# Quantitative comparison of spatial distortions in fast imaging at 3T through nonrigid registration

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## 1. Introduction

Spatial image distortions in fast imaging techniques have previously been corrected through the use of free-form deformations, yielding tissue displacement measures that that compare favorably to field mapping techniques.[1] Image registration techniques have also been extensively applied to describe actual physical changes and inter-subject differences in anatomy, typically described as *deformation tensor morphometry*.[2–4] In this work, we used nonrigid registration to correct spatial distortions in fast imaging methods, and then analyze the estimated distribution of these spatial distortions to compare different acquisition methods. In contrast to computational anatomy studies, we apply these methods to compare image distortions, rather than actual changes in tissue volume. Numerical inversion of the deformation field is used to ensure the validity of a voxel-by-voxel comparison between two imaging techniques. The proposed methodology is demonstrated on a comparison between single-shot fast spin-echo (SSFSE) and spin-echo EPI and partially-parallel sensitivity-encoded (SENSE) and non-parallel gradient-echo echo-planar imaging (EPI) acquisitions. We propose this method as a means of systematically comparing two acquisition methods.

## 2. Materials and Methods

For the comparisons, an undistorted anatomical image (*reference* image) and two distorted images (*test* image) were acquired on a 3T GE Signa EXCITE scanner. For the first comparison, 16 × 4 mm slice SSFSE (TR/TE = 16000 / 74 msec) and spin-echo EPI (TR/TE = 5000 / 84 msec) images served as the test images and a fast spin echo (FSE) (TR/TE = 800 / 84 msec) was the reference image, all with a FOV of 22 × 22 cm$^2$ and a 256 × 128 matrix. This data was acquired with the standard quadrature head coil on six volunteers. For the second comparison, gradient echo EPI images were acquired (TR/TE = 1000 / 54 msec, FOV = 26 × 26 cm$^2$, 128 × 128 matrix, 35° flip) with and without SENSE partially-parallel imaging ($R$ = 2) on eight glioma patients. A T1-weighted 3D-SPGR (TR/TE = 27 / 6 msec, 1 × 1 × 1.5 mm$^3$) served as the reference image. Data for the second comparison were acquired with an 8-channel phased-array head coil.

The test images were registered to the anatomical image by optimizing the positions of a regular grid of B-spline control points.[3] The optimal deformation was found by maximizing the normalized mutual information by a gradient ascent search. From the registration results, for each point ($x$, $y$, $z$) in the test image, a mapping $\mathbf{T}(x, y, z)$ was applied to find the position ($x'$, $y'$, $z'$) in the undistorted reference image. For each point in the distorted image, the mapping $\mathbf{T}$ can be described simply by three vector components $\Delta x = x - x'$, $\Delta y = y - y'$, and $\Delta z = z - z'$, as well as by a magnitude $m = (\Delta x^2 + \Delta y^2 + \Delta z^2)^{1/2}$. We refer to the images of $\Delta x$, $\Delta y$, $\Delta z$, and $m$ as distortion maps. However, to directly compare two distortion maps, it is necessary to be able to select an arbitrary point on the *reference* image and know the displacement of that point in the two distorted test images. Thus, we evaluate the inverse mapping $\mathbf{T}^{-1}(x', y', z') \rightarrow (x, y, z)$ at every point in the reference image. A Nelder-Mead simplex method was used to search for the point ($x$, $y$, $z$) which minimizes the distance between $\mathbf{T}(x, y, z)$ and the desired voxel center in the reference image. Such a search requires only that the mapping $\mathbf{T}$ be a homeomorphism, and not necessarily diffeomorphic. This inversion is thus computationally more expensive but conceptually simpler than other proposed methods.[4] Using the inverted maps $\Delta x'$, $\Delta y'$, $\Delta z'$, and $m'$, points or regions of interest (ROIs) may be defined on the reference image, and the distortions within those regions compared for two imaging methods.

In this study, we considered two possible ROIs that provide indications of the degree of distortion: the whole brain in the reference image and the outer rim of the brain. Both ROIs were restricted to the joint FOV of the test images, transformed into the reference frame, thus eliminating voxels that have no matching points in the test images. These ROIs were selected as examples: in practice, ROIs for specific anatomical structures of interest may be used.

