

Cardiac CINE SSFP Imaging with a 32-Channel Cardiac Coil: Evaluation of SNR Performance for Parallel Imaging at 1.5T

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Introduction: The advent of parallel imaging with the development of SMASH [1] and SENSE [2] has led to tremendous reductions in scan time for a variety of rapid imaging applications including cardiac CINE imaging. Using the sensitivity profiles of the elements of a phased-array to allow under-sampling in the phase encoding direction facilitates reduction of the minimum scan time. Typical acceleration factors range from 2-3 for clinical applications. Acceleration of breath-held cardiac CINE imaging can improve temporal and spatial resolution, or increase the number of acquired slices.

Development of new coil technology with increasing numbers of channels is permitting increases in acceleration factors. Unfortunately, SNR decreases with increasing acceleration factors. For low acceleration factors, SNR decreases with the square root of scan time, however, with further increases in acceleration factors, additional degradation of SNR results from noise amplification, characterized with the geometry, or “g” factor [2]. Unlike non-accelerated acquisitions, where noise is present uniformly throughout the image, there is a strong spatial dependence of noise in accelerated images. Measurement of SNR in parallel imaging is challenging and not as straightforward as SNR measurements in non-accelerated images.

The purpose of this work is to evaluate the SNR performance of a prototype 32-channel cardiac coil, experimentally, in the context of cardiac CINE imaging at 1.5T using SSFP and TSENSE [3]. Practical approaches to measuring SNR performance through the experimental measurement of the g-factor with a “multiple acquisition” and “difference” method are described. The results are used to design a rapid cardiac CINE imaging protocol with high temporal/spatial resolution.

Theory: Images reconstructed from accelerated data sets have a strong spatial dependence on noise, due to spatially dependent ill-conditioning of the unwrapping matrix. This results in noise amplification characterized by the geometry, or “g”-factor. If temporal filtering is not performed for TSENSE, as was the case for our implementation of TSENSE, the g-factor can be then be written

$$g(r, R) = \text{SNR}_o(r) / (\sqrt{R} \text{ SNR}(r)) \quad (1)$$

providing a direct means for calculating the g-factor at position r , for a given acceleration, R .

A “multiple acquisition” method for the measurement of image SNR can be performed through the repeated acquisition of multiple images with identical scan parameters. For each pixel, the mean and standard deviation of the signal is calculated over time, such that $\text{SNR}(\mathbf{r}) = x_i(\mathbf{r}) / \sigma_i(\mathbf{r})$. This method provides an accurate measurement of local SNR on a pixel-by-pixel basis, however, it is time consuming and may be impractical for *in vivo* SNR measurement.

A “difference method” requires only two images (S_1, S_2), from which the local SNR can be determined in a small ROI using the sum and difference of S_1 and S_2 , ie:

$$\text{SNR}_{\text{ROI}} = \text{mean}(S_1 + S_2) / (\sqrt{2} \text{ std}(S_1 - S_2)) \quad (2)$$

where “mean” and “std” represent the mean and std. deviation of $(S_1 + S_2)$ and $(S_1 - S_2)$, respectively, in a small ROI.

Methods: All experiments were performed on a 1.5T scanner with a 32 independent receiver channel receiver array (Magnetom Avanto, Siemens, Erlangen, Germany). A 32-channel prototype cardiac phased array coil (RAPID Biomedical, Rimpfach, Germany) was used for all imaging. A phantom consisting of three plastic bottles of NiCl_2 (1.25g $\text{NiSO}_4 \cdot 6(\text{H}_2\text{O})$ per 1000g water) were positioned within the coil. Imaging was performed using a prospectively-triggered 2D-SSFP CINE pulse sequence with TSENSE. Phase encoding was oriented in both the left-right and up-down directions with $\text{FOV}_x=400\text{mm}$, $\text{FOV}_y=400\text{mm}$, slice=8mm, $N_x=192$, $N_y=192$, $\text{BW}=\pm 89.3\text{kHz}$, $\text{TR}=2.84\text{ms}$, $\text{TE}=1.42\text{ms}$, flip angle=50°, and segmentation=15, and temporal resolution of 42.6ms. The simulated RR interval was 852ms so that 20 non-interpolated phases were acquired. For each acceleration factor ($R=1$ to 7), the acquisition was repeated 11 times for a total of 220 images per acceleration factor.

