Compressed sensing for flow-compensated intra-voxel incoherent motion modeling

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Introduction: The flow-compensated (FC) intra-voxel incoherent motion (IVIM) model has recently been proposed to investigate tissue microvasculature using diffusion weighted (DW) MRI [1]. However, it requires long acquisition times, thus limiting the clinical applicability of this approach. The work at hand proposes to use a compressed sensing (CS) [2] reconstruction exploiting sparsity in the Karhunen-Loève transform (KLT) domain to accelerate the acquisition of abdominal diffusion weighted MR images. Initial results show that FC IVIM parameter maps estimated from 3x accelerated simulations achieve similar quality to those estimated from the original non-accelerated data set.

Methods:

The IVIM model, originally introduced by Le Bihan [3] is a two-compartment model describing the attenuation of the diffusion weighted signal $S(b) = S(0)[e^{-bD}(1-f) + f \cdot F]$. In most of its applications the attenuation F of the perfusion fraction f is expressed using a pseudo-diffusion coefficient $D^* = \frac{\tau v^2}{6}$, i.e. $F \approx e^{-bD^*}$, where τ and v are the characteristic timescale and velocity of the incoherent motion. Since the measured signal attenuation using FC DW gradients indicates that this approximation is not necessarily valid, the FC IVIM model [1] goes a step further, allowing to estimate not only D^*, but both τ and v by measuring $S(b)$ for monopolar and FC DW gradient schemes of varying duration T. Assuming isotropy of the incoherent motion, it uses pre-calculated phase distributions to calculate F as a function of b, τ and v for a given timing of DW gradients, thus allowing one to determine the model parameters from a pixel-wise fit to the signal intensities averaged over directions.

The experimentally obtained data (referred to as 'original') consists of 300 abdominal magnitude images for a single slice, acquired from a healthy female volunteer using the acquisition scheme suggested in [1] with T=40/70/100ms, 8 b-values 10-500 s/mm² (for T=40ms b_{max}=100s/mm²) and 6 DW gradient directions. Images were acquired in expirational breath hold (TR=2.1s, TE=120ms, BW=2000 Hz/px, matrix 100x78, resolution 3.5mm, 7 slices with 5 mm thickness, GRAPPA 2, 1.5 T Siemens Magnetom Avanto).

An 'undersampled' data set was simulated by randomly discarding 4 of the 6 images (3x acceleration) belonging to different DW directions for each combination of b and T, denoted by the index dw_i (i=1…50). To separate the contributions from noise and pulsation from the actual DW signal $\boldsymbol{\sigma_p} = \left[\sigma_{p,dw_1}, \ldots, \sigma_{p,dw_{50}}\right]^T$ for a pixel p, the Karhunen-Loève transform Ψ^T was estimated, based on the observed signal magnitudes measured at the locations x_j of p and its 8 surrounding neighbors for the available number d of DW directions (j=1…9d). The transformation matrix Ψ is the matrix of the eigenvectors of the covariance matrix $C_p = \mathrm{cov}\left(\sigma_{x_j,dw_i}\right) = \sum_{l=1}^{50} \lambda_l \Psi_l \Psi_l^T$ of the observation matrix $\boldsymbol{\sigma}_{x_j}$ (eigenvalues λ). CS reconstruction given by $\min_x \|x\|_1$ s.t. $\|W^{-1}\Psi x - a\|_2 < \varepsilon$ was performed using SPGL1 [4]. Consistency with k-space data a was enforced applying the inverse Fourier transform W^{-1}, yielding a 'reconstructed' data set. The noise contribution ε was estimated from a low intensity region in k-space.

Results:

Parameters maps for D, f, τ, v and the residual r from fitting the FC IVIM model to the original (a), undersampled (b) and reconstructed (c) data sets are shown in Fig. 1. Since DW signal attenuation only deviates little from a mono-exponential decay in the spleen and the renal cortex (low f, white arrows), the value of the reconstruction for estimating τ and v must be judged regarding liver and pancreas (red arrows). Para-meter maps in (b) visually appear to contain more noise than the original set and show an increased residual of the fit in areas close to large blood vessels (yellow arrows, pulsation). CS reconstruction (c) reduces contributions from noise and pulsation, which manifests in lesser residuals compared to (b), but can also be inferred from evaluation of the microscopic parameters τ and v, which is shown for a region of interest (ROI) in the liver (Tab.1).

Conclusion: We were able to show that CS reconstruction exploiting sparsity in the KLT domain could accelerate the acquisition of the abdominal DW images 3x, without affecting the quality of the calculated FC IVIM model parameters D, f, τ and v.

	(a) original	(b) undersampled	(c) reconstructed
D (µm²/ms)	1.65 ± 0.39	1.58 ± 0.40	1.60 ± 0.28
f	0.31 ± 0.12	0.31 ± 0.12	0.31 ± 0.10
tau (ms)	143 ± 27	168 ± 132	145 ± 23
v (mm/s)	5.0 ± 2.4	4.3 ± 2.3	4.8 ± 2.3

Tab.1 Mean and standard deviation from evaluating a ROI in the liver for original, undersampled and reconstructed parameter maps (Fig.1).

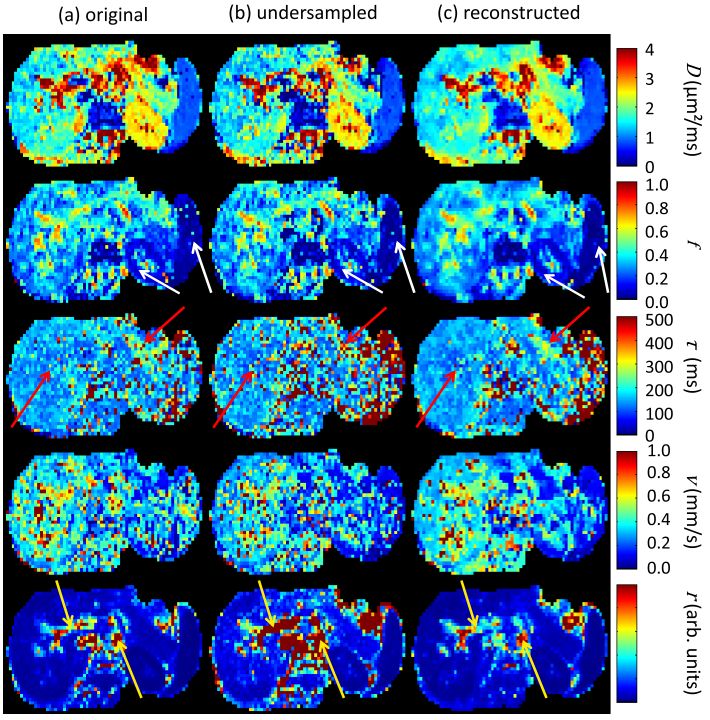

Fig.1 D, f, τ, v and residual maps obtained by fitting the FC IVIM model to the original (a), undersampled (b) and reconstructed (c) data sets.

References: [1] A. Wetscherek *et al.*, *Proc ISMRM 20th Annual Meeting*, p.2012 (2012); [2] D.L. Donoho, *IEEE T Inform Theory* **52**,4 (2006); [3] D. Le Bihan *et al.*, *Radiology* **168**,2 (1988); [4] E. van den Berg *et al.*, *SIAM J Sci Comput* **31**,2 (2008).