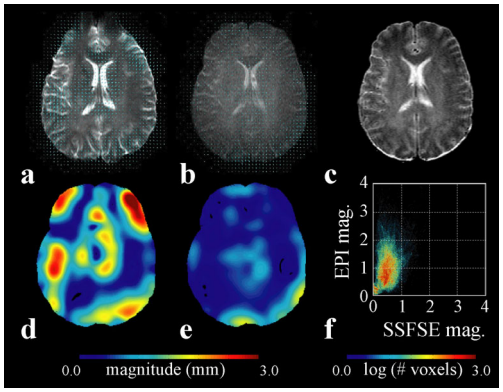


**Figure 1.** (a) EPI and (b) SSFSE with deformation vectors aligning them to (c) FSE. In the FSE frame are distortion maps for (d) EPI and (e) SSFSE. (f) 2D histogram of a voxel-by-voxel comparison of distortion magnitudes.

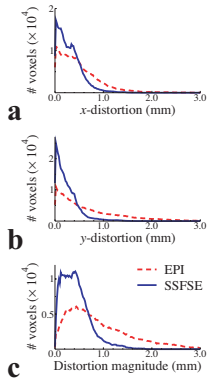


**Figure 2.** Histograms of absolute value of distortions for (a) x, (b) y, and (c) magnitude.

## 3. Results and Discussion

An example of the analysis results for a volunteer study comparing the SSFSE and spin-echo EPI images is shown in figure 1. The deformation maps were put in the frame of the reference image, where the difference in magnitude between the SSFSE and EPI is immediately obvious, as seen in figures 1(d) and (e). A voxel-by-voxel comparison was made of the distortion magnitudes, and a 2D histogram is shown in figure 1(f). The vertical skewing of the histogram suggests that the EPI image is consistently more distorted than the SSFSE image. The same dataset, but without spatial localization information, is shown in figure 2. Here, the shift of the EPI curve towards larger values suggests that as a whole, the EPI image exhibits larger distortion than the SSFSE. Generally, statistical analysis of the moments of both the 2D and 1D histograms will suggest the varying degree of distortion in the two images. In this study of six volunteers, the mean distortion magnitude of the EPI was an average of 2.41 ± 0.22 times larger than the SSFSE.

Visual comparison of the SENSE and standard EPI images suggests a smaller but noticeable reduction in distortion in the SENSE imaging, consistent with theory. Through nonrigid registrations, the mean whole brain distortion magnitudes in eight patients was found to be indistinguishable: the standard acquisition distortion magnitude was 0.99 ± 0.21 times that of the SENSE acquisition. Both the 2D histogram in figure 3 and the 1D histogram in figure 4(a) suggest that there is a "tail" of voxels in which the SENSE acquisition has smaller distortions in this example patient. This "tail" is largely obscured by the large peak of voxels in which no difference exists between the acquisitions. These large distortions were strongly confined to the edges of the brain, within the "rim" ROI. Within this ROI, the standard acquisition mean distortion was 1.15 ± 0.23 times the SENSE acquisition. Histograms for this ROI are shown in figure 4(b).

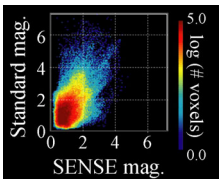


**Figure 3.** 2D histogram comparing distortions in SENSE and standard EPI.

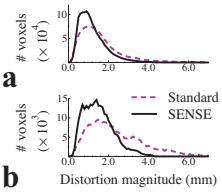


**Figure 4.** Distortion histograms for (a) whole brain and (b) rim ROIs.

## 4. Conclusions

The proposed methods produce quantitative results that are consistent with expectations based upon visual interpretation and imaging physics. We have adopted methods more commonly associated with computational anatomy and applied them to study geometrical changes due to imaging artifacts, rather than actual structural changes. While inherently limited by the accuracy of the image registration, use of automated methods for image distortion comparisons relies upon experimental data, thus including patient geometry effects and technical performance issues that are often difficult to model in theoretical analyses, while removing user bias in direct visual analysis.

[1] Studholme C *et al.* (2000) *IEEE Trans Med Imag.* 19: 1115–27.
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