Combination of Parallel Imaging and a Cut-Corner Acquisition for Neurovascular 4D-flow

T. A. Hope[1], M. D. Hope[2], R. Bammer[3], and M. T. Alley[3]

[1]Department of Medicine, Kaiser San Francisco, San Francisco, CA, United States, [2]Department of Radiology, University of California at San Francisco, San Francisco, CA, [3]Department of Radiology, Stanford University, Stanford, CA

INTRODUCTION: Time-resolved three-dimensional phase-contrast MRI (4D-flow) is a tool for imaging blood flow velocity profiles, but is limited in its clinical applications by its long scan time. In this study, we analyze the effect of the combination of a parallel imaging technique (GRAPPA) with an acquisition that removes the corners of k-space in order to further limit scan time. A previous study has shown that average velocity calculations are preserved when combining GRAPPA with the 4D-flow acquisition (1). We analyzed the effect on both mean velocity across the vessel lumen as well as the difference between individual velocity vectors (Figure 1), as individual vectors may effect calculations of wall shear stress and other variables that depend on the local velocity field (2).

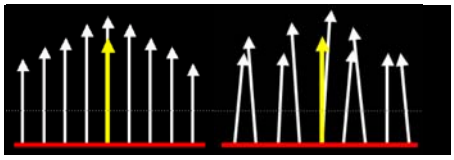

Figure 1: example of the effect of noise. With noise added on the right, average velocity (yellow arrow) may remain unchanged, but individual vectors may differ.

MATERIALS AND METHODS: Time-resolved 3D PC MRI was used to assess neuro-vascular flow patterns in 3 patients at 1.5T and 3.0T (both scanners: Signa LX CNV, 12.0; GE Healthcare, Milwaukee, WI, USA), which has been previously described and validated (1, 3). High-performance gradient systems were used with a maximum gradient strength of 50 mT/m at 1.5T and 40 mT/m at 3.0, and a maximum slew rate of 150 mTs/m. An eight-channel head coil was used (MRI devices, Waukesha, WI, USA), and pulse oximetry was used for cardiac gating. The same parameters were used on both scanners: field of view 180 x 180 mm, 90% rFOV, matrix size 300 x 180 x 30, TR/TE 5.46ms/2.12ms, flip angle 15°, venc = 100 cm/s, and receiver bandwidth of ±65 kHz. Frequency encoding was performed in the anterior-posterior direction. Three slice encodes were repeated in each RR interval, resulting in a temporal resolution of 65.5 msec. Therefore, 1,620 heartbeats were needed to acquire a data set: (PE_x x PE_y x rFOV) / (slice encode repeats per RR), or 180 x 30 x 90% / 3. As previously described, the six fully acquired data sets were subsampled using a variant of GRAPPA with outer reduction factors (ORFs) of 2 and 3 (1, 4). Additionally, the resulting 18 data sets were further subsampled by removing the data in the corners of the ky-kz plane and then zero-filling. The area of the remaining k-space data in this "cut-corner" acquisition is defined by the ellipse $\pi\, k_{zmax}\, k_{ymax}$ (5). To analyze the effect of subsampling, planes were extracted from the left internal carotid artery, the right internal carotid artery and the posterior cerebral artery utilizing 3D visualization software (EnSight, CEI Inc, Apex, NC). The plane from the full acquisition was segmented based on magnitude data, and using the same segmentations on all subsampled data sets, mean velocity across the vessel lumen was calculated using a proprietary program (Aspire2, Stanford, CA). Additionally, individual velocity vectors were compared between the fully acquired data sets and each of the subsampled data sets. The mean of the percent absolute difference between individual velocity vectors from the center of the vessel lumen was calculated. To qualitatively look at the effect of subsampling, streamlines were used to visualize data.

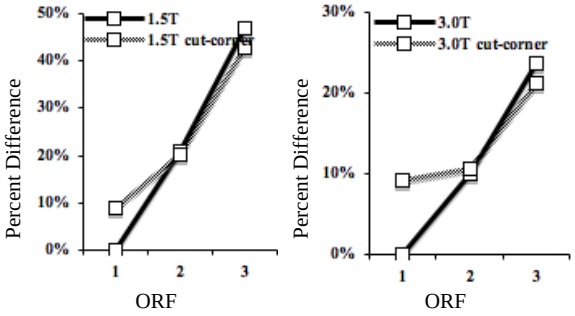

Figure 2: Mean difference between individual velocity vectors between the fully acquired scan and the five subsampled scans at peak systole averaged from the right, left and posterior cerebral arteries at 1.5T (left) and 3T (right). Note that the cut-corner acquisition has roughly the same measurement error when compared to ORFs of 2 and 3.

