703

Model based reconstruction for GRE echo planar imaging

W. Blecher[1], R. Pohmann[2], B. Schölkopf[1], M. Seeger[3];
[1]*Max Planck Institute for Intelligent Systems, Tuebingen, Germany,* [2]*Max Planck Institute for Biological Cbernetics, Tuebingen, Germany,* [3]*School of Computer and Communication Sciences, EPF Lausanne, Lausanne, Switzerland.*

Introduction:
Model based nonlinear image reconstruction methods for MRI [3] are at the heart of modern reconstruction techniques (e.g.compressed sensing [6]). In general, models are expressed as a matrix equation

$$\mathbf{y} = \mathbf{X}\mathbf{u} + e$$

where ***y*** and ***u*** are column vectors of k-space and image data, **X** model matrix and *e* independent noise.
However, solving the corresponding linear system is not tractable. Therefore fast nonlinear algorithms that minimize a function wrt.the unknown image are the method of choice:

$$\min_{\mathbf{u}} \psi(\mathbf{u}) = ||\mathbf{y} - \mathbf{X}\mathbf{u}||$$

In this work a model for gradient echo EPI, is proposed that incorporates N/2 Ghost correction and correction for field inhomogeneities. In addition to reconstruction from full data, the model allows for sparse reconstruction, joint estimation of image, field-, and relaxation-map (like [5,8] for spiral imaging), and improved N/2 ghost correction.

Methods:
N/2 ghosts are a direct consequence of the alternating readout gradient during EPI acquisition, which leads to a pixel-shift in k-space. Correction involves multiplication with a phase-factor that is estimated from additionally acquired lines [4]

$$y(k_x, k_y) = \sum_x e^{-i2\pi k_x x} \left(\sum_y e^{-i2\pi(k_y y + s(k_y)\phi(x,y))} u(x,y) \right)$$

$\phi(x,y)$ encodes phase shift

$s(k_y) = \pm 1$ encoding gradient polarity

B_0 inhomogeneities could be modelled in a comparable fashion [7]:

$$y(k_x, k_y) = \sum_x e^{-i2\pi k_x x} \left(\sum_y e^{-i2\pi(k_y y + t(k_y)\gamma\Delta B(x,y))} u(x,y) \right)$$

$\Delta B(x,y)$ field map

$t(k_y)$ trajectory time course

resulting in the proposed model:

$$\vec{y} = \sum_x e^{-i2\pi k_x x} \left(\sum_y e^{-i2\pi(k_y y + \psi(k_y, x, y))} \vec{u} \right)$$

$$\psi(k_y, x, y) = s(k_y)\phi(x,y) + t(k_y)\gamma\Delta B(x,y)$$

Importantly, **X**=**ZF**, with **F** the Fourier Matrix and $\mathbf{X}^H\mathbf{X}$ is block diagonal, therefore algorithms like ADMM[2] efficiently solve the minimization equation for the unknown image **u**.

Results:

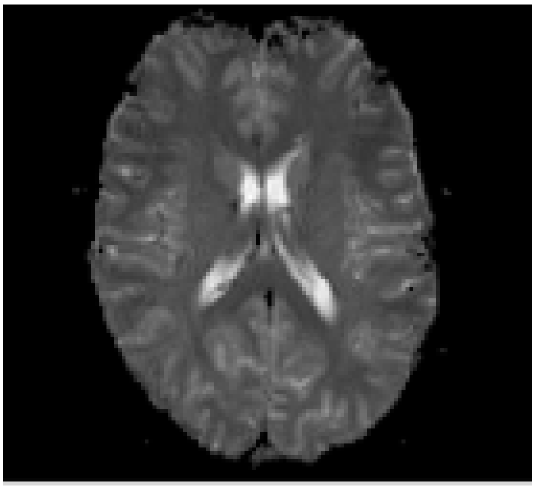

Figure 1: Model based Reconstruction, EPI TE 14ms, 128x128, FOV 200mm

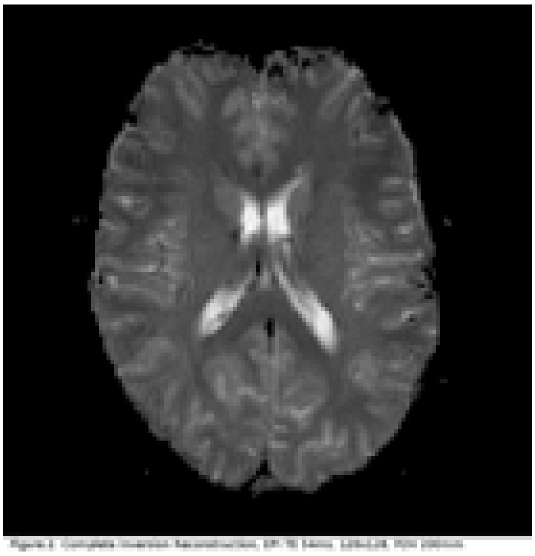

Figure 2: Complete Inversion Reconstruction, EPI TE 14ms, 128x128, FOV 200mm

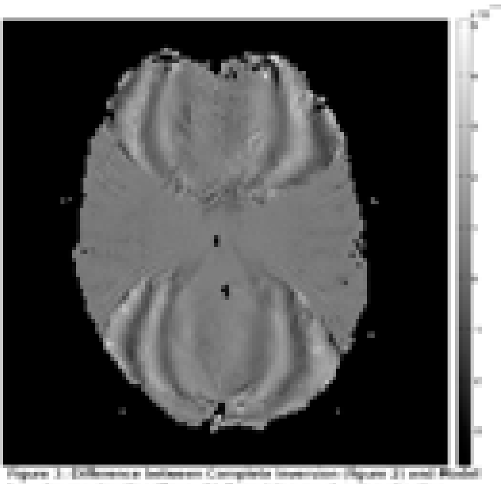

Figure 3: Difference between Complete Inversion (Figure 2) and Model based reconstruction (Figure 1). Remaining overlapping ghosting structures from the EPI image are clearly visible.

