Validation Testing of the MRI Calibration Phantom for the Alzheimer's Disease Neuroimaging Initiative Study

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Introduction: The Alzheimer's Disease Neuroimaging Initiative (ADNI) is a five year natural history study of Mild Cognitive Impairment and Alzheimer's Disease. Eight hundred elderly individuals will be scanned on 90 different MRI systems at 59 sites. Subjects will be scanned serially at roughly 6 month intervals with some variation depending on clinical status. All subjects will be scanned at 1.5T and 25% will also be scanned at 3.0T at each time point. In total over 5000 MRI studies will be performed.

In order to track the performance of scanners involved in the study each enrolling site has received a phantom designed specifically for this study. The phantom will be scanned at the end of the scanning session for each patient at every time point. This guarantees that nearly contemporaneous phantom images will exist for all subject scans. This abstract reports on the stability of measurements from a single phantom and the utility of the phantom-based measurements for correcting linear geometrical distortion in associated human images on a per study basis.

Materials and Methods: The ADNI phantom is a water-filled 20 cm diameter shell with 158 1.0 cm diameter spherical inclusions, two 1.5 cm diameter spherical inclusions, four 3.0 cm diameter spherical inclusions and one 6.0 cm diameter spherical inclusion (Figure 1). The 1.0 and 1.5 cm diameter spheres contain 3.3 mM concentration copper sulfate solution. These "small spheres" are positioned at known locations and are used to measure spatial distortion. Additionally, location and size asymmetries in the layout of the small spheres allow the overall orientation of the phantom to be uniquely determined. The large spheres (3.0 and 6.0 cm diameter) are used for measurements of T1 contrast and signal to noise. The 6.0cm sphere is approximately concentric to the 20cm outer shell while the 3.0cm spheres are approximately 6cm from the center of the phantom. The central laboratory for ADNI MRI quality control is located at the Mayo Clinic. All scans used as input to the analyses presented here were acquired on a GE Signa 1.5T system. Unless otherwise stated a 3D sagittal MP-RAGE sequence (acquisition voxel size 1.2 x 1.2 x 1.2 mm^3) specifically designed for this study was used. An affine (9DOF) transformation may be found which minimizes the sum-of-squared-distances between engineered positions and observed sphere positions. The space scaling parameters of the affine transformation then serve as a measure of linear gradient calibration.

Studies and Results: To explore the robustness of the phantom and the analysis methods several studies were carried out.

We assessed the reproducibility of measurements from the phantom and analysis software in the face of manual phantom placement variability. A sequence of 12 scans on a system with Body Resonance Module (BRM) gradient coils was acquired where the phantom was removed and re-inserted between scans. The standard deviation of the distribution of space scale factors from the affine transformations along each axis was from 0.02% to 0.04%. A similar set of scans in which the phantom was rotated by approximately seven degrees about each axis was also acquired. The standard deviations of scaling factors for the rotated data sets were 0.03% to 0.05%. For comparison, treating the sphere position measurements as independent with 0.1mm design and manufacturing precision one would expect a measurement error for linear scaling factors to be around 0.004%.

The dependence of phantom measurements on the pulse sequence type and acquisition orientation was assessed. Scans using the ADNI MP-RAGE, standard 3D-SPGR, and standard 3D-FSE acquisitions were analyzed and the results of the analysis compared. Images were acquired on a system with TwinSpeed gradient coils and gradient un-warping (gradwarp) was applied in 3D as an offline process. Each sequence was scanned with slice orientation in each cardinal plane yielding nine permutations of orientation and pulse sequence. Analysis results were consistent across orientations within a pulse sequence to within 0.07% with typical standard deviation of 0.01% along a given axis. The standard deviations of the scale factors across all nine scans (i.e., ignoring potential pulse sequence dependence) were 0.04%, 0.07% and 0.09% in the R/L, A/P and S/I directions. No significant pulse sequence dependence was observed on this system.

Figure 1. MP-RAGE images of the ADNI phantom. Small high contrast spheres are used for spatial measurements.

