Signal-to-Noise in TSE-imaging with Incomplete k-Space Coverage: Strategies and Implications for Low-SAR Imaging at High Fields

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Introduction

Data acquisition schemes where only part of the k-space data necessary for image reconstruction are acquired, are commonly used in order to speed up data acquisition. Especially for use at high field strength of 3T and higher such approaches offer the additional benefit to reduce the number of RF-pulses and thus the SAR(=specific absorption rate) applied to the patient. Of particular interest are reconstruction schemes, which at least in principle maintain the image resolution of the full data set. These include partial fourier reconstruction (1) and parallel reconstruction techniques like mSENSE (2) or GRAPPA (3). A common feature of these approaches is the expected loss of SNR with the square root of the reduction factor g, if additional losses by the reconstruction process are neglected and if all k-space data are contributing with equal weight to the final image. The latter condition is violated in TSE-imaging because of the T2-decay of signals along the echo train. The purpose of this paper is to investigate the influence of the inherently weighted sampling in TSE and its consequence on incomplete sampling schemes.

Methods.

For the ideal case and if all k-space samples are treated equally, then the amount of signal grows with the number k of k-lines acquired, whereas incoherent noise grows with the square root of k. For TSE the T2-weighting introduces an exponential attenuation term. SNR for a single point located at the center of k-space will thus be given by:

$$SNR = \sum_{1}^{k} S_0 \exp(-te/T_2)/(N_0\sqrt{k})$$

S_0 represents the spin-density signal without T2-weighting, N_0 is the noise of one k-line. Fig.1 shows the relative SNR as a function of k and the relative T2 scaled to the echo train length ETL. The SNR of the fully acquired image with $k = k_0$ is set to one. It is apparent, that the limiting case of a square root dependence is reached, when 1/T2=0 (**c** in Fig.1). For shorter T2 the SNR-loss will be gradually reduced, in the limiting case (T2=0 (**a** in Fig.1) all signal intensity will be represented by the first k-line, therefore SNR will increase with lower k. **b** represents a practical case, where ETL is corresponds to twice the T2 of tissue. In this case SNR will remain largely independent of the number of k-lines acquired at least up to a reduction factor of about 2. Fig.1 corresponds to a signal from a single point. For finite structures data will be even more focused onto the centre of k-space and thus the relative SNR will be even further increased compared to Fig.1. SNR will then also depend on the phase encoding sampling order.

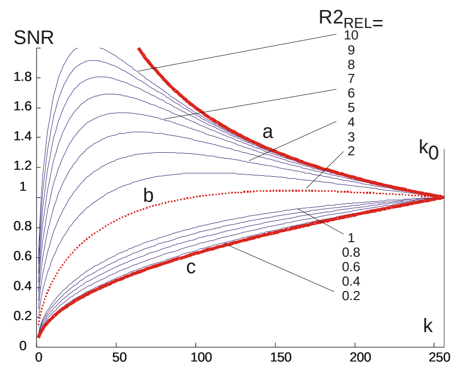

Fig.1 SNR as a function of $R2_{Rel}=1/T2_{Rel}$, where $T2_{Rel}=T2/ETL$.

Experimental

Experiments were performed on a 3T system (Siemens Magnetom Trio) with an 8-channel head receiver coil using a TSE-sequence with ETL=25. Incomplete sampling was accomplished in three different modes: Partial Fourier, where 8/8, 7/8, 6/8, 5/8 and 4/8 of the total k-space data were acquired, GRAPPA with a reduction factor of 2 and 4, where a variable part of fully sampled k-lines symmetrically around the centre was used, and mSENSE with identical parameters as for GRAPPA. Reduction factors were used to reduce the echo train length up to reduction factors of 2, for lower reduction factors the number of excitations was reduced. SNR was measured as usual as mean signal intensity over the standard deviation within a ROI in the background noise. In order to account for the variation of noise inherent to parallel reconstruction techniques, three noise ROIs were chosen: One each besides the object in the readout- and phase-encoding dirtection respectively and one at the corner of the image. Experiments were performed on phantoms as well as on volunteers. Tissue ROIs were placed in grey matter, white matter, CSF and lipid.

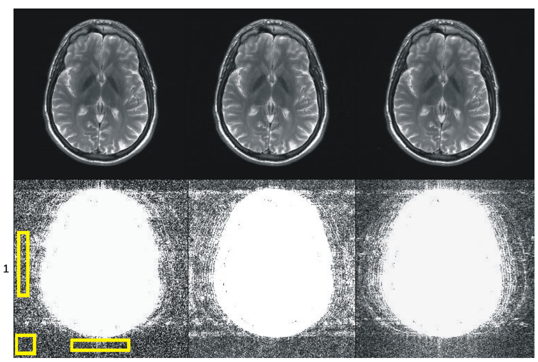

Fig.2 reference image (left), mSENSE (middle) and GRAPPA (right) images with reduction factor of 2 and full sampling of the center 76 of 256 lines. Grey scale in bottom images are inhanced by a factor of 100, yellow boxes correspond to noise ROIs used in the evaluation.

Results

Fig.2 shows images acquired with full sampling, mSENSE and GRAPPA with reduction factor of 2 and full sampling of the 76 lines at the centre. The noise enhanced images show the placement of the noise ROIs and illustrate the somewhat inhomogeneous background. Fig.3 shows the SNR as a function of the number of k-lines acquired. In spite of the considerable differences of T2 between white matter and CSF, the SNR dependence on k is virtually indistinguishable. The same applies to all other tissues. It is demonstrated, that for low k the SNR in parallel reconstruction modes becomes divergent due to the inhomogeneity of background noise. A reduction of SNR with lower k corresponding to Fig.1**c** could only be observed using a resolution phantom with a laminated structure of 1 pixel width filled with pure water.

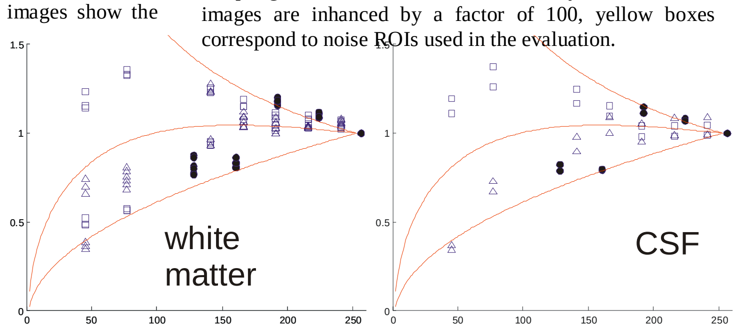

Fig.3 SNR vs. k with partial fourier(full circles), GRAPPA(squares) and mSENSE (triangles) reconstruction. The three red lines correspond to **a,b,c** from Fig.1

Conclusions

It is demonstrated, that incomplete k-space sampling techniques may even lead to an improved SNR. For moderate reduction factors partial Fourier works perfectly well. It would thus appear to be highly recommendable to acquire TSE-images especially at high fields with a combination of partial fourier and parallel imaging techniques leading to faster imaging at lower SAR at identical or even improved SNR compared to standard techniques. Combined with other SAR-saving techniques like asymmetric hyperechoes SAR-savings of a factor 10 appear to be feasible without any compromise (or even improved) image quality.

References: (1) Feinberg DA et al.Radiology 1986 161: 527. (2) Griswold MA et al MRM 2002; 47:1202 (3) Pruessman KP et al MRM1999 42:952 (4) Hennig et al MRM 2003; 49; 527