Distortion Correction Using Simulated Point-Spread Functions

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INTRODUCTION: Field inhomogeneity causes intensity and geometric distortion in magnetic resonance (MR) images. There are several existing approaches to correct for this distortion, including measuring the point-spread function (PSF) with distortion [1]. We propose building a Simulated Point-Spread Function (S-PSF) using knowledge of the k-space trajectory and field map, instead of using a separate acquisition to measure the point-spread function. This method allows for faster, more complete field-corrected image reconstruction, and the sparseness of the system matrix will allow for translation to a parallel software implementation. We will show that the method not only corrects for distortion, but it outperforms other iterative methods for high off-resonance field inhomogeneties if the acquisition trajectory is chosen well.

THEORY: Field inhomogeneities caused by susceptibility differences in the brain, particularly at air-tissue interfaces, lead to distortions in the reconstructed images. We can correct this distortion using information from the k-space trajectory and field map. The S-PSF is estimated by a simple gridding reconstruction of the simulated k-space data from a single point in image space and combining the resulting distorted reconstructions of each point in a matrix. The field map of the object is included in the simulation of the PSF, but field correction is excluded from the gridding reconstruction. By combining the matrix of S-PSF's with sensitivity information from parallel receiver coils, we can form a system matrix, $A_{s\text{-}PSF}$, and set up the following equation to solve for the undistorted image: $f_{distorted} = A_{S-PSF} f_{undistorted}$ (1).

EPI is particularly susceptible to distortion due to its long readout time; the more time spent acquiring each pixel, increasing the echospacing, the worse the distortion. Our method is limited for correcting distortion when the spatial derivative of the field map at a significant number of pixels is greater than the bandwidth per pixel (BWPP) in the phase-encode direction of the acquisition, where BWPP = (1/echospacing) * (1/N) * (R) where N is the size of the image and R is the reduction factor from undersampling. However, we can correct for larger field inhomogeneities with an acquisition of multiple, under-sampled shots with higher effective BWPP.

Equation 1 demonstrates the case of correcting for distortion of a single-shot trajectory. This can be expanded to multi-shot trajectories, where the S-PSF is determined for each shot. When the trajectory is separated into shots, flipping the phase-encode (PE) direction between shots, the composite system matrix A=[A1; A2], where A1, A2 are the S-PSF of each shot combined with the coil sensitivity information, has more information about the distortion. Since the image is distorted in the phase-encode direction, the second shot has opposite distortion information, making the shift of individual pixels separate more clearly.

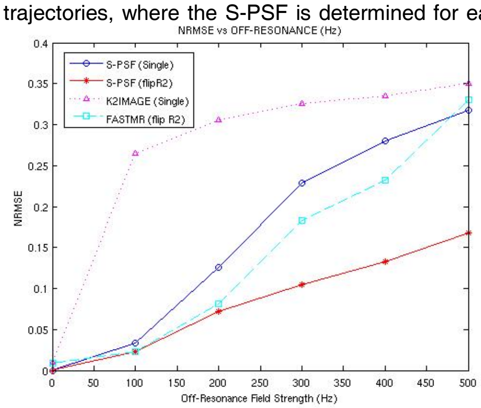

Figure 2 Comparison of S-PSF, Gridding, and Iterative[2] reconstructions of images with bumps in the field map from 0 to 500Hz off-resonance.

METHODS: We simulated distorted 4-coil MRI data in MATLAB for a Shepp-Logan phantom with matrix size of 64x64 by using a field map with a single bump of variable degrees of off-resonance. We compared the distortion-corrected image with the undistorted image, simulated using a homogenous field map. Further, we reconstruct the simulated object using conventional gridding and more common iterative methods [2], in order to quantify the improvement of S-PSF reconstruction over other methods. Comparisons were performed using normalized root mean squared error, NRMSE. In addition to the simulation data, human brain data were acquired using an EPI sequence on a Siemens Trio 3T scanner to verify the viability of the S-PSF reconstruction for a matrix size of 96x96, FOV=240 mm, and 12-channel head coil. Data was acquired from a single shot, fully encoded trajectory and from the combination of two single shot, R=2 acquisitions with TE_1=25ms and TE_2=52ms, with phase encode direction flipped between the two shots.

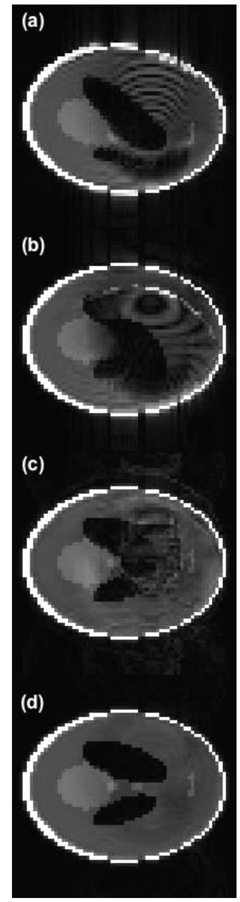

Figure 1: Reconstruction of image with 300Hz off-resonance bump in field map2 R2 Shots, phase encode flipped:
(a) Distorted Image
(b) Gridding
(c) Iterative [2]
(c) S-SPF

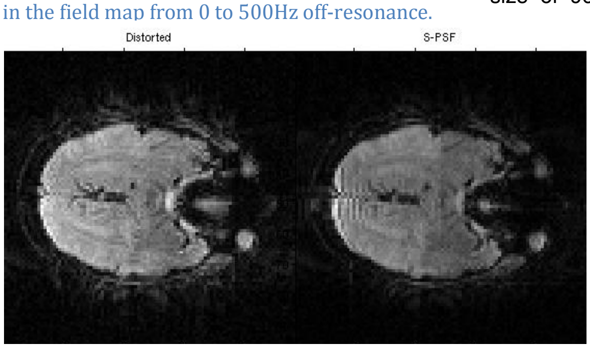

Figure 3: Distorted Image, S-PSF Corrected Image

RESULTS: Figure 1 shows the (a) distorted, uncorrected, (b) corrected, gridded method, (c) corrected, iterative method [2], and (d) corrected, S-PSF reconstructed images of a Shepp-Logan phantom in simulation with a 300Hz off-resonance bump in the field map. The NRMSE of each of the gridded field-corrected (dashed line), iterative method [2] (dotted line), and s-PSF (solid lines of colors varying by trajectory) methods is plotted in Figure 2 against increasing off-resonance field strength. While the error measurements for the three are close at lower fields, the conventional methods range from .18 to .35 NRMSE for conventional methods of correcting for 300-500Hz off-resonance and the S-PSF reconstruction ranges from .10 to .16 NRMSE for the same cases. It is clear that the S-PSF method is the most robust to high field inhomogeneity and that this robustness increases with the reduction factor of each shot of the trajectory. Figure 3 shows the distorted reconstruction of an R=2 single shot acquisition of the human brain and the image corrected using the S-PSF reconstruction. In order to combine two R=2 images for the S-PSF method, the T_2* decay must be accounted for; weighting the image with the longer echo time accordingly accounts for the differences in amplitude between the two images.

CONCLUSIONS: We have concluded that the S-PSF is an effective tool for distortion correction for EPI. The problem of long readout times for EPI in an inhomogenous field can be resolved with the S-PSF method simply by choosing an acquisition trajectory with greater effective BWPP. Thus, the S-PSF method is much more robust than current methods when large inhomogeneities are present.

REFERENCES: [1]Zaitsev, M., *et al.*, *Magn Reson Med*, 2004. 52(5):1156-66, [2]Sutton, B.P., *et al.*, *IEEE Trans Med Imaging*, 2003. 22(2):78-88.

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