Sensitivity encoding and B0 inhomogeneity - A simultaneous reconstruction approach

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INTRODUCTION:

Parallel imaging is increasingly used in the clinical MRI setting. A variety of different reconstruction approaches were proposed (1,2,3) that either work in k-space or image domain. The problem of B_0 artifacts in parallel imaging however is not settled, since 'B_0-encoding' does not happen in either k- or image space but simultaneously in both. This is why a concatenation of unaliasing and traditional B_0-reconstruction fails and a simultaneous reconstruction must be conceived. In the present work the generalization of an iterative B_0-corrected reconstruction (4) toward parallel imaging is described and in-vivo results are presented.

THEORY AND METHODS:

At the reconstruction level, the problem of B_0 inhomogeneity is typically addressed by unwarping (5) or conjugate phase reconstruction (6,7,8). However, the conjugate phase approach relies on the assumption that B_0 varies smoothly in space. Recently, iterative solutions were suggested which can cope with strongly varying B_0 (4, 9). The problem of compensating for in-plane B_0 inhomogeneity in the reconstruction is frequently badly conditioned, leading to numerical instability and local noise enhancement. To address this problem, it has been proposed (4) to minimize the weighted sum of the noise variance and the expected squared signal error on a pixel-by pixel basis. This is achieved with the reconstruction formula:

$$\mathbf{I} = \beta\theta E^H(\beta E\theta E^H + \alpha\psi)^+\mathbf{m} \qquad (1)$$

where $\mathbf{m}$ is a vector of multiple-coil k-space data, $\mathbf{I}$ the vector of the reconstructed pixel values, E the encoding matrix and E^H its Hermitian adjoint. The superscript + denotes the Moore-Penrose pseudoinverse. α and β are coefficients representing the weight of noise and signal variance, respectively, in the joint minimization. Ψ denotes the noise covariance of the sample values and θ denotes the signal covariance, which represents potential prior knowledge about the signal distribution in the object. Note that the set of target pixel positions can be arbitrarily large, corresponding to an arbitrary target image matrix. For the particular case of sensitivity- and Fourier encoding under the influence of B_0 inhomogeneity, the encoding matrix reads

$$E_{\kappa,\rho,\gamma} = s_\gamma(\mathbf{r}_\rho)\exp(-i\mathbf{k}_\kappa\mathbf{r}_\rho - i\omega_\rho t_\kappa), \qquad (2)$$

where $\mathbf{k}_\kappa$ and $\mathbf{r}_\rho$ denote the κ-th sampling position in k-space and the position of the ρ-th pixel in the image domain, respectively. ω_ρ denotes the B_0-induced frequency offset at the position $\mathbf{r}_\rho$, and t_κ denotes the time at which the sample κ is taken. The sensitivity of coil γ at position $\mathbf{r}_\rho$ is given by $s_\gamma(\mathbf{r}_\rho)$. For image reconstruction with Eq. (1), the matrix inversion part is solved by the CG method. Its solution $\mathbf{x}$ can then be pre-multiplied by the left-most part of Eq. (1) to get the reconstructed image: $\mathbf{I}=\beta\theta E^H\mathbf{x}$. The efficiency of each loop in the CG scheme (Fig.2) can be enhanced considerably by performing the matrix-vector multiplications with E and E^H by combining multifrequency interpolation (MFI) (8) with fast Fourier transform, in a scheme related to an approach proposed in Ref. (10) (Figs.1 and 2).

Experiments were performed on a 1.5T Gyroscan Intera system (Philips Medical Systems, Best, The Netherlands), imaging a water bottle and a human head in-vivo with a SENSE head coil. To induce substantial B_0 inhomogeneity in the phantom, four small permanent magnets were placed at the phantom's surface, creating B_0 offsets ranging from –990Hz to +730Hz. In the volunteer a slice right above the sinuses was chosen which naturally shows inhomogeneous B_0. For creating strongly B_0-affected data, a spin echo technique with very small pixel bandwidth of 73 Hz per pixel (resp. 24 Hz pp) was used with a matrix of 256x256. α and β were chosen such as to yield approximately identical noise and artifact levels. θ was set to a diagonal matrix, with ones for pixels inside the object and some fringe region and zeros outside. Since the samples were taken at different times, noise was uncorrelated and Ψ was a multiple of identity, whose scaling could be assessed from the image background.

Fig.1

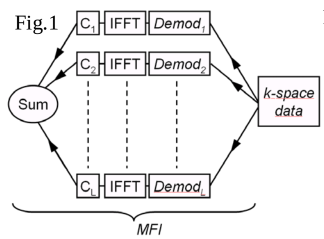

Fig.2

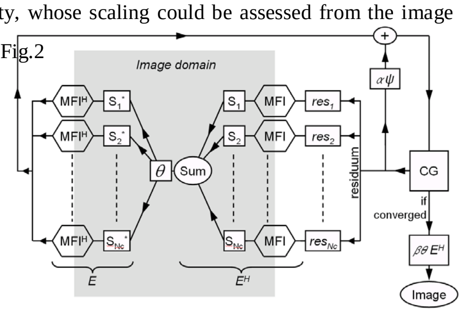

Fig. 1: In the MFI the k-space data is demodulated at L different frequencies ($Demod_i$), inverse Fourier transformed and multiplied by the corresponding coefficient maps (C_i) before being summed up. Gridding is not shown but can readily be incorporated.

