** is an unknown, but can be estimated simply by rigid registration ($\mathbf{R}_{reg}$). To avoid potential registration error due to the distorted MR image, **R** was also estimated based on 15 points (the isocentre and 14 points on a 5cm diameter sphere around the isocenter). Assuming the isocenter region is free from geometric distortion, the mapping of these points from the MR image to the reference can be represented by a rigid-body transformation. The mapping of these points is known from **N** and the transformation ($\mathbf{R}_{est}$) was estimated using Levenberg-Marquardt least-squares minimization. The geometric distortion in the MR image (**M**:mri→phantom) was calculated using both estimates of **R** ($\mathbf{R}_{reg}$ and $\mathbf{R}_{est}$) for each point (**p**), using $\mathbf{M}(\mathbf{p}) = \mathbf{R}^{-1}(\mathbf{N}(\mathbf{p})) - \mathbf{p}$.

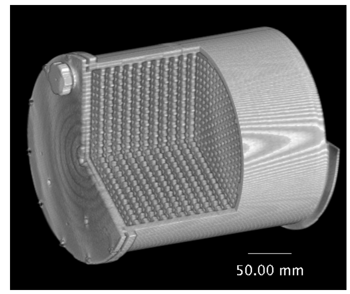
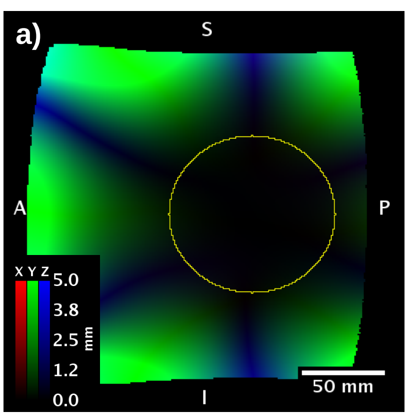


Figure 1. Phantom internal structure (rendered from CT).

To evaluate the reproducibility of the technique, 8 FSPGR images were acquired in a single session with repositioning on a 1.5T GE Signa system. An alignment plate was attached for reproducible positioning in the birdcage RF coil. The manufacturer's gradient warp correction was not applied.

Results: Without gradient non-uniformity correction, geometric distortion is significant outside the isocenter region on the 1.5T Espree (Figure 2a). Geometric accuracy is much improved with the manufacturer's correction applied (Figure 3), however the error still reaches 2mm at 9cm from the isocenter along the Z (S/I) axis. This highlights the importance of imaging at isocenter, especially with short-bore systems. The apparent geometric error over the isocenter region is reduced when **R** is estimated with the isocenter correction (Figure 2b, left), rather than from rigid-body registration (right). The isocenter correction therefore gives a qualitatively improved result reflecting our assumption of minimal geometric error at the isocenter.



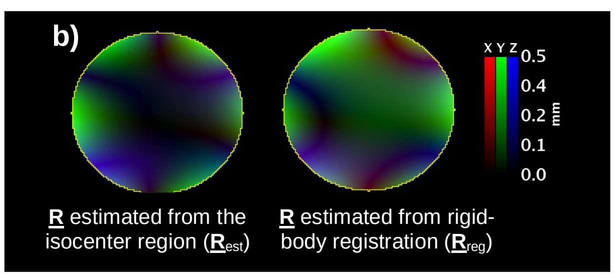


Figure 2. a) Geometric distortion without gradient non-uniformity correction. Red, green and blue indicate X, Y and Z displacement. b) A 10cm diameter region (see Figure 2a) represented with an expanded scale with **R** estimated from the isocenter region (left) and from rigid-body registration (right).

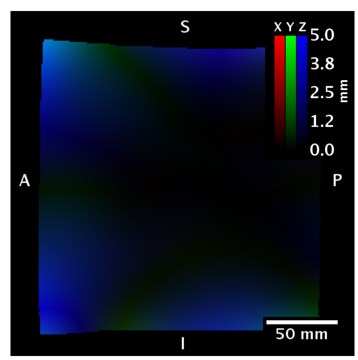


Figure 3. Geometric distortion after gradient non-uniformity correction.

In the reproducibility study, the mean voxel-wise standard error of the magnitude of the geometric distortion, measured across 8 scans with repositioning, was 0.08mm over a 6.2 litre volume. The 95th percentile and maximum were 0.13mm and 0.25mm respectively. The rigid-body transforms estimated by the isocenter correction indicate that phantom repositioning accuracy was worst for translation along the Z axis (std. dev. 0.4mm) and rotation about the Z axis (std. dev. 0.4°). This reflects the constraint imposed on the phantom by the RF coil in the X and Y directions. Further optimisation of the parameters may still improve the reproducibility of the technique.

Conclusions:
1. It is important to scan at the isocenter even when a fully 3D gradient non-uniformity correction is available
2. MRI geometric distortion can be quantified with a detailed 3D phantom and non-linear registration, to a precision of 0.3mm (95% of volume)
3. Estimating the global rigid-body transformation from the distortion-free isocenter region gives an improved result over rigid-body registration

References: [1] A. Janke et al., (2004), 'Use of spherical harmonic deconvolution methods to compensate for nonlinear gradient effects on MRI images', Magn Reson Med 52(1), 115-122. [2] T. Hartkens et al., (2002) 'VTK CISG Registration Toolkit: An open source software package for affine and non-rigid registration of single- and multimodal 3D images.', http://www.bvm-workshop.org/BVM2002, Leipzig, Springer-Verlag, March 2002. [3] J. Goffin, (2005), '3D geometrical phantom for quality assurance and geometrical correction in MRI of the Brain', Master's thesis, University College London.

The authors acknowledge the Wellcome Trust and the Multiple Sclerosis Society of Great Britain and Northern Ireland for support.