Streak Artifact Suppression in Multi-coil MRI with Radial Sampling

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Introduction: Application of radial sampling to myocardial perfusion MRI [1] has been recently proposed to obtain accurate arterial input function (AIF) and tissue enhancement curves. In perfusion imaging it is desirable to cover significant spatial region with high spatial and temporal resolution. This can be achieved by acquiring only a fraction of the radial data required by Nyquist criterion [2]. It was found that artifacts in the perfusion images are mainly caused by streak artifacts in individual coil images and the inability of the sum-of-squares (SoS) algorithm [3] to suppress these artifacts in the combined image. The artifacts may hinder diagnosis by reducing the contrast between regions with normal and reduced perfusion and may make quantitative analysis of perfusion data erroneous due to systematic error (bias) in image intensity. In this abstract, a technique to reduce the artifacts and decrease intensity bias in MR images reconstructed from multi-coil radial data is presented.

Theory and Methods: An image reconstructed from incomplete data can be represented as a convolution of the true image with the PSF characterizing the sampling pattern:

$$J(r) = I(r) * PSF(r)$$

The PSF for radial sampling consists of the central peak surrounded by artifact-free circular region. Streak artifacts appear outside this region. The diameter of the artifact-free region is proportional to the number of radial views included in reconstruction. The PSF can be represented as a sum of delta function (ideal PSF) with a term responsible for the artifact due to incomplete sampling:

$$PSF(r) = \delta(r) + PSF_a(r)$$

A coil image reconstructed from incomplete data can be described by: $J_i(r) = I_i(r) + A_i(r)$, where $I_i(r)$ is the ideal coil image and $A_i(r) = I(r)S_i(r) * PSF_a(r)$ is an artifact in the image acquired by a coil with $S_i(r)$ sensitivity. In the case of radial sampling, streak artifacts in the individual coil images are coil specific. The most noticeable artifacts come out from the image region adjacent to the coil and spread through image FOV including image regions where coil sensitivity is close to zero. The number of radial views can be chosen such that artifacts are mainly outside the FOV defined by the coil sensitivity. When the SoS algorithm is used to combine the coil images the artifacts contribute to the SoS image in the same manner as the image features and without any attenuation:

$$J_{SoS}(r) = \sqrt{\sum |J_i(r)|^2} = \sqrt{\sum |I_i(r)|^2 + 2\operatorname{Re}\{I_i(r)A_i^*(r)\} + |A_i(r)|^2}$$

It is obvious that the SoS algorithm is not optimal in cases when coil images contain artifacts and that the artifacts create intensity bias in the SoS image. The SoS algorithm can be reformulated as a linear product of coil images $J_i(r)$ with weights $\alpha_i(r)$: $J_\alpha(r) = \sum \alpha_i^*(r) J_i(r)$ where $\alpha_i(r) = J_i(r)/J_{SoS}(r)$

In the artifact- and noise-free case, weight $\alpha_i(r)$ is equivalent to the following ratio: $\alpha_i(r) = S_i(r)/S_{SoS}(r)$ and $|\alpha_i(r)| \in [0,1]$. When such ideal weights are used to combine coil images, the artifact in each coil image is multiplied by the corresponding coil sensitivity resulting in artifact suppression. This effect is especially strong when high intensity artifact is located in the regions where the coil sensitivity is close to zero. Typically, coil sensitivities are unknown. Nevertheless, close approximation to ideal weight can be found using low resolution images $J_i^{LR}(r)$. Thus, image artifacts due to incomplete radial sampling can be significantly reduce if the coil images are combined according to

$$J_\alpha(r) = \sum \alpha_i^*(r) J_i(r) \text{ with } \alpha_i(r) = J_i^{LR}(r)/J_{SoS}^{LR}(r)$$

To test the proposed algorithm, myocardial perfusion studies were performed on a 3T Trio MR system (Siemens Medical Solutions, Erlangen, Germany) using a turbo-FLASH pulse sequence with saturation recovery magnetization preparation (TR/TE=1.92/0.97 ms, flip angle=12°, 96 projections with 192 readout points, FOV=340 mm, 8 mm slice thickness). A contrast agent bolus of 0.075 mmol/kg of Gd-DTPA was used. The data sampling scheme was implemented such that each subset of 24 time-adjacent projections covers 180 degrees. The sets of coil images were reconstructed from the data using 24, 48, and 96 projections. The coil images were combined using SoS or the new algorithm. Tissue enhancement curve (TEC) for myocardium and AIF were estimated for each reconstructed set of images.

