Removing Gradient Non-Linearity Effects in Deformation Morphometry of High Field Serial MRI

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Introduction

Deformation tensor morphometry of serial structural MRI data is an important tool in studying subtle, focal changes in tissue volume over time, and has been used to detect characteristic patterns of tissue loss in neurodegenerative disease[1]. Higher MRI field strengths promise to provide increased resolution and contrast to noise to further improve detection of small scale changes in anatomy. However, the data poses challenges from the presence of increased geometric imaging distortions at higher fields. One key source of these distortions arises from gradient non-linearites[2]. This is a significant problem when subjects have not been positioned at the same location within the bore of the magnet for the two images. At high field this can arise when positioning varies by a few centimeters. If the non-linearities are known accurately, their influence on morphometric measurements can be reduced or removed by applying a correction unwarping, using the known location of the image volume within the bore of the magnet. For 3D morphometry this correction must be done on the entire volume in 3D, not simply on a slice by slice basis. It is not always the case that 3D gradient correction is available for a given scanner. In this abstract we examine a simple alternative approach to removing the influence of these distortions from serial MRI morphometry measurements that does not require knowledge of the gradient non-linearity.

Method: Deformation morphometry relies on the estimation of fine scale spatial transformations between scans to capture shape changes. In our approach we assume that gradient non-linearities result in low spatial frequencies in these transformations, and apply a filtering of the displacement fields to remove their effect from the focal atrophy maps. We use a regionalized mutual information driven viscous fluid registration to provide a contrast robust estimate of shape changes between scan pairs, derived from that described in [3]. The estimate of the deformations between images is described by a voxel by voxel vector field $\boldsymbol{u}(\boldsymbol{x})$, such that the coordinates at the second time point $\boldsymbol{y}$ are given in terms of a transformation of locations in the first time point such that $\boldsymbol{y}=\boldsymbol{T}(\boldsymbol{x})=\boldsymbol{x}+\boldsymbol{u}(\boldsymbol{x})$. We then fit a coarse B-Spline basis function to the voxel-wise displacement field $\boldsymbol{u}(\boldsymbol{x})$, using a regular knot spacing. For this work we have used a simple least squares fit of the deformation estimate and experimentally found a coarse knot spacing of 40mm to be suitable. This B-Spline model parameterizes a low resolution deformation estimate $\boldsymbol{u}'(\boldsymbol{x})$, which captures the coarse shape changes only. We then subtract this from the estimated deformation field, to create a modified morphometric deformation field $\boldsymbol{T}'(\boldsymbol{x})=\boldsymbol{T}(\boldsymbol{x})-\boldsymbol{u}'(\boldsymbol{x})$. This is then spatially differentiated to create a map of focal changes in volume between images, given by the determinant of the Jacobian of the corrected transformation $|d\boldsymbol{T}'(\boldsymbol{x})/d\boldsymbol{x}|$.

Experimental Results: We applied the approach retrospectively to data acquired at the CIND in San Francisco. The subjects were studied on a Bruker/Siemens MedSpec 4T system using an eight channel phased-array receive coil. Sagittal, 3D T1-weighted MPRAGE images were acquired with the following parameters: TR/TI = 2300/950 ms, flip angle = 7deg, 1.0 x 1.0 x 1 mm resolution. FOV = 256 x 256 x176 mm, acquisition time = 5:10 min, GRAPPA acceleration = 2. Figure 1 Shows the removal of artifactual tissue losses mapped from a pair of images from a subject with no underlying change in tissue, but who was positioned at locations differing by 3cm's along the bore of the magnet. The colour map shows the estimated volume change pattern before and after removal of low deformation frequencies. Figure 2 shows the use of the same deformation filtering approach where a pattern of underlying tissue loss has occurred between time points in an aging subject. Here the subject was positioning consistently and therefore no gradient effects were present, but application of the filtering still preserves the true tissue loss.

Conclusions: By assuming that gradient non-linearities result in low spatial frequencies in the deformation morphometry data, we can apply a direct filtering of the deformation estimates calculated after non-linear registration, prior to deformation tensor morphometry. Experimental results indicate that significant improvements in the resulting maps are possible, without loss of sensitivity to true focal tissue losses. Further experiments are underway comparing gradient correction techniques to this approach.

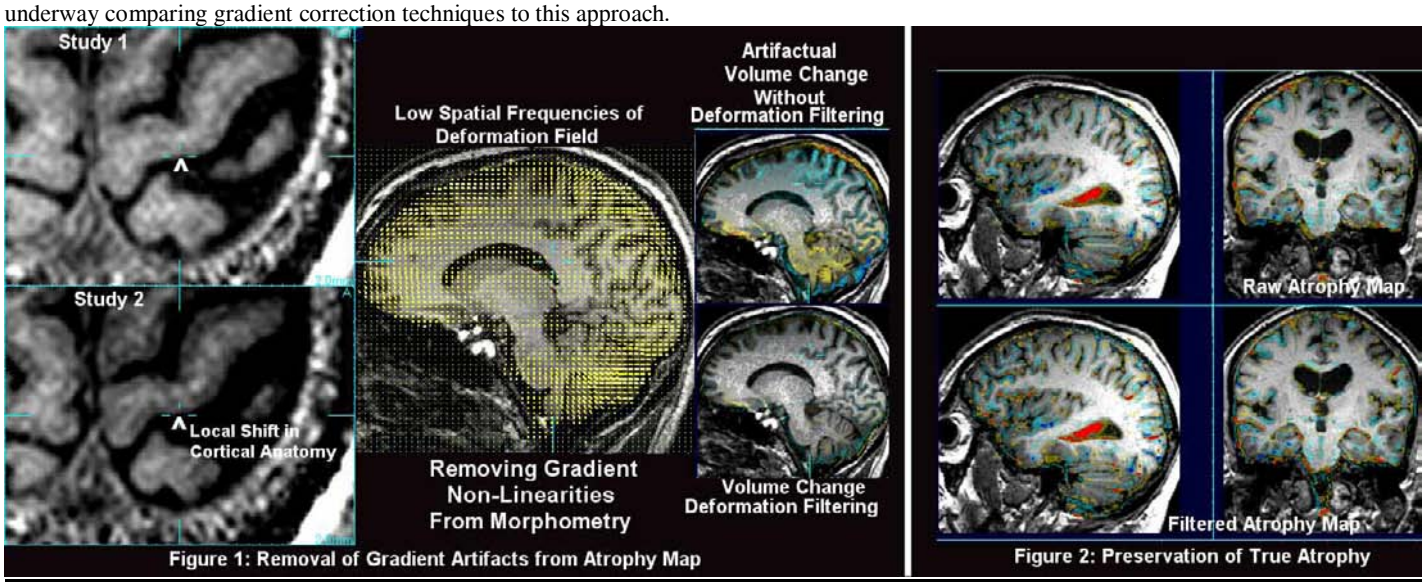

Figure 1: Removal of Gradient Artifacts from Atrophy Map

Figure 2: Preservation of True Atrophy

References

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