RESULTS: For ORFs of 2 and 3, the total number of heartbeats was reduced from 1,620 to 920 (a 43% reduction) and 700 (a 57% reduction). The cut-corner acquisition resulted in a further decrease to 1,222 (a 24% reduction from the fully acquired scan), 724 (a 21% reduction from the ORF 2 scan) and 559 (20% reduction from the ORF 3 scan). Differences in mean velocity measurements across the vessel lumen between subsampled data sets were clinically insignificant (all less than 3%). Figure 2 shows the effect on the mean percent difference between individual velocity vectors at each field strength with ORFs of 1, 2 and 3 as well as with and without the cut-corner acquisition. The mean difference between individual vectors was roughly ten times the difference in the mean velocity across the vessel lumen. The addition of the cut-corner acquisition to the parallel imaging technique changed the measured differences minimally. Figure 3 shows example streamlines at each ORFs, with and without the cut-corner acquisition at 3T, showing the degradation of streamlines as k-space is subsampled.

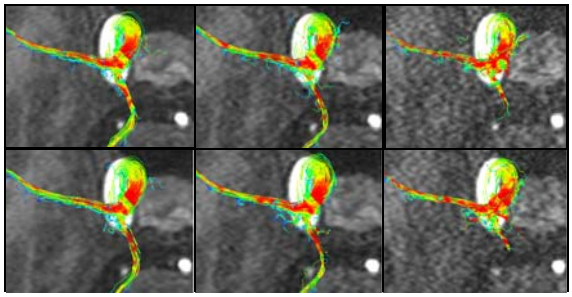

Figure 3: streamlines of the ICA splitting into the left middle cerebral artery and the left posterior communicating artery (column 1 = ORF 1, 2 = ORF 2, and 3 = ORF 3). Top row: fully acquired data; Bottom row: acquisitions with corners cut

CONCLUSION: A limitation of 4D-flow in the clinical setting is lengthy scan time. In an effort to reduce scan time, we studied the effects of two k-space subsampling techniques on velocity data. As previously demonstrated, GRAPPA significantly reduces scan time, but at low field strengths, higher reduction factors result in significant image and velocity degradation. The addition of a cut-corner acquisition to GRAPPA resulted in minimal degradation of velocity data, while decreasing scan time by up to 25%. This can reduce the overall scan time from 20 minutes to 9 minutes when combining GRAPPA with an ORF of 2 and the cut-corner technique. We also have shown that as stochastic noise increases with k-space subsampling, there is preservation of mean velocity, but the error of individual velocity vectors increases. In clinical situations where calculation of mean or peak velocities is important, such as evaluation of arterial stenoses, k-space subsampling is appropriate. However, in situations where the integrity of the local velocity field is needed, such as 4D visualization techniques like streamlines or calculation of wall shear stress, a full acquisition may be more appropriate.

REFERENCES: [1] Bammer R, et al. Time-resolved 3D quantitative flow MRI of the major intracranial vessels: initial experience and comparative evaluation at 1.5T and 3.0T in combination with parallel imaging. Magn Reson Med 2007; 57:127-140. [2] Stalder AF, et al. Advanced Quantitative Flow and Wall Shear Stress Analysis of 3T: 2D Vs 3D time-resolved MR Velocity Mapping. In:International Society for Magnetic Resonance in Medicine. Berlin, Germany, 2007. [3] Markl M, et al. Time-resolved three-dimensional phase-contrast MRI. J Magn Reson Imaging 2003; 17:499-506. [4] Griswold MA, et al. Generalized autocalibrating partially parallel acquisitions (GRAPPA). Magn Reson Med 2002; 47:1202-1210. [5] Bernstein MA, et al. Effect of windowing and zero-filled reconstruction of MRI data on spatial resolution and acquisition strategy. J Magn Reson Imaging 2001; 14:270-280.