Is your system calibrated? MRI gradient system calibration for pre-clinical imaging

James Martin O'Callaghan[1], Jack Wells[1], Simon Richardson[1], Holly Holmes[1], Yichao Yu[1], Bernard Siow[1,2], and Mark Lythgoe[1]
[1]*Centre for Advanced Biomedical Imaging, UCL, London, UK, United Kingdom,* [2]*Centre for Medical Image Computing, UCL, UK, United Kingdom*

Target Audience Preclinical researchers with an interest in small animal imaging, brain morphometry and quality control(QC).

Purpose Confidence in the accuracy and consistency of magnetic resonance imaging (MRI) measurements in small animal studies relies on the performance of quality assurance tests. A protocol, together with an open source calibration phantom (Figure 1, https://www.ucl.ac.uk/cabi/publications/open_source), is presented for the correction of errors caused by imaging gradients. A system calibration is performed to correct errors in gradient scaling parameters and a post processing correction is used to unwarp distortions caused by gradient non-linearity. All post processing software is feely available to download (NifyReg[1-3], http://cmic.cs.ucl.ac.uk/home/software/) making this simple protocol easy to adopt as part of a pre-clinical QC procedure.

Methods *Phantom:* A three dimensional grid phantom, designed to fit into an RF coil of 35mm diameter, was constructed using Fine Polyamide (PA2200) material. The phantom consisted of a cylindrical chamber measuring 60mm in length with a inner diameter of 26mm. A series of regularly spaced rods cross the phantom chamber in the X, Y, and Z axes form a 3D grid structure. The rods are spaced 2.5mm apart and are 0.5mm thick. The phantom was filled with a solution of copper sulphate and sodium chloride in water. *Imaging:* Computed Tomography (CT) images of the phantom were acquired using a Bioscan nanoSPECT/CT with an isotropic voxel resolution of 73µm that could be used as a ground truth. MR data was acquired on a 9.4T Agilent scanner using a 3D gradient echo sequence that had been optimized for imaging ex vivo murine brains(TR = 17ms, TE=4.54ms, flip angle = 51 degrees, 5 averages). The resolution was 100 µm isotropic with a Field of View of 40mm x 40mm x 60mm. The phantom was placed in a 35mm diameter Rapid Biomedical RF volume coil for imaging that was inserted into a gradient set of 60mm inner diameter (SGRAD 115/60/HD/S, Agilent Technologies UK Ltd., Berkshire, UK). *System Calibration:* Scaling values were generated by performing an affine registration(NiftyReg) of the MR data to the CT imaging volume. Imaging and scaling were carried out at monthly intervals over a period of seven months. The scaling factors from the first time point were used to adjust system parameters and stability was measured over the following time points. *Post processing correction:* A displacement field for distortion correction of images was generated by a two step registration process using NiftyReg software. A rigid registration and resampling of the MR data was carried out first to bring it into the CT imaging space. This was followed by a non rigid registration of the resampled MR data to the CT data which produced a displacement field.

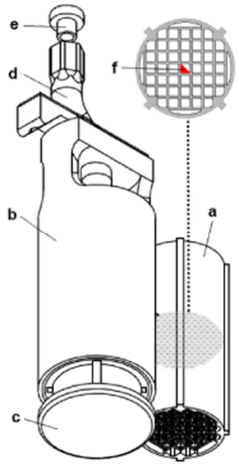

Figure 1 Phantom schematic: a) grid section, b) chamber, c) chamber cap, d) s-bend to trap air bubbles, e) filling cap, f) isocentre marker

Results The mean scaling factor error (relative to unity) across all axes after system calibration (dotted line) was reduced from 2.7% to 0.3%(Figure 2). The mean and standard deviation of the scaling factors measured over the six month period after calibration were 99.7±0.1%, 100±0.2%, and 100.1±0.1% in the X, Y and Z axes. A displacement field was generated from the deformation field output by the non-rigid registration. It shows the magnitude of voxel displacements applied to the MRI data to unwarp and register it to the CT data (Figure 3). The displacements along the Z axis (Fig. 3a) are less than 0.1mm at a distance of less than ±10mm from the isocentre. At a distance of ±20mm and greater, the displacements increase rapidly to more than 0.35mm. At distances greater than ±10mm from the isocentre , there is also a rapid increase to values larger than 0.25mm in the X (Fig3b) and Y(Fig3c) direction within the walls of the grid section of the phantom.