Results: Fig. 1 shows that streak artifacts are substantially reduced in the images reconstructed by the new algorithm in comparison with the SoS images. Change in blood and myocardium intensity is due to the difference in the effective saturation recovery time (eSRT) for the images. Fig. 2a demonstrates that in the SoS images from small number of projections the myocardium signal can be dominated by the bias caused by high intensity streak artifacts from chest wall fat. Notice that the curve for eSRT=37 ms is higher that the curves for eSRT=57 and 103 ms. The bias is significantly smaller for the new reconstruction giving the expected appearance of curves for all eSRTs (Fig. 2b). It is well known that the bias in signal intensity results in underestimation of AIF and especially TEC. Fig. 3 shows that AIF and TEC has lower values for the SoS images from 24 and 48 projections than the corresponding curves from the images reconstructed by the new algorithm.

Discussion: A technique to reduce artifacts in the images reconstructed from multi-coil radial data has been developed. The technique is based on the following facts: 1) in individual coil image high intensity streak artifacts are mainly caused by the image region adjacent to the coil; 2) the artifacts spread through the image and may be dominant signal in the regions where the coil has low sensitivity. In such locations, contribution from the given coil to the SoS image is mainly artifactual. In our algorithm, such contributions to the combined image are strongly suppressed by weighting functions constructed from low resolution coil images. The proposed algorithm is computationally efficient and does not require an accurate knowledge of coil sensitivities which is typically required for parallel imaging techniques.

References: [1] Kholmovski EG, et al. Proc ISMRM 2005, p. 252. [2] Peters DC, et al. MRM 43:91-101. [3] Roemer PB, et al. MRM 1990;16:192-225.
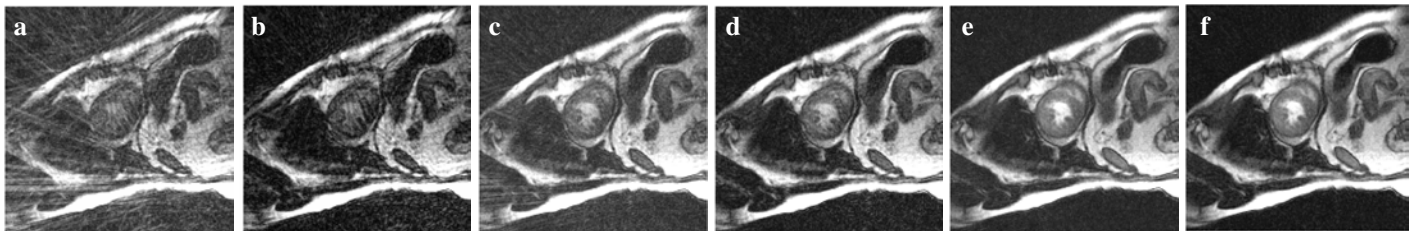
Figure 1. Comparison between SoS and the new algorithm. Images **a, c, e** were reconstructed using SoS. Images **b, d, f** using the new algorithm. The number of projections used in the reconstruction was: 24 for **a** and **b**, 48 for **c** and **d**, 96 for **e** and **f**. eSRT for **a** and **b** was 34 ms, for **c** and **d** - 57 ms, for **e** and **f** – 103 ms.
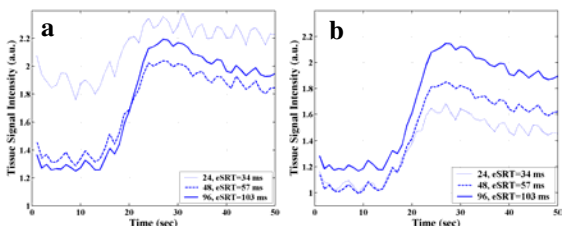

Figure 2. Mean signal intensity for myocardium. **a** from SoS images; **b** from the images reconstructed by the new algorithm.
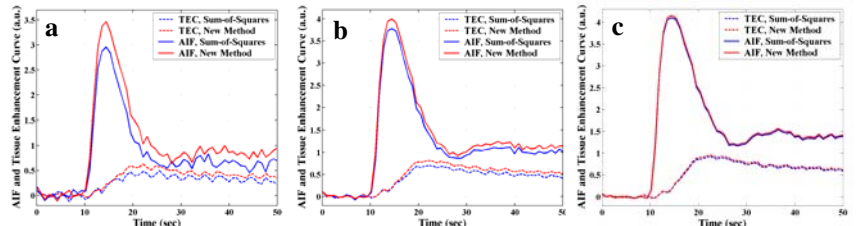
Figure 3. AIF and tissue enhancement curves. **a**: 24 projection, eSRT=34 ms; **b**: 48 projections, eSRT=57 ms; **c**: 96 projections, eSRT=103 ms.