Retrospective Distortion Correction Using the ADNI Phantom to Salvage Unusable Exams

B. Dzyubak[1], J. L. Gunter[1], E. B. Welch[2], R. J. Killiany[3], C. R. Jack[1], and M. A. Bernstein[1]

[1]Radiology, Mayo Clinic, Rochester, MN, United States, [2]Philips Healthcare, Highland Heights, OH, United States, [3]Boston University School of Medicine, Boston, MA, United States

Introduction: The Alzheimer's Disease Neuroimaging Initiative (ADNI) [1, 2] is a six-year, 800-subject observational study to assess how well the combined information obtained from MRI, PET, other biological markers, as well as from clinical and neuropsychological assessment can measure the progression of mild cognitive impairment (MCI) and early Alzheimer's disease (AD). All of the subjects are imaged with 1.5T MRI, and a subset (25%) with 3T MRI. A total of approximately 5500 MRI exams at 59 sites are planned over the course of the study, which is scheduled to be completed in 2010. All of the image data are readily available via the Internet to any researcher. Longitudinal, T1-weighted, 3D volumetric imaging can detect subtle changes in brain volume and morphometry associated with the progression of MCI and AD. Consequently, spatial fidelity is a key performance criterion for the MR data in the ADNI study, which employs a dedicated MR phantom with fiducial markers to accurately characterize local geometric distortions (3). The resulting 3D image sets are analyzed with AQUAL (3) to provide a graphical representation of the gradient distortion error and a global image distortion metric.
One of the 1.5T ADNI scanners suffered a mis-calibration of laser alignment light after an upgrade, causing the on-line distortion correction to be performed about the wrong isocenter, introducing spatial distortion artifacts. The value of the alignment light displacement error (i.e., number of mm S/I) was unknown. The problem was not detected until after additional subject exams were acquired. Images acquired during this period were deemed unusable for tracking changes in brain volume, because the technical error was of the same order of magnitude as the anatomical changes in brain volume due to the progression of disease.

Methods: We developed an off-line gradient distortion correction using a series expansion to model the known spatial dependence of the scanner's gradient field. The gradient distortion correction coefficients depend only on the hardware configuration (e.g. the gradient coil type), and are stored in off-line data files. The correction can be run either in forward mode (i.e., gradient distortion correction) or inverse mode, which undoes a previously-applied gradient distortion correction. ADNI phantom images were acquired on a properly functioning scanner of the same manufacturer and then reconstructed with and without the standard, online distortion correction. These data were used to test the operation of our off-line gradient distortion correction, both in forward and inverse mode. Then ADNI phantom images from the mis-calibrated scanner were inverse distortion corrected, translated by varying amounts of S/I displacement Δz , and then forward-corrected. The spatial distortion error was characterized by AQUAL for each displacement value Δz .

Results: First the operation of the off-line distortion correction was verified. The differences between the commercial on-board distortion correction and our off-line correction were less than 0.01 mm over a 200 mm diameter sphere. Then, by minimizing the spatial distortion metric calculated by AQUAL on the re-corrected images, the actual displacement error of the laser alignment light was estimated. For the aberrant data, a translation by $\Delta z = +40$ mm in the S/I direction minimized the root-mean-square (RMS) distortion after inverse-distortion correction, translation, and forward correction (Figure 1). The RMS displacement error decreased from 0.5162 to 0.3359 mm, and the corresponding maximum error decreased from 1.8 to 0.9 mm as shown on Figure 2, where spatial error (e.g., ΔA/P in mm) is plotted on the horizontal axes. Prior to the upgrade, typical RMS values for this scanner were in the 0.34 to 0.36 mm range.
Images of the same subject acquired before and after the laser light misalignment were registered and subtracted. Figure 3 shows that the re-correction process also reduces the local spatial distortion in vivo. The standard deviation along the indicated line was reduced from 80 to 65 (arb. units) by the re-correction. The re-corrected images are considered usable for ADNI.

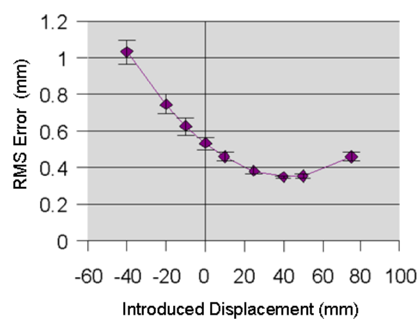

Fig 1. Inverse distortion correction, translation by $\Delta z = +40$ mm and forward distortion correction minimizes spatial error due to gradient distortion.

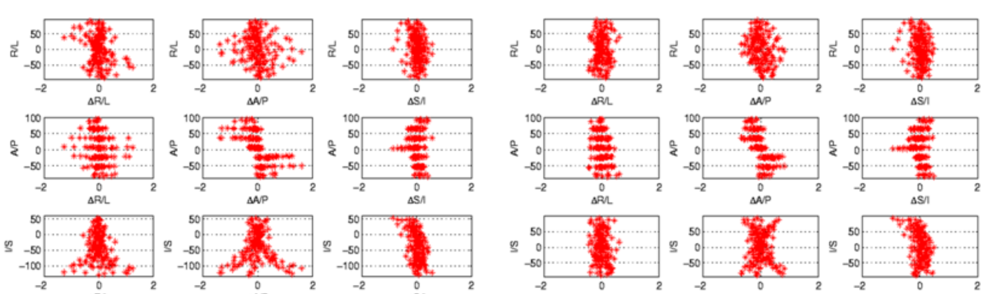

Fig 2 AQUAL plots of spatial distortion error measured with the ADNI phantom acquired on the mis-calibrated scanner before (left) and after (right) correction process described here. The plots on the right show significantly reduced spatial error.

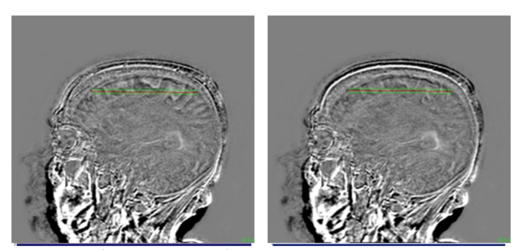

Figure 3: Subtraction of registered pre-and post upgrade (i.e., with the mis-calibrated laser light) images: Right: after re-correction.

Discussion and Conclusion: In longitudinal (i.e., serial) multicenter studies, each subject exam represents a considerable investment of time, money and effort. It was shown here how use of a dedicated phantom, quantitative analysis, and post-processing methods can salvage otherwise unusable subject data.

References: 1. Mueller SG, Weiner MW, Thal LJ, et al. Alzheimers Dement. 2005; 1: 55-66.2. Jack CR, Bernstein MA, Fox NC et al., J Magn Reson Imaging 2008; 27: 685-91. 3. Gunter JL, Bernstein MA, Borowski BJ et al, Med Phys. 2009;36: 2193-205.