Applying Parallel Imaging for SNR Enhancement

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Introduction

In fast MRI (e.g. myocardial perfusion or real time imaging), excitation and gradient switching cover a large percentage of the repetition time. The measurements are usually carried out using a high readout bandwidth (BW) leading to a generally low signal-to-noise ratio (SNR) in the images. Purpose of this work was to optimize imaging protocols by utilizing parallel imaging techniques. By reducing the number of phase encoding steps N_{PE}, simultaneously minimizing the readout bandwidth, and reconstructing the undersampled k-space using parallel imaging techniques, the sampling time and hence the SNR can be significantly raised while the total image acquisition time is maintained.

Material and methods

For fast imaging techniques, the major part of the repetition time TR is required for slice excitation and gradient switching leaving only a short sampling time T_S = 1/BW for signal reception. The SNR being proportional to $\sqrt{N_{PE} \cdot T_S}$ is limited. Undersampling k-space and simultaneously conserving the image acquisition time T_{Acq} allows lengthening of TR and lowering the readout bandwidth. The signal reception time T_S can be disproportionally raised and hence, the SNR increased (Fig. 1). The images are reconstructed using parallel imaging techniques. During the reconstruction, a spatially variant noise enhancement, well known as geometry (g)-factor (1) is introduced. Lowering the SNR gain it has to be taken into account.

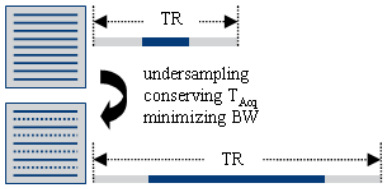

Fig. 1: *Basic concept. The sampling time (blue) can be significantly increased as the time consumed for gradient switching and excitation (gray) is mostly unaffected by lengthening TR.*

The SNR enhancement was investigated performing simulations, phantom and in-vivo measurements. All experiments were carried out on a 1,5T Siemens MAGNETOM Avanto system (Siemens Medical, Erlangen, Germany) using a 32 channel cardiac array (Rapid Biomedical, Rimpar, Germany) for signal reception, and a conventional saturation recovery SSFP sequence (FOV = 350x262 mm^2, matrix = 128x80, TR = 2,5 ms, TI = 110 ms, TE = 1.1 ms, thickness = 8 mm, alpha = 50°, BW = 1260Hz/px) for imaging. Image reconstruction was performed using GRAPPA (2).

According to prevalent acquisition schemes, a peripheral fraction of k-space was undersampled conserving a densely sampled central region (acs-lines) for autocalibrating the reconstruction. Several effective acceleration factors R_{eff} were considered by continuously increasing the peripheral fraction while conserving T_{Acq} and minimizing BW. The acceleration factors R = 2 and R = 3 where employed in the peripheral fraction and the central region was kept to a minimum of 24 acs-lines. Additional noise scans were performed and g-factor maps were calculated according to Breuer et al. (3).

The results were compared to the SNR gain obtained by homogeneously undersampling k-space by a factor of 2 (R_{eff} = 2) and calibrating the reconstruction using an additional reference scan.

In-vivo measurements were performed according to the above protocol using N_{PE} = 52, 24 acs lines, BW = 416Hz/px and R_{eff} = 1.5.

Results

Results of the phantom experiments are depicted in fig. 2. Displayed is the SNR gain obtained while continuously reducing the number of phase encoding steps from 80 (BW = 1260 Hz/px) to 52 (R_{eff} = 1.5, BW = 416 Hz/px) utilizing R = 2 in peripheral k-space, and to 43 (R_{eff} = 1.9, BW = 308 Hz/px) using R = 3. Corresponding g-factor maps are shown in fig. 3a and 3b. While for undersampling peripheral k-space by a factor of 2, reasonable g-factors and a moderate SNR gain of maximum 19% are examined, 3-fold undersampling of peripheral k-space leads to considerable noise enhancement and consequently to a loss of SNR.

However, for undersampling k-space homogeneously by a factor of 2 (R_{eff} = 2, BW = 287 Hz/px) and reconstructing the image using an additional full FOV reference scan, the noise enhancement is negligible with g-factors close to one (fig. 2c). Thus an effective SNR gain of 47% can be realized.

Results of an in-vivo myocardial first pass perfusion measurement are shown in fig. 4. Displayed is the pass of the contrast agent through the myocardium in 3 slices. The reconstruction could be performed without visible reconstruction artifacts.

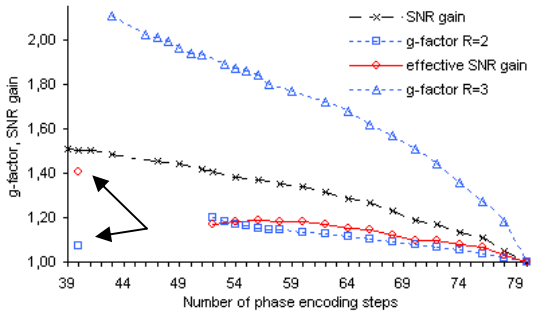

Fig. 2: *SNR gain and average g-factors for the described experimental setup. Shown are the maximum achievable SNR gain (x, chain dotted), the average g-factors of the image reconstructions for undersampling peripheral k-space by R = 2 (Δ, dotted) and R = 3 (□, dotted) and the resulting net SNR gain for R = 2 (o, solid). Not shown is the SNR loss for R = 3. Also depicted are the g-factor and effective SNR gain for homogeneously undersampling k-space by a factor of 2 (arrows).*

Conclusions

The SNR of fast imaging sequences can be improved using parallel acquisition techniques. A crucial investigation of the geometry factors is required, since noise is amplified by the parallel imaging reconstruction. For a given number of phase encoding steps, g-factors are minimized by homogeneously distributing the acquired lines over k-space. Hence in terms of SNR, the use of additional reference scans or techniques like Auto-SENSE (4), TSENSE (5) or TGRAPPA (6) is advantageous.

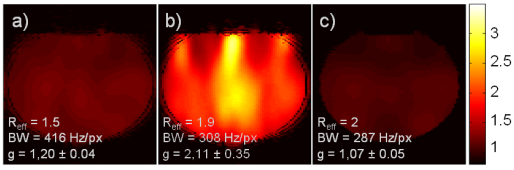

Fig. 3: *g-factor maps calculated for three different setups (description in the text). Average g values and the corresponding standard deviation are given for each map.*

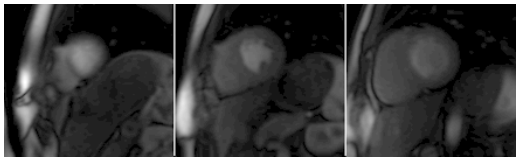

Fig. 4: *Images of an in-vivo myocardial first pass perfusion measurement. Shown is the pass of the contrast agent through the myocardium in three slices.*

References

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