Time-Resolved Non-Contrast Fresh Blood Imaging MRA Using Compressed Sensing Reconstruction

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Target Audience: Researchers and clinicians who are interested in non-contrast enhanced MRA.

Introduction: Fresh blood imaging (FBI), as an ECG-gated non-contrast MR angiography technique[1], enables the visualization of peripheral arterials by utilizing the physiological signal differences between two cardiac phases: systolic (fast flow) and diastolic (slow flow) phases. It is desirable to depict the vascular inflow effects and to obtain more insightful information on vascular physiology by the time-resolved (multiple phases) FBI technique. To achieve a reasonable data acquisition time and also to reduce subject motion artifact in dynamic FBI acquisitions, it is necessary to apply accelerated imaging techniques. Compress sensing has been shown to be especially useful to achieve high acceleration factors in MRA applications due to their inherent sparsity[2]. In this work, we demonstrate the feasibility of an accelerated time-resolved non-contrast FBI MRA technique using compressed sensing, besides, the image reconstruction time was significantly reduced using coil compression.

Methods and Materials: 4D dynamic FBI MRA data of the calves was acquired in a healthy volunteer (71 years old, male) on a 3T clinical whole-body MR scanner (Toshiba Vantage Titan™ 3T, Otawara, Japan) using a 16-element flexible coil array. Datasets at 25 different triggering delays (25 phases) away from the systolic phase were obtained in a step of 20 ms. The data size was 160 (PE) × 256 (RO) × 10 (SE) × 25 (phase) × 16 (coil number). The FBI acquisition parameters were as follows: 3D half-Fourier FASE readout (fast asymmetric spin echo), number of acquired PE lines = 112, TR = 3 RR intervals, TE_{eff} = 80 ms, ETS = 5ms, flip/refocusing angle = 90°/160°, in-plane resolution = 2.3×1.4 mm^2, slice thickness = 5 mm, the acquisition time at each trigging delay = 1:09 minutes, resulting a total acquisition time of 28:45 minutes.

In the reconstruction procedure, we first subtract the data from the first time point, i.e., the systolic phase, in k-space to obtain a vessel-only representation without background signal. Second, in order to reduce the computation time significantly, a coil compression method based on geometric decomposition of the data [3] was utilized to compress the data from 16 channels to 6 virtual coils. For the compressed sensing reconstruction, the undersampling patterns by factors of 4 and 6 in k_y-k_z plane are shown in Fig. 1. Image reconstruction was performed in MATLAB by solving the optimized equation, which combines the spatial and temporal sparsities, $\hat{m} = arg\min_m\{||F_u m - y||_2 + \lambda_1||\Psi_1 m||_1 + \lambda_2||\Psi_2 \frac{\partial m}{\partial t}||_1\}$, where the left-hand term enforces data consistency with the acquired data *y*. F_u is the Fourier transform including the undersampling and half Fourier masks. The second term and the third term enforce spatial and temporal sparsity, respectively. Ψ_1 and Ψ_2 are their corresponding sparsifying transforms.

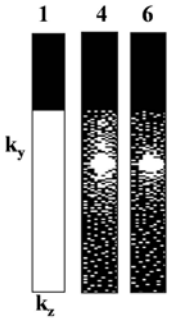

Fig. 1: ky-kz sampling pattern with 1, 4 and 6-fold accelerations (white: sampled positions)