Cardiac triggered SSFP CINE images were obtained in 1 volunteer with similar imaging parameters, such that spatial resolution=1.8 x 2.5 x 8mm³ and temporal resolution=48.2ms. Informed consent was obtained and the study was performed according to the guidelines of our IRB. A set of CINE images consisting of 3 short axis views, and 2-chamber, 3-chamber and 4-chamber views was acquired in a 20sec breath-hold using an acceleration factor of 4.

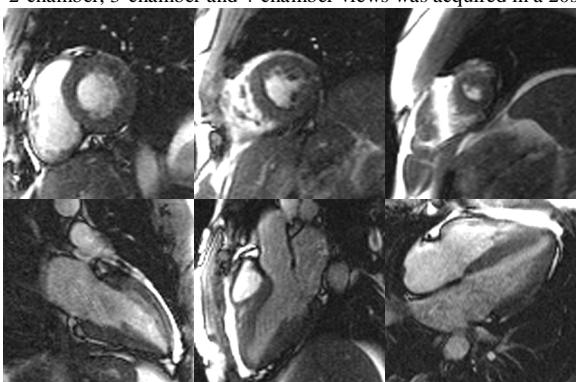


Figure 3: End-systolic CINE SSFP TSENSE images acquired in a 20sec breath-hold with an acceleration of four. Three short axis views (top row), and 2-chamber, 3-chamber, and 4-chamber views (bottom row, left to right).

easily acquired within one breath-hold, representing the acquisition of a complete wall motion study with high spatial and temporal resolution.

References: 1. Sodickson & Manning, MRM, 1997 2. Pruessman et al, MRM, 1999 3. Kellman et al, MRM, 2001

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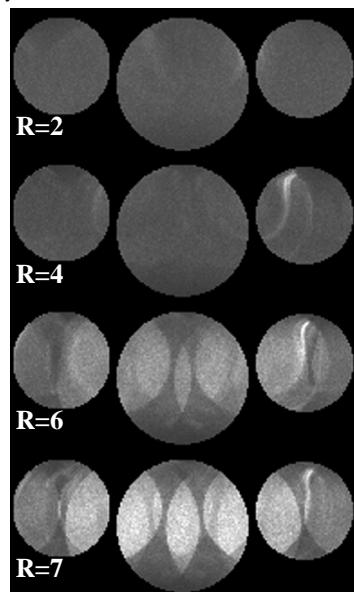


Figure 1: “g”-factor images calculated from images acquired with the multiple acquisition method.

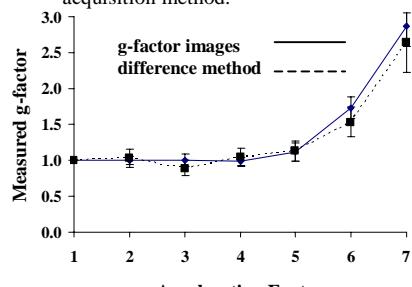


Figure 2: g-factor measured from ROI in g-factor images (fig. 1) and using the difference method.

Results: Fig. 1 shows calculated g-factor images, calculated using the multiple acquisition method, on a pixel-by-pixel basis (eq. 2), for the acquisition with phase encoding in the L-R direction. Fig. 2 plots the g-factor from a small ROI centered in the large bottle, using both the g-factor images (fig. 1) and the difference method. Fig. 3 shows end-systolic CINE images obtained in a single 20sec breath-hold covering 6 separate planes.

Discussion: In this work, we have described two practical experimental approaches for the evaluation of SNR performance of phased-array coils used for parallel imaging. Applied to SSFP CINE imaging with TSENSE to achieve accelerations of seven, very good agreement of measured SNR behavior between the multiple acquisition method and the difference method was observed. In addition, the expected qualitative appearance of g-factor images, with “wedges” and “arcs” that represent distinct regions with elevated noise amplification was also observed. Above factors of 4-5, the g-factor begins to degrade image quality, suggesting an optimal acceleration of 4-5 for clinical protocols using this coil for 1-dimensional accelerations.

Measurement of g-factor images was easily performed with the multiple acquisition method, permitting high resolution evaluation of the g-factor at different acceleration factors. For a phantom experiment, the acquisition of 220 images was relatively rapid, however, this approach is impractical for *in vivo* measurements. The difference method can also be used to make g-factor measurements in small regions over which it is assumed that the g-factor is relatively constant. This approach only requires two images, and could potentially be used for *in vivo* measurements.

With a one-dimensional acceleration factor of four, six separate CINE imaging planes were