Figure 1 shows the image from the presented method and figure 2 the result from the inversion of

$$y(k_x, k_y) = \sum_x \sum_y e^{-i2\pi(k_x x + k_y y + \gamma t(k_x, k_y)\Delta B(x, y))} u'(x, y))$$

$u'(x, y)$Ghost free image

For both reconstruction methods the full B_0 map [1] has been taken into account and [4] has been used to estimate the phase shift factor. With less computational load, the model based method results in superior image quality with a higher ghost reduction and equal geometric distortion correction as shown in figure 3.

Conclusion: A successful implementation of a forward model for the gradient echo EPI imaging sequence has been presented. Fully acquired data and parameter sets have been included in the calculation and better image quality with faster computation has been achieved. In the next steps, reconstruction from sparse k-space acquisition and/or joint estimation of the field inhomogeneity map or the phase correction term are to be implemented.

References:

[1] Aksit P,et.al.2007; 4th IEEE Symposium Biomedical Imaging:From Nano to Macro:141-144

[2] Combettes PL,et.al.2011;Fixed-Point Algorithms for Inverse Problems in Science and Engineering,Proximal Splitting Methods in Signal Processing,Springer,2011

[3] Fessler JA.2010;IEEE Signal Proc Let,27(4):81-89

[4] Heid O.1997;Proceedings ISMRM:2014

[5] Knopp T,et.al.2009; IEEE Trans Med Imaging,28(3):394-404

[6] Lustig M,et.al.2007;Magn Reson Med,58(6):1182-1195

[7] Munger P,et.al.2000;IEEE Trans Med Imaging,19(7):681-689

[8] Sutton BP,et.al.2004;Magn.Reson.Med,51(6):1194-1204

704

Prospective phase correction for diffusion-weighted single shot STEAM with maximised SNR

R. Stirnberg[1], T. Stöcker[1], N. J. Shah[1,2];

[1]Institute of Neuroscience and Medicine - 4, Forschungszentrum Jülich, Jülich, Germany, [2]Faculty of Medicine, Department of Neurology, RWTH Aachen University, JARA, Aachen, Germany.

Purpose/Introduction: A Diffusion-weighted Single shot Stimulated Echo Acquisition Mode (DW Ss-STEAM) sequence has recently been proposed which achieves maximised signal-to-noise ratio (SNR) by coherently refocusing higher-order-echoes together with primary stimulated echoes [1]. Although immune to geometrical distortions, this "odd+even echo" DW Ss-STEAM version (cf. 1a) suffers from artefacts mainly related to motion during DW which cannot be corrected for retrospectively (cf. 1c). Thus, a completely artefact-free "only odd echo" version at the expense of reduced SNR has been proposed [2] (cf. 1b). This work provides one means to overcome the limitations of the "odd+even echo" sequence by applying prospective first-order phase compensation based on a quick "only odd echo" prescan measurement.

Subjects and Methods: One low-resolution artifact-free volume is acquired per DW (~20 slices within TR~5s) for the systematic phase difference to a non-weighted reference. The DW must be identical in the final "odd+even echo" scan. Linear fit to each difference map (after preliminary linear phase-wrap removal using Ahn and Cho's method [3]) yields x/y/z slopes and intercept (best linear phase approximation) to be finally compensated by field gradient blips and RF pulse phases in each readout interval.

A cylindrical white/grey matter gel phantom [4] has been imaged on a 3T TRIO (Siemens) scanner using a 12-channel receive array. The sequences were performed with [both:] b=1000s/mm^2 in slice selection (SS), read-out (RO) and phase encode (PE) direction, 256x176mm^2 transversal FOV; [calibration/final data matrix:] 32x22x15/128x88x60.

Results: Fig. 1 (c-g) each show central, orthogonal slices without (c,d) and with (e,f) prospective linear phase correction on an example of extreme phase variation due to RO-DW. As the ADC map (g) demonstrates, the residual non-linear phase artefacts are negligible. The histograms of residual phases (h) indicate that correction was more successful for SS-DW (blue) and PE-DW (red). The extremely low signal in (c,e) is due to high apparent diffusivity (2.2x10^-3mm^2/s).

Discussion/Conclusion: Fully coherent DW Ss-STEAM is applicable using prospective first-order phase compensation. The artifact-free DW Ss-STEAM calibration scan yields sufficiently accurate estimates in reasonably short time. Correcting for residual non-linear phases using tailored RF pulses, as recently shown for a different sequence [5], may be feasible. To account for brain pulsation both calibration and final in vivo measurements have to be synchronised to the cardiac cycle [6].

References:

[1] Stöcker,T. et al.,2009,MRM,372-380.

[2] Stirnberg,R. et al.,2009,MAGMA,134-135.

[3] 1. Ahn,CB. et al.,1987,IEEE MI,32-26.

[4] Kato,H. et al.,2005,MedPhys.,3199-3208.

[5] Nunes,RG. et al.,2011,Proc.Intl.Soc.MRM,172.

[6] Norris,DG. et al.,2001,MRM,729-733