Results and Discussions: Fig. 2 shows one example of the MIP image from the original 16-channel data (left), the MIP result after coil-compression with 6 virtual coils (middle), and the compression error image amplified by 10 (right). Very little compression loss with 6 virtual coils was observed. The reconstruction time (2 hours and 10 minutes, using a standard PC: Intel single CPU, 3.30 GHz, 16 GB RAM) with 6 virtual coils was reduced to half of the computation time using the original 16-channel data set. Fig. 3 displays and compares the reconstructed MIPs for each undersampling rate at triggering delay of 400 ms from the systolic phase. Fig. 4 is the intensity curves of the blood vessel ROI as indicated by the red circle in Fig. 3. The compressed sensing results with 4- and 6-fold undersampling rates exhibited similar image quality and preserved similar dynamic characteristics of inflow effects to the fully-sampled data. An acceleration factor of 10 was also tested but the branch vessels were suppressed significantly (results not shown here). Note that not many small vessel branches were depicted even in the fully-sampled data due to relatively low through-slice resolution and subject motion during the long acquisition period (~1 min for the fully-sampled data for each delay and ~29 mins for the total acquisition time). This emphasizes the importance of the application of imaging acceleration in time-resolved FBI acquisition.

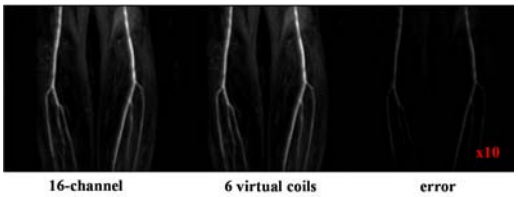

Fig. 2: comparison of MIP images with original 16-channel dataset (left) and 6 virtual coils (middle). The difference image (amplified by 10 times) shows negligible compression loss.

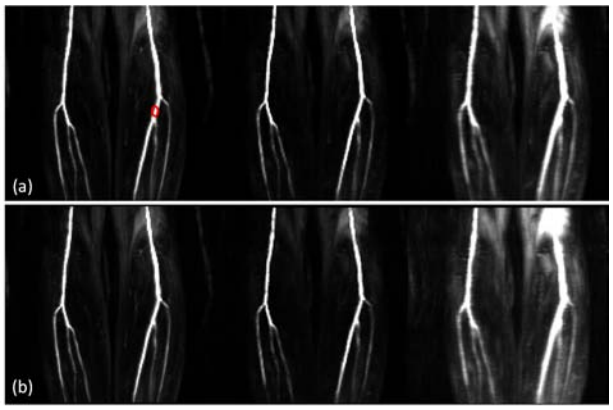

Fig. 3: Reconstructed MIPs at triggering delay of 400 ms from the systolic phase. (a) fully-sampled (left), CS reconstructed image (middle) with 4-fold undersampling rate, and zero-filling reconstruction (right). (b) 6-fold undersampling rate. The red circle indicates the ROI whose signal was plotted in Fig. 4.

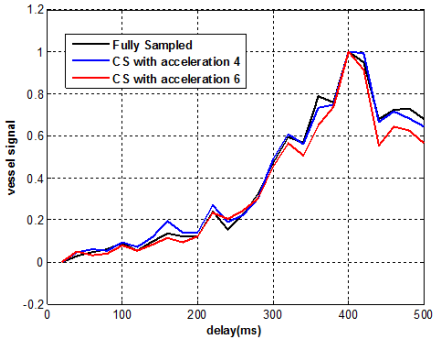

Fig. 4: Comparison of intensity curves among fully sampled reconstruction (black), CS reconstruction with acceleration factor of 4 (blue) and acceleration factor of 6 (red). The red circle in Fig. 3(a) indicates the ROI for this plot.

Conclusions: Reasonable acceleration (4-6, scan time from ~29 minutes down to ~5 minutes) of time-resolved FBI MRA has demonstrated to be feasible by exploiting both the spatial and temporal sparsities using compressed sensing. The computation time can be reduced by applying coil compression algorithms. The total scan time and sensitivity to subject motion artifact decrease with acceleration factors, however, the capability to preserve small vessels decreases with acceleration factors as well, therefore, care must be taken when selecting the appropriate acceleration rate. Clinically, the accelerated dynamic MRA may be helpful for patients with bifurcation diseases, for instance, this technique can gradually show the vessel signal variation at different cardiac phase, and may help identify subtle intensity changes caused by mild stenosis.

References: 1. Miyazaki et al., Radiology, 2008; 2. Lustig et al., MRM 2006; 3. Zhang et al., MRM 2013.