Local Temporal Point Spread Function for CS Reconstructions Exploiting x-f-sparsity

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Introduction: Dynamic MRI can be successfully accelerated by exploiting spatio-temporal sparsity [1,2] where x-f-space (x = spatial domain, f = temporal frequency domain) is used as sparse transform domain. This enables undersampling k-t space and recovering the missing data using compressed sensing (CS, [2], see example in Fig. 1, beating heart). Remarkably, depending on algorithm parameters or sampling patterns, especially the higher harmonics in the frequency domain (see arrows in Fig. 1c) can be underestimated by the model-based CS technique due to their usually lower amplitude. This weighting affects the temporal resolution of the image series. However, a straightforward examination of this effect through one global temporal point spread function is not feasible since CS algorithms are neither linear nor stationary as previously discussed for spatial resolution behavior [3]. In this work, the local point spread function concept [4,5] already utilized to evaluate the spatial resolution of images reconstructed by CS [6] was applied to determine the temporal resolution of CS reconstructed x-f-sparse image series.

Method: A fully sampled in-vivo cardiac cine data set (Magnetom Trio 3.0 T, Siemens, Erlangen, TrueFisp, TR=2.9ms, TE=1.8ms, α=45°, slice thickness=8mm, FOV=380x297mm^2, matrix=256x208, #frames=20, temporal resolution=47ms) and a fully sampled in-vivo cardiac first-pass perfusion data set (Saturation Recovery Flash, TE=1.44ms, TR=2.5ms, TI=175ms, TA=323ms, α=12°, slice thickness=8mm, FOV=320x240mm^2, matrix=120x160, PPA=2, #frames=40) was acquired for two healthy volunteers. The data were retrospectively undersampled (acceleration factors R = 3/5 for cine/perfusion data) according to a random pattern with increased sampling density for central k-space data (Fig. 1b). The CS reconstructions were performed using iterative soft thresholding [7] with x-f-space as sparse transform domain. In order to assess the temporal resolution of the anatomical and the perfusion image series, the procedure described in [3] was applied to the temporal domain of the respective datasets: Pixelwise perturbations were set in one point of x-t-space. The reconstruction was repeated for a perturbation of 1% of the pixel amplitude at this location and additionally for 2% to verify the linear behavior of the reconstruction within the examined range [3,4]. The local temporal point spread function (LTPSF, Fig. 2) was respectively determined by subtracting the reconstruction of the unperturbed dataset from the reconstruction of the perturbed one. The width **w** of its main-lobe at 2/π ≈ 64% of the maximum amplitude was used as a measure of the local temporal resolution (Fig. 2). Maps indicating the blurring in every pixel were created by repeating the procedure for every location in x-t-space.

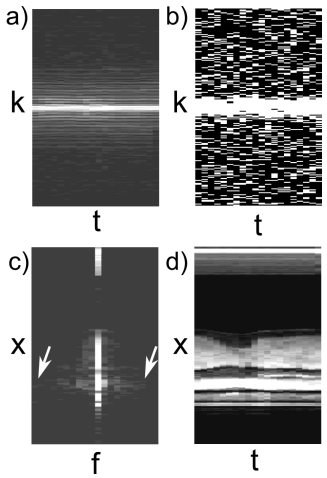

Figure 1: *CS exploiting x-f-sparsity: a) fully sampled k-t acquisition b) undersampling pattern with random variation between timeframes c) sparse x-f-space d) x-t-space of beating heart*

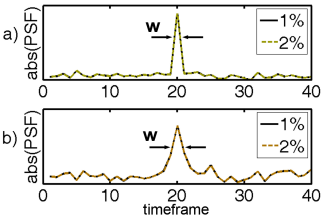

Figure 2: *Normalized LTPSF for single- and double-sized perturbation in timeframe 20 of the perfusion dataset. a) temporal resolution preserved (**w** = 1.0). b) temporal blurring (**w** > 1.0). Both perturbations result in identical LTPSFs indicating linear behavior*

Results: Figure 2 shows the LTPSF for the CS reconstructed perfusion data set at two locations. For each location, the normalized LTPSFs for both perturbations (1% and 2%) proved to be identical. This indicates linear behavior within the examined range and was observed for all investigated locations. The LTPSF in Fig. 2a does not show an acceleration based loss in temporal resolution i.e. the width of the main-lobe corresponds to a Nyquist acquisition (**w** = 1.0). The LTPSF in Fig. 2b features a clear broadening of the main-lobe and therefore a blurring of information across adjacent timeframes. Figure 3a displays the CS reconstruction for cine data (end-systolic timeframe) where only minor artifacts are visible at the lateral wall. In Fig. 3b, the CS reconstruction of the perfusion data set (timeframe prior to contrast agent enhancement in myocardium) is depicted. In Fig. 3c and 3d the images were overlayed by temporal resolution maps for the depicted timeframe representing the color-encoded local temporal resolution (width of the respective LTPSF **w**). Figure 3c points out a correlation between dynamic pixel amplitudes (motion) and degradation of resolution. The maximum broadening of the main-lobe of the LTPSF with respect to a Nyquist acquisition is 42%. The parts of the myocardium which exhibit only low or no fluctuations in amplitude throughout the whole series are not affected by temporal blurring.

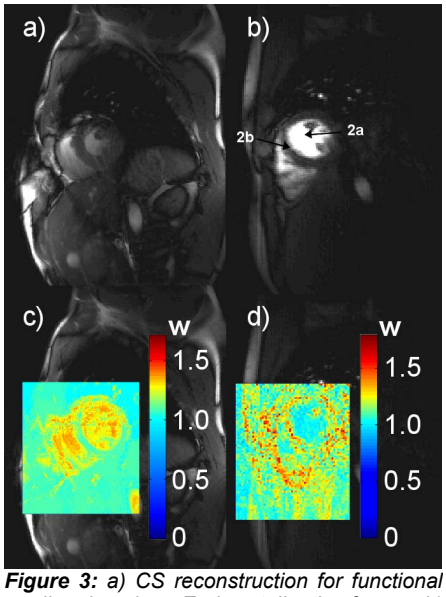

Figure 3: *a) CS reconstruction for functional cardiac imaging. End-systolic timeframe. b) CS reconstruction for perfusion imaging, arrows indicating locations for LTPSFs in Fig. 2 c) map indicating the amount of blurring in the functional image series d) according map for the perfusion image series*

The resolution map for the perfusion image series (Fig. 3d) shows an increase of **w** of up to 66%. Especially areas with low signal amplitudes, but dynamic throughout the time series (e.g. myocardium) suffer from broadening of the LTPSF.

Discussion and Conclusion: The recently presented local point spread function approach used to examine reconstruction induced blurring in the image domain [3,6] analogically can be utilized to examine the temporal resolution of images reconstructed by non-linear and non-stationary algorithms. While temporal smoothing is hardly detectable through a mere visual examination, color maps indicating the amount of blurring (Fig. 3c, d) enable an objective evaluation. Thus, the introduced method allows determining the influence of the choice of regularization parameters or sampling patterns on the temporal resolution and therefore an improvement of model-based reconstructions exploiting spatio-temporal sparsity.

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References: [1] Lustig et al. ISMRM 2420 (2006); [2] Otazo et al. MRM 64:767-76 (2010) [3] Wech et al. ISMRM 73 (2011) [4] Wilson et al. IEEE NSS-MIC 2:1189-93 (1993) [5] Fessler et al. IEEE IP 5(9):1346-58 (1996) [6] Wech et al. ISMRM 4885 (2010) [7] Stern et al. JMR 188:295-300 (2007)