# Temporal Filter Design for Time-resolved VIPR Using an Iterative Density Compensation Algorithm

J. Liu[1], A. Lu[1], A. Alexander[1], J. G. Pipe[2], E. K. Brodsky[1], D. Seeber[3], T. M. Grist[1], W. F. Block[1]

[1]Univ. of Wisconsin-Madison, Madison, WI, United States, [2]Barrow Neurological Institute, Phoenix, AZ, United States, [3]IGC Medical Advances Inc, Milwaukee, WI, United States

## INTRODUCTION

The sampling density in 3DPR, which varies as the inverse square of the k-space radius $k_r$, allows a sliding temporal filter to create time-resolved imaging volumes for Vastly undersampled Isotropic Imaging Projection (VIPR) [1]. Individual time frame images exhibit a significant decrease in image quality and CNR relative to unfiltered reconstructions that use all of the k-space data. The temporal filter design, which determines the tradeoff between temporal resolution and SNR, has been empirically chosen to date. To improve the image quality of the time-resolved reconstruction, a data weighting filter is implemented using a k-space density compensation model. An iterative technique originally developed to weight redundantly acquired data in PROPELLER exams is employed for this purpose [2]. We demonstrate significant increases in image quality and vessel to background contrast in contrast-enhanced MR angiography (CE-MRA).

## MATERIALS AND METHODS

The VIPR projections are interleaved to sub-sample orientations throughout a k-space sphere every 2 seconds. The original density compensation for the upper k-space hemisphere, shown in Fig. 1 for the sixth time frame in a fifteen frame exam, is plotted as an image in gray scale with acquisition time progressing horizontally and the k-space radius plotted vertically. The lower hemisphere is symmetric. This method uses data at low spatial frequencies only from a single time frame and then widens linearly with the k-space radius. Here the density weighting is $k_r^2$/(filter width at $k_r$) to account for the variation in filter width with spatial frequency.

In the iterative density compensation technique, the overall density weighting is first determined without temporal weighting. An initial weight is calculated as $W_0=S/(S\otimes C)$ where S comprises a set of delta functions at the sampled k-space points, $\otimes$ denotes convolution, and C denotes a kernel function. The iterative solution $W_{i+1}=W_i /(W_i\otimes C)$ aims to minimize the error in the point spread function by ensuring that the convolution of the kernel with the weighting function will be unity over sampled k-space. The kernel width can be selected to minimize mean square error when undersampling, at some loss of resolution, or to maintain full resolution, as was chosen here [3]. We refer to the solution without temporal weighting as $W_{nt}$. For each time frame we design a temporal weighting function F that indicates the contribution of time frames. To weight the redundantly sampled data preferentially according to the time function F, we calculate $W_{k+1}=W_k /((W_kF) \otimes C)$ for a few iterations, using $W_{nt}$ as the initial weight. The final time-resolved density weighting, $W_{tr}$ , is $W_{k+1}F$. A sample time function F is shown in Fig. 2 (for the sixth time frame) with the final density correction $W_{tr}$ shown in Fig. 3.

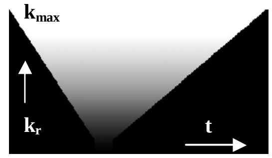


Fig. 1 Original Filter

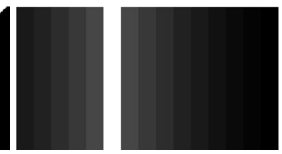

Fig. 2 Time func. F

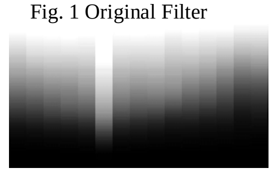

Fig. 3 Iterative filter

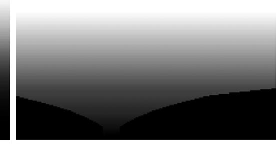

Fig. 4 Approx. filter

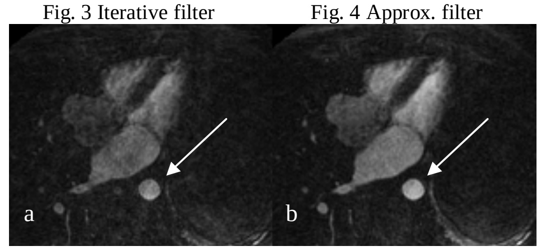


Fig. 5 Axial slice reconstructed with a) original filter, b) iterative filter. Notice improved uniformity in heart chambers in b) and aorta (arrows)

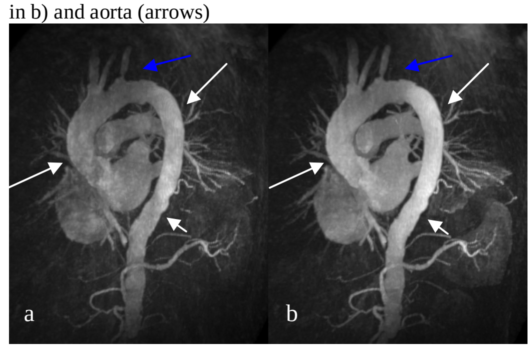


Fig. 6 Coronal oblique MIP reconstructed with a) original filter, b) iterative filter. White arrows show regions of decreased undersampling artifact in b). Vertebral artery stenosis shown by blue arrow.

## RESULTS AND DISCUSSION

All data were acquired on a 1.5 T CVi Signa™ scanner (GE Medical Systems, Milwaukee, WI) with a IGC Medical Advances 12-channel MRA coil. The weighting function $W_{nt}$ was computed with twenty iterations. The time-resolved compensation function $W_{tr}$ was tested on a 30 s CE-MRA exam covering a 44 cm spherical FOV with 1.7 mm isotropic resolution using 40 ml of Omniscan contrast. Figs. 5a and b show axial slices through the ventricles reconstructed using the original filter (Fig. 1) and the new iterative filter (Fig. 3) respectively, with corresponding oblique MIP images shown in Figs. 6a and b. Significant reductions in the undersampling artifacts over the major vessels and chambers are evident, notably as improve uniformity in the ascending aorta, descending aorta, pulmonary arteries and pulmonary veins. The uniformity, measured by the mean/standard dev. over a ROI, is consistently improved by approximately 30% when measured in these areas. The temporal resolution is slightly decreased, however the appearance of some background organs can be useful as a landmark during diagnosis. Vessel to background contrast is also improved.

The algorithm aims to weigh redundant data by the weighting function F while weighting the undersampled data equally. The filter shape agrees with the Nyquist equation for 3D PR, where the number of required projections increases as the square of the readout resolution. An approximate filter design, where the width varies as in Fig. 4, provided similar results while avoiding iterative computation.

## CONCLUSIONS

The temporal filter design for 3DPR imaging has been viewed as a density compensation problem and solved with an iterative technique. Vessel uniformity and contrast in CE-MRA exams are significantly improved through filtering the oversampled data while retaining the undersampled data. Full spatial resolution is retained with a slight compromise in temporal resolution.

## REFERENCES

**1.** A.V. Barger, *et al*., *MRM*, **48**, 297-305, 2002. **2.** J.G. Pipe, *MRM*, **42**(5), 963-969, 1999. **3.** J.G. Pipe, *et al*., *MRM*, **43**(6), 867-875, 2000.