To examine potential heating effects on the observed gradient scaling, continuous scanning with the MP-RAGE sequence was carried out for a two and half hour period. Scanning on a system with TwinSpeed gradient coils was begun "cold" after a full weekend of inactivity. By linear regression the observed time dependence of the scale factors over this 2.5 hour period was approximately 0.01% to 0.03% per hour depending on direction (p<0.002 in all cases). The R/L data are shown in Figure 2. Similar results were observed in the R/L and S/I directions on a system with BRM gradient coils although the time span was shorter (1 hour).

We evaluated the ability of contemporaneous phantom measurements to appropriately rescale human images obtained under conditions of intentional gradient scaling changes. Specifically, scans were acquired at nominal gradient amplitude, and nominal plus 0.50%, 0.25% and 0.125% for the human subject and the phantom. The ratios of scale factors between nominal (baseline) and scaled gradient settings were computed for each gradient scaling step. For completeness, all phantom scans were done in one session with no re-positioning of the phantom within the magnet. Subject scans were also performed in a single session without re-positioning the subject. The reliability of the phantom in recovering the induced scaling was evaluated. The induced scaling was recovered to within 0.02%. Using the phantom derived estimates of the induced scalings, the voxel sizes were modified in the human subject image volumes. The human data was then registered using an affine transformation with AIR[1]. If the phantom perfectly captured the changes in the actual gradient and the same changes affected the human scans in the same ways, then the scaling factors expected for human-human co-registration would be exactly unity. The human-human co-registration scale factors have mean 0.9997, standard deviation 0.0005 and appear independent of the induced scaling and independent of direction.

Summary & Conclusions: The variability of measurements performed with a single ADNI phantom under a variety of conditions is quite small. Gradient drift (ascribed to heating) on a system over several hours were minimal, with a per hour drift similar to the variability in a single phantom measurement when the phantom is removed and replaced in the magnet. Application of phantom based corrections to deliberately scaled human images 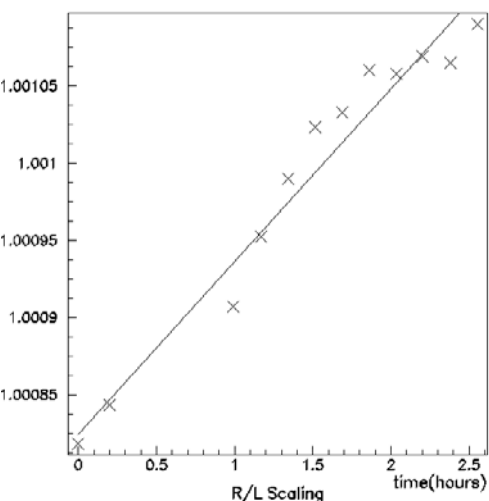can rescale the human images to within 0.05%. In conclusion, systematic errors in brain MRI due to drifts of gradient scaling can be minimized by rescaling the brain images based on associated measurements on the ADNI phantom of linear distortion.
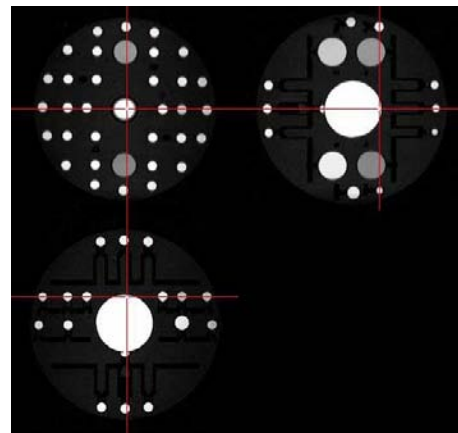

Figure 2. Right/Left scaling as a function of time is shown with a regression line.

[1] Woods RP, Grafton ST, Holmes CJ, Cherry SR, Mazziotta JC. Automated image registration: I. General methods and intrasubject, intramodality validation. Journal of Computer Assisted Tomography 1998;22:139-152.