Fig 2: One iteration of the CG loop is shown where the residual (computed by CG) for every coil is treated by MFI before being multiplied by the sensitivity map of the respective coil (S_i). After the multiplication with θ the data is multiplied by E (the adjoint of E^H). At the end of each loop the original residual, multiplied by $\alpha\Psi$, is added for regularization.

RESULTS:

The measured data was reconstructed using standard DFT reconstruction, conventional MFI, and the proposed reconstruction (30 iterations, reconstructed on a 512x512 grid, 2 hours computation on a 2.8 GHz CPU). MFI recovers the original shape of the phantom. However the aliased portions in the MFI image remain distorted; this illustrates that straightforward SENSE unfolding will not work in this case. Furthermore, MFI yields a false intensity distribution due to steep B_0. The proposed algorithm yielded an accurate image free of aliasing and erroneous intensity variation. In particular unlike MFI it does not assign excessive signal close to the region of strongest B_0-variation.

Fig.3

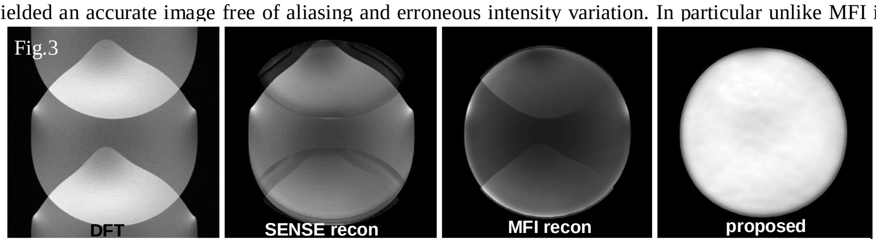

Fig.4

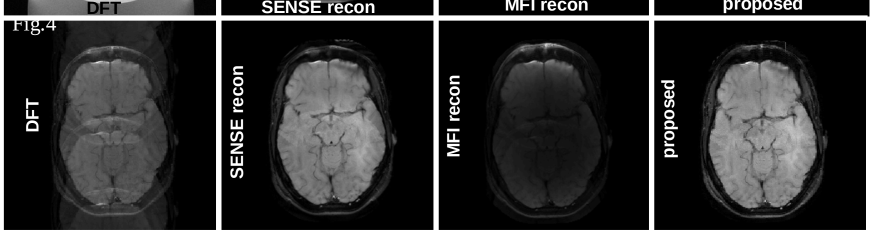

Fig. 3: The low readout BW in combination with the strong B_0-inhomogeneities led to notable distortions of the image. The SENSE reconstruction fails to unalias whereas the MFI results in warped aliasing. Only the simultaneous correction for both artifacts leads to a reconstruction almost free of artifacts.

Fig. 4: The B_0 inhomogeneity in the slice right above the sinuses leads to distortion of the frontal part of the brain. The proposed reconstruction outperformes both the SENSE reconstruction, showing less residual aliasing and the MFI having better intensity correction at the front than MFI.

DISCUSSION AND CONCLUSION:

The proposed method permits simultaneous correction of B_0-corrupted SENSE data without restrictions in terms of local B_0 variation. The reconstruction scheme is readily applicable toEPI data if the k-space trajectory is accurately known, suggesting promising applications in fMRI and DTI. The reconstruction scheme is readily applicable to general k-space trajectories, suggesting promising applications in fMRI and DTI. Ultimately, the reconstruction is limited by the inherent information content of the acquired data. Excessive field inhomogeneity will cause deteriorating conditioning, shifting the balance in the weighted minimization toward unfavorable compromises of the spatial response. Potential means of working against this problem are fast, non-Cartesian sampling patterns. **ACKNOWLEDGEMENT:** This work was supported by the SEP Life Science Grant TH7/02-2.

REFERENCES: [1] Sodickson, DK, MRM 38:591-03, [2] Pruessmann, KP, MRM 42:952-62, [3] Griswold, MA, 47:1202-10, [4] Barmet, C, Proc ISMRM03, p.347 [5] Jezzard, P, MRM 34:65-73, [6] Maeda, A, IEEE TMI 7:26-31, [7] Noll, D, IEEE TMI 10:629-37, [8] Man, LC, MRM 37:785-792, [9] Sutton, B, Proc ISMRM 01, p. 771, [10] Pruessmann, KP, MRM 46:638-51