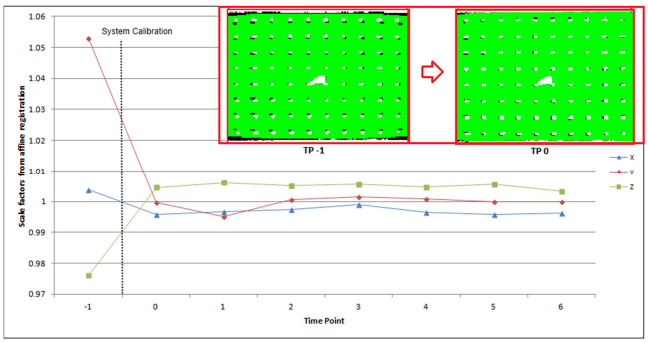

Figure 2 Scaling values plotted versus time point, insert shows alignment of MRI data(black) and CT data (white) before and after calibration

Discussion

The system scaling error has the potential to be a significant confounder to detection of structural volume changes in the mouse brain [4]. Our system calibration reduced the gradient scaling error to 0.3% which may be significant in studies of murine brain volumetrics. Displacements of greater than 100µm were detected in imaging regions affected by gradient non-linearity and corrected through non-rigid registration of the MRI data to the CT data. Scaling values were stable over the six month time period indicating that it may be satisfactory to carry out as few as two system calibrations per annum. The system calibration and post processing correction provide a measure of the magnitude of errors that may result from inaccuracies in applied gradients. The step-by-step protocol implemented could be integrated into any pre-clinical MRI quality assurance protocol to measure and correct for these errors.

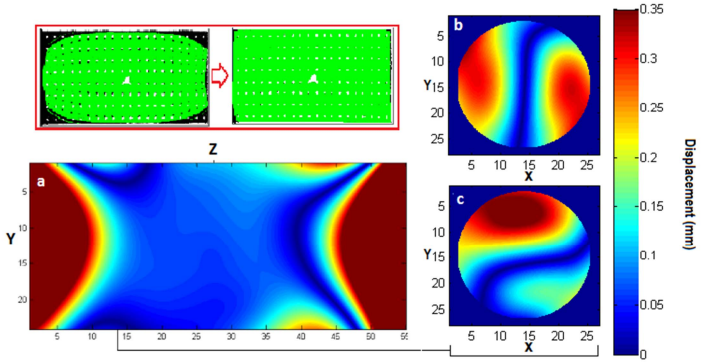

Figure 3 Displacement fields generated by the post processing correction in the z (a), x (b), and y (c) directions. Insert show CT and MRI alignment before and after registraion

References

[1]Ourselin S, Roche A, Subsol G, Pennec X, Ayache N. Reconstructing a 3D structure from serial histological sections. Image and Vision Computing 2001;19(1–2):25-31.
[2]Ourselin S, Stefanescu R, Pennec X. Robust Registration of Multi-modal Images: Towards Real-Time Clinical Applications. In: Dohi T, Kikinis R, editors. Medical Image Computing and Computer-Assisted Intervention — MICCAI 2002. Volume 2489, Lecture Notes in Computer Science: Springer Berlin Heidelberg; 2002. p 140-147.
[3]Modat M, Ridgway GR, Taylor ZA, Lehmann M, Barnes J, Hawkes DJ, Fox NC, Ourselin S. Fast free-form deformation using graphics processing units. Computer Methods and Programs in Biomedicine 2010;98(3):278-284.
[4]Bock NA, Nieman BJ, Bishop JB, Mark Henkelman R. In vivo multiple-mouse MRI at 7 Tesla. Magnetic Resonance in Medicine 2005;54(5):1311-1316.